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Sensitivity of lower-limb joint mechanics to prosthetic forefoot stiffness with a variable stiffness foot in level-ground walking

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A R T I C L E I N F O

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A B S T R A C T

This paper presents the effects of the Variable Stiffness Foot (VSF) on lower-limb joint mechanics in level-ground walking. Persons with transtibial amputations use lower-limb prostheses to restore level-ground walking, and foot stiffness and geometry have been shown to be the main factors for evaluating foot prostheses. Previous studies have validated the semi-active and stiffness modulation capabilities of the VSF. The core aim of this study is to investigate the mechanical effects of adjusting stiffness on knee and ankle mechanics for prosthetic users wearing the VSF. For this study, seven human participants walked with three different stiffnesses (compliant, medium, stiff) of the VSF across two force plates in a motion capture lab. Linear mixed models were utilized to estimate the significance and coefficients of determinations for the regression of stiffness on several biomechanical metrics. A stiffer VSF led to decreased ankle dorsiflexion angle ($p < 0.0001$, $r^2 = 0.90$), increased ankle plantarflexor moment ($p = 0.016$, $r^2 = 0.40$), increased knee extension ($p = 0.021$, $r^2 = 0.37$), increased knee flexor moment ($p = 0.0007$, $r^2 = 0.63$), and decreased magnitudes of prosthetic energy storage ($p < 0.0001$, $r^2 = 0.90$), energy return ($p = 0.0003$, $r^2 = 0.67$), and power ($p < 0.0001$, $r^2 = 0.74$). These results imply lower ankle, knee, and hip moments, and more ankle angle range of motion using a less stiff VSF, which may be advantageous to persons walking with lower-limb prostheses. Responsive modulation of the VSF stiffness, according to these findings, could help overcome gait deviations associated with different slopes, terrain characteristics, or footwear.

1. Introduction

The primary function of lower-limb prostheses for persons with amputations is to restore overground walking. Still, these prosthetic devices are also necessary for standing and walking on stairs, ramps, and uneven terrain. In particular, transtibial prostheses attempt to return walking functionality by mimicking some aspects of the lost human ankle–foot mechanics. These lost mechanics include altered ankle push-off energy and power, and ankle angle motion (Adamczyk et al., 2017; Adamczyk and Kuo, 2015; Fey, 2011). This attempt to restore walking for prosthetic users is difficult as the influence of prosthetic parameters like stiffness and geometry on ankle and knee mechanics is not fully understood, and is variable across terrains and locomotor behaviors. The difficulty in universally recreating the human ankle function in terms of ankle–foot geometry and ankle angle and torque relations instigates a need for improved mechanical adaptability of the prostheses (Adamczyk et al., 2017).

Energy Storage and Return (ESR) prostheses are the standard solutions to offer some passive adaptability and returned functionality to prosthetic users. ESR prostheses absorb mechanical energy using viscoelastic deformation as they are loaded and return that energy when unloaded during push-off. To assess the usefulness of the ESR prostheses, one proposal (AOPA, 2010) created standardized tests to establish essential features to evaluate prosthetic feet for prescription. Another report (Major et al., 2011) put forward that transtibial prostheses can be differentiated by three main factors: normal stiffness (superior-inferior direction to foot) and foot geometry, and to a lesser extent, shear stiffness. (Raschke et al., 2015) found that persons with amputations preferred less stiff (normal stiffness) feet while walking, which was associated with 15 % lower peak sagittal prosthetic socket moments.

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Furthermore, (Fey et al., 2011) found that decreased foot stiffness leads to an increased prosthetic range of motion (ankle), mid-stance energy storage, and late stance energy return. These factors of prosthetic socket moments, ankle angle range, and energy storage and return are affected by changes in foot stiffness, and are important when considering the adaptability of ESR prostheses.

To further explore the effect of foot stiffness on walking mechanics, (Adamczyk et al., 2017) examined the hind/forefoot stiffness effects on prosthetic walking for various stiffnesses and walking speeds. They found that increasing hindfoot stiffness yields decreased prosthesis energy return and increased ground reaction force (GRF) loading rate, knee flexion angle (stance phase), and knee extensor moment. Moreover, they found that increasing forefoot stiffness yields decreased prosthetic ankle and center of mass push-off work, and larger knee extension angle and knee flexor moment in late stance. Also, (Klodd et al., 2010) reported that dorsiflexion angle and ankle plantarflexor moments increased with a stiffer forefoot.

To alter ankle and knee mechanics, the Variable Stiffness Foot (VSF) offers a semi-active foot prosthesis that can modulate its forefoot stiffness during the swing phase. The VSF (Glanzer and Adamczyk, 2018) has a rigid ankle, and its forefoot acts as an overhung beam that can modulate the forefoot stiffness by adjusting a support fulcrum to change the overhang length (parameter a in Fig. 1). The VSF is a semi-active device where power is not supplied to move the body while the foot is on the ground, but minimal power is supplied when the foot is in the air to move the fulcrum, while the motor load is low. This semi-active nature allows the VSF to be lightweight and low power compared to active prostheses, where power is supplied during foot/ground interaction. (Glanzer and Adamczyk, 2018) reported that VSF users displayed greater energy storage and return with lower stiffnesses.

