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Sensitivity of Biomechanical Outcomes to Independent Variations of Hindfoot and Forefoot Stiffness in Foot Prostheses

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Abstract

Many studies have reported the effects of different foot prostheses on gait, but most results cannot be generalized because the prostheses' properties are seldom reported. We varied hindfoot and forefoot stiffness in an experimental foot prosthesis, in increments of 15 N/mm, and tested the parametric effects of these variations on treadmill walking in unilateral transtibial amputees, at speeds from 0.7 to 1.5 m/s. We computed outcomes such as prosthesis energy return, center of mass (COM) mechanics, ground reaction forces, and joint mechanics, and computed their sensitivity to component stiffness. A stiffer hindfoot led to reduced prosthesis energy return, increased ground reaction force (GRF) loading rate, and greater stance-phase knee flexion and knee extensor moment. A stiffer forefoot resulted in reduced prosthetic-side ankle push-off and COM push-off work, and increased knee extension and knee flexor moment in late stance. The sensitivity parameters obtained from these tests may be useful in clinical prescription and further research into compensatory mechanisms of joint function.

Keywords

foot prosthesis; lower limb amputation; gait; forefoot stiffness; hindfoot stiffness; gait asymmetry

1. Introduction

Lower limb amputees have access to a large selection of commercially available prosthetic feet, with designs that allege to provide greater function and comfort to the user. Though many studies have compared the effects of different prosthetic foot designs on gait function and limb comfort, little is known about the direct relationship between the mechanical properties of these devices and the biomechanics of gait. Mechanical properties such as foot stiffness can be very important to the user, affecting daily walking function and quality of

life. Users have been reported to prefer a compliant hindfoot (Lehmann, Price, Boswell-Bessette, Dralle, & Questad, 1993a; Raschke et al., 2015) and a forefoot of low to moderate stiffness (Klodd, Hansen, Fatone, & Edwards, 2010a; Raschke et al., 2015), for reasons such as impact absorption and a general feeling of “springiness.” These observations generally do not translate into energy savings (Barth, Schumacher, & Thomas, 1992; Hafner, Sanders, Czerniecki, & Ferguson, 2002; Klodd et al., 2010a; Schmalz, Blumentritt, & Jarasch, 2002; Torburn, Powers, Guitierrez, & Perry, 1995), but they can affect both prosthetic-side and intact-side kinematics and kinetics (Fey, Klute, & Neptune, 2011; Klodd, Hansen, Fatone, & Edwards, 2010b; Lehmann, Price, Boswell-Bessette, Dralle, Questad, et al., 1993; Lehmann, Price, Boswell-Bessette, Dralle, & Questad, 1993b; Major, Twiste, Kenney, & Howard, 2014; Morgenroth et al., 2011; Perry, Boyd, Rao, & Mulroy, 1997; Segal et al., 2012). Unfortunately, it is rare for biomechanical results to be accompanied by a quantitative description of prosthesis properties, so most results provide little information about the parametric sensitivity of outcomes.

A few researchers have provided some characterization of prosthetic foot properties independent of their human users. Several have characterized stiffness of the forefoot and hindfoot (Geil, 2001; Klute, Berge, & Segal, 2004; Miller & Childress, 1997; Van Jaarsveld, Grootenboer, Vries, & Koopman, 1990). These ideas have grown into a standardized test protocol for classifying prostheses into different categories of functionality, for use in prescription and billing (AOPA, 2010). Other studies have characterized an effective roll-over shape under some nominal loading scenario (Curtze et al., 2009; Hansen, Childress, & Knox, 2000; Sam, Hansen, & Childress, 2004). A recent effort proposed to combine several of these approaches in a comprehensive model (Major, Twiste, Kenney, & Howard, 2011), including normal stiffness, shear stiffness, and normal damping ratios. More recently, a method was presented for characterizing linear stiffness and angular stiffness using simple mechanical tests (Adamczyk, Roland, & Hahn, 2013). All of these approaches are objective, but none have reported typical biomechanical data alongside the mechanical characterization.

A few notable studies have combined objective properties with human testing. One recent study reported the effects of varying angular stiffness, with specified values for dorsiflexion and plantarflexion (Major et al., 2014). Others have varied forefoot stiffness (Klodd et al., 2010a, 2010b) or whole-foot stiffness (Fey et al., 2011; Ventura, Klute, & Neptune, 2011) experimentally, or reported the measured stiffness of commercial devices (Perry et al., 1997). A recent double-blinded study reported stiffness in three broad categories, alongside prosthetic socket moment data and subjective user preferences (Raschke et al., 2015). And, another recent effort used a robotic prosthesis to test the effects of different emulated prosthesis properties on gait in individual amputees (Caputo, Adamczyk, & Collins, 2015; Caputo, Collins, & Adamczyk, 2014). These efforts appear to be converging on a common goal of understanding the influence of each parameter on biomechanical gait outcomes in amputees. One goal of the present study was to combine objective mechanical prosthesis properties with biomechanical human testing to experimentally estimate the sensitivity of gait outcomes to forefoot and hindfoot stiffness of a prosthetic foot, thus providing clinical interpretive value to the associated observations.

Some possible effects of mechanical property changes can be predicted from general principles of human locomotion. Assuming gravitational force to be constant across walking conditions, the relatively static mid-stance phases of gait should be dominated by the force of body weight. If the loading force remains unchanged, then a higher stiffness component will result in lower deflection, reducing total energy stored and subsequently returned (for a given force F and constant stiffness K , energy $E = F^2/(2K)$). If the energy returned from the prosthetic foot is assumed to perform push-off work on the body center of mass (COM), principles of dynamic walking suggest that prosthetic feet with higher energy return could lead to reduced collision at heel strike on the intact side (Donelan, Kram, & Kuo, 2002a; Zelik et al., 2011). It follows that forefoot components with greater energy absorption/return (i.e., lower stiffness) may result in greater COM push-off work and lead to a reduced collision on the intact side (Agrawal, Gailey, O'Toole, Gaunaud, & Finnieston, 2013; Morgenroth et al., 2011; Segal et al., 2012). These concepts have successfully predicted important features of gait in persons with lower limb amputation compared to those without, including asymmetrical center-of-mass mechanics, increased intact-limb loading, and increased work requirements (Adamczyk & Kuo, 2015; Houdijk, Pollmann, Groenewold, Wiggerts, & Polomski, 2009), though the effects also depend on push-off timing (Malcolm, Quesada, Caputo, & Collins, 2015; Ruina, Bertram, & Srinivasan, 2005; Zelik et al., 2011). Additionally, these differences are strongly influenced by walking speed, with various outcomes increasing, decreasing or staying constant across speeds (Adamczyk & Kuo, 2015).

Additional predictions can be made at the level of individual joints. Subject to the requirement to counteract gravity, it may be assumed that one of the body's secondary goals is kinematic invariance (Selles et al., 2004; Srinivasan, Westervelt, & Hansen, 2009), wherein the motion of segments and joints remains relatively constant across walking conditions. Applied to the prosthetic foot, this principle suggests that if foot impact kinematics were constant, then a stiffer hindfoot would lead to greater ground reaction force earlier in stance, thus increasing external knee flexor moment in early stance. Similarly, a stiffer forefoot should increase the external forefoot load more quickly in mid-stance, resulting in an earlier and greater external knee moment acting to extend the knee. Additional changes at the joints have been predicted through computer simulation, such as reduced intact knee loading for stiffer mid-foot and forefoot conditions (Fey, Klute, & Neptune, 2012) and increased energy absorption and decreased propulsion for more compliant prostheses (Fey, Klute, & Neptune, 2013).

Finally, prior experimental work provides additional empirical predictions. For prostheses with different overall stiffness values, it is thought that more compliant feet may lead to higher rearward ground reaction forces and knee extensor moments on both legs, as well as reduced vertical peak forces during prosthetic-side late stance and intact-side early stance (Fey et al., 2011). For prostheses with different rotational stiffness levels, greater dorsiflexion compliance may lead to greater intact knee flexion and lower vertical forces on the prosthetic side during early stance (Major et al., 2014).

Some of these prior findings are inconsistent, likely due to different modeling assumptions and methodologies. For example, force invariance predicts lower energy storage and return

from stiffer forefoot components (as described above), but kinematic invariance (Srinivasan et al., 2009) predicts the opposite (same deflection with stiffer components would yield greater energy storage and return). Current understanding of amputee gait cannot yet resolve these principles (dynamic walking (minimal power and control) *vs.* kinematic invariance *vs.* force invariance *vs.* minimization of muscle activity *vs.* prior related observations) into a complete understanding of the effects of prosthesis compliance on gait. The objective of the current study was to add new experimental evidence by quantifying the sensitivity of prosthesis behavior, COM mechanics, ground forces and joint mechanics in the intact and prosthetic limbs to unit changes in prosthetic component stiffness, and to understand how these relationships are affected by walking speed. To meet these objectives we used a novel multi-component experimental prosthetic foot (Adamczyk et al., 2013) to vary hindfoot and forefoot component stiffness independently, and tested several stiffness combinations across a range of walking speeds.

