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RESEARCH ARTICLE

Prosthetic model, but not stiffness or height, affects the metabolic cost of running for athletes with unilateral transtibial amputations

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Beck ON, Taboga P, Grabowski AM. Prosthetic model, but not stiffness or height, affects the metabolic cost of running for athletes with unilateral transtibial amputations. *J Appl Physiol* 123: 38–48, 2017. First published March 30, 2017; doi:10.1152/japplphysiol.00896.2016.—Running-specific prostheses enable athletes with lower limb amputations to run by emulating the spring-like function of biological legs. Current prosthetic stiffness and height recommendations aim to mitigate kinematic asymmetries for athletes with unilateral transtibial amputations. However, it is unclear how different prosthetic configurations influence the biomechanics and metabolic cost of running. Consequently, we investigated how prosthetic model, stiffness, and height affect the biomechanics and metabolic cost of running. Ten athletes with unilateral transtibial amputations each performed 15 running trials at 2.5 or 3.0 m/s while we measured ground reaction forces and metabolic rates. Athletes ran using three different prosthetic models with five different stiffness category and height combinations per model. Use of an Ottobock 1E90 Sprinter prosthesis reduced metabolic cost by 4.3 and 3.4% compared with use of Freedom Innovations Catapult [fixed effect (β) = -0.177; $P < 0.001$] and Össur Flex-Run ($\beta = -0.139$; $P = 0.002$) prostheses, respectively. Neither prosthetic stiffness ($P \geq 0.180$) nor height ($P = 0.062$) affected the metabolic cost of running. The metabolic cost of running was related to lower peak ($\beta = 0.649$; $P = 0.001$) and stance average ($\beta = 0.772$; $P = 0.018$) vertical ground reaction forces, prolonged ground contact times ($\beta = -4.349$; $P = 0.012$), and decreased leg stiffness ($\beta = 0.071$; $P < 0.001$) averaged from both legs. Metabolic cost was reduced with more symmetric peak vertical ground reaction forces ($\beta = 0.007$; $P = 0.003$) but was unrelated to stride kinematic symmetry ($P \geq 0.636$). Therefore, prosthetic recommendations based on symmetric stride kinematics do not necessarily minimize the metabolic cost of running. Instead, an optimal prosthetic model, which improves overall biomechanics, minimizes the metabolic cost of running for athletes with unilateral transtibial amputations.

NEW & NOTEWORTHY The metabolic cost of running for athletes with unilateral transtibial amputations depends on prosthetic model and is associated with lower peak and stance average vertical ground reaction forces, longer contact times, and reduced leg stiffness. Metabolic cost is unrelated to prosthetic stiffness, height, and stride kinematic symmetry. Unlike nonamputees who decrease leg stiffness with increased in-series surface stiffness, biological limb stiffness for athletes with unilateral transtibial amputations is positively correlated with increased in-series (prosthetic) stiffness.

asymmetry; prosthesis; amputee

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BIOLOGICAL LEGS behave like linear springs during level-ground running (18, 31, 55). From initial ground contact through midstance, tensile forces elongate and store considerable mechanical energy in the elastic structures of the runner's stance leg (i.e., tendons and ligaments) (4, 5, 15, 47, 51, 70). Subsequently, the stored energy is released as the elastic structures recoil and help extend the leg throughout the second half of stance (4). These stance phase running mechanics are well characterized by a spring-mass model, which depicts the stance leg as a massless linear spring supporting a point mass that represents the runner's center of mass (18, 31, 38, 55) (Fig. 1).

The spring-mass model well characterizes running mechanics (18, 31, 38, 55, 57) but fails to explain the metabolic cost of running. Unlike the model's depiction, muscles produce force to allow elastic energy storage, thus consuming metabolic energy (49–51). Furthermore, biological legs do not recycle all of the mechanical energy needed to sustain running (15, 47); therefore, leg muscles change length while producing force, which may constitute a portion of the metabolic cost of running (63). Moreover, athletes modulate their muscular demands by changing running mechanics, accordingly affecting their metabolic cost. For instance, prolonging ground contact time and reducing stance average ground reaction force (GRF) magnitude during running yields more economical muscular force production (9, 26, 49, 50, 64). Muscular force magnitude also depends on the leg's effective mechanical advantage (16, 17, 48), which along with the rate of producing force, is associated with leg stiffness and step frequency (32, 48) at a given running speed. Thus, by changing stride kinematics and kinetics, athletes may be able to minimize their metabolic cost of running and improve their distance running performance (28, 45).

The use of passive-elastic running-specific prostheses (RSPs), which emulate the spring-like function of biological legs, allows athletes with unilateral transtibial amputations to run. An RSP connects in-series to the residual limb via a prosthetic socket. RSPs emulate the mechanics of biological legs by storing and returning elastic energy during running (13, 15, 18, 21, 31, 47, 70). Since the commercialization of RSPs in the 1980s (58), the athletic achievements of athletes with transtibial amputations have improved remarkably (43). Ensuing prosthetic design iterations, such as the removal of the prosthetic "heel" component, have further enhanced running performance (43, 46). Yet, despite the improved performance of athletes with transtibial amputations, the prescription of prosthetic model, stiffness, and height are subjective and may not optimize running performance.

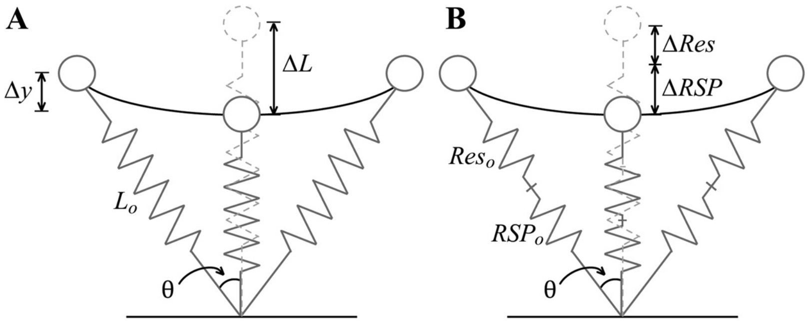

Fig. 1. *A*: illustration of the spring-mass model of running for nonamputees and the unaffected leg of athletes with a transtibial amputation. *B*: illustration of a spring-mass model of running with an in-series leg spring for the affected leg of athletes with a transtibial amputation. Body mass is represented as a point mass (circle), and the touch-down angle is indicated by theta (θ). The stance leg is represented by a massless linear spring (*A*) or two in-series massless linear springs (*B*). The initial leg length (L_0) shortens (ΔL), as its vertical height (Δy) lowers during the stance phase of running. Modeled residual limb length (Res_0) and prosthetic height (RSP_0) compress (ΔRes and ΔRSP) during the stance phase of running.

After prosthetists and athletes arbitrarily select a prosthetic model, which vary in design (35a, 59a, 59b), prosthetists recommend stiffness and height based on the manufacturer's recommendations and their own experience. Prosthetic stiffness recommendations are based on the mass of the athlete, with larger/heavier athletes prescribed stiffer RSPs (13, 35a, 59a, 59b). The recommended prosthetic stiffness (kN/m) for a given user body mass has yet to be standardized across prosthetic models (13); thus, considerable variability exists among recommended prosthetic stiffness. Prosthetic height is recommended so that the affected leg length, which includes the RSP, is 2–5 cm longer than the unaffected leg length. In reality, affected leg length is set at the discretion of the prosthetist and athlete, and has been reported to range 0.3–8.0 cm taller than the unaffected leg (36, 60). Rather than using subjective prosthetic model, stiffness, and height recommendations that may not optimize the metabolic cost of running (13), we aim to determine whether prosthetic model, stiffness, and/or height affect the metabolic cost of running, and, if so, we seek to determine the prosthetic configuration that minimizes metabolic cost, thus optimizing distance running performance (28, 45).