The VSF offers the ability to change forefoot stiffness during movement, so the mechanical effects of the stiffness change and its benefits to different tasks need to be experimentally determined. This study focuses on characterizing the kinetic and kinematic response of participants with transtibial amputation in level-ground walking using the VSF at three different stiffnesses. Based on the findings from fixed component stiffness changes (Adamczyk et al., 2017), we hypothesize that increasing forefoot stiffness will lead to kinematic changes, including decreasing ankle dorsiflexion and increasing stance-phase knee extension angles; joint moment changes, including increasing plantarflexor moment and knee flexor moment; changes in energy and power, including decreasing prosthetic energy storage, energy return, and peak power output; and GRF changes, including increasing second peak of vertical ground reaction force and increasing off-loading rate.

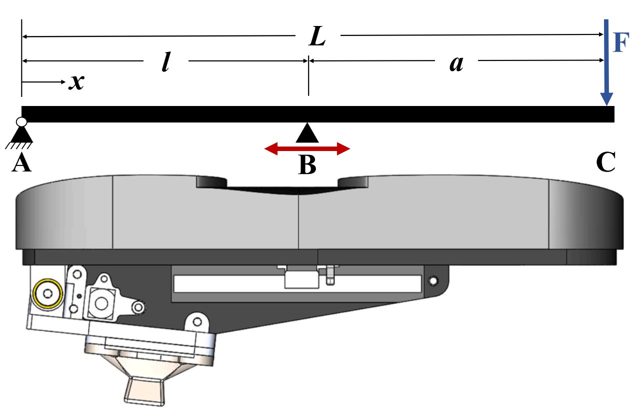

Fig. 1. Side drawing of the Variable Stiffness Foot (VSF) highlighting the cantilever mechanics from Glanzer and Adamczyk, 2018. The ground reaction force on the foot acts at *"F,"* the beam is supported at *"B,"* and pinned at *"A"* © 2018 IEEE.

2. Methods

Seven participants (characteristics shown in Table 1) with unilateral transtibial amputation were included in this study after giving written informed consent according to procedures approved by the University of Wisconsin-Madison Health Sciences Institutional Review Board (protocol #2017–0678). Participants were included under the criteria: having a unilateral transtibial amputation with stable socket fit, being at least six months post-surgery, being able to comfortably walk a minimum of 30 min without aid, and being able to walk comfortably on level ground, stairs, and ramps. Exclusion criteria included the presence of neuromuscular disorders, sores or current injuries, or surgery within the past six months. The VSF was fitted onto the participants and aligned in its Medium stiffness setting (see settings below) by a certified prosthetist. Subjects were given a short (ten-minute) acclimation period in which they walked freely about the laboratory space.

The participants performed three over-ground walking trials for each of the three different VSF stiffness settings. Compliant, Medium, Stiff settings ranged from 13 to 32 N/mm. Body mass was used for scaling the Compliant stiffness to ensure that the minimum stiffness prevented the prosthetic keel from contacting the internal safety limit. The Compliant condition had an estimated mean and standard deviation of 15.4 ± 1.53 N/mm. The Stiff condition was set to be 32 N/mm, and the Medium condition (23.8 ± 0.74 N/mm) was set as the halfway point of the Compliant and Stiff conditions. Body kinematics were recorded using optical motion capture (twelve Optitrack Prime 13 cameras, NaturalPoint, Inc., Corvallis, OR, USA). Marker clusters were placed on each segment of both lower limbs (thighs, shanks, and feet), with additional markers placed on anatomical landmarks (medial and lateral malleoli on the intact ankle, medial and lateral epicondyles bilaterally, bony prominences of the pelvis), and several locations on the VSF prosthesis. Three markers were placed on the pylon/VSF rigid frame, and three more each on the heel and toe areas (center, lateral, and medial). Minor adjustments were made to accommodate differences in individuals' prosthesis componentry. Ground reaction forces were collected using two force plates (Bertec, Inc., Columbus, OH, USA). Motion capture and force plate data were recorded at 200 and 1000 Hz, respectively. All signals were resampled and scaled to 0 to 100 % stance phase, approximately 0 to 60 % stride cycle.

Walking speed was tracked as subjects walked across the force plates using the pelvis segment from motion capture (Fig. 2). At the beginning of the trials, a comfortable target walking speed was selected for each subject (see Table 1), and subjects were asked to maintain that speed consistently. Trials were rejected if the speed exceeded the bounds $\pm$ 0.12 m/s or the feet were not cleanly on force plates. Attempts continued until three successful trials were recorded in each condition. Testing was repeated similarly for the other conditions in randomized order, with five minutes between to rest and then to acclimate to each stiffness setting by walking in the laboratory area. Stiffness settings were concealed from the participant and prosthetist by referring to them only by randomized condition numbers, although subjects could likely perceive the stiffness changes during use.