Hypotheses

Based upon the reviewed literature regarding mechanical prosthesis properties and locomotion, we developed several hypotheses for how forefoot and hindfoot stiffness should influence gait. First, we hypothesized that lower component stiffness would increase keel displacement, and would therefore increase energy storage and return, because prosthesis loading is primarily due to the near-constant force of body weight during stance phase. A secondary hypothesis was that lower forefoot stiffness would result in greater COM push-off work on the prosthetic side and decreased COM collision work on the intact side. Next, we hypothesized that increased component stiffness would affect ground reaction force profiles, as well as knee moments and angles on both sides. Specifically, we hypothesized that a stiffer hindfoot would increase the early ground reaction force (GRF) and cause greater knee flexion, and consequently greater knee internal extensor moment, in early stance. We also hypothesized that a stiffer forefoot would cause the knee to become more extended after mid-stance.

To test these hypotheses, we evaluated a series of relevant outcome measures for numerical sensitivity to component stiffness. These measures included: prosthesis energy storage and return; COM work in prosthetic-side push-off and intact-side collision; vertical GRF, knee angle, and knee moment in early stance; and knee angle in late stance. Because some of these outcomes depend on speed, we incorporated data from several walking speeds and analyzed each hypothesized outcome with linear mixed-effects models based on keel stiffness and speed.

In addition to the hypothesis-driven analyses, we further explored the underlying mechanisms of these changes through analysis of potential precursors or consequences of each hypothesized effect. Because there exist few data and little theory on these mechanisms, many follow-up tests lack clear hypotheses; in these cases, we used statistical models to explore and provide preliminary evidence for previously unknown effects, with the intention of seeding future studies on specific effects of interest.

2. Methods

We designed and built an experimental prosthetic foot, with interchangeable hindfoot and forefoot components of known stiffness and consistent contact point geometry. Unilateral transtibial amputees walked on a split-belt instrumented treadmill at five speeds, using the experimental prosthesis with five different combinations of hindfoot and forefoot stiffness. We recorded body motion, foot keel motion, and ground reaction forces under each leg. We computed center of mass mechanics from ground reaction forces and joint mechanics by inverse dynamics, and then compared outcome metrics including ground reaction force peaks, COM work, peak joint angles and energy storage and return from the prosthesis. Subjects gave their written informed consent as approved by the VA Puget Sound Health Care System Institutional Review Board.

2.1 Experimental Prosthesis

The experimental prosthesis (Fig. 1) consisted of a stiff metal structure with flexible composite hindfoot and forefoot keels of varying stiffness. A bolt-through SACH foot pyramid adapter was attached to a main structural rail (Aluminum 7075-T6) having adapter mounting holes spaced 14 mm apart. Side keel mounting blocks (Aluminum 7075-T6) with multiple holes allowing different keel angles were attached by machine screws to holes in the sides of the main rail, spaced at 10 mm increments. For this study, the side blocks were mounted so that both hindfoot and forefoot keels had a sagittal declination angle of 15 degrees with respect to the main block. Hindfoot and forefoot keels were designed with several stiffness values, in roughly $15 \text{ N}\cdot\text{mm}^{-1}$ increments: hindfoot keels ranged from 22 to $69 \text{ N}\cdot\text{mm}^{-1}$; forefoot keels, from 29 to $90 \text{ N}\cdot\text{mm}^{-1}$, estimated from vertical force and displacement in controlled mechanical tests (Adamczyk et al., 2013). These values were specified to span the range reported in prior literature from commercial prostheses (Geil, 2001; Klute et al., 2004; Van Jaarsveld et al., 1990). Keels were cut from sheets of industrial woven glass-epoxy composite (G10/FR4), having energy storage and return efficiency in bending of approximately 85%, as measured with a material test machine. Hind-foot keels all had the same outline and proximal-end thickness (9.5 mm), with stiffness determined by a linear thickness taper toward the distal end (distal thickness 1 to 9.5 mm). Forefoot keels all had the same outline, but to span a wider range of stiffness, they had different proximal thicknesses (13 or 16 mm) combined with different tapers; spacers were used with thinner keels to keep total proximal thickness constant (16 mm). Forefoot keels extended all the way to the toe, to ensure effective weight support throughout stance (Agrawal et al., 2013; Hansen, Sam, & Childress, 2004; Klodd et al., 2010b).

2.2 Experimental Conditions

Experimental conditions were chosen to test independent variations of hindfoot and forefoot stiffness, and the interaction of these mechanical parameters with the behavioral parameter of walking speed. Foot configurations for each subject were chosen according to the subject's height and weight. The length of the forefoot keel and the location of the pyramid adapter and the side keel mounting blocks along the top rail were adjusted so that foot length was approximately 13% of height and the pyramid adapter was 31% to 34% of foot length forward from the hind end (Fig. 1). Subjects were tested in a Nominal Stiffness (NS)

condition, and in four alternative conditions in which an individual keel was given lower or higher stiffness: Compliant Hindfoot (CH), Compliant Forefoot (CF), Stiff Hindfoot (SH), and Stiff Forefoot (SF) (Table 1, Fig. 2). Nominal stiffness was scaled to each subject's body mass by using the keels closest to a normalized stiffness of $0.55 \text{ N}\cdot\text{mm}^{-1}$ per kg, comparable to the “flexible” or “dynamic” foot category according to industry classifications (AOPA, 2010). Alignment was held constant across conditions, and the interchangeable parts ensured that the contact points at hindfoot and forefoot maintained the same position with respect to leg anatomy. Therefore, the only significant differences among configurations were the experimental variables, hindfoot and forefoot stiffness, and a slight difference in mass (less than 50 g).

We tested human subjects with unilateral transtibial amputation during treadmill walking at several speeds with each configuration of the experimental prosthesis. The subjects tested ($n=6$; K3 or higher mobility rating; Table 1) were experienced users of commercial dynamic response prostheses. Subjects came to the laboratory for an orientation session on a separate day prior to testing, to practice walking for five minutes with each configuration of the experimental prosthesis. On the day of testing, subjects donned a tight-fitting suit of shorts and shirt, and were marked with a modified Helen Hayes set of reflective 12mm markers (Kadaba, Ramakrishnan, Wootten, & others, 1990). Marker trajectories were collected using a 12-camera motion capture system (Vicon, Oxford, UK). Subjects were fitted with the experimental foot in Nominal Stiffness (NS) configuration, which was aligned for walking by an experienced prosthetist. Subjects then walked on a split-belt instrumented treadmill (Bertec Corporation, Columbus, OH, USA; force sensitivity 0.5 N/mV mediolateral (X), 0.5 N/mV anteroposterior (Y), 1 N/mV vertical (Z); moment sensitivity 0.8 N-m/mV (X), 0.25 N-m/mV (Y), 0.4 N-m/mV (Z); natural frequency 220 Hz; center of pressure accuracy ± 1.0 mm) at five speeds ranging from quasi-static to highly dynamic gait (0.7, 0.9, 1.1, 1.3 and $1.5 \text{ m}\cdot\text{s}^{-1}$). Motion and force data were recorded at each speed (at 120 Hz and 1200 Hz, respectively). Following the NS configuration, each of the other configurations (CH, CF, SH, SF) of the experimental foot were tested, in random order, without changing the alignment of the experimental foot. Each time the prosthesis configuration was changed, subjects walked freely in the lab area for 5 minutes to familiarize themselves with the new condition.

2.3 Data Analysis and Outcome Metrics

We combined force, moment and motion data in a standard lower-body inverse dynamics analysis, performed with Visual3D software (C-Motion, Inc., Germantown, MD, USA). Resulting variables included angle, moment and power at each joint of the lower body (Fig. 5). Because the prosthetic-side ankle is not physiological and does not have a true joint center, its standard joint power plot (calculated as the dot product of ankle moment and ankle joint velocity) (Fig. 5) is not accurate. Nevertheless, we include it for comparison against prior literature, using an estimated “ankle” just distal to the foot pyramid adapter (Fig. 5). For better accuracy, we estimated intersegmental power flow into the prosthesis (energy storage) and out of it (energy return) using a deformable-segment model (Prince, Winter, Sjonnesen, & Wheeldon, 1994; Takahashi, Kepple, & Stanhope, 2012; Zelik et al., 2011). This computation estimates the power (P) flowing into and out of the shank segment through the ankle, including both linear and angular contributions (ankle force $\vec{F}$ and linear

velocity $\vec{v}$ ankle moment $\vec{M}$ and shank angular velocity $\vec{\omega}$), without assuming a rigid structure for the foot segment:

$$P = \vec{M} \cdot \vec{\omega} + \vec{F} \cdot \vec{v}$$

Intersegmental power was computed using the same estimated “ankle” (Fig. 5), but the computation is not sensitive to this choice. We computed center of mass (COM) mechanics from ground reaction forces alone (Donelan, Kram, & Kuo, 2002b) to estimate the work and work rate performed by each leg as a whole.

