FLEX CEUs

Running Considerations with Amputation

Characterizing the Mechanical Properties of Running-Specific Prostheses

Abstract

The mechanical stiffness of running-specific prostheses likely affects the functional abilities of athletes with leg amputations. However, each prosthetic manufacturer recommends prostheses based on subjective stiffness categories rather than performance based metrics. The actual mechanical stiffness values of running-specific prostheses (i.e. kN/m) are unknown. Consequently, we sought to characterize and disseminate the stiffness values of running-specific prostheses so that researchers, clinicians, and athletes can objectively evaluate prosthetic function. We characterized the stiffness values of 55 running-specific prostheses across various models, stiffness categories, and heights using forces and angles representative of those measured from athletes with transtibial amputations during running. Characterizing prosthetic force-displacement profiles with a 2nd degree polynomial explained 4.4% more of the variance than a linear function ($p<0.001$). The prosthetic stiffness values of manufacturer recommended stiffness categories varied between prosthetic models ($p<0.001$). Also, prosthetic stiffness was 10% to 39% less at angles typical of running 3 m/s and 6 m/s (10°-25°) compared to neutral (0°) ($p<0.001$). Furthermore, prosthetic stiffness was inversely related to height in J-shaped ($p<0.001$), but not C-shaped, prostheses. Running-specific prostheses should be tested under the demands of the respective activity in order to derive relevant characterizations of stiffness and function. In all, our results indicate that when athletes with leg amputations alter prosthetic model, height, and/or sagittal plane alignment, their prosthetic stiffness profiles also change; therefore variations in comfort, performance, etc. may be indirectly due to altered stiffness.

Introduction

Running is a bouncing gait that is well-characterized by a spring-mass model [1–3]. The spring-mass model portrays the stance leg as a mass-less linear spring supporting a point mass representing the runner's center of mass. Upon ground contact, the leg spring compresses and stores elastic energy until mid-stance, and then returns mechanical energy from mid-stance through the end of ground contact [4]. In this model, the leg spring is completely elastic, however the structures of a biological leg are viscoelastic and therefore only a portion of the stored potential elastic energy is returned (due to hysteresis). The spring-like action of the leg conserves a portion of the runner's mechanical energy, theoretically mitigating the additional muscular force and mechanical energy input necessary to maintain running speed [4,5]. The magnitude of the stored and returned mechanical energy is inversely related to leg stiffness (resistance to compression), and is influenced by the magnitude and orientation of the external force vector acting on the leg [1]. Simply modeled as a linear spring, leg stiffness (k_{leg}) equals the quotient of the peak applied force (F) and the change in leg length (Δl) from touchdown to mid-stance [2]:

$$k_{leg} = \frac{F}{\Delta l} \tag{1}$$

Inspired by the spring-like nature of running, passive-elastic running-specific prostheses (RSPs) were developed to enable athletes with lower-limb amputations to run. These carbon-fiber devices are attached to the sockets that encompass the residual limbs, are in-series with the residual limbs, and mimic the mechanical energy storage and return of tendons during ground contact. Unlike biological ankles, RSPs cannot generate mechanical power anew and only return 63% to 95% of the stored elastic energy during running [6–8]. For context, biological ankles generate mechanical power through use of elastic structures as well as muscles, and thus appear to "return" 241% of the energy stored while running at 2.8 m/s [7].

Athletes with leg amputations may adapt similar leg spring mechanics as non-amputees by using RSPs that emulate biological lower leg stiffness. Individually, non-amputees adopt a constant [2,9,10], metabolically optimal leg stiffness during running [11–13]. Non-amputee runners maintain leg stiffness across speeds by exhibiting constant ankle joint stiffness (sagittal plane torsional stiffness) [14,15]. It has been assumed that prosthetic stiffness is also constant across speeds [8,16–18], which if true, RSPs would act like that of biological ankles [14,15]. Yet, McGowan et al. [16] reported that the affected leg stiffness of athletes with transtibial amputations decreases as speed increases from 3.0 m/s to top speed (the range of top speeds achieved were 7.0 m/s to 10.8 m/s), indicating that prosthetic stiffness and/or affected leg knee stiffness may be inversely related with speed. Moreover, Dyer et al. [19] mechanically tested two Elite Blade RSPs (Chas A Blatchford & Sons Ltd. Basingstoke, UK) in a materials testing machine and reported that the RSPs have curvilinear force-displacement profiles, suggesting that prosthetic stiffness is non-constant and force dependent. Due to conflicting evidence in the literature, coupled with insufficient information provided by manufacturers regarding prosthetic stiffness profiles, it is unknown whether the force-displacement profiles of RSPs are linear, or curvilinear, which would infer that stiffness is contingent upon the applied force magnitude.

Prosthetic manufacturers do not report the stiffness values of RSPs (e.g. in kN/m). Instead, they classify RSPs into predetermined stiffness categories (e.g. categories 1 to 7), which are recommended to users based on body mass and intended activity (slow or fast running) [20–22]. Larger/heavier athletes with amputations are generally prescribed RSPs with numerically greater stiffness categories, which are presumably stiffer than numerically lower stiffness categories. Additionally, some prosthetic models are recommend at greater stiffness categories for fast running than for slow running [20,21], whereas other models are recommended at the same stiffness category irrespective of intended running speed [23,24]. These inconsistencies in prosthetic stiffness recommendations persist despite the potential influence of stiffness on running mechanics and performance. Therefore, it is imperative to quantify and disseminate stiffness values to further understand prosthetic function.

To accurately quantify prosthetic stiffness, it seems obvious to evaluate RSPs using forces and angles indicative of those produced during the respective activity. When athletes with transtibial amputations run, they generate peak vertical ground reaction forces (GRFs) with their affected legs that are 2.1 to 3.3 times body weight at speeds of 2.5 m/s to 10.8 m/s [8,18,25,26]. During running, peak resultant GRFs typically occur around mid-stance and are oriented vertically. At the same instant, the proximal end of the stance leg's RSP is rotated forward in the sagittal plane relative to the peak resultant GRF vector. Therefore, the proximal bending moment acting on shorter RSPs may be less than that on taller RSPs for a given applied force, due to a reduced moment arm length. A smaller moment (torque) associated with shorter RSPs may reduce vertical displacement, and in turn increase prosthetic stiffness. Nonetheless, the peak resultant GRF magnitudes and sagittal plane orientations relative to RSPs are unknown, as is the influence of prosthetic height on stiffness.

Since prosthetic stiffness and hysteresis likely affect running performance, we aimed to 1) characterize the force-displacement profiles of RSPs, 2) quantify and compare prosthetic stiffness and 3) hysteresis values across prosthetic models, stiffness categories, and heights using angles and forces that replicate those exhibited during running, and 4) determine whether prosthetic height affects stiffness. Such information will enable accurate and objective comparisons between RSPs, subsequently allowing for potential improvements in prosthetic design, prescription, and athletic performance. Based on the predominant assumption that prosthetic stiffness is constant during running [8,16–18]; we hypothesized that the force-displacement profiles of RSPs would be linear. We hypothesized that for a given body mass and running speed, manufacturer recommended prosthetic stiffness would be similar between models. We also hypothesized that the magnitude of prosthetic hysteresis would not differ across testing conditions. Lastly, we hypothesized that shorter RSPs would be stiffer than taller RSPs.

Methods

Testing Procedure

We measured GRFs and sagittal plane angles of RSPs relative to the peak resultant GRFs from 11 athletes (5 males and 6 females; mean ± SD; age: 27.8 ± 5.7; standing height: 1.74 ± 0.08 m; body mass: 68.9 ± 15.3 kg) with unilateral transtibial amputations while they ran at 3 m/s and 6 m/s on a force-measuring treadmill. Each athlete used their own personal RSP. 3 m/s represents a typical distance running speed [27–29] and 6 m/s represents the fastest speed that all of our participants could achieve. The Intermountain Healthcare IRB, Colorado Multiple IRB, and the USAMRMC Office of Research Protection, Human Research Protection Office approved this study. Prior to participating, nine athletes provided informed written consent in accordance with the Intermountain Healthcare IRB and two participants provided informed written consent in accordance with the Colorado Multiple IRB and USAMRMC Office of Research Protection, Human Research Protection Office. Data collection took place in two separate labs.

We placed reflective markers on the lateral proximal and distal ends of each RSP's longitudinal axis and measured segment motion during each trial using a motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA, or Vicon Nexus, Oxford, UK) at 240 Hz (lab 1) or 200 Hz (lab 2) and implemented a 4th order low-pass Butterworth filter with a cutoff frequency of 6 Hz (Visual 3D, C-motion, Inc., Germantown, MD, USA) (Fig 1). The longitudinal axis was defined by a line through the center of the pylon connecting each socket to the corresponding C-shaped RSP, and along the center of the proximal, longitudinal section of each J-shaped RSP (Fig 1). Four athletes used a C-shaped RSP, and seven used a J-shaped RSP. We recorded GRFs via force-measuring treadmills (Treadmetrix, Park City, UT, USA) at 2400 Hz (lab 1) or 1000 Hz (lab 2) and applied a 4th order low-pass Butterworth filter with a cutoff frequency of 30 Hz using a custom MATLAB script (MathWorks Inc, Natick, MA, USA). Our

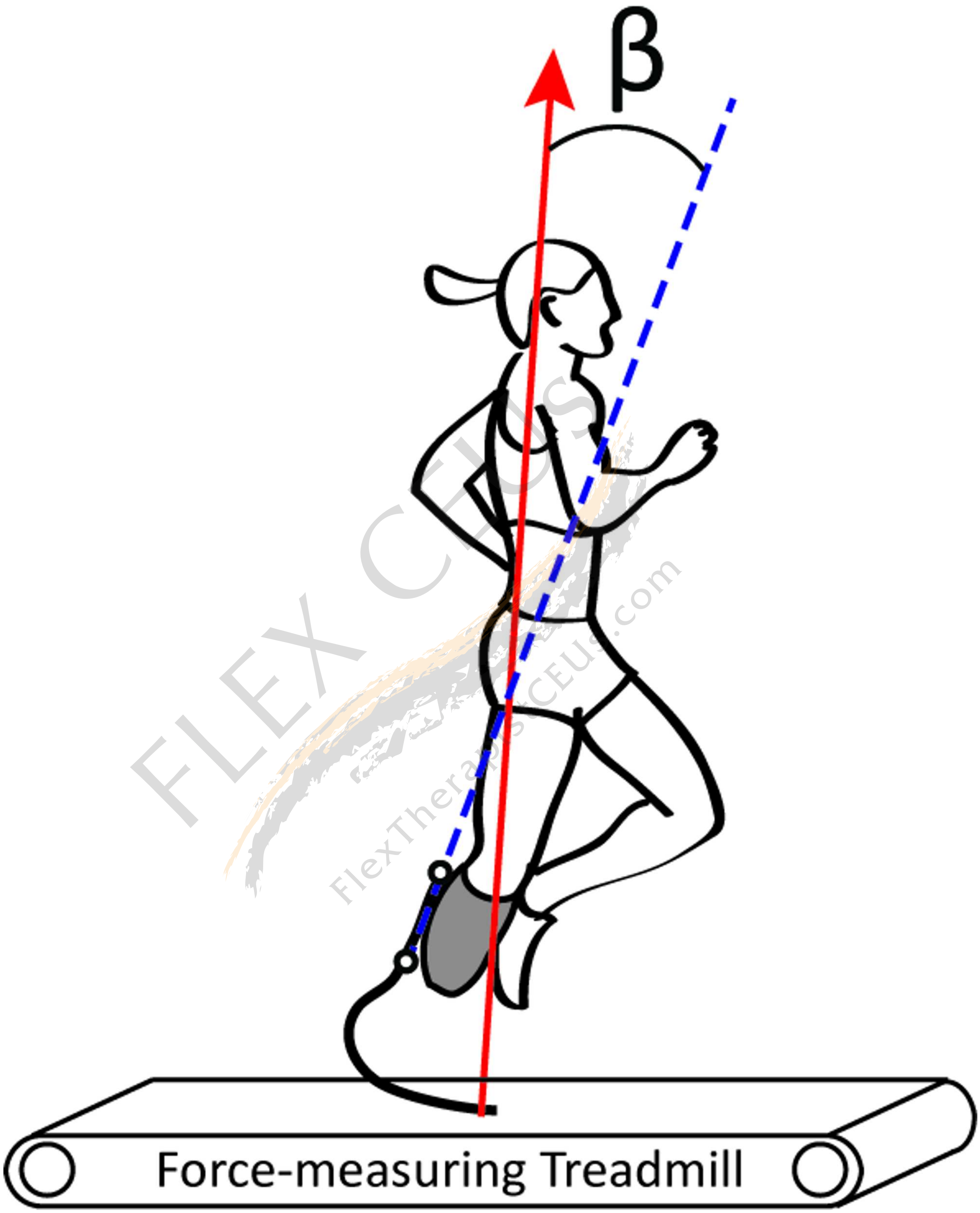

Fig 1. Biomechanics of running. Illustration of the calculated angle (β) between the longitudinal axis of the running-specific prosthesis (dashed blue line) and the peak resultant GRF vector (solid red arrow).

data were comparable because each participant ran both speeds at one lab, and due to the implementation of the same filtering process.

We determined the peak GRF magnitude, as well as the average sagittal plane angle of the longitudinal axis for each athlete's RSP relative to the peak resultant GRF vector from 10 consecutive ground contacts with the affected leg. We assessed the average angles for trials performed with C-shaped RSPs at 3 m/s (α_3) and 6 m/s (α_6), and with J-shaped RSPs at 3 m/s (β_3) and 6 m/s (β_6). When the RSP's longitudinal axis is parallel to the peak resultant GRF vector, the RSP is at 0˚. Positive angles indicate that the proximal longitudinal axis was rotated forward in the sagittal plane relative to the peak resultant GRF vector (Fig 1). Sequentially, we implemented the measured angles (α_3, α_6, β_3, and β_6) and peak resultant GRF magnitudes into our prosthetic testing procedure.

Running-Specific Prostheses

Three prosthetic manufacturers, Össur (Reykjavik, Iceland), Freedom Innovations (Irvine, CA, USA), and Ottobock (Duderstadt, Germany) donated a combined total of 55 RSPs for use in our study. We characterized prosthetic stiffness profiles and hysteresis magnitudes from 14 C-shaped Össur Flex-Run prostheses (stiffness categories 3 low– 7 high), 12 C-shaped Freedom Innovations Catapult FX6 prostheses (stiffness categories 2–7), 14 J-shaped Ottobock 1E90 Sprinter prostheses (stiffness categories 1–5), and 15 J-shaped Össur Cheetah Xtend prostheses (stiffness categories 2–7) (Fig 2) (Table 1). The unique design of the Catapult prosthesis allows for stiffness modifications via interchangeable carbon-fiber supports (PowerSprings) that are designed to supplement overall stiffness [20] (Fig 2). PowerSprings have designated stiffness categories based on the manufacturer's categorization. We tested each Catapult with the PowerSpring of the matching stiffness category (e.g. a category 2 Catapult with a category 2 PowerSpring).

Stiffness Testing

To assess prosthetic stiffness and hysteresis at conditions that matched those of our analyzed running data, we fabricated an aluminum attachment to secure the RSPs on to the force transducer of our materials testing machine (Instron Series 5859, Norwood, MA, USA) (Fig 2). We also constructed an aluminum rotating base and fixed it under each C-shaped RSP at 0˚, α_3, and α_6, as well as under each J-shaped RSP at 0˚, β_3, and β_6 (Fig 2). We applied three successive loading and unloading cycles at 100 N/s on each RSP for each condition. This loading rate was relatively fast and ensured that our materials testing machine operated within the safe speed range, even with our most compliant RSPs. Three compressive loading and unloading cycles matched the number of cycles from Brüggeman et al. [8].