Standard lower-body joint kinematics and inverse dynamics were computed using a lower-body model in Visual 3D (C-Motion, Inc., Germantown, MD, USA). Functional joint centers for the hips and knees were calculated using the Gillette algorithm (Schwartz and Rozumalski, 2005) to establish the rotation axes. The point along that axis representing the knee joint center was determined as a projection from the midpoint of the medial and lateral epicondyles. An estimated anatomical joint center was used for the intact ankle (midpoint between malleoli). A chosen geometric location was used to specify the VSF ankle joint on the prosthetic side (0.1 m above the floor and 0.05 m anterior to the heel marker). This geometric definition of the VSF ankle was arbitrary because it lacks a true joint rotation axis; it was chosen as a systematic way to define the prosthetic ankle joint for all subjects. Leg length was measured from the floor to the greater trochanter. The segments' masses

Table 1
Subject characteristics with mean and standard deviations of age, sex, amputated side, number of years post-amputation, body mass, body height, leg length, and walking speeds for seven subjects.

Subject	Age (years)	Sex	Amputated side	Number of years post- amputation (years)	Body mass (kg)	Body height (m)	Leg length (m)	Walking speed (m/s)
1	70	M	L	14	83.8	1.8	0.978	1.24
2	61	F	R	8	63.8	1.63	0.874	1.22
3	34	M	R	15	77.3	1.81	0.942	1.25
4	46	M	R	5	104	1.9	1.06	1.42
5	51	M	R	8	111	1.75	0.95	1.18
6	44	M	L	19	75	1.72	0.93	0.679
7	56	M	R	3	105	1.83	0.96	0.653
Average	51.7	–	–	10.3	88.6	1.78	0.956	1.09
Standard deviation	11.9	–	–	5.82	18.1	0.087	0.056	0.301

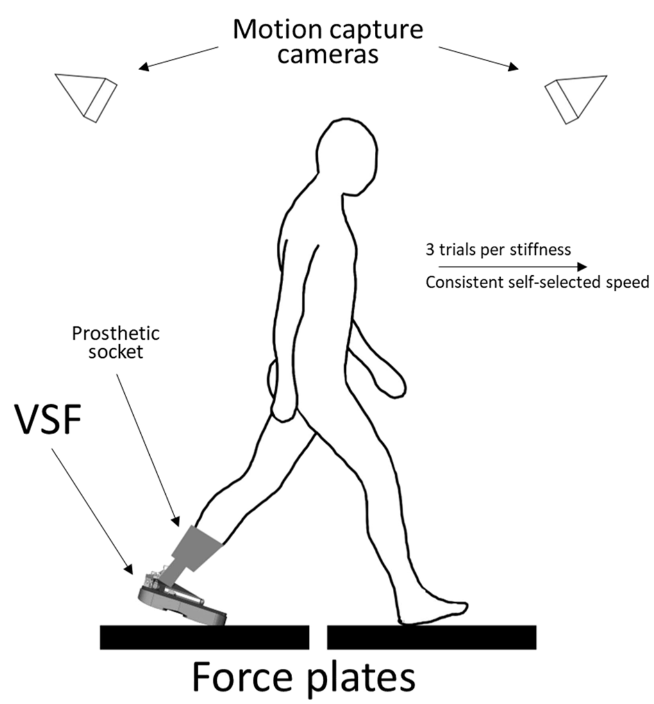

Fig. 2. Side facing diagram showing a person with a transtibial amputation walking with the Variable Stiffness Foot (VSF) across two force plates in a motion capture lab. They walked at a consistent speed for 3 trials per stiffness (total of 9 trials).

were estimated from body mass according to standard anthropometric tables within Visual3D; prosthetic leg mass was not changed from this anthropomorphic assumption. All body segments were modeled as 6-degree-of-freedom rigid bodies (Cappello et al., 1997).

Motion and force data were low-pass filtered using 4th order, bidirectional Butterworth filters with 10 and 25 Hz cutoffs, respectively. Lower-limb joint angles, moments and powers were estimated from standard inverse dynamics calculations. The prosthesis's power and energy absorption and return were calculated using the unified deformable segment model (UD power) (Takahashi et al., 2012) to avoid reliance on the precise ankle joint definition. Energy storage and return in the keel were calculated as the negative and positive parts of the time integral of UD power, excluding initial heel contact (the first 20 % of the stride) to capture only the forefoot contribution. Energy was separated into storage (negative) and return (positive; also known as push-off) regions to investigate these separate consequences of forefoot stiffness modulation.