Previous research of nonamputees indicates that reducing surface stiffness lowers the metabolic cost of running (48, 68); therefore, reducing prosthetic stiffness may lower the metabolic cost of running for athletes with unilateral transtibial amputations. While running, nonamputees adjust their leg stiffness to accommodate different surface stiffness so that the combination of their leg and the surface maintains a constant total stiffness (33, 34, 48). They adapt to compliant surfaces in part by better aligning their leg joints with the resultant GRF vector (56), thereby improving the effective mechanical advantage of their leg joints (16, 17, 48). Additionally, compliant elastic surfaces recycle mechanical energy, theoretically mitigating the required muscular work needed to sustain running velocity (48). Together, these biomechanical adaptations to running on compliant surfaces have been related to a reduced metabolic cost of running (48, 68). For example, Kerdok et al. (48) reported that a 12.5-fold decrease in surface stiffness reduced the metabolic cost of running 3.7 m/s by 12% (48). Hence, reduced prosthetic stiffness compared with the manufacturer recommended stiffness may lower the metabolic cost of running for athletes with unilateral transtibial amputations due to decreased muscular work and increased residual limb stiffness (better effective mechanical advantage).

Prosthetic height may also influence the metabolic cost of running. Increased prosthetic height may prolong the affected leg's ground contact time (73, 74), enabling more economical force production (49, 62). Alternatively, the effective mechanical advantage of the leg joints would worsen with invariant joint angles and taller prostheses. Nonetheless, it is unknown whether prosthetic height affects the metabolic cost of running for athletes with unilateral transtibial amputations.

Athletes with unilateral transtibial amputations exhibit asymmetric stride kinematics and kinetics (10, 12, 22, 27, 36, 41, 42, 54, 61, 66, 75), which are likely a consequence of the RSPs' inability to replicate biological lower leg function. Accordingly, prosthetic manufacturers and prosthetists recommend stiffness and height configurations that mitigate stride kinematic asymmetries between the legs of athletes with unilateral transtibial amputations (35a, 59a, 59b). In line with these recommendations, a preliminary study by Wilson et al. (75) reported that changing prosthetic stiffness and height altered the stride kinematic and kinetic asymmetry for two athletes with unilateral transtibial amputations during running. Perhaps the metabolic cost of running is correlated with the severity of stride kinematic and/or kinetic asymmetry. Previous studies have reported positive associations between stride kinematic and kinetic asymmetries and the metabolic cost of walking for young healthy subjects (29, 35), and for individuals with unilateral transtibial amputations (39). On the other hand, Mattes et al. (53) reported that the metabolic cost of using passive prostheses during walking for individuals with unilateral transtibial amputations is greater when lower limb mass and moments of inertia are symmetric between legs. Yet, it is uncertain if these walking studies translate to running. Seminati et al. (67) reported that nonamputees with slightly asymmetric lower limbs run with more pronounced stride kinematic asymmetries while consuming metabolic energy at the same rate as nonamputees with symmetric lower limbs and biomechanics. Additionally, Brown et al. (20) reported that athletes with unilateral transtibial amputations (with presumably asymmetric biomechanics) consume oxygen at similar rates as age- and fitness-matched nonamputees (with presumably symmetric biomechanics) across running speeds, indicating that asymmetric running biomechanics may not exacerbate the metabolic cost of running for athletes with unilateral transtibial amputations. Due to the current prescription of RSPs, which aim to minimize kinematic asymmetries for athletes with unilateral transtibial amputations, we seek to investigate how different prosthetic configurations affect the stride kinematic and kinetic asymmetries of athletes with unilateral transtibial amputations, and whether these asymmetries are associated with the metabolic cost of running.

The purpose of our study was to determine the prosthetic model, stiffness, and height configuration that minimizes the metabolic cost of running for athletes with unilateral transtibial

Table 1. *Anthropometric measurements and standing metabolic rates of athletes with unilateral transtibial amputations*

Age, yr	33.4 ± 6.1
Height, m	1.77 ± 0.08
Body mass, kg	76.1 ± 14.1
UL leg length, m	0.95 ± 0.05
Rec Catapult AL length, m	1.01 ± 0.07*
Rec Flex-Run AL length, m	1.00 ± 0.07*
Rec 1E90 Sprinter AL length, m	0.98 ± 0.07
Standing metabolic rate, W/kg	1.3 ± 0.1

Data are averages ± SD; $n = 7$ men and 3 women. *Significant difference between recommended (Rec) affected leg (AL) and unaffected leg (UL) lengths ($P < 0.05$), following a Bonferroni corrected paired 2-tailed t-test.

amputations, thus optimizing distance running performance (28, 45). To explain the potential effects of prosthetic configuration on metabolic cost, we also determined the associations between prosthetic model, stiffness, and height on the elicited biomechanics (overall and asymmetric). We also investigated the relationships between running biomechanics (overall and asymmetric) and metabolic cost. We hypothesized that the metabolic cost of running for athletes with unilateral transtibial amputations would be minimized when they used an RSP less stiff than the respective manufacturer's recommended stiffness category and when they used an RSP set at the manufacturer/prosthetist recommended height. We hypothesized that leg stiffness would be invariant across different prosthetic stiffness, thus residual limb stiffness would be inversely correlated with prosthetic stiffness. We also hypothesized that the metabolic cost of running for athletes with unilateral transtibial amputations would be correlated with overall and asymmetric biomechanics. Last, we hypothesized that the prosthetic model, stiffness, and height that minimize metabolic cost would be associated with the biomechanical variables that optimize the metabolic cost of running. Due to their influence on the metabolic cost of running, we investigated the following biomechanical variables: peak and stance average vertical GRFs (49, 50, 71), peak horizontal GRFs (9, 26), ground contact time (49, 50), stride frequency (25, 44, 69), and leg stiffness (25, 32, 44, 69).

METHODS

Subjects. Ten athletes with a unilateral transtibial amputation (7 men and 3 women) participated (Table 1). Each participant had at least one year of experience running using a passive-elastic RSP and gave informed consent according to our protocol, which was approved by the Colorado Multiple Institutional Review Board and the United States Army Medical Research and Materiel Command Office of Research Protection, Human Research Protection Office.

Protocol. Initially, each participant completed an alignment and accommodation session, which entailed a certified prosthetist aligning each participant with three different prosthetic models (Freedom Innovations Catapult FX6, Irvine, CA; Össur Flex-Run, Reykjavik, Iceland; Ottobock 1E90 Sprinter, Duderstadt, Germany) at each manufacturer's recommended stiffness category and ± 1 stiffness category, and at each manufacturer's recommended prosthetic height and ± 2 cm. The Catapult and Flex-Run prostheses are "C" shaped and attach distally to the socket via a connective aluminum pylon (Fig. 2). The 1E90 Sprinter prosthesis is "J" shaped and mounts to the posterior wall of the socket. After the height for a J-shaped prosthesis was established, the device is typically bolted directly to the socket. For this study, we constructed a custom aluminum height adjustment bracket that was bolted to an athlete's socket, allowing us to secure the 1E90 Sprinter prosthesis to the socket, and alter prosthetic height between trials while preserving the RSP (Fig. 2).