We computed discrete outcome metrics from kinematic and kinetic data. Major outcomes included early and late vertical force peaks on each (prosthetic or intact) side; peak positive and negative moments and angles at the knee during prosthetic-side stance; and total energy returned to the leg by the foot prosthesis. Additional outcomes computed to elucidate underlying mechanisms included: early and late anterior force peaks on each side; rate of initial vertical loading and final unloading on the prosthetic side (from 0 to 250 N); peak flexion and peak extension of the prosthetic-side pseudo-ankle; and center of mass work performed in prosthetic push-off and intact collision. We averaged these outcomes across the first seven clean strides recorded for each condition and speed, for each individual subject. A few conditions exhibited significant “crossover stepping” of the feet on the opposite belt, making kinetic analysis impossible in those strides. For these conditions, fewer strides were used as necessary. Also, some subjects could not achieve the lowest or highest speeds with all foot configurations; incomplete conditions were excluded from regression input. For analysis, all parameters and outcome metrics were non-dimensionalized according to subject-specific body mass M , standing leg length L (floor to greater trochanter), and the gravitational acceleration, g . Stiffness was normalized by Mg/L (value for redimensionalization in tables using mean subject parameters: 0.870 N/mm); force by Mg (824 N); work and moment by MgL (781 J or N·m); power by $Mg(gL)^{0.5}$ (2512 W); force rate of change by $Mg(g/L)^{0.5}$ (2652 N·s⁻¹); velocity by $(gL)^{0.5}$ (3.05 m·s⁻¹). Re-dimensionalized coefficients are presented for ease of interpretation for an “average” subject.

2.4 Sensitivity of Outcomes to Parameters

We defined sensitivity as the change in each outcome due to a unit change in each experimental parameter (forefoot stiffness, hindfoot stiffness, and walking speed). We performed a linear mixed-effects regression of each outcome metric versus two independent variables: speed and the stiffness of one keel (either hindfoot or forefoot, never both together). For each outcome metric, we compared trials where a single parameter varied (CH, NS and SH conditions; or CF, NS and SF conditions). We computed the best (least-squares) coefficients to a mixed linear model ($metric = a*stiffness + b*speed + c*(stiffness*speed) + d_i$), along with confidence intervals and P -values for the sensitivities a , b and c . To incorporate results from all subjects in determining one main effect of stiffness, we allowed a different offset d_i for each subject, while constraining common sensitivities across all subjects. Using this approach, the primary result for each outcome metric is the sensitivity a : how much change is expected in an outcome metric for a given change in

hindfoot or forefoot stiffness. The secondary result for each metric is the sensitivity b : how much change is expected in the outcome metric for a given change in walking speed. Finally, the *stiffness*speed* interaction coefficient c shows how sensitivity to each independent variable is influenced by the other. These results are summarized for different outcome metrics in Tables 2–4. Dimensionless results are reported first, along with the same results re-dimensionalized to standard units according to average subject parameters. A significance level of $\alpha = 0.05$ was chosen for convenience, but because this choice is arbitrary, true P -values are reported (down to 0.001) to allow alternative interpretations. To limit the probability of false-positive results due to multiple-comparisons, statistical tests were not performed where the parameter was expected not to influence the outcome metric (for example, sensitivity to heel stiffness for a metric sampled when the heel is not touching the ground).

Because the mixed linear model is difficult to interpret directly, we also present results in terms of the average net effect of each independent variable. Sensitivity to stiffness is presented as a “forefoot increment difference” (FID) and/or “hindfoot increment difference” (HID), the expected difference in outcome for an average subject due to a one-condition increase in keel stiffness ($15 \text{ N}\cdot\text{mm}^{-1}$; dimensionless 17.24). These quantities are evaluated from the full regression assuming average nominal conditions ($1.1 \text{ m}\cdot\text{s}^{-1}$, 45 N/mm forefoot and hindfoot stiffness). Net sensitivity to walking speed is similarly estimated as a “velocity increment difference” (VID), assuming an increase of $0.2 \text{ m}\cdot\text{s}^{-1}$, evaluated using the same nominal conditions and the mean of the interaction terms fit with hindfoot stiffness and with fore-foot stiffness. The HID, FID and VID thus estimated are similar in principle to the sensitivity term in a simple main-effect model with only stiffness or speed (not both).

3. Results

Nominal stiffness values (Table 1), non-dimensionalized to each subject’s parameters, ranged from 43 to 59 at the hindfoot, and 51 to 58 at the forefoot, with summed vertical stiffness of 95 to 116. The experimental change from nominal hindfoot stiffness ranged from -17% to -38% for the CH condition, and $+8\%$ to $+61\%$ for the SH condition. At the forefoot, changes ranged from -17% to -27% for the CF condition and $+20\%$ to $+36\%$ for the SF condition. The range of percentage changes in each condition was due to the discrete stiffness increments of the pre-built components; this resulted in different percentage changes according to the different nominal stiffness values for subjects of different sizes.

Prosthesis Energy Return

It was hypothesized that increasing stiffness would lead to decreasing energy return in the prosthesis. Results showed that total energy return from the prosthesis in each stride decreased with increasing hindfoot stiffness ($P = 0.01$), speed ($P = 0.009$), and the interaction between them ($P < 0.001$) but was not statistically sensitive to forefoot stiffness in the mixed linear model (Table 2; Fig. 3; see Discussion). The net difference expected due to a one-condition increment in hindfoot stiffness from nominal conditions (hindfoot increment difference, HID) was -1.0 J , and the net difference expected due to a one-condition increment in speed (velocity increment difference, VID) was $+0.74 \text{ J}$.

Exploring further, we found that energy storage and return were predominantly affected by hindfoot stiffness during the first half of stance, and by forefoot stiffness during push-off (Tables 2, A1; Fig 5). Prosthesis energy return specifically in the push-off phase (Adamczyk & Kuo, 2009) was sensitive specifically to forefoot stiffness ($P = 0.01$; one-condition fore-foot increment difference (FID) of -1.0 J) and speed ($P < 0.001$; VID -1.3 J). The total positive work output of the prosthetic foot was greater than that of the intact foot-ankle complex at $0.7 \text{ m}\cdot\text{s}^{-1}$ (two-tailed t-test: $P = 0.005$, mean difference across all configurations 0.0049 or 3.8 J), but not at $1.5 \text{ m}\cdot\text{s}^{-1}$ ($P = 0.16$, mean difference -0.022 or -1.7 J). Total positive foot-ankle work in the Nominal Stiffness condition ranged from 0.0148 (11.6 J) at $0.7 \text{ m}\cdot\text{s}^{-1}$ to 0.0198 (15.5 J) at $1.5 \text{ m}\cdot\text{s}^{-1}$ on the prosthetic side, and from 0.0123 (9.6 J) at $0.7 \text{ m}\cdot\text{s}^{-1}$ to 0.0242 (19.0 J) at $1.5 \text{ m}\cdot\text{s}^{-1}$ on the intact side.

COM Work

It was hypothesized that increasing forefoot stiffness would decrease COM push-off work on the prosthetic side and increase COM collision work on the intact side, according to COM step-to-step transition mechanics. Results showed (Table 2) that positive COM push-off work from the prosthetic-side leg decreased with a stiffer forefoot through an interaction effect with speed ($P = 0.01$; FID -1.5 J), and negative collision work on the intact side tended to increase in magnitude (more negative) with a stiffer forefoot, again through interaction with speed (non-significant $P = 0.08$; FID -1.7 J).

Exploring further, we found that all COM work metrics tested increased in magnitude with a main effect of increasing speed (Table 2, “Slope vs Speed” column), including COM push-off work and collision work on both sides, and total (whole-body, full-stride) positive COM work.

Ground Reaction Forces

It was hypothesized that increasing hindfoot stiffness would lead to a higher early force peak on the prosthetic side. Results (Table 3; Fig. 4) showed a borderline positive main effect with increasing hindfoot stiffness ($P = 0.03$), and a stronger negative hindfoot stiffness-speed interaction effect ($P = 0.004$), for a resulting HID of -15 N, or -1.6% .

Exploring further, we found that the late peak vertical force on the prosthetic side trended slightly greater with hindfoot stiffness (non-significant $P = 0.07$; HID $+6$ N), but neither peak changed with fore-foot stiffness (Table 3; Fig. 4). On the intact side, the first peak GRF trended higher with forefoot stiffness-speed interaction, but also did not reach significance ($P = 0.06$; FID $+22$ N; VID $+18$ N). Finally, the second intact peak GRF trended higher with speed, but again did not reach significance ($P = 0.06$; VID $+21$ N).