To determine the peak GRF magnitude applied on each RSP, we considered the heaviest manufacturer recommended body weight for each prosthetic stiffness category, then multiplied it by 3.0 to replicate the upper limit of peak GRFs typically produced by affected legs while running 3 m/s [16], and by 3.5 to replicate the upper limit of peak GRFs produced by affected legs while running 6 m/s [16]. We compared the effects of testing angle and prosthetic height on stiffness and hysteresis by evaluating prosthetic compression with an applied peak resultant GRF of 3.0 times the largest recommended body weight for each RSP. We minimized shearing forces by using a low-friction roller-system beneath each RSP that allowed anterior and posterior translation while maintaining the angle of the applied force relative to the longitudinal axis (Fig 2) [30]. We set the threshold for force detection at 10 N. We recorded applied force magnitudes and prosthetic displacement measurements at 10 Hz, which, when combined

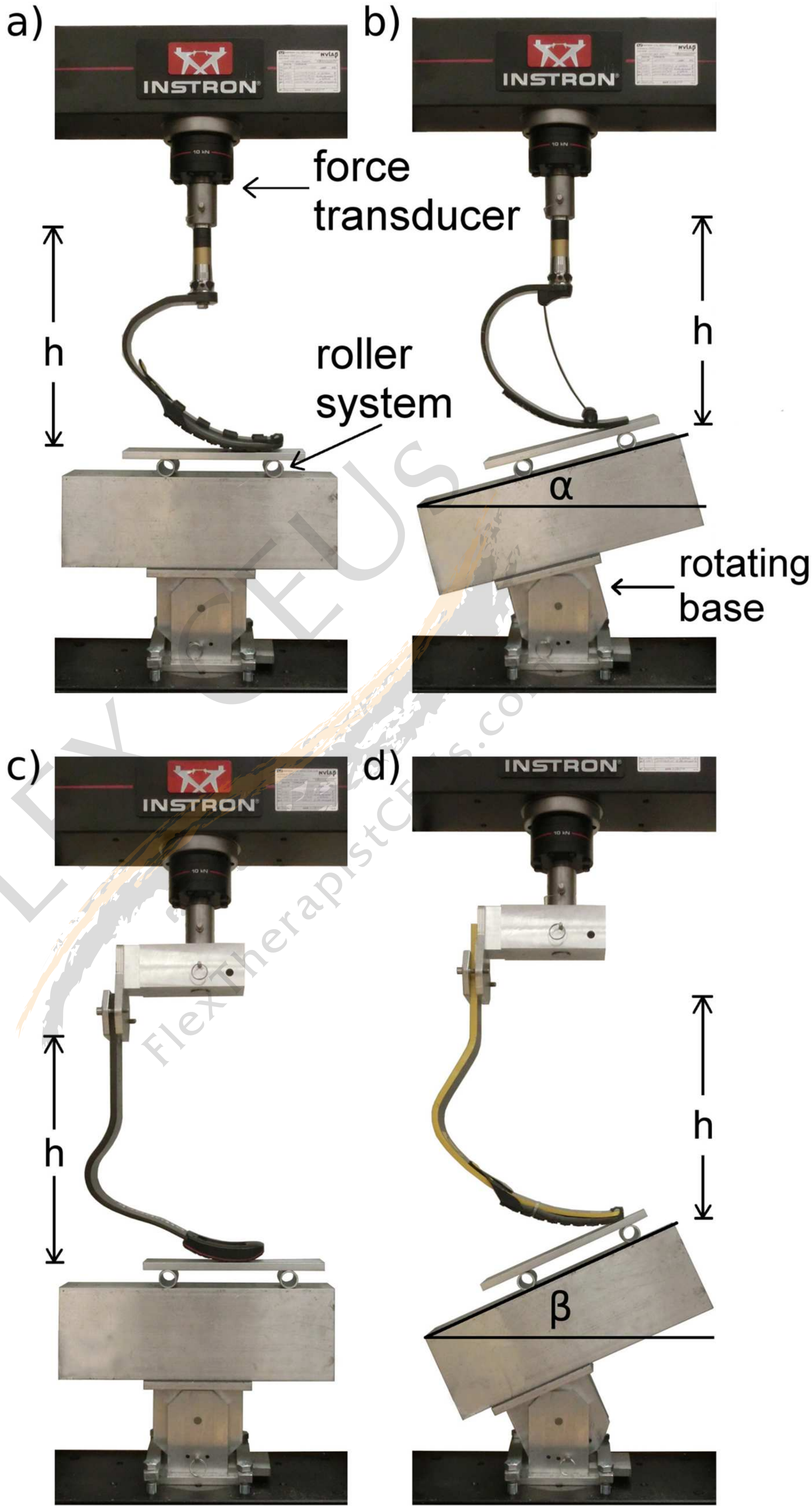
a)
b)
INSTRON
force
transducer
h
roller
system
α
rotating
base
c)
d)
β

Fig 2. Material testing setup with each running specific-prosthetic model. Each running specific-prosthesis (RSP) was tested with the respective manufacturer's rubber sole (Össur Cheetah Xtend prosthesis was equipped with an Össur Flex-Run's sole), using our rotating base, and low-friction roller system. a) An Össur Flex-Run prosthesis (C-shaped) tested at 0˚. b) A Freedom Innovations Catapult prosthesis (C-shaped) tested at α˚ (3 m/s). c) An Ottobock 1E90 Sprinter prosthesis (J-shaped) tested at neutral (0˚). d) An Össur Cheetah Xtend prosthesis (J-shaped) tested at β˚ (6 m/s). h indicates prosthetic height.

with the loading rate (100 N/s), allowed the measurement of force-displacement data from every 10 N of applied force; ~150 to 400 data points per loading cycle.

To determine the effect of prosthetic height on the stiffness of C-shaped RSPs, we tested the Catapult and Flex-Run prostheses at 38.2 cm and 69.7 cm by altering the aluminum pylon height. To determine the effect of height on the stiffness of J-shaped RSPs, we tested the 1E90 Sprinter prostheses at 25.0, 31.5, and 38.0 cm, and the Cheetah Xtend prostheses at 31.5, 38.0, and 41.5 cm. Prosthetic height was measured vertically from the ground to the base of our height adjustment attachment in an unloaded state (Fig 2). We chose to test C-shaped RSPs across the largest possible height range given our components. We tested J-shaped RSPs at heights that spanned the largest possible range while allowing matched height comparisons (31.5 cm and 38.0 cm) between different models.

Table 1. The manufacturer recommended running-specific prosthesis (RSP) stiffness categories with the corresponding body mass for distance running and sprinting, plus the quantity of RSPs tested.

RSP Model	Stiffness Category	Body Mass (kg)		Quantity of RSPs
		Distance Running	Sprinting	
Össur Flex-Run	3 Low	53–56	N/A	1
	3 High	56–59		1
	4 Low	60–64.5		1
	4 High	64.5–68		2
	5 Low	69–73		1
	5 High	73–77		2
	6 Low	78–83		1
	6 High	83–88		2
	7 Low	89–94.5		1
	7 High	94.5–100		2
Freedom Innovations Catapult FX6	2	53–59	N/A	2
	3	60–68		2
	4	69–77		2
	5	78–88		2
	6	89–100		2
	7	101–116		2
Ottobock 1E90 Sprinter	1	40–59	40–52	3
	2	60–70	53–63	3
	3	71–86	64–79	3
	4	87–102	80–95	3
	5	103–118	96–111	3
Össur Cheetah Xtend	2	53–59	53–59	2
	3	60–68	60–68	2
	4	69–77	69–77	3
	5	78–88	78–88	3
	6	89–100	89–100	3
	7	101–116	101–116	2

Analyses

To characterize prosthetic stiffness, we calculated the average coefficients of determination (R^2) for linear and curvilinear characterizations of the applied force relative to the vertical displacement for each 3-cycle trial. Next, we averaged R^2 values within and across trials for a given prosthetic model, stiffness category, height, and testing angle combination. Furthermore, we calculated average prosthetic stiffness for each model across stiffness categories using the force-displacement function during simulated running conditions.

For every cycle, we calculated hysteresis as the ratio of energy lost during recoil relative to the energy stored during compression, then expressed it as a percentage:

$$Hysteresis = \frac{\int_o^H F(h)dh - \int_H^o F(h)dh}{\int_o^H F(h)dh} \times 100 \tag{2}$$

where F is the applied force as a function of the change in prosthetic height (h) and peak change in prosthetic height (H) of the corresponding cycle. Hysteresis was averaged for each 3-cycle trial, and averaged across trials of the same prosthetic model, stiffness category, height, and testing angle. We measured prosthetic stiffness and hysteresis with the respective manufacturers supplied rubber sole. We also measured the stiffness and hysteresis of the highest stiffness category from each model at 0° without the rubber sole.

Statistical Analyses

We used paired two-tailed t-tests to compare average R^2 values from linear and curvilinear force-displacement functions across prosthetic models and to compare the manufacturer recommended stiffness across prosthetic models for athletes at body masses of 55 kg to 100 kg in 5 kg increments using the average angles and peak applied force magnitudes produced at 3 m/s (α_3 and β_3) from the C- and J-shaped RSPs, respectively. We also used paired two-tailed t-tests to compare the prescribed stiffness of different prosthetic models for athletes at body masses of 55 kg to 100 kg in 5 kg increments using the average angles and peak applied force magnitudes produced at 6 m/s (α_6 and β_6) from the C- and J-shaped RSPs, respectively. The recommended stiffness values for J-shaped RSPs were calculated using the tallest mutual height (38 cm).

Moreover, for C-shaped RSPs, we used linear mixed models to compare 1) prosthetic stiffness and 2) hysteresis for each prosthetic model across stiffness categories, testing angles, and interaction effects. For the J-shaped RSPs we included prosthetic height as an independent variable and used two linear mixed models to compare 1) prosthetic stiffness and 2) hysteresis for each prosthetic model across stiffness categories, testing angles, and heights, in addition to their interactions. We performed paired two-tailed t-tests to assess the influence of the prosthetic sole on stiffness and hysteresis. We carried out our statistical analyses using R-studio (Boston, MA, USA) software. Significance was set at $p<0.05$. When applicable, we implemented the Bonferroni correction to account for multiple comparisons.

Results

Subject Data

When participants used C-shaped RSPs to run 3 m/s, the average angle of their RSP's longitudinal axis relative to the peak resultant GRF was 15.1° ± 4.8° and the mean peak resultant GRF was 2.5 ± 0.3 times body weight. At 6 m/s the average angle was 10.0° ± 4.2° and the peak resultant GRF was 2.7 ± 0.3 times body weight. When participants used a J-shaped RSP to run 3 m/s, the average angle of their RSP's longitudinal axis relative to the peak resultant GRF was 20.9°

± 8.9˚ while the average peak resultant GRF was 2.6 ± 0.3 times body weight. At 6 m/s, the average angle was 24.2˚ ± 9.3˚ and the average peak resultant GRF magnitude was 2.8 ± 0.3 times body weight. Since our custom base was constructed to rotate in incremental steps, we used the following values for RSP testing: α_3 = 15.0˚, α_6 = 10.0˚, β_3 = 20.0˚, and β_6 = 25.0˚.

Prosthetic force-displacement characteristics

Overall, characterizing the slope of the force-displacement curves with a 2nd degree polynomial explained 4.4% more of the variance than a linear function using angles indicative of 3 m/s and 6 m/s ($p<0.001$) (Fig 3). At a testing angle of 0˚, a 2nd degree polynomial explained 5.0% more of the variance than using a linear function ($p<0.001$). We did not explore functions beyond a 2nd degree polynomial due to its impeccable fit (average R^2 = 0.998).

Prosthetic Prescription

Using the peak resultant GRFs and angles produced at 3 m/s, the actual stiffness of the manufacturer recommended Cheetah Xtend, which is prescribed based on user body mass, was 4% to 15% stiffer than the Flex-Run ($p<0.001$), 7% to 19% stiffer than the Catapult ($p<0.001$), and 20% to 28% stiffer than the 1E90 Sprinter ($p<0.001$) prostheses across matched user body masses (Fig 4). Using the peak resultant GRFs and angles produced at 6 m/s, the manufacturer recommended Cheetah Xtend prostheses were the same stiffness as the Flex-Run ($p = 0.166$), 0% to 22% less stiff than the Catapult ($p = 0.001$), and 3% to 21% stiffer than the 1E90 Sprinter ($p<0.001$) prostheses at matched user body masses (Fig 4). The Flex-Run and Catapult prostheses are not specifically recommended for fast running/sprinting; therefore we used manufacturer recommended stiffness categories for distance running at 6 m/s.

Prosthetic stiffness depends on peak GRF magnitude; hence we calculated the average 2nd order polynomial equations for each prosthetic model and stiffness category (S1-S4) so that prosthetists can predict an athlete's prosthetic stiffness from the amount of force they apply on the ground and/or prosthetic compression. For those unable to quantify force magnitudes or compression, and because of the relatively linear force-displacement relationships (average R^2 = 0.956), we also report average linear stiffness values (Table 2).

Hysteresis

The percentage of mechanical energy lost per cycle for C-shaped RSPs across conditions averaged 5.14% (SD: 0.70%). For every 1˚ increase in testing angle, the hysteresis magnitude decreased 0.04% ($p<0.001$). The average hysteresis for J-shaped RSPs across conditions was 4.28% (SD: 0.65%), which was lower than that of the C-shaped RSPs ($p<0.001$). Furthermore, testing angle affected the hysteresis of J-shaped RSPs ($p<0.001$), while height had no effect ($p = 0.215$). For every 1˚ increase in testing angle, the hysteresis of the 1E90 Sprinter and Cheetah Xtend prostheses decreased 0.01% and 0.08%, respectively ($p<0.001$). Additionally, removing the rubber soles from C- and J-shaped RSPs reduced the hysteresis magnitudes by 42% ($p<0.001$).

Effect of angle and height on prosthetic stiffness

While controlling for prosthetic height, every 1˚ increase in testing angle decreased the stiffness of the Flex-Run and Catapult prostheses by 0.41 kN/m ($p<0.001$) and 0.79 kN/m ($p<0.001$), respectively (Fig 3). Every 1˚ increase in testing angle decreased the stiffness of the 1E90 Sprinter and Cheetah Xtend prostheses by 0.45 kN/m ($p<0.001$) and 0.76 kN/m ($p<0.001$), respectively. Moreover, at a fixed testing angle, every 1 cm increase in height decreased the stiffness of both J-shaped RSPs by 0.27 kN/m ($p<0.001$). Despite a drastic pylon height difference

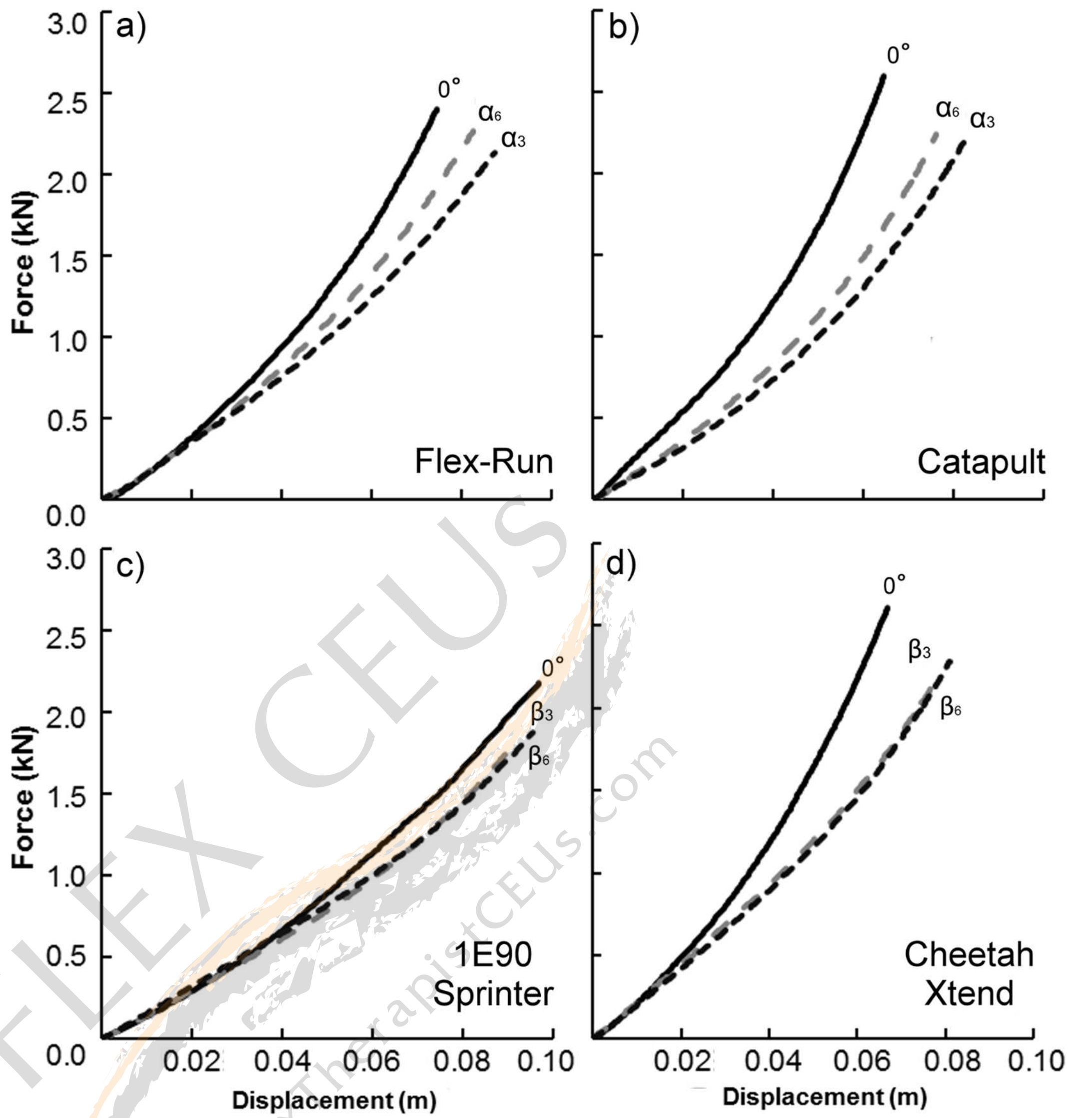

Fig 3. Representative force-displacement profiles for running-specific prosthetic models at each testing angle. Each running-specific prosthesis (RSP) is the manufacturer recommended stiffness category for a 70 kg distance runner. α_3 and β_3 indicate the measured angle between the RSP and peak resultant ground reaction force (GRF) vector while running 3 m/s using the C-shaped RSPs (Flex-Run and Catapult) and J-shaped RSPs (1E90 Sprinter and Cheetah Xtend), respectively. α_6 and β_6 indicate the measured angles between the RSP and peak resultant GRF vector while running 6 m/s using the C-shaped RSPs and J-shaped RSPs, respectively. a) The Flex-Run prosthesis at testing angles of 0˚, α_3, and α_6, b) the Catapult prosthesis at testing angles of 0˚, α_3, and α_6, c) the 1E90 Sprinter prosthesis at testing angles of 0˚, β_3, and β_6, and d) the Cheetah Xtend prosthesis at testing angles of 0˚, β_3, and β_6

(31.5 cm), preliminary testing revealed no effect of height on the stiffness of C-shaped RSPs; therefore we did not further test the effect of height across C-shaped RSPs. Furthermore, removing the rubber soles did not affect prosthetic stiffness across models (p = 0.151).