The results were sorted, processed, and graphed using MATLAB (The Mathworks, Natick, MA USA) in the form of hip, knee, and ankle angles, moments, and powers (Adamczyk et al., 2017). Peak values on the prosthetic side were recorded for the following variables: ankle dorsiflexion angle, ankle plantarflexion moment, midstance knee extension angle, midstance knee flexor moment, UD Power, the second peak of vertical Ground Reaction Force (vGRF), prosthetic energy storage, prosthetic energy return, and vGRF off-loading rate (slope of vGRF vs time between the time-point of 250 N to the 0 N time-point (Adamczyk et al., 2017)). The "midstance" peaks were chosen between 30 % and 55 % of the stride cycle. Each metric was averaged across all trials for each stiffness setting to give one mean value per stiffness for each subject.

Linear mixed-effects (LME) models (Leestma et al., 2021) were used to estimate the sensitivity of these prosthetic-side metrics to stiffness settings, their significance (p-value), and the coefficients of determination (r^2) of their regressions. To investigate the participant-independent effects of the stiffness changes on joint mechanics, the LME model treated the stiffnesses as the fixed effects. The differing offset across participants was modeled as a random effect. Coefficients of determination were adjusted to account only for the explanatory value of the linear term and not the individualized offsets (random participant effects). All outcome measures were redimensionalized (Adamczyk et al., 2017) using an average subject's body mass ($M = 88.6kg$), standing leg length ($L = 0.956m$, greater trochanter to floor), and gravitational acceleration (g, 9.807 m/s/s). We used Mg for forces, MgL for work and moment, $Mg\sqrt{gL}$ for power, and $Mg\sqrt{\frac{g}{L}}$ for force rate of change.

3. Results

Representative joint mechanics data are shown in Fig. 3, and comprehensive statistical results of the LME regression are shown in Table 2 and Fig. 4.

A software fix was implemented to synchronize the data for three subjects due to an unexpected sampling rate change in the instrumentation, where the motion capture and force plate lost synchronization. Based on observed toe-off events from the other subjects, the first vertical ground reaction force frame before toe-off with a magnitude below 10 N was synced with the maximum inferior displacement of the toe marker relative to the foot reference frame, which occurred due to recoil of the elastic keel just as the foot left the ground.

Some trials were rejected during post-processing due to participants' incomplete contact with force plates, missing markers, and faults in the VSF (Subject 6). Subjects 2 – 4 and 6 were missing a trial at the highest stiffness, Subject 6 was missing a trial at the lowest stiffness, and Subject 4 did not complete testing for the lowest stiffness. In total, 55 of the original 63 trials were kept for statistical analyses.

Joint Angle: Results showed that during late stance, the peak dorsiflexion angle significantly decreased with increasing stiffness ($p < 0.0001$, $r^2 = 0.90$). During mid-stance, the peak knee extension angle significantly increased (less negative) with increasing stiffness ($p = 0.021$, $r^2 = 0.37$).

Joint Moment: Results showed that during late stance, the peak plantarflexor ankle moment increased in magnitude (more negative)

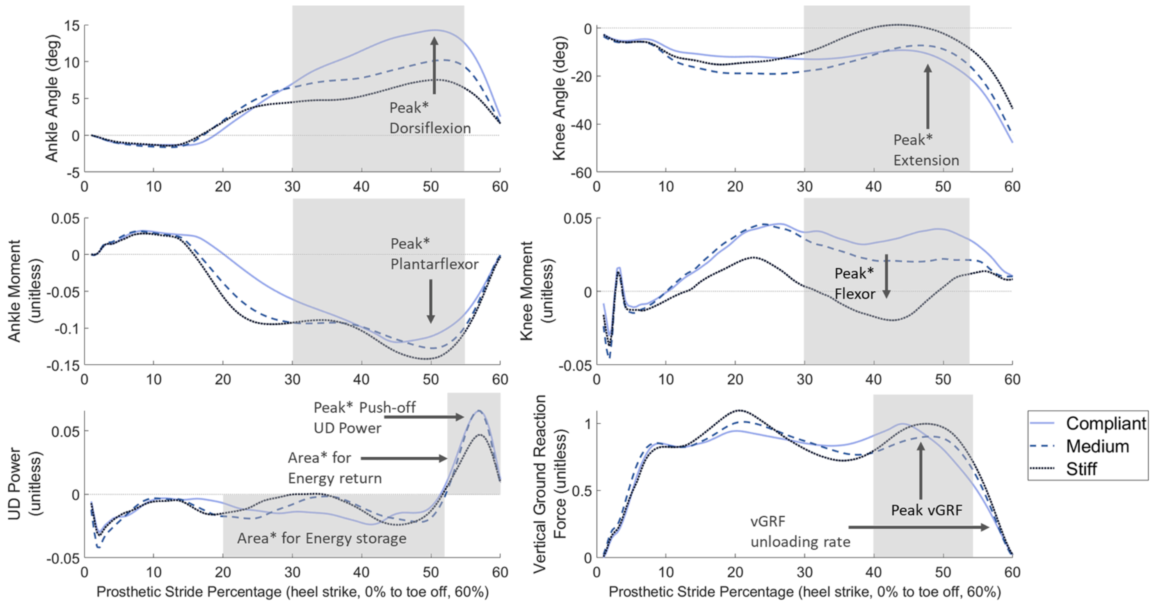

Fig. 3. Prosthetic-side joint mechanics for a representative subject walking with the Variable Stiffness Foot (VSF). First row is prosthetic angles for ankle and knee; second row is the prosthetic moments for the ankle and knee; the third row is the Unified Deformable-body (UD) push-off power and vertical ground reaction force for the prosthetic side. Mean curves across trials for each stiffness condition are plotted, and data is shown for the prosthetic stance phase from heel strike to toe-off. Grey rectangles represent the region of concern for each variable where the peaks or areas are considered. Asterisks (*) were included after the words, “area” and “peak” to highlight the variables which had statistical significance with changes in stiffness.