Initial loading rate of the prosthetic-side vertical force increased substantially with hindfoot stiffness-speed interaction ($P < 0.001$; HID $+1254 \text{ N}\cdot\text{s}^{-1}$), and trended higher with speed independently ($P = 0.07$; VID $+1828 \text{ N}\cdot\text{s}^{-1}$). Final unloading rate of the prosthetic-side vertical force increased in magnitude (more negative) with increasing speed ($P < 0.001$; VID $-1645 \text{ N}\cdot\text{s}^{-1}$), but was not similarly influenced by forefoot stiffness.

Anterior-posterior (A-P) GRF on the prosthetic side were not strongly affected by keel stiffness (Table 3; Fig. 4). The early negative peak (deceleration) exhibited a significant main effect of hindfoot stiffness (more negative, $P = 0.005$), but a larger, opposite effect with hindfoot stiffness-speed interaction ($P = 0.004$), such that overall sensitivity to hindfoot stiffness was negligible at nominal speed (HID +0.61 N). The late positive peak (acceleration) was not sensitive to either keel stiffness, but increased with a main effect of speed ($P < 0.001$; VID +20 N). On the intact side, A-P GRF peaks were not affected by prosthetic component stiffness at nominal speed ($P = 0.04$ for sensitivity of the deceleration peak to forefoot stiffness and to forefoot-speed interaction, but with canceling effects; FID -0.73 N), but both peaks increased in magnitude with a main effect of speed ($P < 0.001$).

Joint Angle and Moment

It was hypothesized that a stiffer hindfoot would cause greater knee flexion and internal moment in early stance, and that a stiffer forefoot would cause greater knee extension after mid-stance. Results showed (Table 4; Fig. 5) that peak knee flexion angle in early stance increased in magnitude (knee angle became more negative) with both hindfoot stiffness ($P < 0.001$; HID -1.9 deg) and speed ($P = 0.002$; VID -0.41 deg), with an attenuating interaction between the two ($P = 0.01$). Peak knee internal extensor moment in early stance was similarly sensitive to both hindfoot stiffness ($P = 0.02$; HID +3.9 N-m) and speed ($P = 0.02$; VID +2.4 N-m). Peak knee extension angle in mid-stance increased with a main effect of fore-foot stiffness ($P = 0.004$; FID +1.4 deg).

Exploring further, we found that peak knee internal flexor moment in mid-stance also increased in magnitude (more negative) with a main effect of forefoot stiffness ($P = 0.002$; FID -5.0 N-m). On the intact side, peak knee flexion angle in early stance increased in magnitude (more negative) in response to speed ($P = 0.02$; VID -2.6 deg), but was not sensitive to prosthetic forefoot stiffness. The intact knee's peak extensor moment in early stance trended toward an increase with speed, but the effect did not reach significance ($P = 0.052$).

Hip joint peak moments on both sides were sensitive only to main effects in speed, or not sensitive at all (Table 4; Fig. 5). On the prosthetic side, peak hip extensor (early stance; $P = 0.004$; VID -6.2 N-m) and flexor (late-stance; $P = 0.003$; VID +7.7 N-m) moments increased in magnitude with speed, but were not affected by either component stiffness. Intact hip peak extensor moment (early stance) increased in magnitude with speed ($P < 0.001$; VID -9.4 N-m), while peak flexor moment was not sensitive to any variable.

Prosthetic ankle angle was affected by different keels in different parts of the gait cycle (Table 4, Fig. 5). Ankle peak plantarflexion angle was decreased in magnitude (ankle angle less negative) with hindfoot stiffness, dominated by a hindfoot stiffness-speed interaction ($P = 0.01$; HID +1.9 deg). Plantarflexion angle also increased in magnitude (ankle angle more negative) with a main effect of speed ($P < 0.001$; VID -0.8 deg). Ankle peak dorsiflexion angle, which occurs in mid-late stance, decreased with a main effect of forefoot stiffness ($P = 0.004$; FID -1.1 deg). Note that these results are approximate, as the rigid body model used to define joint angles is not fully appropriate for a flexible prosthesis.

4. Discussion

This study was intended to characterize the sensitivity of kinematic and kinetic gait performance to the forefoot and hindfoot stiffness of a foot prosthesis. Past studies have quantified the available values of forefoot and hindfoot stiffness (Geil, 2001; Klute et al., 2004; Van Jaarsveld et al., 1990) or have investigated the effects of overall changes in a foot's stiffness (Fey et al., 2011). However, commercially available feet often allow independent selection of keels and bumpers for the hindfoot and forefoot, which interact with foot alignment during the fitting process (Hansen, Meier, Sam, Childress, & Edwards, 2003). We undertook to quantify the effects of independent changes in forefoot and hindfoot stiffness, in isolation from alignment changes, as a step toward understanding the differential effects of these two adjustment mechanisms. To assess these effects, we hypothesized that lower component stiffness would increase keel displacement, and would therefore increase energy storage and return. A secondary hypothesis was that lower forefoot stiffness would result in greater COM push-off work on the prosthetic side and decreased COM collision work on the intact side. Finally, we hypothesized that increased component stiffness would affect ground reaction force profiles, knee moments and angles on both limbs. Where appropriate, we further explored possible mechanisms of change, which may be interpreted as either precursors to, or consequences of the observed differences. The following section discusses the primary findings of this study, and the possible mechanisms by which the observed changes may have come about.

Our results indicated that the amount of energy returned by the foot prosthesis decreased with a stiffer hindfoot. This effect may explain the preference of highly active amputees for very “soft” prostheses - a softer spring deflects more under a given load, providing greater energy storage and return (ESR). This result suggests that as conditions change, ground reaction forces are more closely conserved than prosthetic ankle kinematics: if kinematics (specifically keel deflections) were conserved (Srinivasan et al., 2009), then stiffer keels would cycle more energy, not less. The dominance of forces rather than kinematics suggests that it is necessary to change stiffness directly in order to influence ESR behavior, and that clinical adjustment of alignment is unlikely to change it. This idea should be tested in future work, to help distinguish the benefits achievable through alignment from those requiring mechanical property adjustments.

It was surprising that overall energy return did not show sensitivity to forefoot stiffness in this analysis, as we expected the forefoot to have an effect similar to the hindfoot. Further exploration indicated that energy return specifically in push-off was dominated by forefoot stiffness as expected (FID -1.0 J, or -26%), as was peak push-off power (FID -32 W, or -33% ; Table A1). Furthermore, in an alternative regression excluding interaction terms, the effect of forefoot stiffness on total energy return is highly significant ($P = 3e-9$; FID -2.1 J, or -15%). It may be that a real effect is obscured by specific statistical methods, perhaps due to the small sample size. However, if there is in fact no effect as our main result suggests, such a result would suggest that the observed reduction in push-off energy return with increasing fore-foot stiffness is offset by greater energy return at another time. One possible mechanism could be a tendency for a stiff forefoot to keep more load on the hindfoot, increasing its energy storage and return. The timing of energy storage and return in specific

prosthesis components has not previously been reported to our knowledge, so the generality of this result cannot be verified. These parametric effects of forefoot and hindfoot stiffness on overall energy return should be investigated further, as this property is considered crucial to prosthesis performance.

The effect of component stiffness on the whole-body task of walking can be studied through center of mass mechanics. COM push-off work from the prosthetic side decreased with increasing forefoot stiffness (Table 2), and this led to an increasing trend in intact-side COM collision work, as predicted from principles of dynamic walking (Adamczyk & Kuo, 2015). Decreased push-off likely increased contralateral collision by redirecting the body COM less than normal (Adamczyk & Kuo, 2009; Donelan et al., 2002b; Ruina et al., 2005), a trend observed in past studies of amputees (Segal et al., 2012) and non-amputees (Collins & Kuo, 2010). However, the sensitivity of intact collision to forefoot stiffness was only weakly observed here ($P = 0.08$ for forefoot-speed interaction only; FID 13% collision increase; Table 2), and was not observed at all in another past study (Zelik et al., 2011). The details of how push-off work is applied, such as timing (Malcolm et al., 2015; Ruina et al., 2005) and center of pressure location, may involve important nuances that can dramatically alter this interaction between the limbs. There may also be important trade-offs between push-off energy in a compliant forefoot and quasi-static weight support in a stiff forefoot, for example to prevent a “fall-off” effect on a soft forefoot, and the resulting increase in contralateral collision (Adamczyk, Collins, & Kuo, 2006; Adamczyk & Kuo, 2013; Klodd et al., 2010a). These effects are worth further investigation, as the magnitude of COM collisions can affect loading at the intact knee, with potential consequences for overuse injuries such as knee osteoarthritis (Morgen-roth et al., 2011).