During the accommodation session, participants ran using each prosthetic model on a treadmill at self-selected speeds until both the participant and prosthetist were satisfied with the recommended height and the alignment at each height. Generally, athletes accommodated to each prosthetic model at the recommended stiffness category and height. For the C-shaped RSPs, the athletes ran at all three heights (recommended and ± 2 cm) to determine proper alignment for each connective pylon. For the J-shaped RSP, the alignment of the custom bracket that connected the RSP to the socket remained unaltered across height conditions; thus, athletes did not typically accommodate to the nonrecommended heights before the experimental sessions.

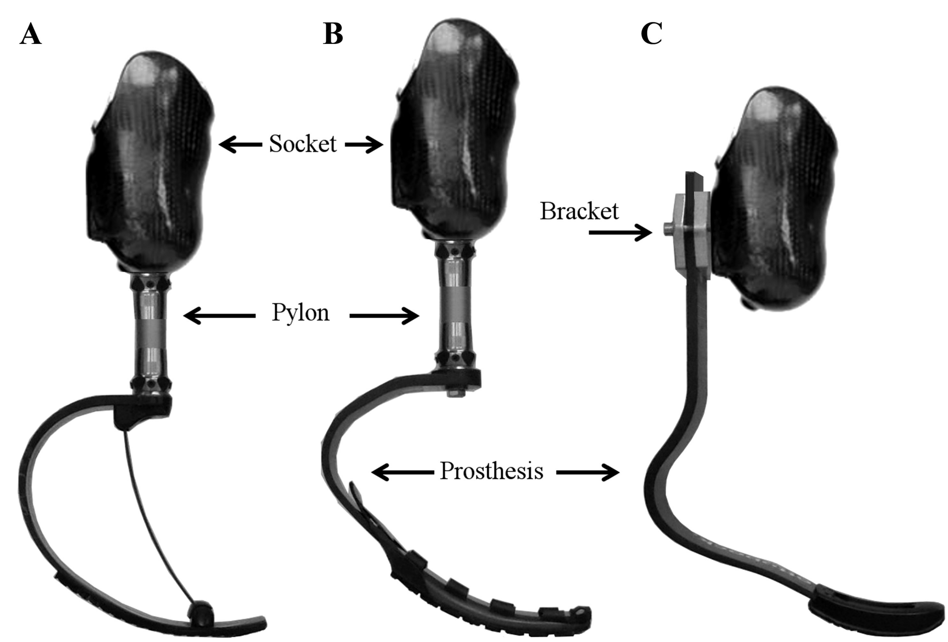

Fig. 2. *A*: Freedom Innovations Catapult FX6 prosthesis (C-shaped) at a representative recommended height. *B*: Össur Flex-Run prosthesis (C-shaped) at a representative height of +2 cm. *C*: Ottobock 1E90 Sprinter prosthesis (J-shaped) at a representative height of −2 cm. The C-shaped prostheses are connected beneath the socket via an aluminum pylon, and the J-shaped prosthesis is connected behind the socket via a custom aluminum bracket.

On subsequent days, participants performed a 5-min standing trial (using their personal walking prosthesis) and up to six 5-min running trials/session with at least 5 min of rest between trials. Participants ran on a 3D force-measuring treadmill (Treadmetrix, Park City, UT) at 3 m/s. If a participant was unable to maintain primarily oxidative metabolism at 3 m/s, indicated by a respiratory exchange ratio >1.0, running speed was reduced to 2.5 m/s. All running trials for a participant were performed at the same speed; therefore, if running speed was reduced to 2.5 m/s, all of the trials for the respective participant were (re)tested at 2.5 m/s.

Each participant ran using 15 different combinations of prosthetic model, stiffness category, and height. Initially, participants ran using each prosthetic model at three different stiffness categories (recommended and ±1) and the recommended affected leg length (9 trials, 3/prosthetic model). The stiffness category for each prosthetic model that elicited the lowest net metabolic cost of transport (CoT in $J \cdot kg^{-1} \cdot m^{-1}$) was deemed optimal. Subsequently, participants ran using the optimal stiffness category of each prosthetic model at two additional affected leg lengths (±2 cm) (6 additional trials). We randomized the trial order beginning with the nine prosthetic model and stiffness category combinations at the recommended affected leg length. Once a participant completed trials at all three stiffness categories for a prosthetic model at the recommended affected leg length, the height alteration trials for the respective prosthetic model at the optimal stiffness category were randomly inserted in the trial order. We tested each participant at the same time of day for all of their sessions to minimize any potential day-to-day variability.

Prosthetic stiffness. The recommended prosthetic stiffness (kN/m) for each prosthetic model varies (13). Accordingly, we evaluated the influence of each manufacturer's recommended prosthetic stiffness category, as well as the influence of actual prosthetic stiffness (kN/m) on the net CoT during running using recently published data from our laboratory (13). Concisely, we calculated prosthetic stiffness using the mean peak vertical GRF magnitude measured from the affected leg during each trial (present study) and estimated prosthetic displacement using the force-displacement equations from Beck et al. (13). Subsequently, we divided the measured peak GRF magnitude by the respective RSP displacements to yield prosthetic stiffness.

Biomechanics. Participants ran on a 3D force-measuring treadmill. We collected and analyzed the vertical and anterior-posterior components of the GRFs during the last two minutes of each trial. We collected GRFs at 1,000 Hz and filtered them using a fourth-order low-pass Butterworth filter with a 30-Hz cutoff. We used the filtered data to calculate peak and stance average vertical GRFs, peak horizontal (braking and propulsive) GRFs, in addition to ground contact time, step frequency, and leg stiffness values from 10 consecutive strides (10 affected leg steps and 10 unaffected leg steps) with a custom MATLAB script (Mathworks, Natick, MA). To detect periods of ground contact, we set the vertical GRF threshold to 1% of participant body weight.

Leg stiffness (k_{leg}) was computed as the quotient of peak vertical GRF (F_{peak}) and the maximum compression of the leg spring (ΔL) during ground contact (Fig. 1), as per Farley et al. (31).

$$k_{leg} = \frac{F_{peak}}{\Delta L} \quad (1)$$

To calculate the maximum compression of the leg spring (ΔL), we measured initial unaffected leg length (L_0) from the greater trochanter to the floor during standing, and affected leg length (L_0) as the distance from the greater trochanter to the distal end of the unloaded RSP (14, 36, 54). Next, we used initial leg lengths to calculate theta (θ), which is the angle of the leg spring at initial ground contact relative to vertical.

$$\theta = \sin^{-1}\left(\frac{v t_c}{2 L_0}\right) \quad (2)$$

Mathematically, θ equals half the angle swept by the stance leg, as determined from running velocity (v), ground contact time (t_c), and initial leg length (L_0). ΔL was calculated using *Eq. 3*,

$$\Delta L = \Delta y + L_0\left(1 - \cos\theta\right) \quad (3)$$

which incorporates the peak vertical displacement of the center of mass during ground contact (Δy), calculated by twice integrating the vertical acceleration of the center of mass with respect to time (23). The instantaneous vertical acceleration of the center of mass was calculated by subtracting the participant's body weight from the vertical GRF magnitude (net force) and dividing by body mass (23).

Because biological legs and RSPs act as relatively linear springs (13, 30, 31, 37), we modeled the affected leg stiffness as two in-series linear springs comprised of the RSP and residual limb (Fig. 1 and *Eq. 4*).

$$\frac{1}{k_{leg}} = \frac{1}{k_{RSP}} + \frac{1}{k_{res}} \quad (4)$$

Thus, we used the measured leg stiffness (k_{leg}) and the calculated prosthetic stiffness (k_{RSP}) to solve for the residual limb stiffness (k_{res}) during running.