with increased stiffness ($p = 0.016$, $r^2 = 0.40$). Likewise, peak knee flexor moment during midstance increased in magnitude (more negative) with increasing stiffness ($p = 0.0007$, $r^2 = 0.63$).

Prosthesis energy storage, energy return, and power flow: Results showed significantly decreased magnitudes of energy storage ($p < 0.0001$, $r^2 = 0.90$), energy return ($p = 0.0003$, $r^2 = 0.67$), and peak UD power ($p < 0.0001$, $r^2 = 0.74$) with increasing stiffness.

Ground reaction forces: Results showed no significant dependence on stiffness for the 2nd peak of vGRF ($p = 0.23$, $r^2 = 0.12$) and the off-loading rate of the prosthetic foot ($p = 0.085$, $r^2 = 0.23$).

Hip mechanics: Additional to the hypotheses, prosthetic-side hip mechanics were also calculated to observe the effect of VSF forefoot stiffness. With increasing VSF stiffness, there was a trend toward increasing peak magnitude of hip flexor moment ($p = 0.045$, $r^2 = 0.30$). There were no significant trends for early-stance hip flexion angle, late-stance hip extension angle, late-stance hip power, and net stance-phase hip work.

4. Discussion

The main purpose of this study is to examine the influence of modulating forefoot stiffness on knee and ankle mechanics for prosthetic users wearing the Variable Stiffness Foot (VSF). Results (Fig. 4) showed that increased VSF stiffness led to decreased peak ankle dorsiflexion angle, increased peak plantarflexor moment, increased peak flexor knee moment, increased peak knee extension angle, decreased magnitudes of energy storage, energy return, and peak UD power. These results supported most of our hypotheses. These results imply lower ankle, knee, and hip moments and more ankle angle range of motion using a less stiff VSF. According to these findings, responsive modulation of VSF stiffness could help overcome gait deviations associated with different slopes (e.g. knee hyperextension on uphills or instability on downhills), terrain characteristics (e.g. soft ground), or footwear. In addition, persons with transtibial amputations tend to have problems with knee hyperextension (Adamczyk and Kuo, 2015), so a less stiff VSF can help reduce knee extension.

Previous studies (Adamczyk et al., 2017; Raschke et al., 2015) proposed that active people with transtibial amputations tend to prefer soft (less stiff) prostheses due to the dominance of kinetics over kinematics in ESR interactions, compared to stiffer settings of an adjustable stiffness prosthetic foot (Adamczyk et al., 2017) and other commercial prostheses (Raschke et al., 2015). Also, softer forefoot associated with higher energy return could help with ground clearance due to increased leg swing acceleration (Darter and Wilken, 2014; Wu and Kuo, 2016; Zelik and Adamczyk, 2016).

In the ESR prosthetic literature, three (Glanzer and Adamczyk, 2018; Lecomte et al., 2021; Shepherd and Rouse, 2017) semi-active devices specialize in giving prosthetic users the customizability to change the ankle and forefoot stiffness: Variable Stiffness Prosthetic Ankle (VSPA), Variable Stiffness Ankle (VSA), and Variable Stiffness Foot (VSF). VSPA (Clites et al., 2020; Shepherd and Rouse, 2017) utilizes a cam-based transmission to allow selectable nonlinear ankle torque–angle curves to mimic intact gait and modulate forefoot stiffness through motorized leaf spring configurations. Similarly, the VSA (Lecomte et al., 2021) used a variable leaf-spring design within an ankle module atop a standard flexible prosthesis. Gait testing with both VSPA (Shepherd and Rouse, 2017) and VSA (Ármannsdóttir et al., 2021) demonstrated decreased peak dorsiflexion angles, and the VSPA showed increased peak plantarflexor moment with increased stiffness. Together with the present study and prior studies of fixed-component stiffness changes (Adamczyk et al., 2017), this evidence builds up the mechanistic understanding of how parametric changes in prosthesis properties affect typical walking.