Walking speed also has important effects on ankle push-off energy return and COM mechanics, which interact with forefoot stiffness in unexpected ways (Table 2). First, increasing speed actually leads to decreasing ankle energy return in push-off (VID -34%). This result we cannot yet explain, but it likely contributes to the finding that increasing speed has only a small effect on prosthetic-side COM push-off work (VID +4%), as observed previously (Adamczyk & Kuo, 2015). Evidently the other joints of the leg add work to overcome the loss. This behavior still falls short of the intact leg, which increases its push-off work dramatically with speed (VID +23%). Considering these results together with the sensitivity of COM push-off to forefoot-speed interaction (Table 2), it appears that it may be beneficial to use lower forefoot stiffness for higher speeds, which may offset the speed effect with a stiffness effect.

The minimal effect of keel stiffness on ground reaction forces is an unexpected result. The reduction in prosthetic-side early vertical GRF peak with increasing hindfoot stiffness appears to be a redistribution phenomenon, in which the initial loading rate increases, reducing downward momentum of the center of mass and allowing lower force in the major peak (see Fig. 4). Nevertheless, the net change expected from an increment in hindfoot stiffness (HID) is a negligible 1.6% of the peak's mean value. Thus, the effect appears clinically insignificant compared to past results showing meaningful changes of 7–10% (Klodd et al., 2010a, 2010b). Similarly small effects of hindfoot stiffness on the 2nd prosthetic-side peak (HID < 1%) and forefoot stiffness on the 1st intact-side peak (FID

2.4%) also suggest that the effect of stiffness on the main ground reaction force peaks is not important, even if the effects are statistically real. In the anterior-posterior (A-P) direction, a similar analysis suggests that hindfoot stiffness does not affect peak forces at nominal speed, as the effect is either statistically zero or practically negligible (effects around 1%, Table 3). Sensitivity of anterior-posterior peak forces to forefoot stiffness is worth further investigation, as the apparent effect (FID) was to bias peak forces posteriorly by 6% to 11% of peak values. These effects did not reach statistical significance with the regression model used here. But, if an alternative regression without interaction terms is used, the result suggests strong significance ($P < 0.001$). If confirmed in future research, this effect could explain the feeling of “walking uphill” that patients sometimes report, which is commonly compensated in clinical practice by realigning the foot in a more dorsiflexed pose (Boone et al., 2013).

In contrast to the unaffected major GRF peaks, prosthetic-side initial loading rate was strongly influenced by the interaction effect of hindfoot keel stiffness and walking speed. This structure of this interaction - stiffness times velocity - gives units of loading rate (Ns^{-1}), and has the same form as the expected loading rate when an object of finite stiffness collides with a rigid substrate. Thus, the interaction suggests that initial loading rate may be governed by impact mechanics of the heel on the ground, until the later acceptance of body weight (Fig. 4). This phenomenon is similar to the “impact peak” commonly observed in normal walking and running, and could have consequences for comfort and risk of injury to the residual limb (Lieberman et al., 2010).

With each disturbance to the mechanical properties of the foot prosthesis, the major compensation appears to take place at the next proximal joint, the knee. The knee flexes with a braking function during weight acceptance in early stance, then extends into a weight-support posture through mid-stance, and finally flexes again during terminal stance/pre-swing (Fig. 5). Early knee angle and knee moment were sensitive to changes in both hindfoot keel stiffness and walking speed. Increasing the stiffness of the hindfoot keel led to increased knee flexion during weight acceptance ($\text{HID} -1.9$ deg, roughly 15% of stance-phase knee range of motion (ROM) in the NS 1.1 condition; Table 4). This more flexed knee posture requires a greater extensor moment (Table 4, Fig. 5; $\text{HID} +3.9$ N-m, or +9% of peak) to straighten the leg again. The increase in knee flexion may be caused initially by the increased hindfoot loading rate discussed above (Table 3, Fig. 4), which leads to a higher rate of ankle moment loading as well (Table A1, Fig. 5). These two effects suggest an increase in the external moment tending to flex the knee (Lehmann, Price, Boswell-Bessette, Dralle, Questad, et al., 1993). Additionally, the more flexed knee posture in early stance may contribute to the slight reduction in the early GRF peak (as opposed to the hypothesized increase). A flexed knee results in a mechanical disadvantage for the knee extensors in transmitting force to the torso, thus it follows that the force itself would not increase. The difference between our hypothesis and this finding may depend on timing: it is not the main GRF peak that causes the knee to bend in this case, but rather the early impact phase of force generation.

A similar chain appears to affect knee mechanics in mid-stance, which were sensitive to forefoot stiffness. Increasing forefoot stiffness led to a straighter knee in mid-to-late stance

(FID 1.4 deg, 11% of ROM, Table 4), and an increase in flexor moment (FID -5 N, 13% of peak), presumably to prevent knee hyperextension. We did not observe an increased ankle plantar flexor moment to explain this effect (Table 4, Fig. 5), but it could result from a more subtle change such as an earlier extension while the ankle moment is still low.

Clinically, these changes in knee kinematics due to component stiffness indicate possible design or prescription changes that could be used to influence an individual's gait in favorable ways. Increasing the hindfoot stiffness of a prosthesis may encourage knee flexion in amputees who exhibit an excessively straight knee during stance phases. However, the greater knee flexion angle also leads to a greater knee internal extensor moment near the time of peak GRF (Table 4, Fig. 5), which could cause socket discomfort if it becomes too high (Boone et al., 2013; Pinzur et al., 1995; Seelen, Anemaat, Janssen, & Deckers, 2003). Increasing the forefoot stiffness may help amputees who experience instability in the prosthetic-side knee, by helping the knee stay extended (Table 4, Fig. 5). However, this straight-knee posture leads to a requirement for greater knee internal flexor moment in late stance, to prevent knee hyperextension. Such differences in mechanical properties are normally mimicked in practice by changing sagittal alignment (Boone et al., 2013; Schmalz et al., 2002), if the knee is felt to be thrust forward (with a foot too dorsiflexed) or hyper-extended (with a foot too plantar flexed). But, the alignment approach directly compromises hindfoot and forefoot function against each other. In contrast, directly changing properties could provide specific benefits from both keels, without the trade-off. The consequences of using component changes vs. alignment changes to achieve a desired outcome should be carefully considered in clinical practice.

It appears that the knee is the primary compensator for changes in distal mechanics, as hip mechanics in both legs were not sensitive to either hindfoot or forefoot stiffness. Hip moment peaks changed only with speed, and showed no indication of any trend with stiffness (Table 4). Component stiffness also had no effect on hip angle or power (not shown). From this analysis, it seems that hip behavior is maintained in spite of changes in the foot and ankle. Prior data from a similar study also show minimal differences at the hip (Fey et al., 2011).

The generality of this study's results is both enhanced and limited by the use of feet with a particular geometry, held invariant across conditions. This experimental design allowed isolation of each individual component's stiffness as an independent variable, and interrogation of its unique effects. Because commercially available prostheses use a wide range of configurations, varying such parameters as internal or external bumpers, contact locations, foot shape, keel length, and multi-axis motion, quantitative results may be different for each design. Nevertheless, qualitative results (positive vs. negative sensitivity to stiffness) are likely generalizable across a wide range of passive prostheses that approximate spring-damper function.

The experimental prosthesis used in this study was designed according to established scientific principles and findings, but aspects of the implementation may not have been ideal. Foot length, ankle position and keel stiffness matched documented ranges, but ground contact geometry and interface mechanics were different. First, we tested the experimental

device without a foot shell or a shoe (to eliminate these uncontrolled variables), whereas other prostheses are usually worn with both. Second, the points of load transfer to the structural keels in the experimental device were approximately 12 mm from each end of the foot; many other prostheses include curved portions yielding contact under more proximal portions of the heel and the metatarsophalangeal joint. These designs may have different effects on motion of the center of pressure, and on the effective angular stiffness of the prosthesis (Adamczyk et al., 2013).

This property of angular stiffness, viewed separately from linear hindfoot and forefoot stiffness, is an aspect of foot prostheses that should be studied further. In the present experiment, some of the stiffer conditions caused subjects to complain of a “flat spot” phenomenon during stance phase. This may be similar to the “dead spot” sometimes reported clinically in commercial prostheses. We understood this to be a feeling that the center of pressure (COP) moves forward too rapidly (rather than smoothly and gradually), producing a sudden change in moment and/or force when the forefoot contacts the ground. We speculate that this sensation was caused not by the keel stiffness per se (which was within reported ranges), but by high angular stiffness, which relates to keel stiffness by the square of the distance between the two keels’ contact points (Adamczyk et al., 2013). According to a mechanical model of this relationship, the angular stiffness of the conditions tested in this study ranged from 13.2 to 33.6 N·m·deg⁻¹, which is toward the higher side of ankle quasi-stiffness estimated from natural ankles (3.5 to 24.4 N·m·deg⁻¹) and from the stiffness and geometry of commercial prostheses (4 to 28 N·m·deg⁻¹) (Adamczyk et al., 2013; Major et al., 2014; Shamaei, Sawicki, & Dollar, 2013; Singer, Ishai, & Kimmel, 1995). Further work directly quantifying angular stiffness and its effects in commercial prostheses would provide valuable insight into this aspect of gait biomechanics.