To assess interlimb symmetry, we used the absolute value of the symmetry index (40, 65, 75) expressed as a percentage (*Eq. 5*). Taking the absolute value of the symmetry index is necessary to discern symmetry from asymmetry using linear statistical models. Perfect interlimb symmetry is equal to 0%.

$$\left|\frac{UL - AL}{0.5\left(UL + AL\right)}\right| \times 100 \quad (5)$$

Due to the potential association between RSP mechanical energy return and the metabolic cost of running, we calculated absolute mechanical energy return per affected leg step ($\dot{E}_{step}$),

$$\dot{E}_{step} = \frac{1}{2} k_{RSP} (\Delta d)^2 \left(1 - Hyst_{RSP}/100\right) \quad (6)$$

determined from k_{RSP}, peak prosthetic displacement (Δd), and percent prosthetic hysteresis ($Hyst_{RSP}$) (13). Next, we divided absolute mechanical energy return ($\dot{E}_{RSP}$) by user body mass (m) and stride length (L_{stride}) to calculate normalized $\dot{E}_{RSP}$ per stride ($J \cdot kg^{-1} \cdot m^{-1}$).

$$\dot{E}_{RSP} = \frac{\dot{E}_{step}}{m\left(L_{stride}\right)} \quad (7)$$

Metabolic cost of transport. We instructed participants to fast for at least 3 h before testing. We measured each participant's rate of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) using open-circuit expired gas analysis (TrueOne 2400; ParvoMedic, Sandy, UT) throughout each trial and averaged these rates during the last two minutes of each trial to calculate steady-state metabolic power (W) using a standard equation (19). Next, we subtracted the average metabolic power consumed during standing of the corresponding day from each running trial to yield net metabolic power. We normalized net metabolic power by the mass of each participant, which included running gear (e.g., RSP, socket, shoe, and clothes) for each respective trial. Finally, to combine data from 3.0 and 2.5 m/s, we divided net metabolic power by running velocity to calculate the net metabolic cost of transport (CoT) in Joules per kilogram per meter.

Statistical analyses. We used a linear mixed model to evaluate the effects of using different prosthetic models, stiffness categories, and heights on net CoT. We used a second linear mixed model to evaluate the effects of using different prosthetic models, actual prosthetic stiffness (kN/m), and heights on net CoT. We used linear regressions to assess the independent relationships between affected leg stiffness, prosthetic stiffness, and residual limb stiffness. We also tested whether affected and unaffected leg stiffness are correlated, and if net CoT is influenced by the absolute RSP mechanical energy return per step (*Eq. 6*) and/or per unit distance traveled (*Eq. 7*) with linear regressions.

Four participants ran at 3.0 m/s, and six ran at 2.5 m/s. For this reason we used additional linear mixed models to control for speed while independently testing the associations between overall (affected leg and unaffected leg averaged) and asymmetric GRFs (stance average vertical GRF, peak vertical GRF, and peak horizontal braking and propulsive GRFs) stride kinematics (ground contact time and step frequency), and leg stiffness on net CoT. We also performed a linear mixed model to evaluate the relationship between prosthetic model, stiffness, and height on the overall and asymmetric biomechanical variables that influence net CoT.

We used paired two-tailed *t*-tests to assess leg length discrepancies between affected and unaffected legs and implemented Bonferroni corrections when appropriate. For the linear mixed models and linear regressions, we report the fixed effect (β) from each statistically significant association (dependent variable = β independent variable + intercept). We tested all potential independent variable interactions with linear mixed models. If independent variables or interactions were nonsignificant, they were dropped from the model for the interpretation of the significant variables and interactions. We set the level of significance at $\alpha = 0.05$ and performed all statistical analyses using R-studio software (Boston, MA).

RESULTS

All prosthetic models were set at statistically similar recommended heights ($P \geq 0.053$) (Table 1). The recommended affected leg lengths were statistically longer than the corresponding unaffected leg lengths when using the Catapult (1.01 ± 0.07 m; $P < 0.001$) and Flex-Run (1.00 ± 0.07 m; $P = 0.001$) prostheses but not when using the 1E90 Sprinter prosthesis (0.98 ± 0.07 m; Bonferroni corrected $P = 0.080$). Furthermore, our highest stiffness category Flex-Run prosthesis was the manufacturer recommended stiffness category for two participants. Hence, these participants were tested at the stiffness categories of recommended, −1, and −2 with the Flex-Run prosthesis. Because of residual leg lengths and component heights, we were unable to perfectly match prosthetic heights at −2 cm for five participants. Therefore, the actual prosthetic heights for the shortest condition for five participants were −1.2, −1.3, −2.6, −0.5, and −1.2 cm with the C-shaped RSPs. We accounted for these disparities with our statistical analyses. Additionally, because of RSP component and logistical limitations, we were unable to complete four trials for three different participants; hence, our results include 146 trials (Table 2) rather than 150.

While controlling for covariates (i.e., controlling for two of the following while assessing the third: prosthetic model, stiffness, and height), the net CoT for athletes with unilateral transtibial amputations was independent of prosthetic stiffness category ($P = 0.180$), actual prosthetic stiffness ($P = 0.327$) (Fig. 3), and height ($P = 0.062$). In contrast, prosthetic model had a significant effect on net CoT. Use of a 1E90 Sprinter prosthesis resulted in 4.3 and 3.4% lower net CoT compared with use of the Catapult ($\beta = -0.177$; $P < 0.001$) and Flex-Run ($\beta = -0.139$; $P = 0.002$) prostheses, respectively. Net CoT was similar with use of the Catapult vs. Flex-Run prosthesis ($P = 0.393$) (Fig. 3). There were no prosthetic model, stiffness, or height interactions affecting net CoT ($P \geq 0.151$).

The affected leg stiffness of athletes with unilateral transtibial amputations was positively correlated with prosthetic stiffness ($P < 0.001$; $R^2 = 0.708$; affected leg stiffness = 0.558 prosthetic stiffness + 0.814) (Fig. 4) and residual limb stiffness ($P < 0.001$; $R^2 = 0.728$; affected leg stiffness = 0.196 residual limb stiffness + 6.777). Increased prosthetic stiffness was associated with increased residual limb stiffness ($P < 0.001$; $R^2 = 0.212$; residual limb stiffness = 1.333 prosthetic stiffness + 4.399) (Fig. 4). Unaffected