Discussion

Despite well-characterizing the force-displacement relationships of the RSPs (average R^2 = 0.956), a linear function did not fit quite as well as a 2nd degree polynomial function

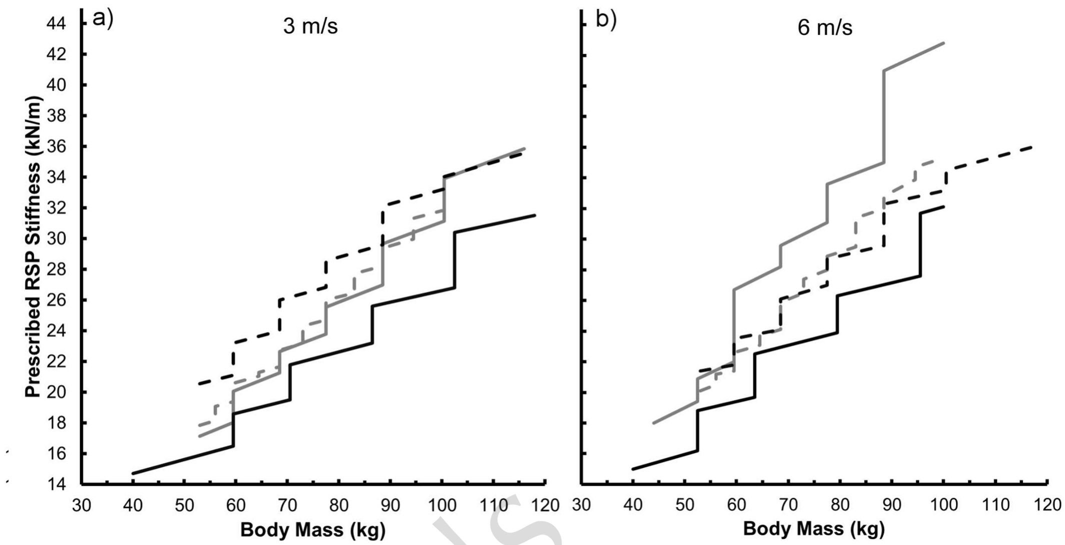

Fig 4. Prescribed prosthetic stiffness. The average stiffness (kN/m) of each running-specific prosthesis (RSP) as a function of the respective manufacturer's recommended user body mass (kg) at running speeds of 3 m/s (a), and 6 m/s (b). The stiffness of each RSP was calculated using peak applied force magnitudes that simulated running 3 m/s (α_3 and β_3) and 6 m/s (α_6 and β_6). We then calculated displacement using the mean curvilinear force-displacement profiles with the appropriate applied force magnitudes. See S1–S4 Tables.

($p<0.001$), leading us to partially reject our initial hypothesis. Contrary to the notion that prosthetic stiffness is invariant during running [8,16–18], our data suggest that as athletes exert greater forces on the ground and/or adjust the angle between the peak resultant GRF and their RSP during stance, prosthetic stiffness is altered. For example, a 70 kg athlete that produces peak resultant GRFs of 2.2, 2.6, 3.0, 3.4 times body weight with their affected leg using a manufacturer recommended Cheetah Xtend prosthesis (height: 38 cm; angle: 25.0°) would exhibit stiffness values of 25.1, 26.1, 27.1, and 28.1 kN/m, respectively. Yet, if the 70 kg athlete increased the angle of their RSP with respect to the resultant GRF from 15° to 30° in 5° increments, the aforementioned prosthetic stiffness values would change to 32.7, 29.9, 27.1, 24.3 kN/m. It is possible that the inverse relationship between affected leg stiffness and running

Table 2. The manufacturer recommended average prosthetic stiffness across models based on running 3 m/s and 6 m/s. All values include the rubber sole that comes with the prosthetic model, with the exception of the Össur Cheetah Xtend, which was equipped with the Össur Flex-Run's rubber sole.

Users Mass (kg)	3 m/s				6 m/s			
	Flex-Run (kN/m)	Catapult (kN/m)	1E90 Sprinter (kN/m)	Cheetah Xtend (kN/m)	Flex-Run (kN/m)	Catapult (kN/m)	1E90 Sprinter (kN/m)	Cheetah Xtend (kN/m)
55	18.0	17.4	16.2	20.7	20.4	20.4	19.0	21.5
60	20.6	20.1	18.6	23.2	22.6	25.8	19.5	23.5
65	22.1	20.8	19.1	23.7	23.7	27.6	22.7	23.9
70	22.9	22.8	21.8	26.1	26.1	29.9	23.1	26.4
75	23.7	23.5	22.2	26.6	27.7	30.7	23.5	26.8
80	26.2	25.9	22.7	28.8	29.2	33.7	26.4	28.9
85	26.1	26.5	23.2	29.3	31.3	34.5	26.8	29.4
90	29.5	29.9	25.9	32.3	33.4	41.2	27.2	32.4
95	31.4	30.5	26.3	32.7	34.7	42.0	27.6	32.8
100	31.8	31.1	26.7	33.2	35.3	42.8	32.1	33.1

speed found in McGowan et al. [16] can be attributed to decreased prosthetic stiffness via increased angles between the resultant GRF vectors and RSPs at faster speeds.

Overall, mechanically testing RSPs at 0˚ overestimates prosthetic stiffness (linear) by 10% to 39% compared to using angles utilized by athletes with transtibial amputations while running at 3 m/s and 6 m/s. Previous studies have tested the stiffness of RSPs at 0˚ [8], and 30˚ [19]. We compared our methodology to that of Brüggeman et al. [8] by acquiring the same prosthetic model (Össur Cheetah) as the previous study, replicating their protocol (applied force: 1500 N, testing angle: 0˚, loading velocity: 1 m/min), and then using our method (applied force: 2724 N, testing angle: 25˚, loading velocity: 100 N/s) to determine stiffness. Brüggeman et al.'s protocol resulted in a prosthetic stiffness (linear) of 34.2 kN/m, whereas our protocol resulted in a linear stiffness of 29.2 kN/m. These discrepancies suggest that prosthetic stiffness testing procedures should be standardized.

We reject our second hypothesis; manufacturer recommended prosthetic stiffness varies across models for a given user body mass and activity. Additionally, we compared manufacturer recommended prosthetic stiffness during running at 6 m/s versus at 3 m/s. At a given body mass (prosthetic height of 38 cm), the manufacturer recommended 1E90 Sprinter prostheses were 11% stiffer at 6 m/s compared to 3 m/s across a 45 kg span in user body mass ($p = 0.003$). Also, the recommended Catapult prosthetic stiffness increased 32% due to a greater recommended prosthetic stiffness category and reduced angle between the RSP and peak resultant GRF (Fig 4). Conversely, the Cheetah Xtend prostheses are recommended at the same stiffness categories for 3 m/s and 6 m/s [24], and thus the stiffness values varied by <1% (Bonferroni corrected p-value: $p = 0.080$). Prosthetic stiffness requirements may be different for running at various speeds due to the different mechanical demands of the respective tasks. Future studies are needed to assess the effects of prosthetic stiffness on distance running and sprinting performance.

Since testing angle affected hysteresis, we also reject our third hypothesis stating that prosthetic hysteresis would be invariant across testing conditions. Intriguingly, RSPs dissipate less energy when their proximal end is rotated forward with respect to the applied force. Future studies are needed to examine prosthetic designs and decipher why RSPs display less hysteresis when rotated forward. Due to the importance of mechanical energy return on running and sprinting performance [4,5], the designs of future RSPs should be developed to mitigate mechanical energy dissipation.

Moreover, prosthetic hysteresis was 42% lower when we removed the rubber soles, indicating that the rubber soles were responsible for almost half of the dissipated energy. Athletes with leg amputations should use soles with minimal damping to maximize the mechanical energy return of RSPs. In addition to the sole, energy dissipation probably occurs at the residual limb/socket interface. To our knowledge, no study has quantified the mechanical behavior of the residual limb and socket interface while running. Improving socket design by enhancing the connection between athletes and their RSPs may allow better utilization of the returned mechanical energy and potentially improve running performance.

Pylon height does not affect the stiffness of C-shaped RSPs; therefore, we reject our final hypothesis. The aluminum pylon of C-shaped RSPs has an annular section (i.e. an empty cylinder) and appears less prone to bending due to the perpendicular components of the applied compression forces, and due to a higher area moment of inertia [31] compared to the rectangular section of J-shaped RSPs. Increasing the overall length of the aluminum pylon technically reduces its overall stiffness, but the lengths used in our measurements were not enough to elicit a measurable difference. The height of RSPs needs to be within a relatively narrow range for athletes with unilateral amputations due to their unaffected leg length. Therefore prosthetic stiffness adjustments would primarily be accomplished by changing stiffness category or sagittal plane angle. On the other hand, athletes with bilateral amputations can consider a wide range of heights and stiffness categories to achieve a specified prosthetic stiffness; however, height and stiffness may affect running performance in different ways. In addition to stiffness, the effects of prosthetic height and alignment on performance warrant future research.

We assumed that the C-shaped RSPs were perpendicular to the respective pylons. Yet, the sagittal plane RSP-pylon alignment may have been slightly altered due to individual preference, thus our reported angles between the C-shaped RSPs and resultant GRF vectors may have been over/underestimated by a few degrees. We collected prosthetic angles and peak resultant GRFs from a cohort of exceptional athletes with unilateral transtibial amputations at 3 m/s and 6 m/s. Conceivably, less athletic individuals with amputations, or athletes with different amputation levels may not utilize the same prosthetic angles and/or generate the same resultant GRFs compared to those exhibited by our participants, and consequently prosthetic stiffness may differ. For example, athletes with transfemoral amputations with pylons connecting their RSPs to their sockets can use our reported values at 0˚, as it is a fair approximation of their RSP-peak GRF angle to determine the prosthetic stiffness and hysteresis.

Our methodology does not account for the rotation of the RSP with respect to the resultant GRF throughout ground contact. It may be that RSPs are stiffer at initial and terminal ground contact than at mid-stance due to a smaller angle between the RSP and resultant GRF vector. On the other hand, as applied force accrues RSPs become stiffer, implying that RSPs are stiffest at mid-stance. The influence of angle and force may counteract each other, exhibiting a constant prosthetic stiffness throughout stance; perhaps a deliberate design choice of prosthetic manufacturers. Future studies are warranted to include a rotational component to the mechanical stiffness testing of RSPs. Furthermore, we tested our RSPs with a loading rate (100 N/s) that is much lower than that recorded during running (over 4000 N/s [16,18]). However, our low loading rate (100 N/s) enabled us to record force-displacement data from every 10 N of applied force, thus presenting ~150 to 400 data points per loading cycle. When athletes with an amputation run 6 m/s, they have a ground contact time of ~0.2 seconds [18,25]. If ground reaction forces were recorded at 2000 Hz, then 200 data points would have been collected from initial ground contact to mid-stance/peak GRF, which coincides with our material testing machines sampling versus loading rate data. Nevertheless, it is ideal for prosthetic testing to mimic the loading/unloading rates of those recorded during running; unfortunately these rates are beyond the capability of our equipment.

Conclusions

We assessed prosthetic stiffness and hysteresis across a wide range of models, stiffness categories, and heights, at forces and angles that simulate those exhibited by athletes with transtibial amputations running at 3 m/s and 6 m/s. We found that the force-displacement profiles of RSPs are curvilinear, indicating that prosthetic stiffness varies with the magnitude of applied force. Yet, a linear force-displacement characterization is strongly predictive. We also found that manufacturer recommended prosthetic stiffness varies between models, and that the height of J-shaped RSPs is inversely related to stiffness. Moreover, we provide evidence that prosthetic stiffness is much greater at 0˚ than at angles representative of those that occur during running.

When athletes with leg amputations change prosthetic models, height, and/or sagittal plane alignment, prosthetic stiffness also changes; therefore variations in comfort, performance, etc. may be indirectly due to altered stiffness. We propose that prosthetic stiffness should be assessed under conditions that simulate the demands of the respective activity, and that manufacturers should provide the stiffness values of each RSP at specific heights. Until then, our study provides reference for the stiffness values of various prosthetic models across multiple stiffness categories and heights, and provides a foundation for future research to understand the potential effects of prosthetic stiffness on performance during distance running and sprinting.

Supporting Information

S1 Table. The stiffness and hysteresis characteristics for Össur Flex-Run prostheses at each testing condition. The equations indicate prosthetic displacement in meters (h) used to calculate the applied force in kN. Stiffness equals applied force divided by displacement. a and b are constants. All prostheses were tested with the manufacturer supplied sole, with the exception of stiffness category 7 High No Sole.
(DOCX)

S2 Table. The stiffness and hysteresis characteristics for Freedom Innovations Catapult FX6 prostheses at each testing condition. The equations indicate prosthetic displacement in meters (h) used to calculate the applied force in kN. Stiffness equals applied force divided by displacement. a and b are constants. All prostheses were tested with the manufacturer supplied sole, with the exception of stiffness category 7 No Sole.
(DOCX)

S3 Table. The stiffness and hysteresis characteristics for Ottobock 1E90 Sprinter prostheses at each testing condition. The equations indicate prosthetic displacement in meters (h) used to calculate the applied force in kN. Stiffness equals applied force divided by displacement. a and b are constants. All prostheses were tested with the manufacturer supplied sole, with the exception of stiffness category 5 No Sole.
(DOCX)

S4 Table. The stiffness and hysteresis characteristics for the Össur Cheetah Xtend prostheses at each testing condition. The equations indicate prosthetic displacement in meters (h) used to calculate the applied force in kN. Stiffness equals applied force divided by displacement. a and b are constants. All RSPs were tested with the supplied sole from the Össur Flex-Run prostheses, with the exception of stiffness category 7 No Sole.
(DOCX)

Effect of Running Speed and Leg Prostheses on Mediolateral Foot Placement and Its Variability

Abstract

This study examined the effects of speed and leg prostheses on mediolateral (ML) foot placement and its variability in sprinters with and without transtibial amputations. We hypothesized that ML foot placement variability would: 1. increase with running speed up to maximum speed and 2. be symmetrical between the legs of non-amputee sprinters but asymmetrically greater for the affected leg of sprinters with a unilateral transtibial amputation. We measured the midline of the body (kinematic data) and center of pressure (kinetic data) in the ML direction while 12 non-amputee sprinters and 7 Paralympic sprinters with transtibial amputations (6 unilateral, 1 bilateral) ran across a range of speeds up to maximum speed on a high-speed force measuring treadmill. We quantified ML foot placement relative to the body's midline and its variability. We interpret our results with respect to a hypothesized relation between ML foot placement variability and lateral balance. We infer that greater ML foot placement variability indicates greater challenges with maintaining lateral balance. In non-amputee sprinters, ML foot placement variability for each leg increased substantially and symmetrically across speed. In sprinters with a unilateral amputation, ML foot placement variability for the affected and unaffected leg also increased substantially, but was asymmetric across speeds. In general, ML foot placement variability for sprinters with a unilateral amputation was within the range observed in non-amputee sprinters. For the sprinter with bilateral amputations, both affected legs exhibited the greatest increase in ML foot placement variability with speed. Overall, we find that maintaining lateral balance becomes increasingly challenging at faster speeds up to maximum speed but was equally challenging for sprinters with and without a unilateral transtibial amputation. Finally, when compared to all other sprinters in our subject pool, maintaining lateral balance appears to be the most challenging for the Paralympic sprinter with bilateral transtibial amputations.

Introduction

When running and sprinting, individuals with transtibial amputations face unique biomechanical constraints when using running-specific prostheses. When compared to the biological leg, the affected leg fitted with a running-specific prosthesis generates lower ground forces and exhibits less stiffness [1–3]. Additionally, using these passive-elastic leg prostheses likely challenges a person's ability to maintain balance while running and sprinting because the design of running-specific prostheses primarily facilitates forward sagittal plane motion. Running-specific prostheses are passive devices made of carbon fiber that can only store and return elastic energy. In addition, running-specific prostheses have a fixed stiffness, whereas the stiffness of the human leg can be neurally modulated and thus non-amputees can adapt to changes in surface stiffness and speed [2, 4]. Further, unlike an intact human foot, running-specific prostheses cannot provide direct sensory information about ground contact and the distal end of the residual limb lacks the neural specialization that the plantar surface of the foot provides. As a final point, running-specific prostheses do not provide any proprioceptive feedback about "ankle" joint or foot position. In light of these observations, we were curious to explore how well sprinters can modulate foot placement from step-to-step when using running-specific prostheses.

Our recent studies of runners with two biological legs have demonstrated that adjusting step width from step-to-step is an effective strategy for maintaining lateral balance [5, 6]. We challenged a runners' ability to maintain lateral balance by having them run with a range of unnaturally large step widths [5]. Running with non-preferred step widths increased step width variability and increased metabolic cost, thus increasing the effort to maintain lateral balance. In a follow up study [6], we reduced the muscular effort needed to maintain lateral balance during running by providing external lateral support with mechanical springs attached to a waist harness. Providing external lateral support decreased step width variability and decreased metabolic cost, thus reducing the effort to maintain lateral balance. Taken together, these studies suggest that mediolateral (ML) foot placement variability can be used as an indicator of lateral balance during running.