This project was limited by small sample size and restricted experimentation time with each participant, similar to most studies with lower limb amputees. The result is that the outcomes are most applicable to active unilateral transtibial amputees (K-level = 3 or better), and the range of stiffness values and walking speeds tested. Also, greater acclimation time for each condition would be preferred. Findings may differ in the wider population of lower limb amputees or in chronic settings where slower adaptations may develop.

Another limitation of this study was the use of a fixed set of keels for all subjects. More precise scaling of stiffness to subject mass would likely improve the results. In the present study, time, expense, and achievable stiffness resolution did not allow for custom keels for each subject. Future studies should consider a more subject-specific approach, and should confirm these findings in a broader sample, including bilateral transtibial amputees and transfemoral amputees.

The outcomes of this study re-confirm the importance of balancing the stiffness properties between the hindfoot and forefoot portions of the foot-ankle system. It would be useful to further examine the effect of heel height and relative stiffness in different types shoes, e.g. to determine an ideal hindfoot stiffness for accommodating a variety of footwear, or how a forefoot component can compensate for heel height changes. Finally, further work should

investigate how stiffness changes and alignment changes interact, and how each should be used in clinical practice.

5. Conclusions

Prosthetic component stiffness can have a substantial effect on energy storage and return, whole-body COM work performed during step-to-step transitions, and knee mechanics. Our data suggest that less stiff components (specifically the forefoot) would be well-suited to the more active patient with below knee amputation. For patients with specific gait deviations, specific changes to the stiffness of a component may be useful for correcting problems, such as excessive or inadequate knee flexion. These observations may be useful to prosthetic foot designers, clinicians and researchers interpreting the risks and benefits of utilizing the variety of prosthetic feet commercially available today.

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Appendix:: Sensitivity of Clinical Outcomes to Changes in Hindfoot and Forefoot Stiffness

In addition to basic gait mechanics, we compiled several results for clinically relevant outcomes. Table A1 includes details of prosthetic-side ankle and knee kinetics. These results may help prosthetics practitioners use the results of this study to address specific gait deviations commonly observed in the clinic.

Table A1:

Additional explanatory results in dimensionless and redimensionalized units.

Supplementary Results - Dimensionless										Redimensionalized					Increment Effects		
Quantity	Mean $\pm$ SD (NS 1.1)	Slope vs. K_{Fore} P_{val}	Slope vs. K_{Hind} P	Coef. K_{Hind}^{*v} P	Slope ^e vs. Speed v P	Mean $\pm$ SD (NS 1.1)	Slope vs. K_{Fore} (units)	Coef. K_{Fore}^{*v} (units)	Slope vs. K_{Hind} (units)	Coef. K_{Hind}^{*v} (units)	Slope ^e vs. Speed ^e (units)	FID +15 N/mm (units)	HID +15 N/mm (units)	VID +0.2 m/s (units)			
Prosth. Side Ankle [†]	-0.154 $\pm$ 0.013	-1.8E-4	-	-	-0.13	-120 $\pm$ 10 (N-m)	-0.16 (N-m per N/mm)	0.26 (N-m per N/mm per m/s)	-	-	-32 (N-m per m/s)	1.8 (N-m)	-	-4.1 (N-m)			
	Peak Plantarflexor Moment (Late)	0.41	0.16	-	<0.001	69.2 $\pm$ 190 (N-m/s)	-	-	-5.6 (N-m per N/mm)	11 (N-m per N/mm per m/s)	220 (N-m/s per m/s)	-	100 (N-m/s)	145 (N-m/s)			
Prosth. Side Ankle [†]	0.275 $\pm$ 0.075	-	-1.9E-3	1.2E-2	0.27	-58.4 $\pm$ 21.2 (W)	-0.37 (W per N/mm)	0.59 (W per N/mm per m/s)	5.1E-2 (W per N/mm)	-0.21 (W per N/mm per m/s)	-61 (W per m/s)	4.2 (W)	-2.6 (W)	-11 (W)			
Mean Inverse Rate (Initial Loading)		-	0.33	0.03	0.35	56.7 $\pm$ 21.8 (W)	0.96 (W per N/mm)	-0.62 (W per N/mm per m/s)	1.0 (W per N/mm)	-0.66 (W per N/mm per m/s)	57 (W per m/s)	4.3 (W)	4.4 (W)	5.7 (W)			
Prosth. Side Knee	-2.32E-2 $\pm$ 8.4E-3	-1.3E-4	1.8E-5	-2.2E-4	-7.5E-2	-50.0 $\pm$ 18.4 (W)	-	-	-1.0 (W per N/mm)	1.6 (W per N/mm per m/s)	-171 (W per m/s)	-	11 (W)	-20 (W)			
Peak Neg. Power (Early Stance)		0.48	0.92	0.64	0.003	53.0 $\pm$ 7.1 (W)	0.44 (W per N/mm)	-6.5E-2 (W per N/mm per m/s)	0.80 (W per N/mm)	-1.2 (W per N/mm per m/s)	83 (W per m/s)	5.6 (W)	-7.3 (W)	11 (W)			
Prosth. Side Knee	2.26E-2 $\pm$ 8.7E-3	3.3E-4	3.6E-4	-6.5E-4	7.0E-2	-51.6 $\pm$ 12.9 (W)	-0.13 (W per N/mm)	0.86 (W per N/mm per m/s)	-0.78 (W per N/mm)	0.57 (W per N/mm per m/s)	-81 (W per m/s)	12 (W)	-2.3 (W)	-9.8 (W)			
Peak Pos. Power (Mid Stance)		0.10	0.04	0.13	0.03	96.5 $\pm$ 25.7 (W)	0.24 (W per N/mm)	-2.1 (W per N/mm per m/s)	-	-	171 (W per m/s)	-32 (W)	-	15 (W)			
Prosth. Side Ankle [†]	-1.99E-2 $\pm$ 7.3E-3	-	-	-	-0.21												
Peak Neg. Power (Early Stance)		-	0.053	<0.001	<0.001												
Prosth. Side Ankle [†]	2.11E-2 $\pm$ 2.8E-3	1.5E-4	2.8E-4	-1.2E-3	1.0E-1												
Peak Pos. Power (Mid Stance)		0.51	0.07	0.003	0.006												
Prosth. Side Ankle [†]	-2.05E-2 $\pm$ 5.1E-3	-4.5E-5	-2.7E-4	6.0E-4	-9.9E-2												
Peak Neg. Power (Late Stance)		0.80	0.04	0.09	<0.001												
Prosth. Side Ankle [†]	3.84E-2 $\pm$ 1.0E-2	8.3E-5	-	-	0.21												
Peak Pos. Power (Terminal Stance)		0.67		-	<0.001												

P -values and Slopes vs. Speed are estimated conservatively. P -values are the greater of the P -values of the Speed coefficient in mixed linear regressions with K_{Fore} or K_{Hind} foot. Slope values are the lesser of the two Speed coefficients.

$\ddagger$ Indicates statistical test not performed.

$\ddagger$ Ankle definitions for the prosthetic side are approximate, as the system lacks a true ankle joint. NS 1.1: Result for Nominal Stiffness condition at 1.1 m·s⁻¹.

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Figure 1:

Schematic and photograph of the experimental prosthesis with interchangeable components. A central mounting rail allows linear repositioning of the pyramid adapter, and linear and angular repositioning of the keel mounting blocks. Forefoot (red) and hindfoot (blue) keels of different stiffness are achieved by tapering the thickness of each keel; these keels are interchanged to increase or decrease the stiffness of each component. Settings were chosen for each subject according to height and body mass, to select appropriate nominal stiffness (NS) conditions and maintain normal ratios of foot length to height and ankle position with respect to the heel.

Figure 2:

Schematic representation of the experimental protocol. A nominal stiffness (NS) condition was chosen according to body weight, used for prosthetic alignment, and tested at five walking speeds (0.7, 0.9, 1.1, 1.3, 1.5 $\text{m}\cdot\text{s}^{-1}$). Then hindfoot and forefoot keels were changed individually to alter the stiffness of each component (CH, SH, CF, SF conditions), and tested at the same five speeds.

Figure 3:

Energy storage and return (mean energy return shown) from the prosthesis decreases with increasing stiffness of the hindfoot, and possibly the forefoot. This result supports the principle of gravity force-driven energy storage. If instead kinematics were invariant, energy storage and return should have increased with higher stiffness.

Figure 4:
Example Vertical and Anterior-Posterior (A-P) Ground Reaction Forces (GRF), for one subject at $1.3 \text{ m}\cdot\text{s}^{-1}$.

Figure 5: Example prosthetic side ankle, knee and hip joint mechanics (angle, internal moment, and power output) for one subject at $1.3 \text{ m}\cdot\text{s}^{-1}$. Systematic changes with stiffness are shown with gray arrows indicating direction. Ankle power is also shown with a deformable-body formulation (intersegmental power) more appropriate for prostheses. Scales are matched across columns.