Another important consideration is the VSF’s potential contribution to the musculoskeletal health of its users. The higher push-off work of the lower VSF stiffnesses could contribute to lower first peaks of intact-side knee external adductor moment, potentially leading to a reduced risk of knee osteoarthritis (Baliunas et al., 2002; Morgenroth et al., 2011), which is a major issue for persons with unilateral amputation (Struyf et al., 2009). Prosthesis users also face difficulties with gait symmetry (Dingwell et al., 1996). Asymmetric gait for persons with unilateral amputation may be unavoidable because passive prostheses cannot completely replace the lost prosthetic-side push-off work (Adamczyk and Kuo, 2015); but, fortunately, these compensations can be reduced. Preferred stiffness using the VSF could increase the kinematic symmetry between the intact and prosthetic sides, as shown with the VSPA (Clites et al., 2021). Finally, the VSF could have beneficial

Table 2
Assessed biomechanical metrics with their associated means for each stiffness (Compliant, Medium, Stiff), slope of the Linear Mixed Effects regression against stiffness (the slope indicates the change in the relevant quantity per level of stiffness), mean intercept across subjects, adjusted r^2, and p-value. GRF represents Ground Reaction Force, SD represents Standard deviation, and UD represents Unified Deformable-body. Results are presented in dimensionless form and redimensionalized according to average subject characteristics.

Prosthetic Side		Non-dimensionlized										Redimensionlized								
	Peak	Comp. Mean	Comp. SD	Med. Mean	Med. SD	Stiff Mean	Stiff SD	Slope	Intercept	Adjusted r2	p-val of slope	Units	Comp. Mean	Comp. SD	Med. Mean	Med. SD	Stiff Mean	Stiff SD	Slope	Intercept
Ankle	Angle (dorsiflexion +)	9.41	2.35	7.8	2.04	5.46	1.9	−2.2	12.1	0.899	**<0.0001**	Deg	9.41	2.35	7.8	2.04	5.46	1.9	−2.2	12.1
	Moment (dorsiflexor +)	−0.148	0.0218	−0.148	0.0229	−0.154	0.0224	−0.00582	−0.136	0.396	**0.016**	Nm	−123	18.1	−123	19	−128	18.6	−4.84	−113
Knee	Angle (extension +)	−5.19	4.76	−3.7	6.28	−1.82	6.72	1.95	−7.63	0.372	**0.0205**	Deg	−5.19	4.76	−3.7	6.28	−1.82	6.72	1.95	−7.63
	Moment (extensor +)	0.00079	0.0176	−0.00806	0.0199	−0.0199	0.0235	−0.0114	0.0144	0.632	**0.0007**	Nm	0.653	14.6	−6.69	16.6	−16.5	19.5	−9.45	12
GRF	Vertical 2nd peak	0.959	0.0263	0.935	0.0667	0.959	0.0832	0.0124	0.918	0.12	0.225	N	833	22.8	812	57.9	833	72.3	10.8	797
	Vertical Off-loading Rate	−7.46	2.82	−8.02	2.73	−8.4	3.21	−0.33	−7.39	0.227	0.0848	N/s	−20800	7850	−22300	7590	−23400	8930	−918	−20600
UD	Energy Storage	−0.0133	0.00387	−0.00937	0.0043	−0.00671	0.0031	0.0029	−0.0153	0.903	**<0.0001**	J	−36.9	3.21	−7.78	3.53	−5.57	2.56	2.4	−12.7
	Energy Return	0.0174	0.0055	0.0177	0.0059	0.0142	0.0055	−0.00232	0.0216	0.672	**0.0003**	J	48.3	4.57	14.7	4.93	11.8	4.54	−1.93	17.9
	Push-off Power	0.0559	0.0203	0.0568	0.0259	0.0453	0.0245	−0.00921	0.0736	0.739	**<0.0001**	W	156	56.5	158	72.1	126	68.2	−25.6	205

effects in mitigating knee hyperextension or instability on up- or down-slopes, respectively. Knee hyperextension can be caused by high socket moments when the uphill ground moves the center of pressure (COP) toward the toes (Leestma et al., 2021) and can be uncomfortable and potentially injurious to the prosthetic-side knee. Softening the VSF may help to limit this effect. On the other hand, down-slope gait can keep the COP too near the heel (Leestma et al., 2021), forcing the knee into flexion, which many persons with amputation find difficult to control and which may lead to falls (Williams et al., 2006). Stiffening the VSF forefoot can help stabilize against such instability. These potential benefits require not just prostheses of different stiffnesses, but rather the ability to actively modulate stiffness as different environments are encountered, as enabled by the VSF and related technologies.

Effective control of the stiffness modulation can be connected to the Dynamic Mean Ankle Moment Arm (DMAMA) (Adamczyk, 2020). DMAMA represents the ratio of sagittal ankle moment impulse to sagittal ground reaction force impulse in stance phase, capturing how forefoot-dominated the ground interaction is. Tests (Adamczyk, 2020) of DMAMA showed that in natural gait, GRF impulse moves closer to the ankle (less forward) with increased walking speed. As a follow-up, (Leestma et al., 2021) analyzed companion data from level-walking, ramps, and stairs in a subset of the present study's subjects, and found a positive linear sensitivity of DMAMA (more forefoot dominated) to both forefoot stiffness and ground incline while using the VSF. Thus, stiffness control of the VSF could target DMAMA as a control loop variable to adapt to walking on level-ground, ramps, stairs, and potentially standing.