Although step width—the ML distance between the left and right foot during subsequent steps—provides a general measure of foot placement, it is insensitive to foot placement asymmetries that may exist between the legs of the same individual. Because runners with a unilateral transtibial amputation exhibit a variety of inter-limb biomechanical asymmetries [1, 7–11], we reasoned that measuring ML placement of each foot relative to the midline of the body, as opposed to step width, would be a more appropriate metric for elucidating any asymmetries that may exist between the individual legs. Thus, in the present study, we quantified ML foot placement relative to the body's midline and its variability.

Our previous experiments [5, 6] have focused on only a single, modest speed of running (3.0 m/s); however, the effects of running speed on lateral balance are not known. We quantified how changes in running speed up to maximum sprint speed affect both ML foot placement and ML foot placement variability in non-amputee sprinters and sprinters with transtibial amputations. Since we were curious to understand how speed might affect one's ability to properly modulate ML foot placement from step-to-step, we present and discuss this data in the text. However, our hypothesis concerns only ML foot placement variability—our indicator of lateral balance. We interpret our results with respect to a hypothesized relation between ML foot placement variability and lateral balance. The rationale for our hypothesis was that in order to attain faster running speeds, an individual must elicit briefer ground contact times and more rapid leg swings [1, 3, 12]. Because there is less time to adjust ML foot placement from step-to-step, it is likely that maintaining lateral balance becomes more challenging. We hypothesized that ML foot placement variability would increase across running speed up to maximum sprint speed in sprinters with and without transtibial amputations. We further hypothesized that ML foot placement variability would be symmetrical between the right and left legs of non-amputee sprinters but asymmetrically greater for the affected leg compared to the unaffected leg of sprinters with a unilateral transtibial amputation.

Materials and Methods

Subjects

Twelve non-amputee recreational sprinters and seven elite Paralympic sprinters with transtibial amputations (6 unilateral and 1 bilateral) volunteered to participate in this study. We selected non-amputee recreational sprinters who had similar maximal speeds to the Paralympic sprinters. The study was reviewed and approved by the Intermountain Healthcare Urban Central Region Institutional Review Board before the study began. Prior to experimental data collection, each subject read and signed the study's written informed consent document. All data were collected at the Biomechanics Laboratory of the Orthopedic Specialty Hospital (Murray, Utah).

Anthropometric Measurements

We measured height and body mass. The body mass of subjects with transtibial amputations included their running-specific prosthesis and socket mass (see S1 Table, which details the anthropometric and biomechanical characteristics for each sprinter that participated in this study).

Experimental Protocol

After emulating their pre-race warm-up (i.e. jogging, stretching, and short sprints), each subject performed brief running trials on a high-speed 3D force-sensing motorized treadmill (Fig. 1; Treadmetrix, Park City, UT). The treadmill is a custom designed lightweight treadmill with 3D force transducers (MC3A AMTI, Watertown, MA) at each corner that interface with amplifiers (MSA6 AMTI, Watertown, MA). During each trial, we simultaneously collected 3D ground reaction forces and moments (2400 Hz) and whole body kinematics from the 3D positions of reflective markers placed on the body (200 Hz; Motion Analysis Corporation, Santa Rosa, CA). We used a full-body custom marker set to define the position of the subject's head, trunk, arms, legs, and running-specific prosthesis. Each subject began the series of trials at 3 m/s and we incremented the speed by 1 m/s until subjects approached their maximum speed. We then used smaller speed increments until the subject reached maximum speed, defined as the fastest speed at which the subject could maintain the same position on the treadmill for at least 20 consecutive steps. All subjects had experience with treadmill running and sprinting and were familiar with this task. During each trial, subjects lowered themselves from handrails onto the treadmill belt, which was moving at the testing speed. Handrails were placed along the front and sides of the treadmill and each subject had practice holding and then releasing the handrails when achieving maximum treadmill sprinting speeds (for clarity, handrails are not shown in Fig. 1).

Data Analysis

We measured ML foot placement relative to the midline of the body and its variability from step-to-step using center of pressure (COP) data calculated from the force measuring treadmill (Fig. 2). Net ground reaction forces and COP were calculated using a calibration matrix and

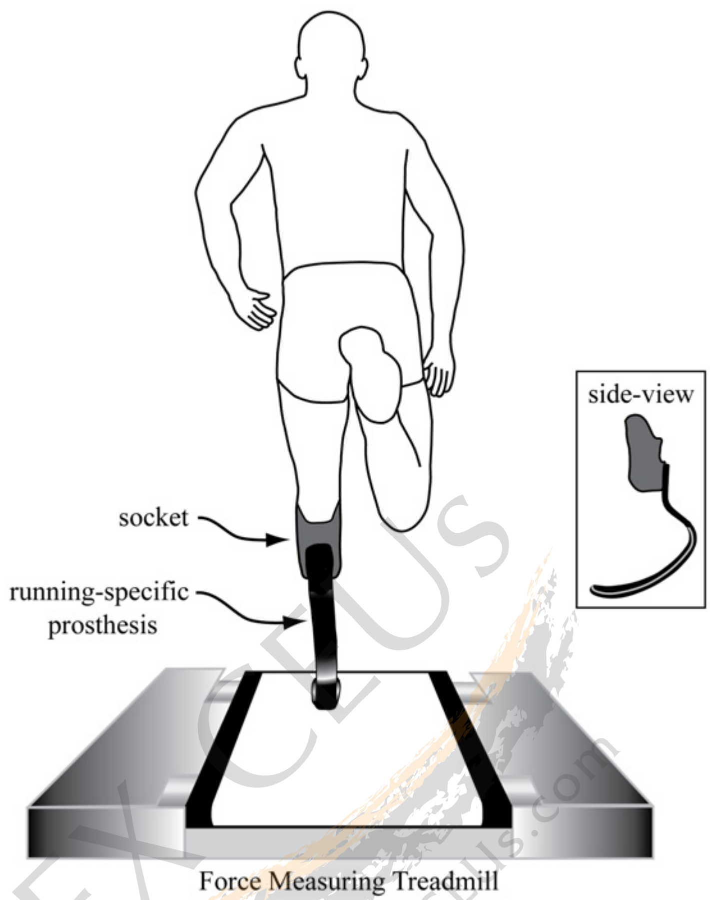

Figure 1. (a) Rear view of a sprinter with a left-side transtibial amputation running on a force-measuring treadmill. A side-view illustration of the running-specific prosthesis is provided as an inset. For measurements of ML foot placement relative to the body's midline, we measured the position of the center of pressure at peak vertical ground reaction force for both biological legs and the legs using the running-specific prosthesis.

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treadmill dimensions provided from the manufacturer. Force and COP signals were low pass filtered (4^{th} order, recursive Butterworth filter) at 20 Hz, 20 Hz, and 25 Hz for the AP, ML, and Vertical direction, respectively. The COP obtained from the treadmill was validated during a separate procedure by comparing the value to a known point of force application during running. This was accomplished by running with a rigid-sole shoe with a small bolt screwed to the bottom of the sole resulting in a contact point that was 1 cm in diameter. Markers were placed on the top of the shoe and on the contact point of the bolt. Once the location of the point was determined relative to the shoe markers, the marker on the contact point was removed. Our results show that while running at 3.1 m/s, the COP at peak vertical ground reaction force (~1600 N) differed from the contact point by 0.6 ± 1.5 mm in the ML direction.

Following a similar convention as McClay and Cavanagh [13], we defined the midline of the body as the ML position of the pelvis center of mass (COM*p*). The COM*p* was modeled as an elliptical cylinder [14] with dimensions defined by markers placed bilaterally on the iliac crests, anterior-superior iliac spines, greater trochanters, posterior-superior iliac spines, and sacrum. The location of the COM*p* was defined as the geometric center of the cylinder (Visual 3D, C-Motion Inc., Germantown MD.). We defined ML foot placement relative to the midline of the body as the distance between the COP and the COM*p* at the instance of the peak vertical

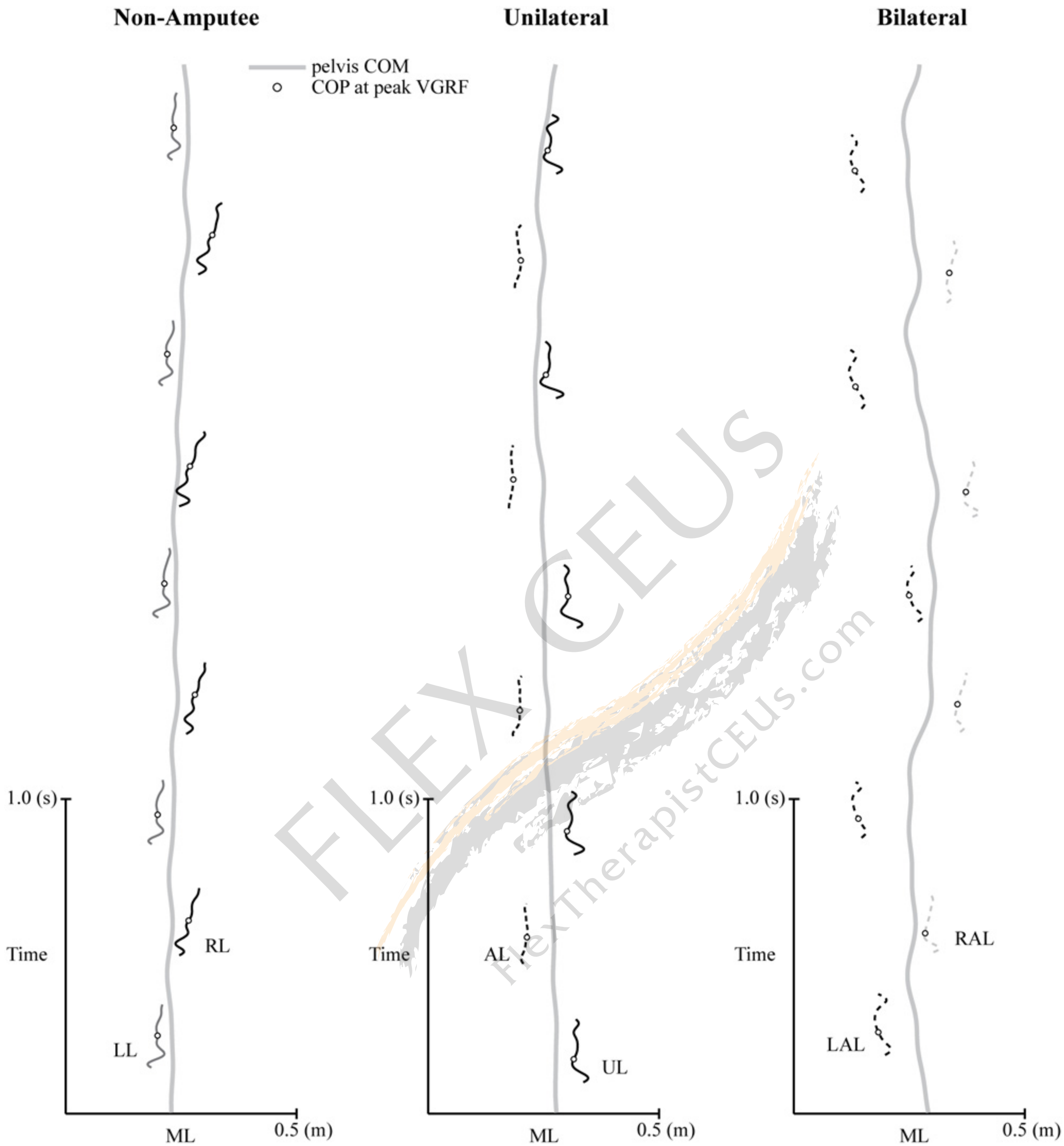

Figure 2. Representative time-series trajectories for the pelvis center of mass (COM) and center of pressure (COP) trajectories while running at 3 m/s for a non-amputee sprinter (Non-Amputee), a sprinter with a unilateral transtibial amputation (Unilateral), and a sprinter with bilateral transtibial amputations (Bilateral). For clarity, we use acronyms to denote the right leg (RL), left leg (LL), affected leg (AL), unaffected leg (UL), left affected leg (LAL), and right affected leg (RAL) for each respective sprinter. The ML position of the COP (open circles) indicates the instance of peak vertical ground reaction force (VGRF) during mid support for each step while running. Note that the Bilateral sprinter exhibits greater variation in ML foot placement from step-to-step when compared to the Non-Amputee and Unilateral sprinter.

ground reaction force that occurs during mid support for each step while running [15]. Our sign convention is that placing the left foot to the left of the midline is considered a positive foot placement, to the right of the midline a negative foot placement, placement on the line as zero, and vice versa.

Each subject performed each trial once. To control for the number of steps and to avoid bias in our variability measurements, we calculated the average ML foot placement during 20 consecutive steps that occurred during the middle of each trial. ML foot placement variability was defined as the standard deviation. From a total of 153 trials, we discarded 19 trials because they consisted of fewer than 20 consecutive steps or there was a large amount of marker drop out that could not be sufficiently corrected with gap filling and/or filtering techniques. Some of the discarded trials included the maximum speed for 5 non-amputee sprinters and 2 sprinters with a unilateral amputation. While a greater number of steps would be ideal [5], athletes can only sprint at their maximum speed for a short duration. Thus, we were inherently limited by the number of steps that we could incorporate into our variability analysis.

Statistical Analysis

We normalized speed to each subject's maximum speed prior to making comparisons between the legs of the same individual. We chose to normalize speed in this manner so that we could evaluate the change in ML foot placement and its variability across individuals with different absolute sprinting capabilities. Therefore, using an individual's relative speed allows one to compare between sprinters using a standard scale that ranges from 0.3 to 1.0, where 1.0 is maximal speed. In contrast, we chose not to normalize M-L foot placement and its variability for two reasons. First, we were interested in comparing our data in absolute units, making it convenient for the reader to interpret by using physical measurements that are easily understandable. Second, normalizing ML foot placement and its variability to a maximum value of 1.0 would not allow us to illustrate differences between individual sprinters. In this normalized, hypothetical scenario, the maximum normalized value for both sprinters would be equal to 1.0 and one might interpret this to mean that both sprinters exhibited similar foot placement variability values. The absolute values; however, might be drastically different. For example, the maximum variability in sprinter #1 and #2 might be 1.0 cm and 2.0 cm, respectively. Therefore, expressing ML foot placement and its variability in absolute units makes it clear that the variability was greater in sprinter #2.

With respect to ML foot placement and its variability, we performed separate repeated measures MANOVAs to compare between the right and left leg of the non-amputee group and the unaffected and affected leg of the group with a unilateral amputation. We defined the "leg" as the within subjects fixed factor and "normalized speed" as a covariate. We avoided performing a between subjects comparison for two reasons: 1. the sample sizes between groups were unequal (non-amputee: $n = 12$ vs. unilateral: $n = 6$) and 2. our analysis was focused on comparing two dependent regression lines [16]. For example, we were interested in testing whether the regression line for ML foot placement variability of the right leg vs. normalized speed was significantly different from the regression line for ML foot placement variability of the left leg vs. normalized speed in non-amputee sprinters. We followed the same approach for the sprinters with a unilateral amputation.

To determine the direction and strength of the relation between variables, we followed the repeated measures MANOVAs with separate linear regression and correlation analyses for each dependent variable (i.e. ML foot placement and ML foot placement variability) with normalized speed as the independent variable. Linear regression and correlation (Pearson's r) analyses were performed separately for non-amputee sprinters ($n = 12$) and sprinters with a unilateral amputation ($n = 6$). To elucidate the individual and group trends between ML foot placement, ML foot placement variability, and normalized speed, we present the linear regression equations denoting the slope, intercept, Pearson's r values, and range (Figs. 3, 4, and 5). The range is defined in brackets, expressed in units of centimeters, and denotes the ML foot placement (or its variability) value at the individual's minimum and maximum normalized speed. Statistical significance for all analyses was set at an α level = 0.05 (SPSS Inc., Chicago, IL).

It is important to note that the manner in which we incrementally increased running speed as we approached a sprinter's maximum speed resulted in more data points closer to a normalized speed of 1.0 (see Figs. 3 and 4). Thus, the data are not evenly spread across normalized speed, which means that the overall magnitude and direction of the correlation coefficient may be heavily biased by the data points closer to 1.0. To be cautious that the strength of the correlation coefficients was not dramatically influenced by this phenomenon, we systematically compared the magnitude of the correlation coefficient with and without the cluster of data points. Although this resulted in more evenly spread data points across normalized speed, we found that the magnitude and direction of the correlation coefficients were roughly the same, therefore, we decided to include all the data in our individual and average regression analyses. Furthermore, since maximum speed data for some individuals was not available, we note that our regression analyses should be interpreted with this limitation in mind.

For our single sprinter with bilateral transtibial amputations, we report the regression equation, correlation values, and range to demonstrate the relation between the dependent variables and normalized speed. Because $n = 1$, we did not attempt statistical inference.

Since we were interested in highlighting any notable asymmetries that exist between the individual legs, we analyze and present individual as well as group outcomes. All the data underlying the findings in this study are freely available in the S1 Appendix.

Results

Non-Amputee Sprinters (n = 12)

Individual analyses. As indicated by the negative magnitude of the Pearson's r values (Fig. 3, left column), 10 of 12 non-amputee sprinters exhibited a 0.9–4.6 cm decrease in ML foot placement with a two-fold increase in normalized speed. Two non-amputee sprinters (#4 and #5) did not decrease ML foot placement with speed. With respect to ML foot placement variability, 6 of 12 non-amputee sprinters exhibited an increase of 0.2–1.7 cm across the two-fold increase in normalized speed. The other 6 non-amputee sprinters exhibited a 1.4–2.1 cm increase in ML foot placement variability. Both of these trends are indicated by the positive magnitude of the Pearson's r values (Fig. 4, left column).