Table 1:

Characteristics of human subjects and experimental conditions.

Subject Number	Mass kg	Leg Length [*] m	Amputation Side	Prescribed Prosthesis	CH N/mm	NSH N/mm	SH N/mm	CF N/mm	NSF N/mm	SF N/mm
1	71.9	0.951	Left	Flex-Foot Naser	21.9	33.0	53.1	29.2	39.6	47.5
2	73.1	0.945	Left	Highlander	21.9	33.0	53.1	29.2	39.6	47.5
3	90.0	0.976	Left	Vari-Flex ESR	33.0	53.1	63.8	39.6	47.5	64.5
4	74.9	0.961	Left	Elite Blade	21.9	33.0	53.1	29.2	39.6	47.5
5	111.9	0.975	Right	Multi-axial	53.1	63.8	69.0	47.5	64.5	90.3
6	82.7	0.877	Right	Renegade	33.0	53.1	63.8	39.6	47.5	64.5

^{*} Leg Length was defined as the height from the floor to the greater trochanter.

Abbreviations: C: Compliant; NS: Nominal Stiffness; S: Stiff; H: Hindfoot; F: Forefoot

Table 2:

Sensitivity of Ankle energy return and Center of Mass work to component stiffness and speed.

Ankle and Center of Mass Work - Dimensionless							Redimensionalized						Increment Effects		
Quantity	Mean ± SD (NS 1.1)	Slope vs. K_{Fore} P -val	Coef. K_{Fore}^{*v} P	Slope vs. K_{Hind} P	Coef. K_{Hind}^{*v} P	Slope* vs. Speed v P	Mean ± SD (NS 1.1)	Slope vs. K_{Fore} (units)	Coef. K_F^{*v} (units)	Slope vs. K_{Hind} (units)	Coef. K_H^{*v} (units)	Slope* vs. Speed v (units)	FID +15 N/mm (units)	HID +15 N/mm (units)	V/D +0.2 m/s (units)
Prosth. Side Ankle	1.75E-2±4.5E-3	-6.8E-5	-2.6E-4	1.5E-4	-6.2E-4	3.7E-2	13.7±3.5 (J)	-6.1E-2 (J per N/mm)	-7.5E-2 (J per N/mm per m/s)	0.13 (J per N/mm)	-0.18 (J per N/mm per m/s)	9.5 (J per m/s)	-2.2 (J)	-1.0 (J)	0.74 (J)
Total Energy Return (Intersegmental)		0.45	0.32	0.01	<0.001	0.009									
Prosth. Side Ankle	4.94E-3±2.3E-3	-1.8E-4	3.0E-4	-	-	-4.1E-2	3.86±1.79 (J)	-0.16 (J per N/mm)	8.8E-2 (J per N/mm per m/s)	-	-	-10 (J per m/s)	-1.0 (J)	-	-1.3 (J)
Push-off Energy Ret. (Intersegmental)		0.01	0.15	-	-	<0.001									
Prosth. Side	-8.45E-3±4.2E-3	-	-	-1.0E-4	3.9E-4	-6.0E-2	-6.60±3.27 (J)	-	-	-9.0E-2 (J per N/mm)	0.11 (J per N/mm per m/s)	-15 (J per m/s)	-	0.54 (J)	-2.0 (J)
Negative COM Work In Collision		-	-	0.21	0.08	<0.001									
Prosth. Side	1.86E-2±2.7E-3	1.2E-4	-6.6E-4	-	-	4.6E-2	14.5±2.1 (J)	0.11 (J per N/mm)	-0.19 (J per N/mm per m/s)	-	-	12 (J per m/s)	-1.5 (J)	-	0.61 (J)
Positive COM Work In Push-Off		0.18	0.01	-	-	0.002									
Intact Side	-1.64E-2±8.6E-3	2.3E-4	-1.0E-3	-	-	-7.5E-2	-12.8±6.7 (J)	0.21 (J per N/mm)	-0.29 (J per N/mm per m/s)	-	-	-19 (J per m/s)	-1.7 (J)	-	-6.5 (J)
Negative COM Work In Collision		0.26	0.08	-	-	0.02									
Intact Side	2.01E-2±6.5E-3	-	-	1.2E-4	-2.6E-4	8.3E-2	15.7±5.1 (J)	-	-	0.11 (J per N/mm)	-7.6E-2 (J per N/mm per m/s)	21 (J per m/s)	-	0.38 (J)	3.6 (J)
Positive COM Work In Push-Off		-	-	0.21	0.32	<0.001									
Total (Both Legs)	5.90E-2±6.1E-3	4.0E-4	-9.2E-4	2.1E-4	-3.9E-4	0.14	46.1±4.8 (J)	0.36 (J per N/mm)	-0.27 (J per N/mm per m/s)	0.19 (J per N/mm)	-0.11 (J per N/mm per m/s)	35 (J per m/s)	0.86 (J)	0.88 (J)	5.3 (J)
Positive COM Work (Whole Stride)		0.15	0.23	0.19	0.36	<0.001									

P -values and Slopes vs. Speed are estimated conservatively. P -values are the greater of the P -values of the Speed coefficient in mixed linear regressions with K_{Fore} foot or K_{Hind} foot. Slope values are the lesser of the two Speed coefficients.

[†]Indicates statistical test not performed.

NS 1.1: Result for Nominal Stiffness condition at 1.1 m·s⁻¹.

Table 3:

Sensitivity of ground reaction forces to component stiffness and speed.

Ground Reaction Force Outcomes - Dimensionless							Redimensionalized						Increment Effects		
Quantity	Mean ± SD (NS 1.1)	Slope vs. K_{Fore} <i>P-val</i>	Coef. K_{Fore}^{*v} <i>P</i>	Slope vs. K_{Hind} <i>P</i>	Coef. K_{Hind}^{*v} <i>P</i>	Slope* vs. Speed <i>P</i>	Mean ± SD (NS 1.1)	Slope vs. K_{Fore} (units)	Coef. K_{Hind}^{*v} (units)	Slope vs. K_{Hind} (units)	Coef. K_H^{*v} (units)	Slope* vs. Speed <i>v</i> (units)	FID +15 N/mm (units)	HID +15 N/mm (units)	VID +0.2 m/s (units)
Prosth. Side	1.13±0.08	3.3E-3	-7.1E-3	3.4E-3	-1.2E-2	0.47	932±63 (N)	3.2 (N per N/mm)	-2.2 (N per N/mm per m/s)	3.2 (N per N/mm)	-3.9 (N per N/mm per m/s)	126 (N per m/s)	11 (N)	-15 (N)	-2.1 (N)
Fz Peak 1	1.07±0.06	2.2E-4	-8.4E-4	1.5E-3	-2.9E-3	0.20	883±49 (N)	0.21 (N per N/mm)	-0.26 (N per N/mm per m/s)	1.4 (N per N/mm)	-0.90 (N per N/mm per m/s)	54 (N per m/s)	-1.2 (N)	6.0 (N)	5.5 (N)
Fz Peak 2	1.13±0.05	-1.9E-3	9.5E-3	4.8E-4	-1.7E-3	0.14	931±39 (N)	-1.8 (N per N/mm)	3.0 (N per N/mm per m/s)	0.46 (N per N/mm)	-0.52 (N per N/mm per m/s)	38 (N per m/s)	22 (N)	-1.7 (N)	18 (N)
Fz Peak 1	0.986±0.026	4.4E-4	-1.0E-3	1.0E-3	-2.2E-3	0.47	812±21 (N)	0.42 (N per N/mm)	-0.32 (N per N/mm per m/s)	0.94 (N per N/mm)	-0.68 (N per N/mm per m/s)	127 (N per m/s)	0.93 (N)	2.9 (N)	21 (N)
Fz Peak 2	2.96±0.54	-	-	-2.1E-2	0.13	3.6	7860±1440 (N/s)	-	-	-64 (N/s per N/mm)	134 (N/s per N/mm per m/s)	3110 (N/s per m/s)	-	1254 (N/s)	1828 (N/s)
Fz Loading Rate	-3.39±0.42	3.9E-3	3.7E-3	-	-	-9.7	-8980±1100 (N/s)	12 (N/s per N/mm)	3.7 (N/s per N/mm per m/s)	-	-	-8400 (N/s per m/s)	240 (N/s)	-	-1645 (N/s)
Fz Unloading Rate	-0.115±0.033	-9.7E-4	7.1E-4	-	-	<0.001	-94.6±27.5 (N)	-0.92 (N per N/mm)	0.22 (N per N/mm per m/s)	-1.3 (N per N/mm)	1.2 (N per N/mm per m/s)	-53 (N per m/s)	-10 (N)	0.61 (N)	-4.3 (N)
Fy Peak Decel.	0.155±0.021	4.8E-4	-2.9E-3	4.0E-4	-8.2E-4	0.46	128±18 (N)	0.45 (N per N/mm)	-0.89 (N per N/mm per m/s)	0.38 (N per N/mm)	-0.25 (N per N/mm per m/s)	125 (N per m/s)	-7.9 (N)	1.5 (N)	20 (N)
Fy Peak Accel.	-0.161±0.039	1.3E-3	-3.7E-3	-2.7E-4	6.0E-4	-0.39	-133±32 (N)	1.2 (N per N/mm)	-1.2 (N per N/mm per m/s)	-0.25 (N per N/mm)	0.19 (N per N/mm per m/s)	-106 (N per m/s)	-0.73 (N)	-0.70 (N)	-25 (N)
Fy Peak Decel.	0.159±0.034	5.3E-4	-8.6E-4	3.4E-4	-8.8E-4	0.55	131±28 (N)	0.50 (N per N/mm)	-0.27 (N per N/mm per m/s)	0.32 (N per N/mm)	-0.27 (N per N/mm per m/s)	150 (N per m/s)	3.1 (N)	0.24 (N)	27 (N)

P-values and Slopes vs. Speed are estimated conservatively. *P*-values are the greater of the *P*-values of the Speed coefficient in mixed linear regressions with *K*Forefoot or *K*Hindfoot. Slope values are the lesser of the two Speed coefficients.