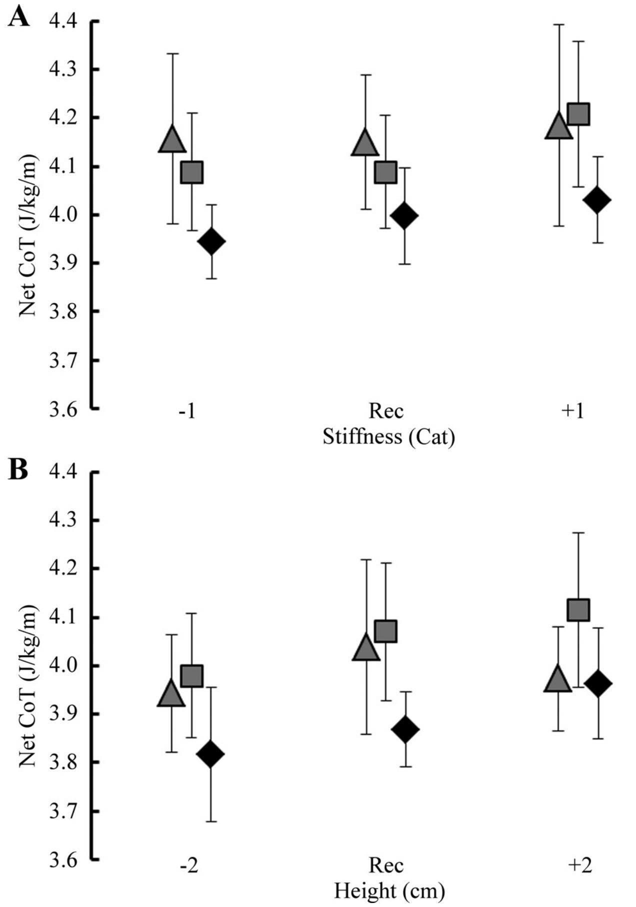

Fig. 3. *A*: average (±SE) net cost of transport (CoT) across prosthetic stiffness categories (Cat) for each prosthetic model. *B*: average (±SE) net CoT across recommended (Rec) and ± 2 cm prosthetic height alterations. Triangles, Catapult prostheses; squares, Flex-Run prostheses; diamonds, 1E90 Sprinter prostheses. Symbols are offset for visual representation. See Table 2 for sample size in visual depiction. We performed linear mixed models from all of our collected data to determine that there was no effect of stiffness category ($P = 0.180$) or height ($P = 0.062$) on net CoT, and that net CoT was reduced when participants used the 1E90 Sprinter prosthesis compared with using the Catapult ($P < 0.001$) or Flex-Run ($P = 0.002$) prosthesis.

Table 2. *Number of participants for each prosthetic model*

	Catapult			Flex-Run			1E90 Sprinter		
	−1 Cat	Rec Cat	+1 Cat	−1 Cat	Rec Cat	+1 Cat	−1 Cat	Rec Cat	+1 Cat
+2 cm	5	2	2	4	5	1	5	2	2
Rec Ht	10	10	10	10	10	8	10	10	10
−2 cm	5	2	2	4	4	1	6	2	2

Data are the no. of participants for each prosthetic model, at recommended and ±1 stiffness categories, and Rec and ±2 cm height (Ht) configurations.

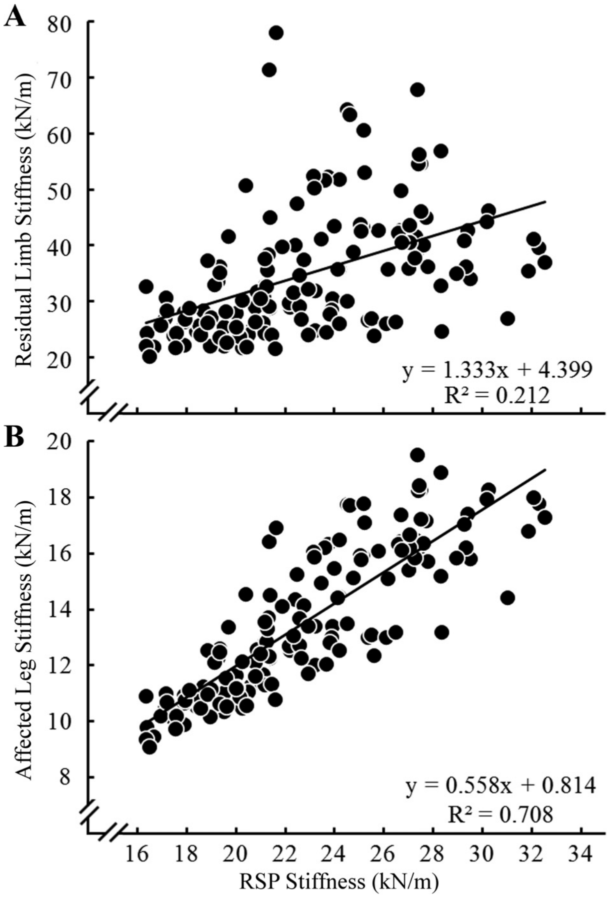

Fig. 4. *A*: residual limb stiffness compared with running-specific prosthetic (RSP) stiffness ($P < 0.001$). *B*: affected leg stiffness compared with prosthetic stiffness. We performed linear regressions across all collected data to determine significant correlations between residual limb stiffness and RSP stiffness ($P < 0.001$) (*A*) and between affected leg stiffness and RSP stiffness ($P < 0.001$) (*B*).

leg stiffness was positively correlated with affected leg stiffness ($P < 0.001$; $R^2 = 0.509$; unaffected leg stiffness = 0.693 affected leg stiffness + 5.270) and prosthetic stiffness ($P < 0.001$; $R^2 = 0.381$; unaffected leg stiffness = 0.398 prosthetic stiffness + 5.583).

The majority of overall (affected leg and unaffected leg average) biomechanical parameters affected net CoT. Namely, for every 0.1 times body weight reduction in peak ($\beta = 0.649$; $P = 0.001$) and stance average vertical ($\beta = 0.772$; $P = 0.018$) GRF, net CoT decreased 2.6%. For every 0.1-s increase in ground contact time, net CoT decreased 8.4% ($\beta = -0.435$; $P = 0.012$). For every 1 kN/m reduction in leg stiffness, net CoT decreased 2.3% ($\beta = 0.071$; $P < 0.001$). Net CoT was not affected by peak horizontal braking ($P = 0.502$) or propulsive ($P = 0.899$) GRFs, nor step frequency ($P = 0.773$). Additionally, neither the amount of RSP mechanical energy returned per step nor per unit distance traveled influenced net CoT ($P \geq 0.060$).

Of the investigated stride kinematic and kinetic asymmetries, net CoT was only related to peak vertical GRF asymmetry ($\beta = 0.007$; $P = 0.003$) (Fig. 5). Across all prosthetic configurations, for every 10.0% reduction in peak vertical GRF asymmetry, net CoT decreased 1.9%. For perspective, if the mean elicited peak vertical GRF asymmetry (15.7%) between the affected and unaffected legs became perfectly symmetric (0.0%), net CoT would decrease 3.0%. The elicited net CoT was independent of the following asymmetries: stance average vertical GRF ($P = 0.410$), peak braking ($P = 0.119$), and peak propulsive ($P = 0.917$) horizontal GRF, ground contact time ($P = 0.867$), step frequency ($P = 0.754$), and leg stiffness ($P = 0.636$). Within our protocol, running speed did not alter the influence of biomechanics on metabolic cost ($P \geq 0.170$).

Increased prosthetic stiffness (kN/m) resulted in greater stance average vertical GRFs ($\beta = 0.007$; $P < 0.001$), shorter ground contact times ($\beta = -0.002$; $P < 0.001$) (Table 3 and Fig. 6), and greater leg stiffness ($\beta = 0.194$; $P < 0.001$) (Table 3). Increased prosthetic height resulted in more asymmetric peak vertical GRFs ($\beta = 4.062$; $P < 0.001$) (Table 4 and Fig. 7). The 1E90 Sprinter prosthesis resulted in greater stance average vertical GRF compared with the Catapult ($\beta = 0.033$; $P = 0.001$) but not Flex-Run ($P = 0.137$) prosthesis, longer ground contact time ($\beta = 0.008$; $P < 0.001$) compared with the Flex-Run but not the Catapult ($P = 0.395$) prosthesis, and lower leg stiffness compared with both C-shaped RSPs ($\beta \geq -0.556$; $P < 0.001$). The 1E90 Sprinter prosthesis resulted in 8.3–8.7% (symmetry index percentage) more symmetric peak vertical GRFs compared with the use of the C-shaped RSPs ($P < 0.001$) (Figs. 6 and 7). Neither prosthetic model, stiffness, nor height affected overall peak vertical GRF magnitude ($P \geq 0.050$). Prosthetic stiffness was independent of peak vertical GRF asymmetry ($P = 0.108$) (Table 3 and Fig. 6), and prosthetic height was independent of stance average vertical GRF ($P = 0.959$), ground contact time ($P = 0.353$), and leg stiffness ($P = 0.348$).