Group analyses. Our regression analyses revealed that as non-amputee sprinters ran faster, they placed each foot closer to the body's midline ($r_{RL} = -0.50$, $p < 0.001$ and $r_{LL} = -0.45$, $p < 0.001$; Fig. 5). At the slowest speed, non-amputee sprinters placed their feet ~ 4.1 cm lateral to the midline, whereas at maximum speed, they placed their feet 1.5 cm lateral to the midline, a 65% reduction. In contrast, ML foot placement variability of the right leg and left leg increased by 152% at maximum speed, compared to the slowest speed ($r_{RL} = 0.61$, $p < 0.001$ and $r_{LL} = 0.62$, $p < 0.001$). The average regression lines quantifying the changes in ML foot placement and ML foot placement variability across the two-fold increase in normalized speed were not significantly different between the right and left leg (Table 1, $p = 0.569$). As illustrated in Fig. 5 (left column), the average regression lines for ML foot placement and its variability for the right and left leg are similar in slope and intercept.

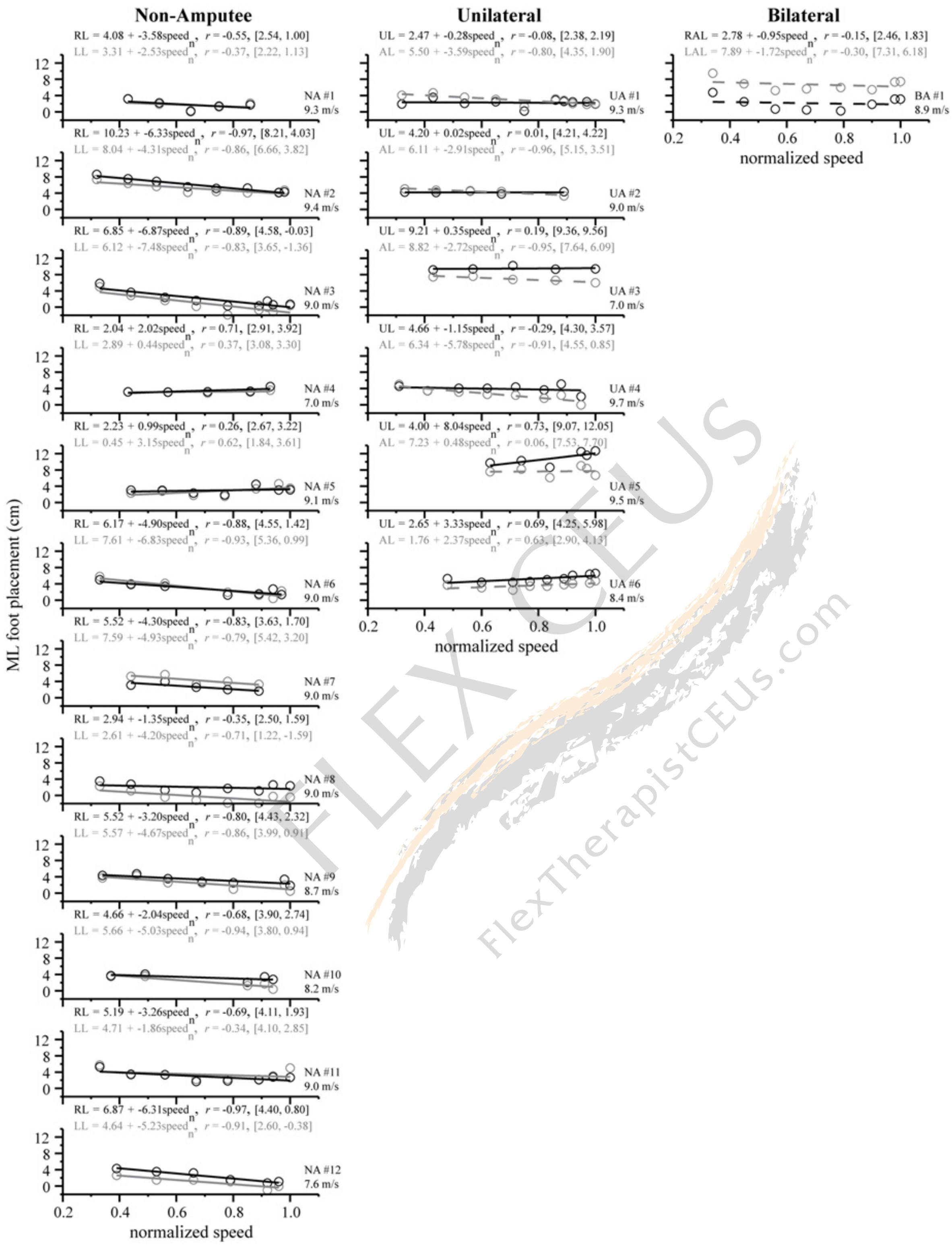

Figure 3. Individual linear regression lines for ML foot placement across normalized speed for non-amputee sprinters (Non-Amputee), sprinters with a unilateral transtibial amputation (Unilateral), and a sprinter with bilateral transtibial amputations (Bilateral). For clarity, we label each subplot with a "#" in the bottom, right corner that corresponds to each sprinter. Just below, we also include the absolute maximum speed (m/s) achieved by each sprinter. In general, non-amputee sprinters exhibited symmetrical changes in ML foot placement between the right leg (RL, black line) and left leg (LL, gray line) across normalized speed. In contrast to non-amputee sprinters, sprinters with a transtibial amputation exhibited varying degrees of asymmetry, as noted

by differences in ML foot placement between the unaffected leg (UL, black line) and affected leg (AL, dashed gray line) and between the right affected leg (RAL, dashed black line) and left affected leg (LAL, dashed gray line). Note that for each sprinter's leg, we present the regression equation, correlation coefficient, and range denoting the ML foot placement value achieved at minimum and maximum speeds.

Sprinters with Unilateral Amputations (n = 6)

Individual analyses. We found that changes in ML foot placement across normalized speed were different between the unaffected and affected leg of sprinters with unilateral amputations (Fig. 3, middle column). Unilateral sprinters #1, 2, 3, and 4 exhibited a 1.6–3.7 cm decrease in ML foot placement of the affected leg (r_{AL} = -0.80 to -0.95) while maintaining the same position of the unaffected leg (r_{UL} = -0.01 to -0.29) across normalized speed. In contrast, unilateral sprinter #5 exhibited a 3.0 cm increase in ML foot placement of the unaffected leg ($r_{UL} = 0.73$) while maintaining the same position of the affected leg ($r_{AL} = 0.06$). Only one unilateral sprinter (#6) showed similar increases in ML foot placement for both their unaffected leg (1.7 cm) and affected leg (1.2 cm) across normalized speed, although the unaffected leg was consistently placed further away from the midline.

We also found that changes in ML foot placement variability across normalized speed were different between the unaffected and affected leg. In unilateral sprinters #1 and 6, the variability of the unaffected leg remained the same while the variability of the AL exhibited a 1.3 cm increase across normalized speed (Fig. 4, middle column). In unilateral sprinter #2, the variability of the unaffected leg exhibited a 1.3 cm increase while the variability of the affected leg remained the same across normalized speed. In unilateral sprinter #3, the variability of both the unaffected and affected leg exhibited a less than 1 cm increase across normalized speed. In unilateral sprinter #4, the unaffected and affected leg exhibited a 1.7 cm and 1.2 cm increase in variability across normalized speed, respectively. In unilateral sprinter #5, the variability of the unaffected leg exhibited a 1.2 cm increase while the variability of the affected leg exhibited a 2.5 cm increase across normalized speed. To emphasize these dramatic asymmetries a bit further, the Pearson's r values for the affected leg in three unilateral sprinters (#1, 5, and 6) was notably higher than the unaffected leg, indicating that across normalized speed, the variability of the affected leg increased to a greater extent than the unaffected leg. In contrast, the Pearson's r value for the unaffected leg in one unilateral sprinter (#2) was notably higher than the affected leg, indicating that across normalized speed, the variability of the unaffected leg increased to a greater extent than the affected leg. Unilateral sprinters #3 and #4 showed similar increases in the variability of the unaffected and affected legs across normalized speed. For unilateral sprinter #3, however, the variability of the unaffected leg was consistently higher than the AL.

Group analyses. In contrast to non-amputees, there was no significant relation for ML foot placement of the unaffected or affected leg across normalized speed ($r_{UL} = 0.235$, $p = 0.065$ and $r_{AL} = -0.085$, $p = 0.295$; Fig. 5, middle column), although the regression lines were significantly different between the unaffected and affected leg ($p < 0.001$). ML foot placement variability of the unaffected and affected leg increased across normalized speed; however, the regression lines were statistically different in slope and intercept (Table 2, $p = 0.020$). As indicated by the slopes, a unit increase in normalized speed coincides with an ~ 1 cm and 2 cm increase in ML foot placement variability of the unaffected and affected leg, respectively. Most notably, the unaffected and affected leg exhibited a 57% and 190% increase in ML foot placement variability, respectively ($r_{UL} = 0.377$, $p = 0.006$ and $r_{AL} = 0.586$, $p < 0.001$).

Sprinter with Bilateral Amputations (n = 1)

Similar to sprinters with a unilateral amputation, we found that our regression analyses did not reveal strong trends between ML foot placement of the right or left affected leg and normalized

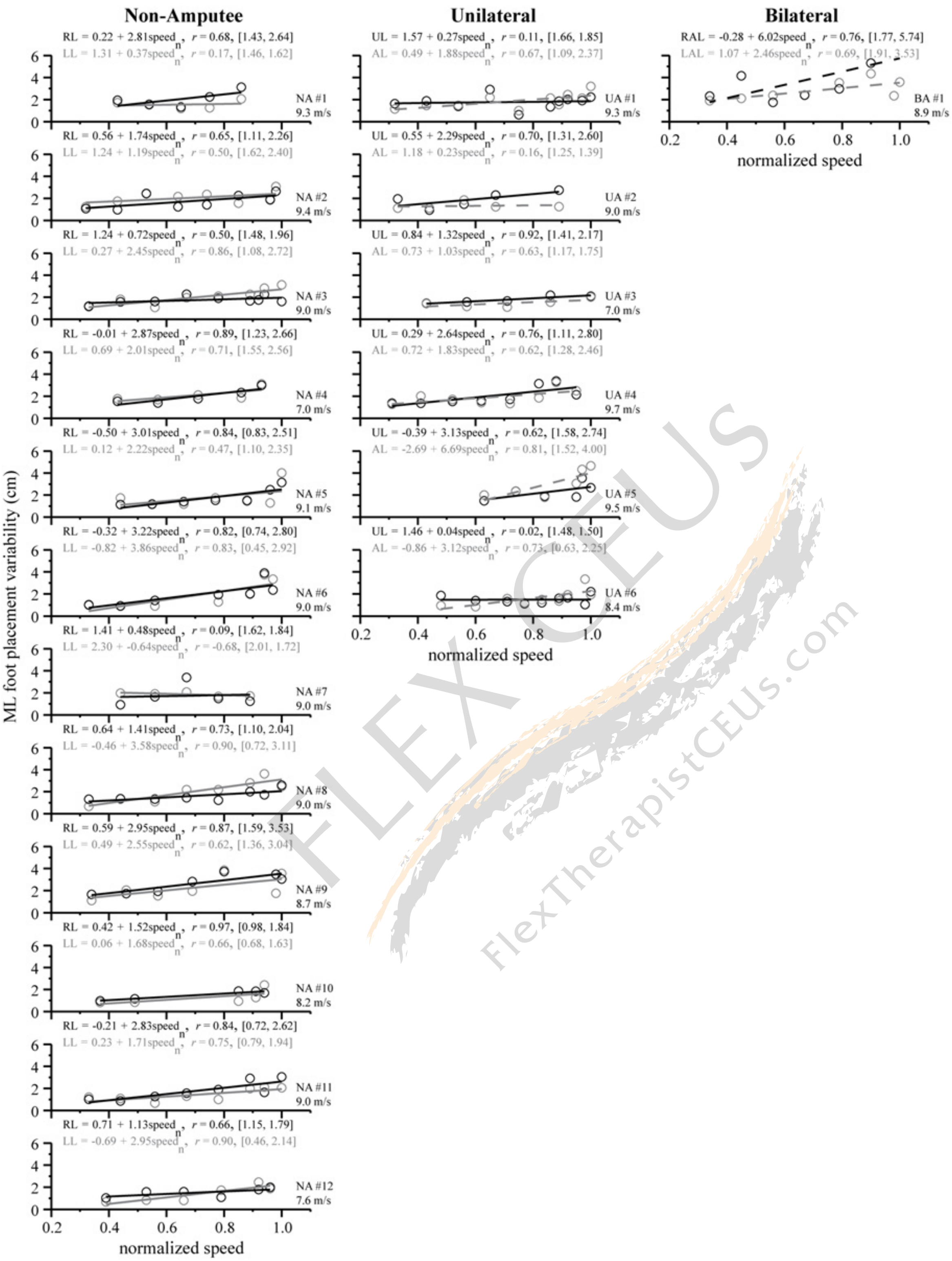

Figure 4. Individual linear regression lines for ML foot placement variability across normalized speed (graphical layout same as Fig. 3). In general, ML foot placement variability tended to increase across normalized speed, with the exception of the right leg (RL) and left leg (LL) of one non-amputee sprinter (#7) and the unaffected leg (UL) of one sprinter with a unilateral amputation (#6). Non-amputee sprinters also exhibited symmetrical changes in ML foot placement variability between the RL and LL across normalized speed. Similar to the ML foot placement trends (Fig. 3), sprinters with a transtibial amputation also exhibited varying degrees of asymmetry in ML foot placement variability, as illustrated by differences between the unaffected leg (UL) and

affected leg (AL) and between the right affected leg (RAL) and left affected leg (LAL). Note that for each sprinter's leg, we present the regression equation, correlation coefficient, and range denoting the ML foot placement variability value achieved at minimum and maximum speeds.

speed in the sprinter with bilateral amputations (r_{RAL} = -0.15 and r_{LAL} = -0.30; Fig. 5, right column). However, ML foot placement variability of the right affected leg exhibited a 4.0 cm increase while the left affected leg exhibited a 1.6 cm increase across normalized speed. When comparing his maximum speed to his slowest speed of 3 m/s, the bilateral sprinter exhibited a 276% and 95% increase in ML foot placement variability for the right and left affected leg, respectively (r_{RAL} = 0.76 and r_{LAL} = 0.69). Across normalized speed, ML foot placement of the left affected leg (6.7 ± 1.4 cm) was substantially greater than the right affected leg (2.1 ± 1.6 cm) but both ML foot placements were within the range observed in sprinters with a unilateral transtibial amputation. ML foot placement variability of the right affected leg (4.0 ± 2.0 cm) was slightly greater than the left affected leg (2.8 ± 0.9 cm) but was on average 89% greater than the affected leg of sprinters with a unilateral amputation.

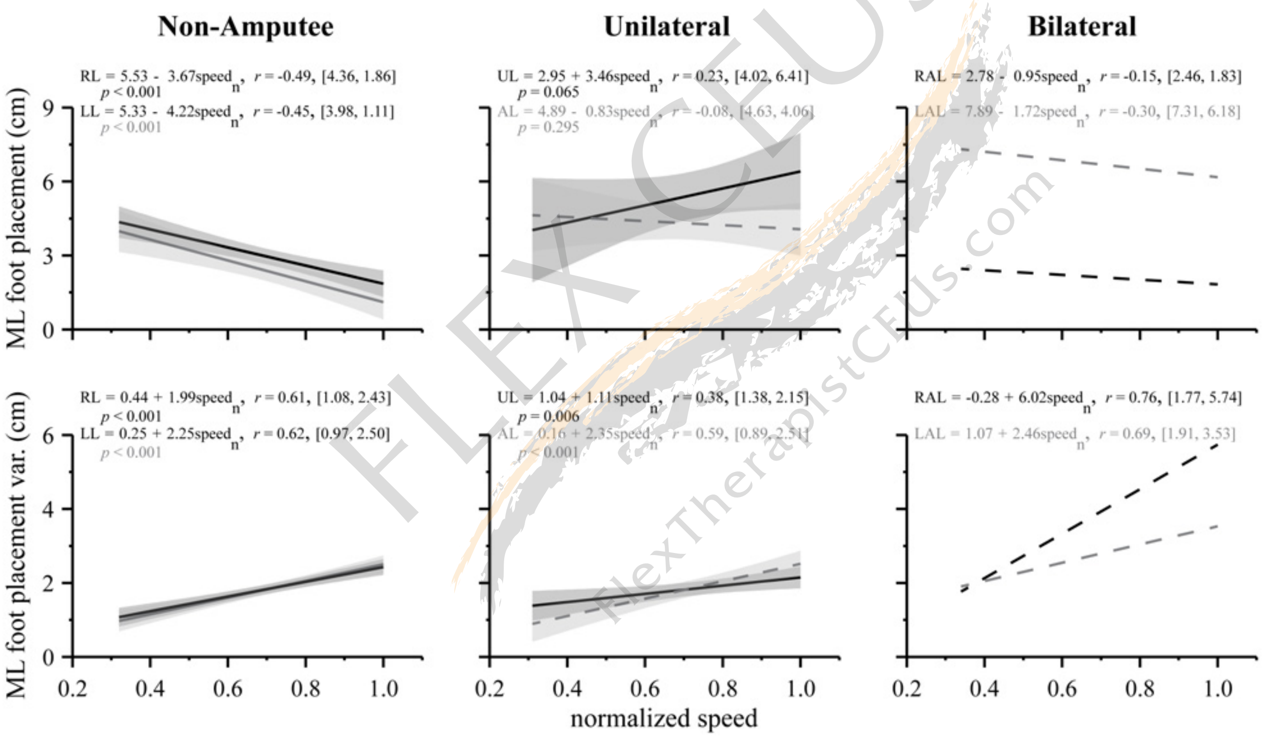

Figure 5. Average linear regression lines for ML foot placement (top) and ML foot placement variability (bottom) across normalized speed. Across speed, non-amputees exhibited symmetrical foot placement patterns between their right leg (RL) and left leg (LL). As non-amputee sprinters approached maximum speed, they placed their feet closer to the midline of the body; however, each leg exhibited greater ML foot placement variability (left column). In contrast, sprinters with unilateral transtibial amputations exhibited asymmetrical foot placement patterns between the affected leg (AL) and unaffected leg (UL). They tended to place the AL closer to the midline as compared to the UL, but the AL exhibited greater ML foot placement variability (middle column). The sprinter with bilateral transtibial amputations also exhibited asymmetrical foot placement patterns. Although he tended to place the right affected leg (RAL) closer to the midline of the body as compared to the left affected leg (LAL), ML foot placement was similar across speed. Both the LAL and RAL exhibited dramatic increases in ML foot placement variability at faster running speeds up to maximum sprint speed (right column). For each variable, we present the equation and *r* values from a least-squares linear regression analysis that included the individual data points across normalized speed for each sprinter. For the Non-Amputee and Unilateral group, the shaded regions represent the 95% lower and upper confidence bands. Since *n* = 1, we do not include the confidence bands for the Bilateral case. For clarity, individual data points are not shown.

