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RESEARCH ARTICLE

Reduced prosthetic stiffness lowers the metabolic cost of running for athletes with bilateral transtibial amputations

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Beck ON, Taboga P, Grabowski AM. Reduced prosthetic stiffness lowers the metabolic cost of running for athletes with bilateral transtibial amputations. *J Appl Physiol* 122: 976–984, 2017. First published January 19, 2017; doi:10.1152/japplphysiol.00587.2016.—Inspired by the springlike action of biological legs, running-specific prostheses are designed to enable athletes with lower-limb amputations to run. However, manufacturer's recommendations for prosthetic stiffness and height may not optimize running performance. Therefore, we investigated the effects of using different prosthetic configurations on the metabolic cost and biomechanics of running. Five athletes with bilateral transtibial amputations each performed 15 trials on a force-measuring treadmill at 2.5 or 3.0 m/s. Athletes ran using each of 3 different prosthetic models (Freedom Innovations Catapult FX6, Össur Flex-Run, and Ottobock 1E90 Sprinter) with 5 combinations of stiffness categories (manufacturer's recommended and ± 1) and heights (International Paralympic Committee's maximum competition height and ± 2 cm) while we measured metabolic rates and ground reaction forces. Overall, prosthetic stiffness [fixed effect (β) = 0.036; $P = 0.008$] but not height ($P \geq 0.089$) affected the net metabolic cost of transport; less stiff prostheses reduced metabolic cost. While controlling for prosthetic stiffness (in kilonewtons per meter), using the Flex-Run (β = −0.139; $P = 0.044$) and 1E90 Sprinter prostheses (β = −0.176; $P = 0.009$) reduced net metabolic costs by 4.3–4.9% compared with using the Catapult prostheses. The metabolic cost of running improved when athletes used prosthetic configurations that decreased peak horizontal braking ground reaction forces (β = 2.786; $P = 0.001$), stride frequencies (β = 0.911; $P <$ 0.001), and leg stiffness values (β = 0.053; $P = 0.009$). Remarkably, athletes did not maintain overall leg stiffness across prosthetic stiffness conditions. Rather, the in-series prosthetic stiffness governed overall leg stiffness. The metabolic cost of running in athletes with bilateral transtibial amputations is influenced by prosthetic model and stiffness but not height.

NEW & NOTEWORTHY We measured the metabolic rates and biomechanics of five athletes with bilateral transtibial amputations while running with different prosthetic configurations. The metabolic cost of running for these athletes is minimized by using an optimal prosthetic model and reducing prosthetic stiffness. The metabolic cost of running was independent of prosthetic height, suggesting that longer legs are not advantageous for distance running. Moreover, the in-series prosthetic stiffness governs the leg stiffness of athletes with bilateral leg amputations.

amputee; prosthesis; biomechanics; prescription; economy

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RUNNING IS A BOUNCING GAIT that is mechanically well-characterized by a spring-mass model, which depicts the stance leg as a massless linear spring and the body as a point mass (Fig. 1; Refs. 15, 25, 47). In the model, the leg spring compresses and stores elastic energy during the first half of the stance phase. Subsequently, the leg spring releases energy as it lengthens from midstance through the end of ground contact (5). During running, elastic elements such as tendons and ligaments act as springs that stretch and recoil (5, 12, 38, 55). Inspired by the springlike action of biological legs, passive-elastic carbon-fiber running-specific prostheses (RSPs) are designed to enable athletes with lower-limb amputations to run. RSPs are shaped like the uppercase letters "C" or "J," attach in-series to residual limbs (Fig. 2), and emulate the springlike function of biological legs during level-ground running (5, 12, 38, 55) by storing and returning elastic energy during ground contact (11, 18, 51). Since conserving mechanical energy via elastic mechanisms theoretically reduces the metabolic cost of running (5, 12, 38, 55), the elastic function of RSPs likely contributes to the 14% lower metabolic cost of running for athletes with transtibial amputations using RSPs compared with using relatively rigid, conventional walking prostheses (17).