DISCUSSION

Within the study's parameters, neither prosthetic stiffness nor height affected the net CoT during running for athletes with unilateral transtibial amputations; therefore, we reject our initial hypothesis. Unlike prosthetic stiffness and height, net CoT was affected by prosthetic model. The use of the J-shaped

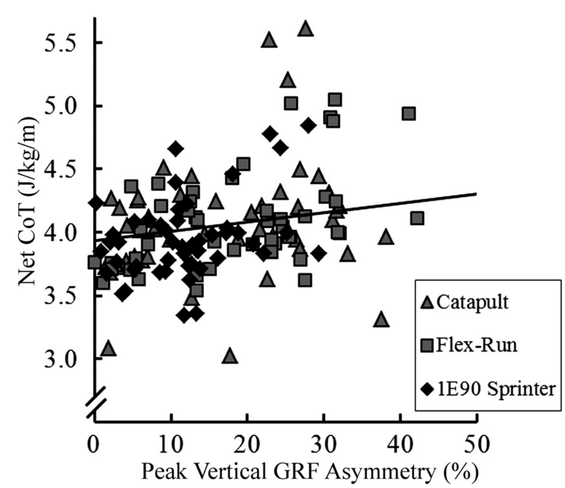

Fig. 5. Individual net CoT values plotted as a function of absolute peak vertical ground reaction force (GRF) asymmetry. Using all of our collected data, we performed a linear mixed model to determine that reducing peak vertical GRF asymmetry lowered net CoT (net CoT = 0.007 peak vertical GRF asymmetry + 3.933).

Table 3. *Biomechanical variables that influenced net CoT*

Biomechanics	Catapult			Flex-Run			1E90 Sprinter		
	−1 Cat	Rec Cat	+1 Cat	−1 Cat	Rec Cat	+1 Cat	−1 Cat	Rec Cat	+1 Cat
Peak vGRF, body wt	2.40	2.39	2.40	2.38	2.40	2.43	2.37	2.37	2.40
Avg vGRF, body wt*	1.32	1.34	1.35	1.38	1.39	1.41	1.34	1.35	1.39
t_c, s*	0.27	0.26	0.25	0.25	0.25	0.24	0.27	0.26	0.25
k_{leg}, kN/m*	14.0	14.4	15.0	14.3	14.5	14.3	13.2	13.6	14.5
Peak vGRF, SI	15.0	18.4	18.8	14.6	16.9	17.6	9.2	10.3	13.1

Biomechanical variables that influenced net cost of transport (CoT): overall peak vertical ground reaction force (vGRF), stance average vGRF (Avg vGRF), ground contact time (t_c), leg stiffness (k_{leg}), and peak vGRF asymmetry at each stiffness category (recommended and ±1 category) for each prosthetic model at the recommended height. SI, symmetry index (%). *Significant effect of prosthetic stiffness (kN/m) on the biomechanical variable across all of our data using linear mixed-model analyses.

1E90 Sprinter prosthesis lowered the metabolic cost of running compared with the use of the C-shaped Catapult and Flex-Run prostheses; the 1E90 Sprinter prosthesis was metabolically optimal for 9 out of 10 athletes. These results occurred despite the heavier custom bracket used for the 1E90 Sprinter prosthesis compared with the typical J-shaped RSP configuration. Rather than bolting the prosthesis directly to the socket, we used a relatively large bracket (~400 g) to connect the RSP to the socket (Fig. 2). As a result, the combined mass of the 1E90 Sprinter prosthesis and attachment was ~425 g greater than that of the Catapult and Flex-Run prostheses. Previous nonamputee running studies demonstrate that adding 100 g to each foot increases the metabolic cost of running by ~1% (24, 45, 52), indicating that our testing configuration for the 1E90 Sprinter prosthesis may have artificially increased the metabolic cost of running. Thus, the lower metabolic cost while using the J-shaped 1E90 Sprinter prosthesis vs. the use of C-shaped prostheses would have likely been further reduced through the use of a typical, lightweight, configuration.

The best prosthetic configuration (model, stiffness, and height combination) for each participant resulted in an 18.9% lower net CoT compared with the worst configuration (paired *t*-test; $P < 0.001$; 3.65 ± 0.37 vs. 4.50 ± 0.45 $J \cdot kg^{-1} \cdot m^{-1}$). Our results coincide with previous research demonstrating the sensitivity of the metabolic cost of running to prosthetic model for athletes with unilateral transtibial amputations (46, 72). In 1999, Hsu et al. (46) reported that athletes with unilateral transtibial amputations consumed oxygen at 8–11% greater rates while running at 2.01–2.45 m/s using a solid-ankle cushioned heel (SACH) prosthesis compared with using a passive-elastic Re-Flex Vertical Shock Pylon prosthesis. The SACH prosthesis uses a static rigid design, whereas the Re-Flex Vertical Shock Pylon prosthesis uses a vertical leaf spring and piston-cylinder pylon design (46). In 2009, Brown et al. (20) reported that athletes with transtibial amputations consumed 14% less oxygen while running at 2.23 m/s using RSPs (the athlete's personal RSP), similar to those used in the present study, compared with using relatively rigid passive-elastic walking prostheses that have an incorporated heel component. Remarkably, the most- and least-economical RSPs for each participant in the present study elicited a wider range of metabolic costs compared with the previous research that compared the use of RSPs with walking prostheses (20, 46). This may be because of inconsistent sagittal plane alignment, the use of different sockets, and/or the faster running speeds used in the present study compared with previous investigations (20, 46). Altogether, prosthetic model strongly influences the metabolic cost of running for athletes with unilateral transtibial amputations.

We reject our second hypothesis because residual limb stiffness was positively correlated with prosthetic stiffness (Fig. 4). This positive correlation accentuated the leg stiffness changes of our participants with altered in-series (prosthetic) stiffness, contrasting that of nonamputee runners (33, 34, 48). Consequently, running mechanics and center of mass dynamics of athletes with unilateral transtibial amputations may be affected by the in-series (RSP or surface) stiffness (Table 3 and Fig. 6). The residual limb stiffness of athletes with bilateral transtibial amputations is also positively correlated with prosthetic stiffness (14), indicating that the absence of biological lower legs may yield novel biomechanical adaptations to in-series stiffness changes.

Because of the effects of different biomechanical parameters on the metabolic cost of running, we accept our third hypothesis. Regarding overall biomechanics, the metabolic cost of running was reduced with lower peak and stance average vertical GRFs, longer ground contact times, and decreased leg stiffness (Tables 3 and 4). Thus, it is likely that the optimal

Table 4. *Biomechanical variables that influenced net CoT*

Biomechanics	Catapult			Flex-Run			1E90 Sprinter		
	−2 cm	Rec Ht	+2 cm	−2 cm	Rec Ht	+2 cm	−2 cm	Rec Ht	+2 cm
Peak vGRF, body wt	2.40	2.40	2.36	2.51	2.38	2.37	2.35	2.38	2.38
Avg vGRF, body wt	1.33	1.33	1.30	1.45	1.38	1.36	1.33	1.36	1.34
t_c, s	0.26	0.26	0.27	0.25	0.25	0.26	0.27	0.26	0.27
k_{leg}, kN/m	14.6	14.4	14.3	15.3	14.5	14.4	13.2	13.8	13.4
Peak vGRF, SI*	8.8	17.4	23.8	13.8	16.6	27.1	8.9	10.8	19.0

Biomechanical variables that influenced net CoT: vGRF, Avg vGRF, t_c, k_{leg}, and peak vGRF asymmetry at each prosthetic height (recommended and ±2 cm) for every prosthetic model across stiffness categories. *Significant effect of prosthetic height on biomechanical variable across all of our data using linear mixed-model analyses.