Table 1. Mean, standard deviation, and linear regression results based on a repeated measures MANOVA[a] comparing ML foot placement and ML foot placement variability across normalized speed for the Non-Amputee group (n = 12).

Main Effect: *leg*						
	RL		**LL**			
	Mean	SD	Mean	SD	Univariate ANOVA	*p* value
ML foot placement, cm	2.9	1.6	2.4	2.1	–	–
ML foot placement variability, cm	1.8	0.7	1.8	0.8	–	–
Interaction Effect: *leg*normalized speed*						
	Slope	Intercept	Slope	Intercept	Univariate ANOVA	*p* value
ML foot placement, cm	-3.67	5.53	-4.22	5.33	–	–
ML foot placement variability, cm	1.99	0.44	2.25	0.25	–	–

[a]A one-way repeated measures MANOVA with normalized speed as a covariate revealed no significant main effect for "leg", indicating that ML foot placement and ML foot placement variability between the RL and LL was not statistically different: $F_{2,77} = 0.43$, $p = 0.650$, Wilk's $\lambda = 0.99$. In addition, our statistical analysis revealed no significant interaction effect between "leg" and "normalized speed", indicating that the dependent regression lines quantifying the linear relation of ML foot placement or ML foot placement variability across normalized speed for the RL and LL were not statistically different: $F_{2,77} = 0.57$, $p = 0.569$, Wilk's $\lambda = 0.98$. Since significance was not detected, we did not follow up with univariate ANOVAs.

Discussion

Our data support our 1st hypothesis that ML foot placement variability in sprinters with and without transtibial amputations generally increases with running speed up to maximum sprint speed. Our data also support our 2nd hypothesis that ML foot placement variability is symmetrical between the right and left legs of non-amputee sprinters and asymmetrically greater for the affected leg (with a running specific prosthesis) compared to the unaffected leg of sprinters with a unilateral transtibial amputation. Upon close inspection, we also found that increases in ML foot placement variability across speed differed between the affected and unaffected leg, highlighting the fact that the legs of each sprinter with a unilateral amputation showed a modest, yet asymmetrical response to increases in running speed up to maximum speed.

Table 2. Mean, standard deviation, and linear regression results based on a repeated measures MANOVA[b] comparing ML foot placement and ML foot placement variability across normalized speed for the Unilateral group (n = 6).

Main Effect: *leg*						
	AL		**UL**			
	Mean	SD	Mean	SD	Univariate ANOVA	*p* value
ML foot placement, cm	4.3	2.1	5.5	3.2	$F_{1,41} = 6.77$	0.013
ML foot placement variability, cm	1.9	0.9	1.8	0.6	$F_{1,41} = 5.02$	0.030
Interaction Effect: *leg*normalized speed*						
	Slope	Intercept	Slope	Intercept	Univariate ANOVA	*p* value
ML foot placement, cm	-0.83	4.89	3.46	2.95	$F_{1,41} = 19.07$	< 0.001
ML foot placement variability, cm	2.35	0.16	1.11	1.04	$F_{1,41} = 5.83$	0.020

[b]A one-way repeated measures MANOVA with normalized speed as a covariate revealed a significant main effect for "leg", indicating that ML foot placement and ML foot placement variability between the AL and UL were statistically different: $F_{2,40} = 5.10$, $p = 0.011$, Wilk's $\lambda = 0.80$. In addition, our statistical analysis revealed a significant interaction effect between "leg" and "normalized speed", indicating that the dependent regression lines quantifying the linear relation of ML foot placement or ML foot placement variability across normalized speed for the AL and UL were statistically different: $F_{2,40} = 10.99$, $p < 0.001$, Wilk's $\lambda = 0.65$ Since significance was detected, we followed up with univariate ANOVAs.

Modulation of ML foot placement across speed

Although we did not attempt to statistically compare between groups, we would be remiss, given the unique nature of our data set, if we did not make some general observations. Overall, all sprinters adopted a positive foot placement while running and sprinting, i.e. placing the foot lateral to the body's midline. As they approached maximum speed, non-amputee sprinters tended to place their feet closer to the body's midline. On the contrary, sprinters with unilateral or bilateral transtibial amputations did not show a systematic tendency to place their feet closer to the midline. One explanation for this could be that modulating foot placement across speed might be difficult for sprinters with an amputation because they lack tactile exteroreception and proprioception in their affected leg. Indeed, a combination of afferent input arising from the leg and foot plays an important role in proper foot placement during running, which is required for controlling the body's center of mass trajectory [17]. Another possibility is that sprinters with a unilateral amputation place each foot slightly away from the midline to compensate for inertial and ground force asymmetries that exist between the affected and unaffected legs [1]. For the sprinter with bilateral transtibial amputations, ML foot placement between the right and left affected leg exhibited the highest asymmetry across speed. For the same reasons mentioned above, it seems reasonable to postulate that the loss of proprioception and fine muscular control in both legs would make it more difficult to modulate foot placement from step-to-step.

Increase in ML foot placement variability across speed

In general, a steady increase in speed coincided with steady increases in ML foot placement variability. This modest, but steady increase in ML foot placement variability suggests that maintaining lateral balance becomes more challenging at faster running speeds up to a maximum normalized sprint speed of 1.0 [5, 6]. While the increase in ML foot placement variability between the right and left legs was similar for non-amputee sprinters, the variability of the affected leg in sprinters with a unilateral amputation generally increased to a greater extent than the unaffected leg across the two-fold increase in normalized speed (Fig. 5). However, one must be cautious in generalizing these average regression trends because the changes in our foot placement data were highly variable between sprinters with a unilateral transtibial amputation. For example, ML foot placement variability of the affected leg increased more dramatically than the unaffected leg across normalized speed for 3 out of the 6 sprinters with a unilateral amputation. The major finding to take away from our average regression analysis is that increases in ML foot placement variability across speed were asymmetric between the affected and unaffected leg.

Symmetry vs. Asymmetry

As expected, we found similar changes in ML foot placement and its variability between the right and left legs across speed, reflecting a high degree of symmetry between the biological legs of non-amputee sprinters. Statistically speaking, the average correlation coefficient expressing the strength of the linear trend between ML foot placement variability and normalized speed ($r = 0.61$) falls within a moderate to large magnitude [18], suggesting a moderate to substantial effect of increased speed on increased ML foot placement variability. In contrast, the large differences in the values of the correlation coefficients (r_{UL} vs. r_{AL}) demonstrate that changes in ML foot placement and its variability across speed were asymmetric between the affected and unaffected leg of sprinters with a unilateral transtibial amputation. Previous studies of individuals with a transtibial amputation have identified important asymmetries in sagittal plane kinematics and kinetics while running and sprinting [1, 2, 7, 8, 10, 11]. Here, we demonstrate that sprinters with a unilateral transtibial amputation also exhibit biomechanical asymmetries in the frontal plane. In general, however, the average increase in ML foot placement variability across speed in sprinters with a unilateral amputation fell within the range observed in non-amputee sprinters. Our data suggest that when running at faster speeds up to maximum sprint speed, sprinters with a unilateral transtibial amputation found maintaining lateral balance just as challenging as non-amputee sprinters.

Perhaps our most intriguing data are from the sprinter with bilateral transtibial amputations, who exhibited the greatest increases in ML foot placement variability with speed. Our supplementary videos provide an immediate and intuitive appreciation of our empirical data (see S1–S12 Videos, which demonstrates a non-amputee sprinter, a sprinter with a unilateral transtibial amputation, and a sprinter with bilateral transtibial amputations running at their slowest speed and sprinting at their maximum speed). Video representations of the running and sprinting kinematics of these sprinters in the sagittal plane (side-view) appear very similar. However, it is only when one examines the running and sprinting performance from the frontal plane (rear-view) that obvious differences visually emerge in ML foot placement from step-to-step. This affect can be appreciated when comparing the sprinter with bilateral transtibial amputations to a non-amputee sprinter and a sprinter with a unilateral transtibial amputation. When compared to all other sprinters in this study, our data suggest that the sprinter with bilateral transtibial amputations experienced the greatest challenge with maintaining lateral balance, especially at his maximum sprint speed. However, we interpret these findings with caution because we only collected data on one elite sprinter with bilateral transtibial amputations who at the time had only two years of experience using running-specific prostheses. In our future work, we hope to investigate ML foot placement and its variability in additional sprinters with bilateral transtibial amputations and to determine if more experience with the use of running-specific prostheses improves the ability to modulate foot placement from step-to-step.

Although we use measurements of ML foot placement variability as an indicator of lateral balance, some may consider our simple approach to be a limitation. We recognize that foot placement variability measures alone do not quantify dynamic stability nor do they provide rigorous insights into the control of balance from a first-principles approach. We emphasize here that we did not attempt to quantify dynamic stability or undertake a first-principles approach. Although it was beyond the scope of our present analyses, we have recently pursued nonlinear measures of dynamic stability that have been previously used to study human running [19]. In brief, we quantified the maximal Lyapunov exponent (LyE) from the leg joint dynamics of sprinters without and with a unilateral transtibial amputation [20]. We found that the maximal LyE increased as running speed increased, indicating that sprinters were less stable as running speed increased. Furthermore, while dynamic stability was similar between the biological legs of non-amputee sprinters, we found that the dynamics of the affected leg were less stable than the unaffected leg in sprinters with a unilateral transtibial amputation, revealing notable asymmetries. Taken together, our simple foot placement variability measures reveal similar trends as the relatively complex nonlinear measures of dynamic stability.

Other potential limitations of our study include our relatively small sample size, using the COM*p* as a proxy for the body's midline, the use of a treadmill, and differences in maximum speed performances between subjects. Because we chose to focus on elite Paralympic sprinters, our sample size of sprinters with transtibial amputations was small, however, we believe that our unique population allowed us to understand how the use of running-specific prosthesis affects ML foot placement and its variability. We used the COM*p* to define the body's midline, but ideally would have measured ML foot placement relative to the whole body COM (COM*wb*). We did not attempt to calculate COM*wb* because of potential inaccuracies in modeling the running-specific prostheses used by the sprinters with transtibial amputations. We found that the COM*p* is a good reference for the body's midline, and moreover, using the COM*p* as the body's midline kept the point of reference the same for all groups. While there may be differences between the two (COM*p* vs. COM*wb*), they are likely small and consistent in the ML direction, thereby having little influence on our results. All subjects had experience with treadmill running and sprinting prior to participating in the experiment, but differences in ML foot placement and its variability may exist between treadmill and over-ground running and sprinting. In addition, as opposed to using a counterbalanced or random design, it is possible that incrementally increasing treadmill speed up to each sprinters maximum speed might have biased our results. To minimize these potential effects, we allowed subjects a full recovery *ad libitum* following each bout of running or sprinting. Finally, the training and best performances of these athletes may be relevant to one's ability to properly modulate ML foot placement from step-to-step and as an important consequence, lateral balance. As noted in Figs. 3 and 4, the maximum speed achieved by each sprinter ranged from 7.0–9.7 m/s. It would be worthwhile in the future to compare athletes based on different levels of sprinting experience and training background. Nonetheless, we feel that the maximum sprint speed performances of all athletes in this experiment gave an accurate description of their sprinting ability.

Conclusions

Because the use of running-specific leg prostheses are becoming more common, measures of ML foot placement and its variability may be useful for those seeking simple metrics to assess potential balance problems in runners and sprinters. Overall, our study highlights several important findings about sprinters with and without transtibial amputations. We infer from our results that 1) when compared to slow speeds, maintaining lateral balance is more challenging at faster running speeds up to maximum sprint speed and 2) sprinters with a unilateral transtibial amputation found maintaining lateral balance just as challenging as non-amputee sprinters. Furthermore, the apparent asymmetries in ML foot placement and its variability suggest that the use of running-specific prostheses results in a compensatory foot placement strategy for maintaining lateral balance in sprinters with a unilateral transtibial amputation. Finally, when compared to all other sprinters in our subject pool, the sprinter with bilateral transtibial amputations exhibited the greatest increases in ML foot placement variability across normalized speed, indicating that maintaining lateral balance was the most challenging for this Paralympic sprinter.

Supporting Information

S1 Table. Table that details the anthropometric and biomechanical characteristics for each sprinter that participated in this study.
(DOCX)

S1 Appendix. The data underlying the findings in this study.
(XLS)

S1 Video. Video from the side-view showing a non-amputee sprinter at his slowest speed of 3.0 m/s.
(MOV)

S2 Video. Video from the rear-view showing a non-amputee sprinter at his slowest speed of 3.0 m/s.
(MOV)

S3 Video. Video from the side-view showing a non-amputee sprinter at his maximum speed of 9.0 m/s.
(MOV)

S4 Video. Video from the rear-view showing a non-amputee sprinter at his maximum speed of 9.0 m/s.
(MOV)

S5 Video. Video from the side-view showing a sprinter with a unilateral transtibial amputation at her slowest speed of 3.0 m/s.
(MOV)

S6 Video. Video from the rear-view showing a sprinter with a unilateral transtibial amputation at her slowest speed of 3.0 m/s.
(MOV)

S7 Video. Video from the side-view showing a sprinter with a unilateral transtibial amputation at her maximum speed of 9.0 m/s.
(MOV)

S8 Video. Video from the rear-view showing a sprinter with a unilateral transtibial amputation at her maximum speed of 9.0 m/s.
(MOV)

S9 Video. Video from the side-view showing a sprinter with bilateral transtibial amputations at his slowest speed of 3.0 m/s.
(MOV)

S10 Video. Video from the rear-view showing a sprinter with bilateral transtibial amputations at his slowest speed of 3.0 m/s.
(MOV)

S11 Video. Video from the side-view showing a sprinter with bilateral transtibial amputations at his maximum speed of 8.9 m/s.
(MOV)

S12 Video. Video from the rear-view showing a sprinter with bilateral transtibial amputations at his maximum speed of 8.9 m/s.
(MOV)

Spatiotemporal Parameters of 100-m Sprint in Different Levels of Sprinters with Unilateral Transtibial Amputation

Abstract

The aim of this study was to investigate differences of the spatiotemporal parameters in a 100-m sprint among elite, sub-elite, and non-elite sprinters with a unilateral transtibial amputation. Using publicly available Internet broadcasts, we analyzed 125, 19, and 33 records from 30 elite, 12 sub-elite, and 22 non-elite sprinters, respectively. For each sprinter's run, the average velocity, step frequency, and step length were calculated using the number of steps in conjunction with the official race time. Average velocity was greatest in elite sprinters (8.71±0.32 m/s), followed by the sub-elite (8.09±0.06 m/s) and non-elite groups (7.72±0.27 m/s). Although there was a significant difference in average step frequency between the three groups, the effect size was small and the relative difference among the three groups was 3.1%. Statistical analysis also revealed that the average step length was longest in elite sprinters, followed by the sub-elite and non-elite groups. These results suggest that the differences in sprint performance between the three groups is mainly due to the average step length rather than step frequency.

Introduction

Running-specific prostheses (RSPs) with energy storing capabilities have attracted more and more individuals with lower extremity amputations to running as a form of exercise and athletic competition. More recently, RSPs have allowed amputee runners to compete at athletic levels achieved never before [1, 2]. Theoretically, the average velocity during a 100-m sprint is the product of the average step frequency and average step length. Although both parameters are inversely correlated, an increase in one factor will result in an improvement in sprint velocity, as long as the other factor does not undergo a proportionately similar or larger decrease. Because spatiotemporal parameters are modifiable by sprint training sessions [3], identifying factors affecting these parameters of 100-m sprints in unilateral transtibial amputees will provide coaches and practitioners with a basis for better evaluation of sprint performance and aid in the development of more effective training methods for amputee sprinters.