It is important to investigate whether the more capable weight support of a stiffer foot or larger energy return of a compliant foot are of contrasting benefit when deciding optimal stiffness in walking. While the energy return is usually considered good and users prefer more compliant prostheses (Fey et al., 2011), there can also be a "drop-off" effect if the foot is too compliant (Klodd et al., 2010). This effect can appear as a low "effective foot length ratio" (EFLR) that fails to mimic an intact limb appropriately. For example, prostheses with short or very soft internal keels have a low EFLR, while those with longer or stiffer keels achieve EFLR closer to the physiological value (Hansen et al., 2004). This trade-off is fundamental to the mechanics of a passive compliant prosthesis, so the right choice or balance may come down to individuals' preferences or the use of semi-active, adaptable prostheses like the VSF to adjust the behavior for different circumstances. Simultaneously achieving both the low impedance and high energy return of a compliant prosthesis and the firm weight support of a stiffer foot may fundamentally require a powered foot–ankle system or a more complex mechanism such as energy recycling (Collins and Kuo, 2010; Segal et al., 2011; Shepherd and Rouse, 2017).

Beyond the prosthetic energy flow, whole body metrics like energy expenditure and center of mass mechanics (COM) must also be considered. (Zelik and Adamczyk, 2016) showed a unified effect of ankle energy return that contributes to both leg swing and COM acceleration. Ankle energy return is a defining characteristic of the VSF since stiffness directly affects push-off power and work. (Zelik et al., 2011) and (Segal et al., 2012) found that lower stiffnesses led to higher energy storage, energy return, and prosthetic limb center of mass (COM) push-off work in a semi-active energy-recycling prosthesis, the controlled energy storage and return (CESR) prosthesis. They showed that using CESR with low stiffness increased prosthetic push-off (energy return) and decreased intact limb COM collision work compared to using a conventional foot. Walking energy expenditure in their participants with TTA was lowest for intermediate stiffness, suggesting biomechanical disadvantages to low stiffness despite higher energy return. Agreeing with previous results from (Adamczyk and Kuo, 2015), people with TTA showed higher hip work with lower energy transfer from the prosthesis to the COM, which could be attributed to higher energy dissipation at the knee. These results showed that spring stiffness influences push-off but has a co-occurring biomechanical trade-off that limits how push-off can benefit