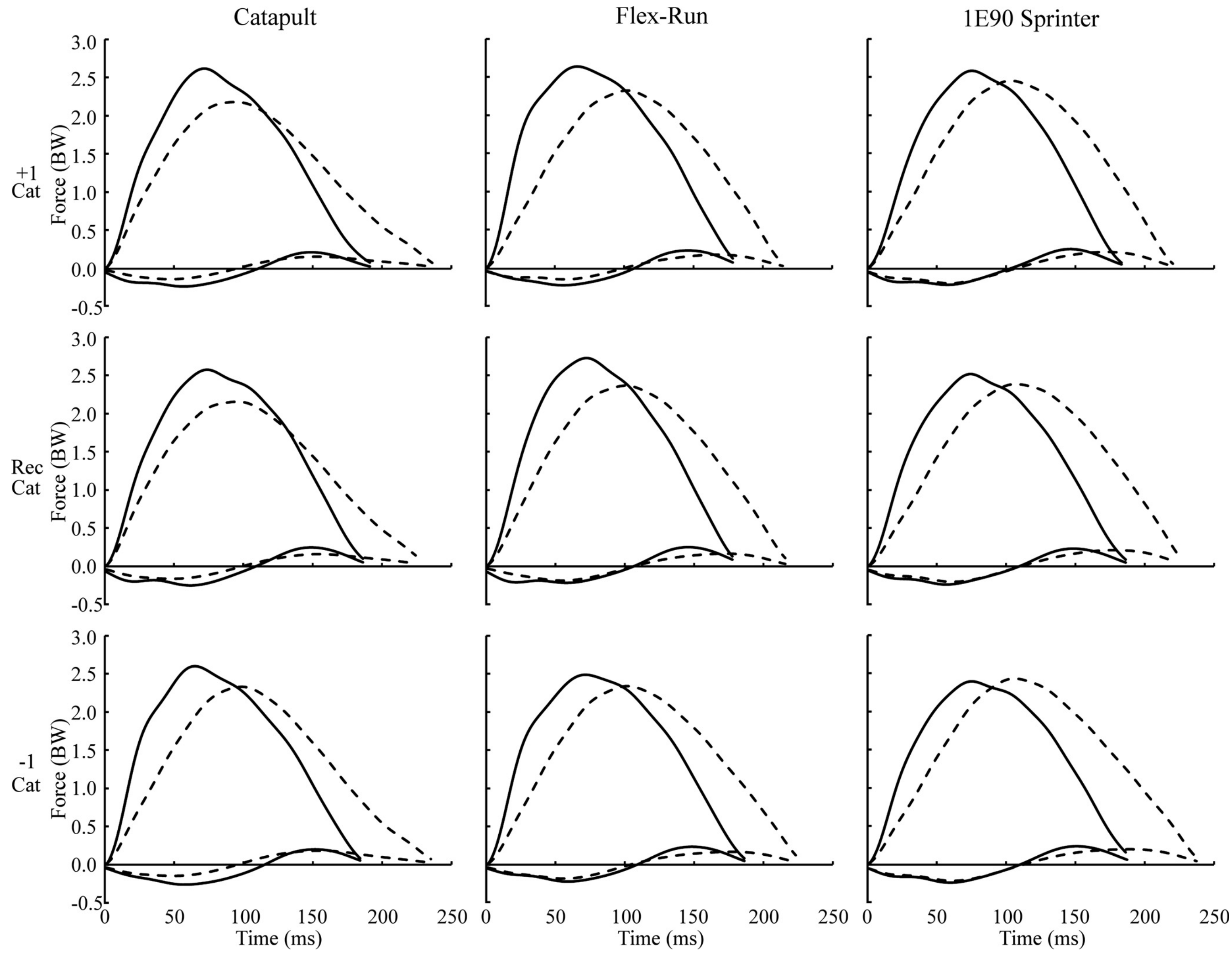

Fig. 6. Mean vertical and horizontal GRFs from 10 consecutive affected (broken line) and unaffected (solid line) leg steps from a representative participant running at 3 m/s. Columns *left* to *right* indicate the Freedom Innovations Catapult FX6, Össur Flex-Run, and Ottobock 1E90 Sprinter prostheses. Rows *top* to *bottom* indicate prosthetic stiffness category: −1, recommended (Rec), and +1.

combination of these biomechanical variables minimizes the metabolic cost of running for athletes with unilateral transtibial amputations. For instance, in our study, running with compliant leg springs resulted in prolonged ground contact time and decreased stance average vertical GRFs. Longer ground contact time extends the duration that athletes are able to generate force on the ground, enabling the recruitment of slower more economical muscle fibers (49, 62). Lower stance average vertical GRFs reduce the number of active ATP-consuming actin-myosin cross bridges needed to sustain running (49, 62). However, reduced leg stiffness relates with a decreased effective mechanical advantage of the leg joints. Thus, there is likely an optimal leg stiffness that elicits the ideal combination of the rate and magnitude of muscular force production. Furthermore, reduced peak vertical GRF asymmetries resulted in an improved metabolic cost of running for athletes with unilateral transtibial amputations. Within the range of observed asymmetries, peak vertical GRF asymmetry was the only such parameter that changed net CoT. Six of the seven observed asymmetries had no effect on the metabolic cost of running, including all of the measured stride kinematics. Therefore, current prosthetic prescriptions, which aim to mitigate stride kinematic asymmetries (35a, 59a, 59b), may not necessarily minimize the metabolic cost of running. Rather, prosthetic prescriptions focused on both legs' biomechanics may optimize the distance running performance of athletes with unilateral transtibial amputations.

Our last hypothesis was supported because the J-shaped prosthetic model that minimized the metabolic cost of running for athletes with unilateral transtibial amputations was associated with reduced leg stiffness and more symmetric peak vertical GRFs compared with the use of the C-shaped RSPs ($P < 0.001$). In addition, the use of the 1E90 Sprinter prosthesis may have led to enhanced sagittal plane alignment and/or improved lateral balance during running compared with the C-shaped RSPs. The sagittal plane alignment of the 1E90 Sprinter prosthesis may have yielded shorter GRF-leg joint moment arms, mitigating joint moments and the muscular force requirements during running (16, 17). Moreover, through a series of studies Arellano and Kram (6–9) demonstrated that there is a measureable metabolic cost associated with maintaining lateral balance during running. Hence, the wider (0–2.5