Several studies have compared biomechanical characteristics between able-bodied sprinters and amputee runners using RSPs during maximal and submaximal running [4–7]. The results of these studies are useful to aid in developing ideal coaching and training regimes by identifying the underlying differences between the two groups. On the other hand, despite the fact that examining sprint performance among different levels of sprinters is useful for training-conditioning programs and the design of effective talent development, less research attention has been given to examining differences in biomechanical characteristics in amputee sprinters using RSPs.

In a previous study, Gajer et al. [8] compared sprint performance between faster and slower groups in able-bodied athletes. The authors found that the faster group had a longer stride length during the entire 100-m race than the slower group. Further, Hunter et al. [9] and Weyand et al. [10] also suggested that step length has strong association with the sprint velocity in able-bodied athletes. However, it is unknown if this longer step length of elite group can be observed in amputee sprinters using RSPs.

The aim of this study was to investigate the differences in the spatiotemporal parameters of a 100-m sprint among elite, sub-elite, and non-elite sprinters with a unilateral transtibial amputation. We hypothesized that the differences in sprint performance between the three groups would mainly be due to the average step frequency rather than step length.

Methods

Data collection

Since our data set was obtained from publicly-available Internet broadcasts, we did not obtain informed consent. Institutional review board (Environment and Safety Headquarters, Safety Management Division, AIST) approval was obtained prior to the study. We analyzed 177 races of 64 sprinters with lower extremity amputations from publicly available Internet broadcasts. Based on the classification system created by the International Paralympic Committee (IPC; http://www.paralympic.org/), we included the Men's T44 class (defined as any athlete with lower limb impairment/s that meets the minimum disability criteria for lower limb deficiency, impaired lower limb passive range of motion, impaired lower limb muscle power, or leg length difference). These races included several Paralympics, the IPC Athletics World Championships, and other national- and international-level competitions from 2004 to 2015 (Table 1). Individual races were excluded from the analysis if the athlete did not complete the race or the athlete's body was not visible throughout the entire race. T44 sprinters who did not use RSPs were also excluded. Analyzing publicly available data from sport competitions for research purposes were performed by Salo et al. [11] for sprint running using 52 able-bodied sprinters, and by Hobara et al. [12] for prosthetic sprinting using 36 able-bodied and 42 amputee sprinters (25 unilateral and 17 bilateral transtibial amputees).

In the present study, we separated the whole population into three groups based on qualification standards. The elite group (EL) consisted of sprinters who satisfied the A-Qualification Standards of the Men's T44 class (12.20 s) in the analyzed races (London 2012 Paralympic Games Qualification Guide-Athletics, 2011). The sub-elite group (SEL) consisted of sprinters who could not reach the A-Qualification Standards, but satisfied B-Qualification Standards (12.50 s). The non-elite group (NEL) consisted of sprinters who could not reach the B-Qualification Standards. Consequently, the EL, SEL, and NEL groups consisted of 125 (30 sprinters), 19 (12 sprinters), and 33 (22 sprinters) data, respectively.

Data analyses

As stated in previous studies [12–14], we determined the average velocity (V_{100}) of each individual by dividing the official race distance (100 m) by the official race times (t_{race}), which were

Table 1. Summary of the competitions analyzed. EL, SEL and NEL indicates elite-, sub-elite and non-elite group, respectively.

		Number of Subjects		
Year	**Competitions**	**EL**	**SEL**	**NEL**
2015	ASEAN Paralympic Games			5
2015	IPC Athletics	11	2	
2015	Parapan 2015	1		
2015	IPC Grandprix London	6		
2015	Mano a Mano Challenge	3		
2015	Jogos Paralimpicos Falta 1 Ano	1		
2015	SEIKO Super Athetics	3	1	1
2015	Shizuoka International	1	1	2
2015	7th Fazza IPC Grand Prix Athletics Competition Dubai	1		
2014	IPC Athletics Grand Prix, Berlin	3		
2014	IPC European Athletics	4		
2014	Meeting De Montreuil	4		
2014	IAAF Diamond League Glasgow 2014	7		
2014	Japan Nationals	1	1	2
2014	Great City Games Manchester	2		
2014	Kumamoto Challenges			1
2014	Kanto Championship	1		
2014	Shizuoka International		3	
2013	IWAS			2
2013	Japan Paralympic			2
2013	Sainsbury's Anniversary Games	6		
2013	Great North City Games	3	1	
2013	T-Meeting Tilburg	1		
2013	IPC Athletics	18		
2013	Shizuoka International	2		1
2012	London DAC	1		1
2012	IDM Leichtathletik German Open Athletics	1		
2012	London Paralympic	11		1
2012	IPC European Championship	4		
2012	Mt. Sac Relays	3		
2012	Occidental Oxy Invite	1		
2012	US National	1		
2012	JPN National	2	2	
2011	Oita Athletics	1	2	
2011	IDM Singen	1		
2011	IWAS			3
2011	US National	2		
2011	Japan Paralympic	1	1	2
2011	Japan Nationals			2
2011	IPC Athletics	5	1	1
2011	Kyushu Challenge Athletics			1
2010	Asia Paralympic		2	2
2009	Manchester BT Paralympic World Cup	1		
2008	Beijing Paralympic	7		
2007	Parapan 2007	1		
2004	Athens Paralympic	3	2	4
	total	**125**	**19**	**33**

obtained from each competition's official website; thus,

$$V_{100} = 100/t_{race.} \tag{1}$$

In the present study, we calculated average step frequency (f_{step}) as

$$f_{step} = N_{step}/t_{race,} \tag{2}$$

where N_{step} is the number of steps, which was manually counted by the authors. Because V_{100} is the product of f_{step} and average step length (L_{step}), we calculated the L_{step} by

$$L_{step} = V_{100}/f_{step}. \tag{3}$$

If we could not count the number of steps, we excluded the data from our analyses. The last step before the finish line was considered to be the last step. If an athlete's foot was located on the finish line, we considered it as a step [15].

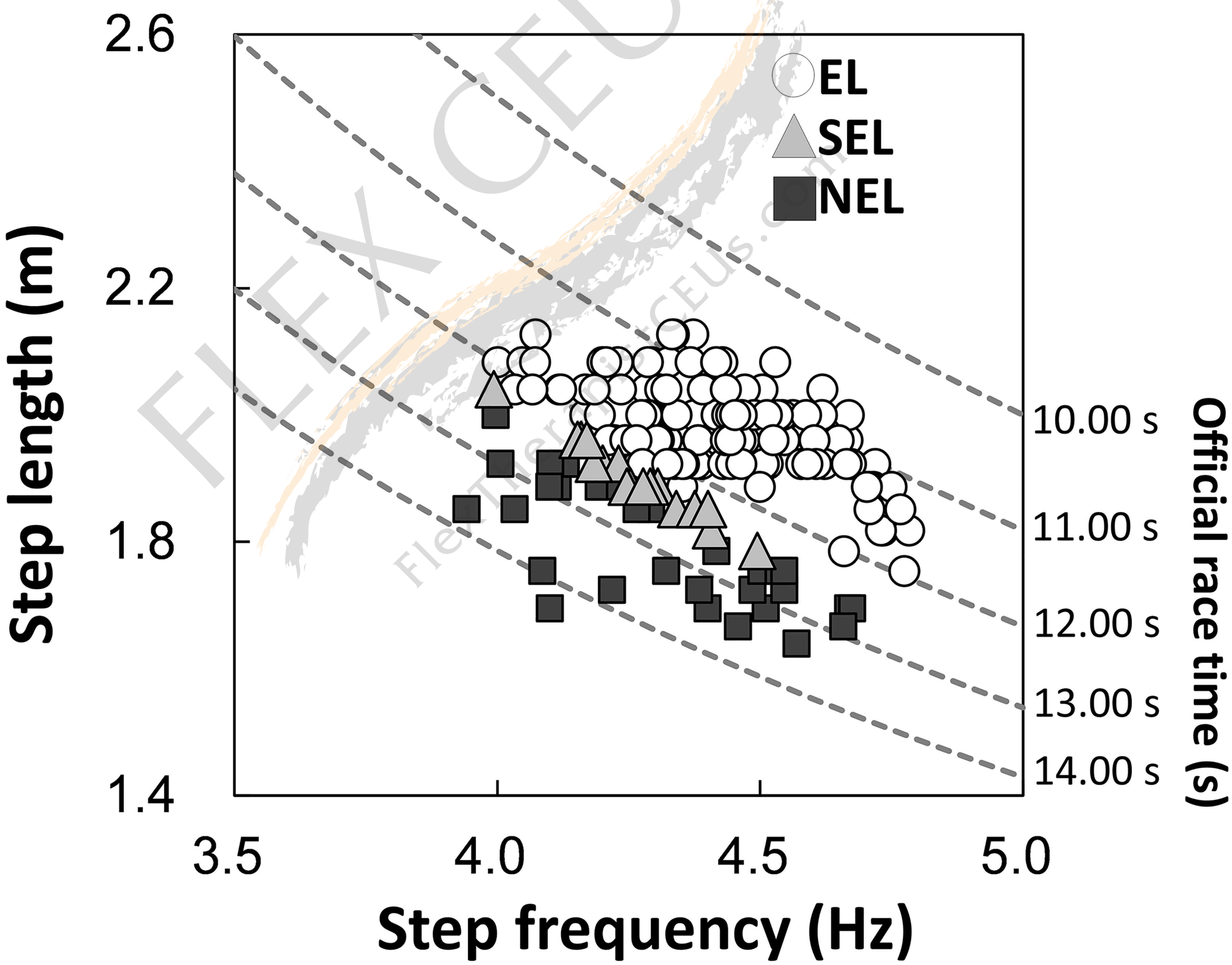

Fig 1. Relationship between f_{step} and L_{step} for the three groups. Unfilled circles, gray triangles, and filled squares indicate the data for elite (EL), sub-elite (SEL) and non-elite (NEL) groups, respectively. Dotted lines denote the official race times calculated using the combination of f_{step} and L_{step}.

Statistics

Before the statistical analyses, Levene's test was performed to ensure that the assumptions of normality and homogeneity of variance were met. Since the assumptions were violated in our data, the Kruskal-Wallis test was used to compare V_{100}, f_{step}, and L_{step} among the EL, SEL, and NEL groups. We also calculated the effect size (ES) for the Kruskal-Wallis test using Cramer's V. From this effect size calculation, the results were interpreted as small (0.1 to 0.3), medium (0.3 to 0.5), or large (higher than 0.5). If a significant main effect was observed, the Mann-Whitney U test with a Bonferroni correction as post hoc multiple comparison was repeated for all combinations in each variables. Because there were three Mann-Whitney U tests in each variable, the alpha levels were set at 0.016 (0.05/3) and 0.003 (0.01/3). Statistical significance was set at $P < 0.05$. These statistical analyses were executed using SPSS version 19 (IBM SPSS Statistics Version 19, SPSS Inc., Chicago, IL).

Results

Fig 1 shows the f_{step}–L_{step} plot for all the individuals in the three groups. Dotted lines indicate the times predicted using the combination of f_{step} and L_{step}. As shown in Fig 2, V_{100} exhibited a

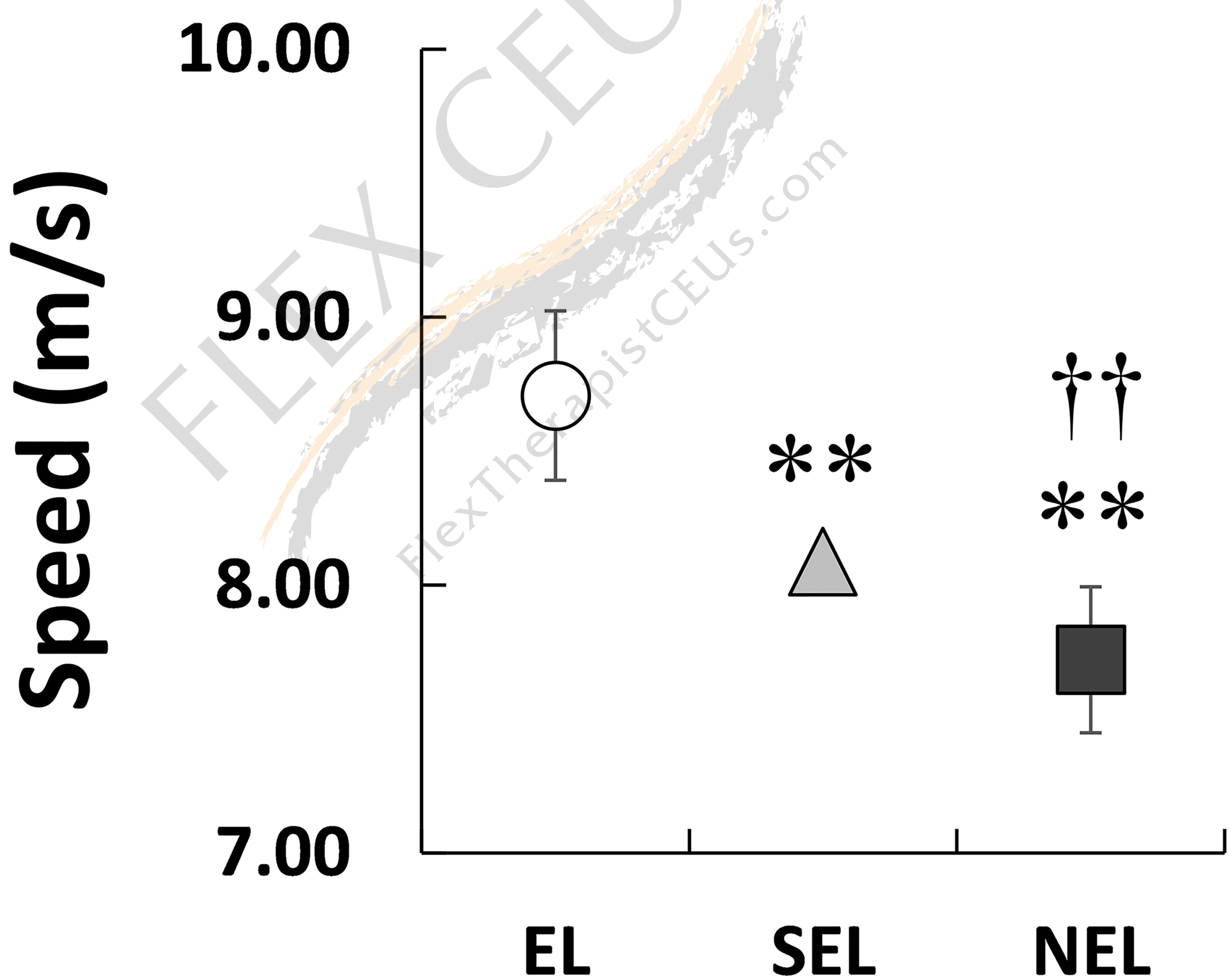

Fig 2. Comparisons of averaged velocity among three groups. Asterisks (**) indicate significant differences with EL at $P < 0.01$. Daggers (††) indicate significant differences with SEL at $P < 0.01$.

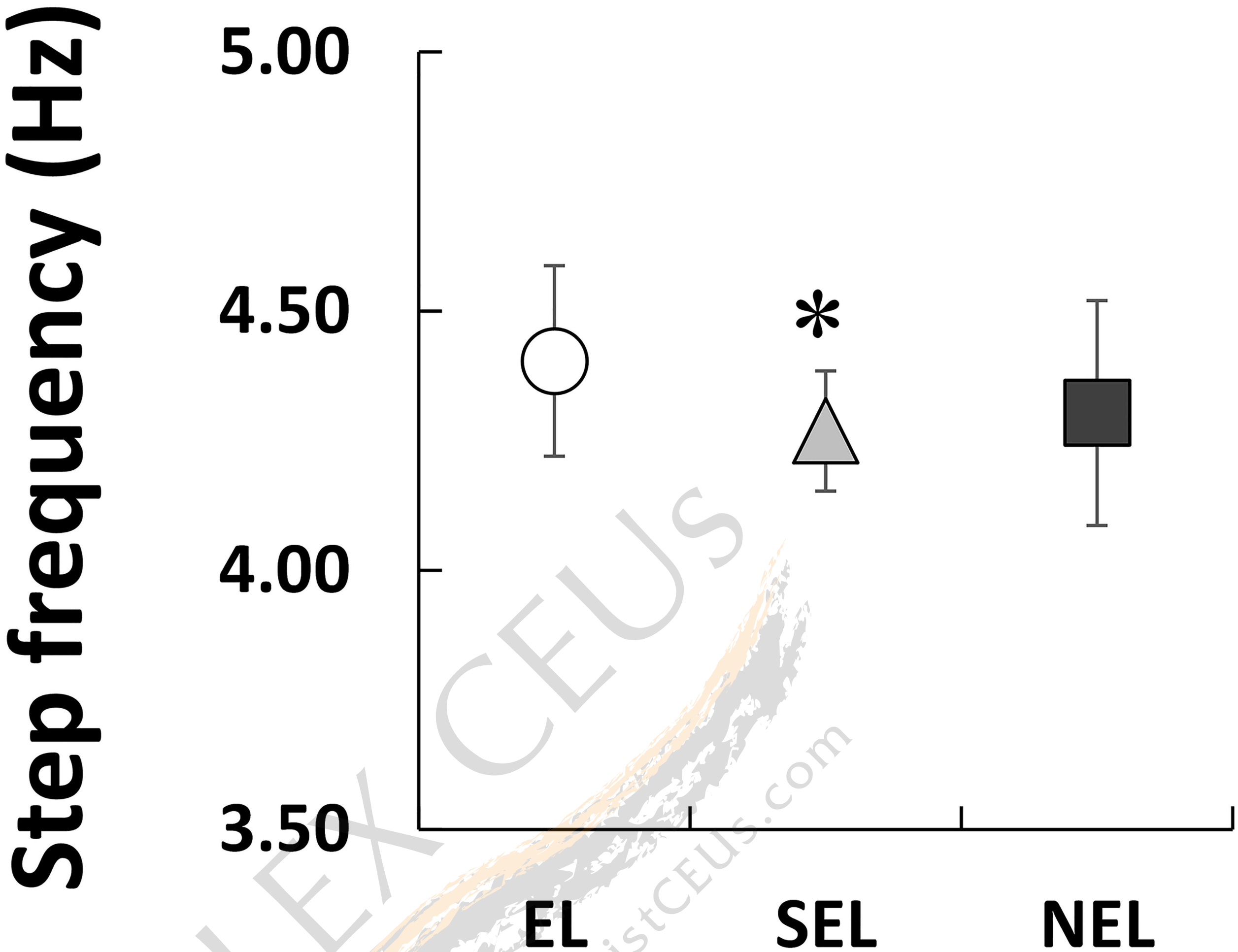

Fig 3. Comparisons of averaged step frequency among three groups. An asterisk (*) indicates a significant difference with EL at $P < 0.05$.

significant main effect on the groups ($X^2(2) = 112.66$, $P < 0.01$, ES = 0.57). V_{100} was greatest in EL, followed by SEL and NEL ($P < 0.003$ for all comparisons). Statistical analyses revealed that f_{step} had a significant effect on the groups (Fig 3; $X^2(2) = 13.184$, $P < 0.01$), and f_{step} in EL was significantly higher than SEL ($P < 0.016$) but not NEL. However, the relative difference between EL and SEL was 3.1%, and the ES of the Kruskal-Wallis test was 0.19 (small). L_{step} also displayed a significant effect on the groups (Fig 4; $X^2(2) = 72.58$, $P < 0.01$, ES = 0.45). Statistical analyses also revealed that L_{step} was longest in EL, followed by SEL and NEL ($P < 0.003$ for all comparisons).