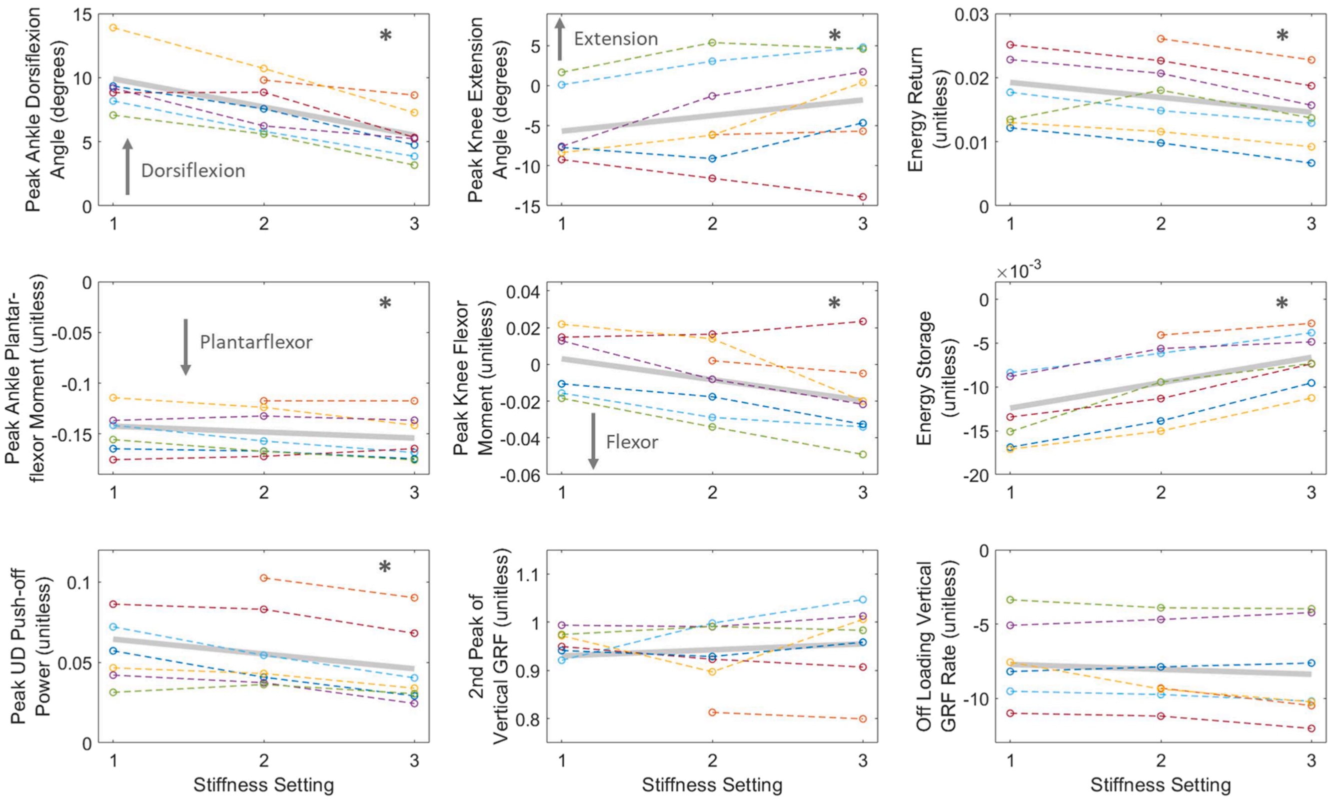

Fig. 4. Linear mixed-effect regressions for ankle and knee angles and moments, Unified Deformable-body (UD) power, prosthesis energy storage and energy return, vertical Ground Reaction Force (vGRF), and vGRF offloading rate for the non-dimensionalized data, with respect to stiffness settings. Stiffness settings 1, 2, 3 relate to Compliant, Medium, and Stiff stiffnesses, respectively. The grey bar represents the linear mixed effects fit of all subjects. One subject did not have data for the lowest stiffness setting. Asterisks (*) were included in the top right area for the variables which had statistical significance with changes in stiffness.

walking economy. Additionally,(Clites et al., 2020) found that subject-preferred stiffness did not correlate with energy expenditure but tended to be lower for self-selected speed than fast or slow speeds.

The ability to control properties without varying components may allow the VSF and similar systems to support the development of predictive computational models of gait mechanics. Through a combination of mechanical modeling and user testing in catch-trial and adaptation protocols, models could be developed to anticipate how users will respond to changes and perhaps how they will select optimal settings. Already, we have incorporated the VSF into a computational contact model (McGeehan et al., 2021a), a study predicting the resultant and center of pressure of the ground reaction force (McGeehan et al., 2021b) under stiffness changes, and a study that predicts the normal pressure and shear stress of the residual limb-socket interface with changing stiffness (McGeehan et al., 2022).

Improvements to the experimental design of this research study could include increasing the number of participants, having more trials per stiffness category, and giving participants more acclimation time with the VSF for each stiffness. Another practical limitation is the fixed foot length of the VSF; the same foot length was used for all subjects despite varying body masses and heights. Technologically, the VSF achieved a very high range of stiffness modulation, but only at the forefoot; a future step may be to build a second keel and carriage to allow for heel stiffness modulation, improving heel strike and energy storage in the early stance. Also, combining the VSF function with a repositionable ankle (e.g., Ossur Proprio) could help with swing-phase ankle dorsiflexion to improve toe clearance and minimize tripping. A further limitation involves the wide range of our participants' self-selected walking speeds and the potential influence of walking speed on lower body mechanics. An alternative experiment with controlled speeds (Adamczyk et al., 2017) would have emphasized mechanistic effects, but we used self-selected speeds to observe the effects on participants' most realistic use conditions. The analysis using a Linear Mixed-Effect Model partially accounts for this variation by accommodating speed-related shifts in the outcome measures. Where possible, future research on the effects of prosthesis properties should include both controlled and self-selected speeds to include both perspectives.

These improvements will depend on control, so further work must focus on real-time algorithms for gait measurement and stiffness modulation across various locomotive conditions. Using biomimetic, experimentally-optimized target values of DMAMA, or the joint mechanics measured in this study, embedded sensors (e.g. inertial sensors (Glanzer and Adamczyk, 2018) or load cells (Leestma et al., 2021)) could feed a control loop that iteratively modulates stiffness for optimal gait mechanics. Upgrades to the VSF feedback controller could include detecting previously validated kinematic variations (Kitagawa and Ogihara, 2016; Li et al., 2021; Washabaugh et al., 2017) while walking on and transitioning among various speeds, ramps, stairs, and standing still.

5. Conclusion

The variable stiffness foot (VSF) enables biomechanical knee and ankle adjustments through controlled modulations of forefoot stiffness. Mainly, ankle dorsiflexion angle and plantarflexor moment; knee extension angle and extensor moment; and unified deformable energy storage, energy return, and power were affected by changes in forefoot stiffness of the VSF. Future closed-loop control of VSF stiffness could be used to achieve continuous real-time modulation for optimized gait mechanics.

CRediT authorship contribution statement

Kieran M. Nichols: Visualization, Validation, Supervision, Software, Project administration, Methodology, Investigation, Formal analysis, Data curation, Writing – original draft, Writing – review & editing. **Peter G. Adamczyk:** Visualization, Validation, Supervision, Software, Project administration, Methodology, Investigation, Formal analysis, Data curation, Writing – original draft, Writing – review & editing.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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