NS 1.1: Result for Nominal Stiffness condition at 1.1 m·s⁻¹.

Indicates statistical test not performed.

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Table 4:

Sensitivity of ankle and knee mechanics to component stiffness and speed.

Peak Joint Angles and Moments - Dimensionless										Redimensionalized				Increment Effects		
Quantity	Mean ± SD (NS 1.1)	Slope vs. K_{Core} P -val	Coef. K_{Fore}^{*v} P	Slope vs. K_{Hind} P	Coef. K_{Hind}^{*v} P	Slope* vs. Speed v P	Mean ± SD (NS 1.1)	Slope vs. K_{Fore} (units)	Coef. K_F^{*v} (units)	Slope vs. K_{Hind} (units)	Coef. K_H^{*v} (units)	Slope* vs. Speed v (units)	FID +15 N/mm (units)	HID +15 N/mm (units)	VID +0.2 m/s (units)	
Prosth. Side Ankle [‡]	−6.84±1.17	-	-	4.5E−2	0.17	−21	−6.84±1.17 (deg)	-	-	5.2E−2 (deg per N/mm)	-	−6.9 (deg per m/s)	-	1.9 (deg)	−0.80 (deg)	
		-	-	0.06	0.01	<0.001										
Peak Plantarflexion																
Prosth. Side Ankle [‡]	4.54±0.49	−4.1E−2	−5.7E−2	-	-	3.3	4.54±0.49 (deg)	−4.7E−2 (deg per N/mm)	−2.2E−2 N/mm per m/s)	-	-	1.1 (deg per m/s)	−1.1 (deg)	-	2.3E−2 (deg)	
Peak Dorsiflexion		0.004	0.14	-	-	0.12										
Prosth. Side Knee	−10.4±7	-	-	−0.26	0.42	−28	−10.4±6.7 (deg)	-	-	−0.30 (deg per N/mm)	0.16 (deg per N/mm per m/s)	−9.1 (deg per m/s)	-	−1.9 (deg)	−0.41 (deg)	
Peak Flexion (Early Stance)		-	-	<0.001	0.01	0.002										
Prosth Side Knee	2.60±6.79	0.16	−0.23	-	-	14	2.60±6.79 (deg)	0.19 (deg per N/mm)	−8.7E−2 N/mm per m/s)	-	-	4.4 (deg per m/s)	1.4 (deg)	-	0.10 (deg)	
Peak Extension (Mid Stance)		0.004	0.14	-	-	0.12										
Prosth. Side Knee	5.66E−2±3.9E−2	-	-	7.3E−4	−1.2E−3	0.11	44.2±30.3 (N-m)	-	-	0.65 (N-m per N/mm)	−0.36 (N-m per N/mm per m/s)	28 (N-m per m/s)	-	3.9 (N-m)	2.4 (N-m)	
Peak Extensor Mom. (Early Stance)		-	-	0.02	0.16	0.02										
Prosth. Side Knee	−5.10E−2±2.4E−2	−9.6E−4	1.6E−3	-	-	−8.8E−2	−39.8±18.7 (N-m)	−0.86 (N-m per N/mm)	0.48 (N-m per N/mm per m/s)	-	-	−23 (N-m per m/s)	−5.0 (N-m)	-	−0.19 (N-m)	
Peak Flexor Mom. (Mid Stance)		0.002	0.06	-	-	0.06										
Intact Side Knee	−12.0±3	8.6E−2	−0.22	-	-	−29	−12.0±3.4 (deg)	9.8E−2 (deg per N/mm)	−8.2E−2 (deg per N/mm per m/s)	-	-	−9.5 (deg per m/s)	0.12 (deg)	-	−2.6 (deg)	
Peak Flexion (Early Stance)		0.28	0.32	-	-	0.02										
Intact Side Knee	5.70E−2 ±2.0E−2	−6.6E−4	2.1E−3	-	-	0.15	44.5±15.2 (N-m)	−0.59 (N-m per N/mm)	0.61 (N-m per N/mm per m/s)	-	-	38 (N-m per m/s)	1.3 (N-m)	-	13 (N-m)	
Peak Extensor Mom. (Early Stance)		0.18	0.13	-	-	0.052										
Prosth. Side Hip	−6.45E−2±2.1E−2	−3.5E−4	6.6E−4	−1.0E−4	−2.4E−4	−0.13	−50.3±16.4 (N-m)	−0.31 (N-m per N/mm)	0.19 (N-m per N/mm per m/s)	−9.2E−2 (N-m per N/mm)	−7.1E−2 (N-m per N/mm per m/s)	−34 (N-m per m/s)	−1.5 (N-m)	−2.5 (N-m)	−6.2 (N-m)	

Peak Joint Angles and Moments - Dimensionless							Redimensionalized						Increment Effects		
Quantity	Mean ± SD (NS 1.1)	Slope vs. K_{Core} P -val	Coef. K_{Fore}^{*v} P	Slope vs. K_{Hind} P	Coef. K_{Hind}^{*v} P	Slope* vs. Speed v P	Mean ± SD (NS 1.1)	Slope vs. K_{Fore} (units)	Coef. K_F^{*v} (units)	Slope vs. K_{Hind} (units)	Coef. K_H^{*v} (units)	Slope* vs. Speed v (units)	FID +15 N/mm (units)	HUD +15 N/mm (units)	VID +0.2 m/s (units)
Peak Extensor Mom. (Early Stance)		0.35	0.52	0.74	0.77	0.004									
Prosth. Side Hip	6.66E-2±1.9E-2	6.7E-5	-9.3E-5	9.0E-5	-1.7E-4	0.16	52.0±15.2 (N-m)	6.0E-2 (N-m per N/mm)	-2.7E-2 (N-m per N/mm per m/s)	8.1E-2 (N-m per N/mm)	-5.0E-2 (N-m per N/mm per m/s)	41 (N-m per m/s)	0.46 (N-m)	0.40 (N-m)	7.7 (N-m)
Peak Flexor Mom. (Late Stance)		0.84	0.92	0.70	0.79	0.003									
Intact Side Hip	-9.68E-2±2.4E-2	-2.6E-4	3.1E-4	-1.5E-4	1.1E-4	-0.20	-75.6±18.5 (N-m)	-0.23 (N-m per N/mm)	9.1E-2 (N-m per N/mm per m/s)	-0.13 (N-m per N/mm)	3.4E-2 (N-m per N/mm per m/s)	-50 (N-m per m/s)	-1.9 (N-m)	-1.5 (N-m)	-9.4 (N-m)
Peak Extensor Mom. (Early Stance)		0.60	0.82	0.63	0.89	<0.001									
Intact Side Hip	5.73E-2±3.0E-2	-3.8E-4	5.1E-4	-8.5E-5	5.2E-4	6.8E-2	44.8±23.3	-0.34 (N-m per N/mm)	0.15 (N-m per N/mm per m/s)	-7.7E-2 (N-m per N/mm)	0.15 (N-m per N/mm per m/s)	17 (N-m per m/s)	-2.6 (N-m)	1.4 (N-m)	4.8 (N-m)
Peak Flexor Mom. (Late Stance)		0.32	0.63	0.77	0.52	0.21									

P -values and Slopes vs. Speed are estimated conservatively. P -values are the greater of the P -values of the Speed coefficient in mixed linear regressions with K_{Fore} foot or K_{Hind} foot. Slope values are the lesser of the two Speed coefficients.

~ Indicates statistical test not performed.

[‡] Ankle definitions for the prosthetic side are approximate, as the system lacks a true ankle joint.

NS 1.1: Result for Nominal Stiffness condition at 1.1 m·s⁻¹.