Discussion

The aim of this study was to investigate the differences in the spatiotemporal parameters of a 100-m sprint among elite, sub-elite, and non-elite sprinters with a unilateral transtibial amputation. In the present study, V_{100} was greatest in EL sprinters, followed by SEL and NEL (Fig 2). Although a statistically significant difference in f_{step} between the three groups was identified (Fig 3), the ES for this effect was small (0.19). On the other hand, L_{step} was the longest in EL, followed by SEL and NEL (ES = 0.45, medium; Fig 4). Therefore, the results of the present study support our initial hypothesis that the differences in sprint performance between the three groups would mainly be due to the L_{step} rather than the f_{step}.

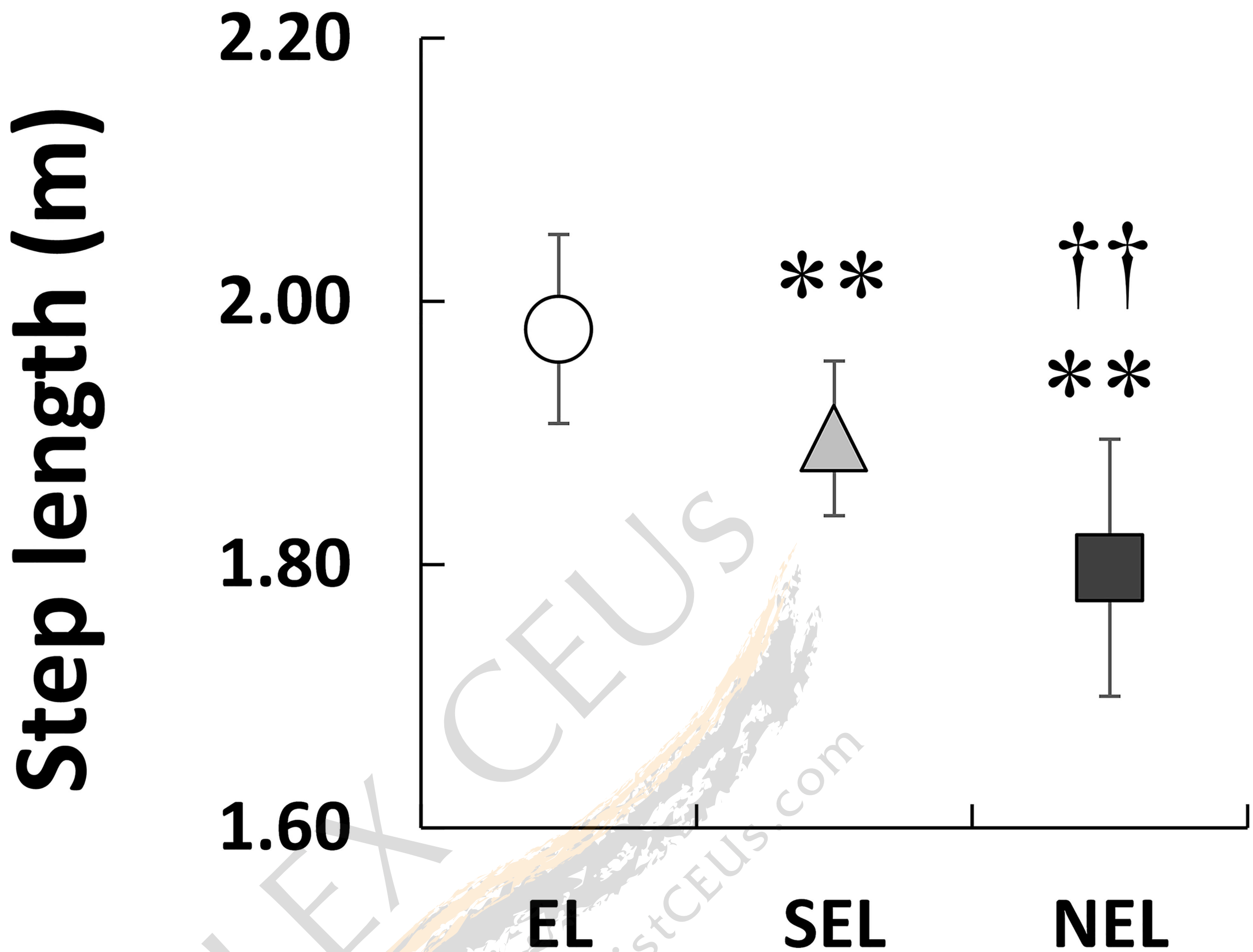

Fig 4. Comparisons of averaged step length among three groups. Asterisks (**) indicate significant differences with EL at $P < 0.01$. Daggers (††) indicate significant differences with SEL at $P < 0.01$.

In a previous study, Hunter et al. [9] introduced a deterministic model for sprint running, especially for both the f_{step} and L_{step}. Based on the deterministic model, determinants of f_{step} and L_{step} could be partly explained by the relative horizontal and vertical ground reaction force impulse, segment positions, segment inertial parameters, and air resistance. For the ground reaction forces, Rabita et al. [16] found that elite able-bodied sprinters in their study had a 9.7% greater force production capacity than sub-elite able-bodied sprinters. Furthermore, Fortier et al. [17] and Slawinski et al. [18] reported that elite able-bodied sprinters showed better force production capacity during the sprint start and subsequent steps than sub-elite able-bodied sprinters. Therefore, the differences in L_{step} among elite, sub-elite, and non-elite sprinters with a unilateral transtibial amputation in our study may be attributed to force production capacity during sprinting.

There are certain considerations that must be acknowledged when interpreting the results of the current study. First, we assumed the athletes completed a step exactly at the 100-m mark [12–14]. However, a previous study [11] subtracted a distance of 0.55 m and a time of 0.52 s from the calculations of averaged step frequency and step length based on their pilot test. Thus, the reliability and accuracy of the current data should be interpreted carefully. Second, we calculated the average step length using the number of steps taken and the official race time as data. However, Nagahara et al. [19] demonstrated that not all the steps in a 100-m sprint have the same length and frequency, indicating that the average step frequency and step length in the present study may not necessarily be representative of any particular part of the sprint. Thus, continuous changes in spatiotemporal parameters during 100-m sprint in the three groups should be investigated. Third, we only investigated athletes with unilateral transtibial amputations who participated in official competitions, such as the Paralympic Games, IPC Athletics World Championship, IPC European Championship, and other national and international competitions. Thus, caution should be exercised in interpretation and generalization of these findings to other classes, such as transfemoral amputees and bilateral transtibial amputees. Finally, although most of T44 sprinters from 2004 to 2015 generally used Flex-Foot Cheetah® (Össur), Cheetah® Xtreme™ (Össur), or 1E90 Sprinter (Ottobock) RSPs, we did not determine each individual's RSPs, which might indirectly influence the spatiotemporal parameters during sprinting [20–21]. Consequently, the differences in RSPs used within each of the groups may affect the spatiotemporal parameters of the 100-m sprint. Further research is required on whether and how sprint performance changes among different levels of sprinters with unilateral transtibial amputations.

Conclusion

In summary, we investigated differences in the spatiotemporal parameters of a 100-m sprint among elite, sub-elite, and non-elite sprinters with a unilateral transtibial amputation. The results of the present study suggest that the differences in sprint performance between the three groups is mainly due to the L_{step} rather than the f_{step}.

Supporting Information

S1 Appendix. The data underlying the findings in this study.
(XLSX)

Does amputation side influence sprint performances in athletes using running-specific prostheses?

Abstract

Background: For athletes using running-specific prostheses (RSPs), current Paralympic guidelines for track events are generally based on level of amputation, not side of amputation. Although 200- and 400-m sprint races are performed in a counterclockwise direction, little is known about the effects of amputation side on race performance in athletes with unilateral lower limb amputation. The study aim was to test whether athletes using RSPs on their left side have slower race times than those using RSPs on their right side.

Findings: Athletes with unilateral lower limb amputation (N = 59 in total) participating in elite-level 200-m races were analyzed from publicly available Internet broadcasts. These races included the 2008 Beijing and 2012 London Paralympics, and the International Paralympic Committee Athletics World Championships in 2011 and 2013. For each athlete the official race time and amputation side were determined. There was no significant difference in number of participants and race time between left and right side amputees in T42 men, T44 men, and T44 women.

Conclusion: The results of the present study suggest that sprint performance of athletes using RSPs is not affected by amputation side on a standard 400-m track.

Keywords: Prosthetic sprinting, Paralympic, Curved track, Regulation

Background

Recent technical developments in carbon fiber running-specific prostheses (RSPs) have helped amputee athletes to regain the functional capability of running. For athletes using RSPs, current Paralympic guidelines for track events (Tweedy 2010) are generally based on level of amputation, such as T42 (unilateral or bilateral transfemoral), T43 (bilateral transtibial), and T44 (unilateral transtibial), rather than side of amputation.

In 200-m sprint events, races are performed in a counterclockwise direction, beginning on the curve and ending on the home straight. A previous study (Chang and Kram 2007) demonstrated that during sprinting on a curved track, the inner leg consistently generates smaller peak forces than the outer leg, leading to a reduction of maximum performance of the entire locomotive system. Furthermore, several studies have suggested that the ground reaction forces of the RSPs are smaller than those of the intact leg during straight running (Hobara et al. 2013, 2014; Grabowski et al. 2010). Therefore, the goal of this study was to test the hypotheses that athletes using RSPs on their left leg would have slower race times than those using RSPs on their right leg.

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Methods

Male and female athletes with unilateral lower limb amputation (N = 59 in total) participating in elite-level 200-m races were analyzed from publicly available Internet broadcasts. The distribution of participants is shown in Table 1. Every performance at every competition was considered individually. The races included the 2008 Beijing and 2012 London Paralympics, and the International

Table 1 Number of participants included in this study and results of Chi square test

	Left	Right	Chi square test
T42 Men	7	9	$\chi^2 = 0.25, df = 1, p = 0.62$
T44 Men	7	13	$\chi^2 = 1.80, df = 1, p = 0.18$
T44 Women	13	10	$\chi^2 = 0.39, df = 1, p = 0.53$
Total	27	32	

Paralympic Committee Athletics World Championships in 2011 and 2013. A similar approach to analyze publicly available data from sports competitions for research purposes has previously been used in several studies (Salo et al. 2011; Hobara et al. 2015; Dyer et al. 2014). Institutional review board approval was obtained prior to study initiation.

First, we determined the race times from the official website of the Paralympic movement (http://www.paralympic.org/). Race times of 200-m sprints of tournament finals were collected. Second, we investigated the amputation side in each athlete through publicly available Internet broadcasts and other sources. Finally, all athletes were categorized into four classes (T42 men, T44 men, and T44 women) and side of amputation (left or right). In the case of T42 men, bilateral transfemoral amputees were excluded from analysis. Further, we also excluded athletes who did not use RSPs.

An independent *t* test was performed for race times to determine significant differences between sides of amputation in each event. Further, to determine whether there is a significant difference in the number of participants in each event, the Chi square test was used. Statistical significance was set at $p < 0.05$. All statistical analyses were performed using SPSS version 19 (IBM Corp., Armonk, NY, USA).

Results

Statistical analysis revealed there were no significant differences in race times between left and right side amputees during any event (Fig. 1). Further, it was found that there were no significant differences in the number of participants between two groups during any event (Table 1).

Discussion

As shown in Fig. 1, there were no significant differences in race times between left and right side amputees during any event. These results contrast with our initial hypothesis that athletes with left side amputations would have slower race times than those with right side amputations. A possible explanation for the similar race times between left and right side amputees may be the radius of curvature of a standard 400-m track. It has shown that the inside leg consistently generates smaller peak forces compared with the outside leg during curve sprinting (Chang and Kram 2007) and this leads to a reduction of maximum performance on the curved track. However, this previous study investigated circular tracks with radii of 1, 2, 3, 4, and 6 m (Chang and Kram 2007), whereas a standard 400-m track has a radius of 36.5 m (IAAF Track and Field Facilities Manual 2008 Edition). Therefore, the radius of a 400-m track might be too large to observe the same effect as in a previous study (Chang and Kram 2007).

Current Paralympic sports guidelines (Tweedy 2010) are based on level of amputation, such as unilateral transfemoral amputation (T42) and bilateral (T43) and unilateral (T44) transtibial amputation, not on the side of amputation. Despite the fact that 200- and 400-m sprint races are completed in a counterclockwise direction regardless of amputation side, only a few studies have investigated the effect of amputation side on race performance in athletes using RSPs. As shown in Fig. 1, there were no clear relationships between amputation side and

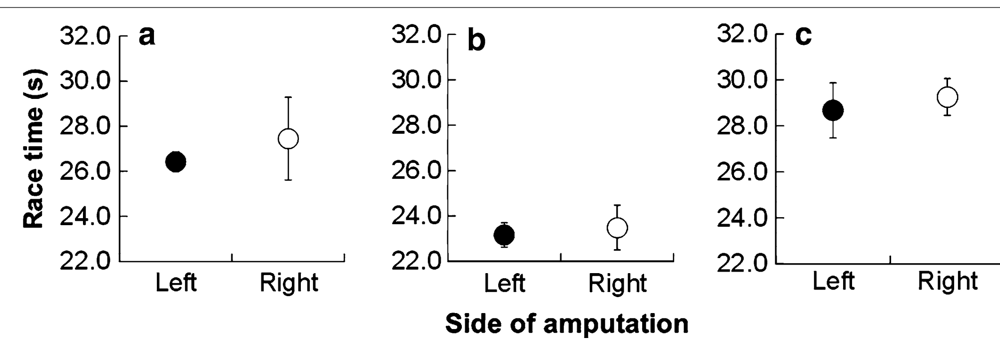

Fig. 1 Comparison of race times between left and right side amputations in men's 200-m T42 (**a**), men's 200-m T44 (**b**), and women's 200-m T44 (**c**). *filled* and *unfilled circles* indicate left and right side amputations in each event, respectively

race performance in unilateral amputees using RSPs. Further, there were no significant differences in the number of participants between two groups during any event (Table 1), indicating that there would not be any inherent bias against amputation side during a race. Hence, the results of the current study suggest that amputation side is a factor that does not need to be taken into consideration to ensure fairness in 200- and 400-m sprint events.

There are some limitations to this study. First, we only investigated athletes with unilateral amputation who participated in elite-level competitions, such as the Paralympics and the International Paralympic Committee Athletics World Championship. Thus, caution needs to be taken regarding interpretation and generalization of these findings toward novice and non-expert level athletes. Second, our results are based on retrospective, not experimental, observations. A recent study (Funken et al. 2014) demonstrated that one left sided unilateral transfemoral amputee can run faster in a clockwise direction than in a counterclockwise direction on a curved track due to the asymmetric kinetics. Further, in the present study, stump length in each class and prosthetic knee joint used in Men T42 may vary within each class of amputees. In addition, bilateral differences in strength, elasticity, leg length and balance control in unilateral amputees may induce excessive fatigue in sprinting on a curved track which might cause uneven energy consumption between the legs. Since these factors may affect 200-m sprint performance, further research is required on whether and how sprint performance on a curved track can be affected by amputation side.

Conclusion

In this study, we tested the hypotheses that athletes using RSPs on their left leg have slower race times than those using RSPs on their right leg. The results of the present study suggest that (1) 200-m sprint performance on a standard 400-m track in athletes using RSPs is not affected by amputation side, and (2) there are no inherent bias against amputation side in athletes using RSPs.