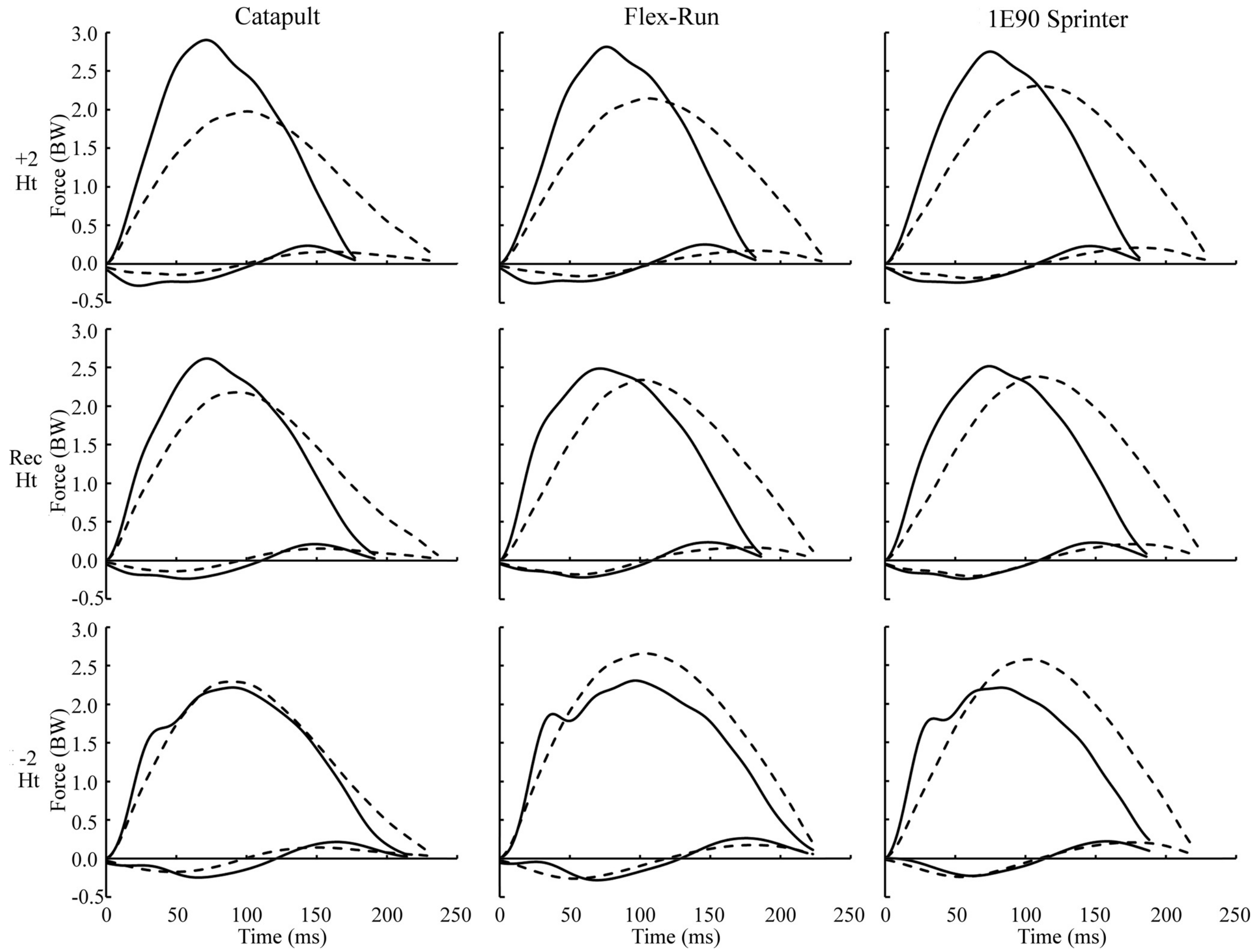

Fig. 7. Mean vertical and horizontal GRFs from 10 consecutive affected (broken line) and unaffected (solid line) leg steps from a representative participant running at 3 m/s. Columns *left* to *right* indicate the Freedom Innovations Catapult FX6, Össur Flex-Run, and Ottobock 1E90 Sprinter prostheses. Rows *top* to *bottom* indicate prosthetic height: +2 cm, recommended (Rec), and −2 cm.

cm) and thicker (0.1–0.9 cm) design of the 1E90 Sprinter prosthesis vs. the C-shaped RSPs at each segment (i.e., proximal, medial, distal) (35a, 59a, 59b) may have improved lateral balance, consequently reducing the metabolic cost of running.

It has been widely accepted that athletes with unilateral transtibial amputations generate lower peak and stance average vertical GRFs with their affected leg compared with their unaffected leg (11, 36, 41, 54, 61). Our study supports this notion; the unaffected leg of our participants averaged 15.4% greater peak vertical GRFs than those of the affected leg. Lower affected leg peak vertical GRFs have been attributed to residual limb discomfort (61), weakness (36), and the lack of net positive RSP mechanical power (36, 54). However, our data indicate that peak vertical GRF asymmetry occurred because of unequal leg lengths. The affected leg's peak vertical GRF production is inversely correlated with its relative length [linear regression; $P < 0.001$; $R^2 = 0.417$; peak vertical GRFs = −0.052 relative affected leg length (cm) + 2.449] (Fig. 7). Unaffected leg peak vertical GRFs were independent of affected leg length (linear regression; $P = 0.052$). Of our study's 18 trials (spanning 5 participants) where affected leg length was shorter or equal to unaffected leg length, the peak ($P = 0.421$) and stance average ($P = 0.686$) vertical GRFs were statistically similar between legs. Simply stated, reducing affected leg length, by decreasing prosthetic height, yields more symmetric peak and stance average vertical GRFs between the legs of athletes with unilateral transtibial amputations (Fig. 7).

Future studies are needed to optimize RSP configuration across multiple amputation levels (e.g., transfemoral, transtibial, etc.) and over a broad range of athletic endeavors (e.g., sprinting, cycling, and jumping). Socket design may also influence the metabolic cost of running. In the current study, our participants used two different sockets to complete the protocol (one for C-shaped RSPs and one for the J-shaped RSP). As a result, there may have been unequal residual limb movement within the different sockets, potentially leading to varying levels of muscle activation, which may have affected the metabolic cost of running (59). Additionally, the use of two separate testing speeds may have limited our study; however, we verified that running speed did not affect net CoT ($P = 0.454$) or any of the investigated biomechanical parameters ($P \geq 0.170$) using linear mixed models. We risk reporting type 1 errors resulting from our procedure of assessing each dependent variable with a separate statistical test. In addition, the effect of prosthetic height may have been confounded by our pseudorandomized trial order. Ideally, height alteration trials would have been inserted in the initial randomized trial order rather than after all the prosthetic stiffness category trials at the recommended height for each of the respective prosthetic models.

Conclusions. Prosthetic model, but not stiffness or height, affected the metabolic cost of running for athletes with unilateral transtibial amputations. The use of a J-shaped 1E90 Sprinter prosthesis elicited lower metabolic costs during running compared with the use of C-shaped prostheses. Furthermore, athletes with transtibial amputations appear to modulate biological leg stiffness with altered in-series stiffness differently than nonamputees. As such, changes to in-series prosthetic stiffness and surface stiffness likely alter the running mechanics of athletes with unilateral transtibial amputations. Despite the current prescriptions of running-specific prostheses, which aim to mitigate kinematic asymmetries between the affected and unaffected legs of athletes with unilateral transtibial amputations, the metabolic cost of running was independent of stride kinematic asymmetries, and only related to one kinetic asymmetry (peak vertical GRFs). Instead, the metabolic cost of running was reduced with decreased overall (affected and unaffected leg average) peak and stance average vertical GRFs, prolonged ground contact times, and reduced leg stiffness. Therefore, current prosthetic manufacturer recommendations do not necessarily reduce the metabolic cost of running (or optimize distance-running performance). Instead, recommendations based on prosthetic design and the affected and unaffected leg's average biomechanics, rather than asymmetries, likely optimize distance-running performance for athletes with unilateral transtibial amputations.

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GRANTS

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DISCLOSURES

The authors have no conflicts of interest to disclose. All of the running-specific prostheses used in our study were donated from the respective manufacturer.

AUTHOR CONTRIBUTIONS

A.M.G. conceived and designed the research; O.N.B., P.T., and A.M.G. performed experiments; O.N.B., P.T., and A.M.G. analyzed data; O.N.B., P.T., and A.M.G. interpreted results of experiments; O.N.B., P.T., and A.M.G. prepared figures; O.N.B., P.T., and A.M.G. drafted manuscript; O.N.B. and A.M.G. edited and revised manuscript; O.N.B., P.T., and A.M.G. approved final version of manuscript.

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