Despite reducing the metabolic cost of running (17) and improving athletic performances compared with the use of previous prosthetic designs (34), current manufacturer's recommendations for prosthetic stiffness may not optimize the running performance of athletes with bilateral transtibial amputations. For athletes with unilateral amputations, the aim of the current manufacturer's recommended prosthetic configurations is to mitigate stride kinematic asymmetries between the affected and unaffected legs (e.g., asymmetric ground contact times; Ref. 51a). For athletes with bilateral amputations, prosthetists simply match the left- and right-leg RSPs at the manufacturer's recommended stiffness category, which is based on the same prosthetic-stiffness-to-body-mass ratio as athletes with unilateral amputations (30a, 51a, 51b).

Surface stiffness, which is in-series with the stance leg, affects the running performance of nonamputees (39, 47). For example, Kerdok et al. (39) reported that changing surface stiffness from 945 to 75 kN/m decreased the metabolic cost of running in nonamputees by 12%. This decreased metabolic cost was primarily attributed to the greater mechanical energy return from the compliant surface to the runner. Furthermore, when surface compliance changes, nonamputees maintain a constant overall surface plus leg stiffness by altering leg joint stiffness and/or segment geometries during running (29, 30, 39). Straighter limb posture generally results in lower joint

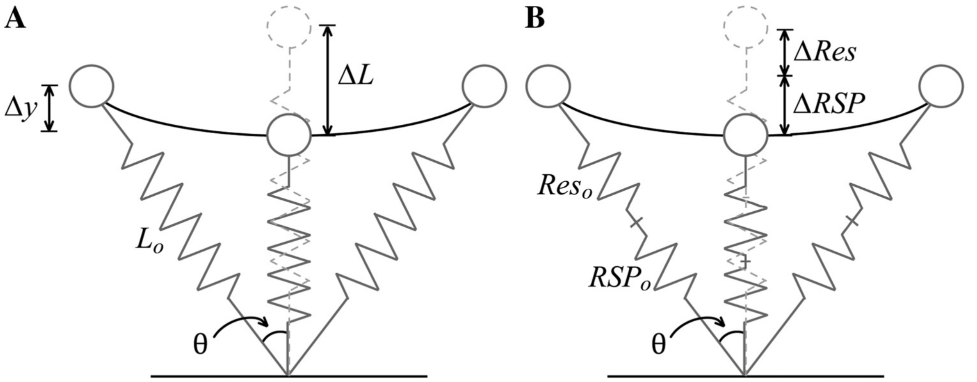

Fig. 1. Illustration of a spring-mass model of running (*A*) and a spring-mass model of running with an in-series leg spring (*B*). Body mass is represented as a point mass (circle), and the touch-down angle is indicated by θ. The stance leg is represented by a massless linear spring for nonamputees (*A*) or 2 in-series massless linear springs for athletes with bilateral amputations (*B*). The initial leg length (L_0) shortens (ΔL) as does its vertical height (Δy) during the stance phase of running. Modeled residual limb length (Res_0) and prosthetic height (RSP_0) compress and extend (ΔRes and ΔRSP) during the stance phase of running.

moments and in turn reduces the muscular force needed to support body weight (13, 14), which is the primary determinant of the metabolic cost of running (9, 40, 41, 43, 53). These previous studies suggest that decreasing prosthetic stiffness will reduce the metabolic cost of running. However, the effects of prosthetic stiffness on overall leg stiffness and metabolic cost during running have yet to be determined.

Analogous to prosthetic stiffness recommendations, current prosthetic height recommendations may not optimize distance running performance. Prosthetic height is set at the discretion of the athlete and/or their prosthetist and/or in accordance with the International Paralympic Committee (IPC) guidelines (37a). Anecdotally, the potential effects of increased prosthetic height were brought to light at the 2012 Paralympic Games when it appeared that athletes with bilateral transtibial amputations improved their sprinting performance by using taller RSPs. Hypothetically, longer legs could improve running speed by increasing the forward distance traveled during ground contact while accounting for step frequency and the stance average vertical ground reaction force (GRF; Ref. 58). Previous research indicates that the metabolic cost of running is poorly associated with the leg lengths of nonamputees (60); however, simple correlations fail to account for potential covariates such as increased lower limb mass with longer legs. No study has systemically altered prosthetic height for athletes with bilateral leg amputations and assessed its influence on distance running performance.

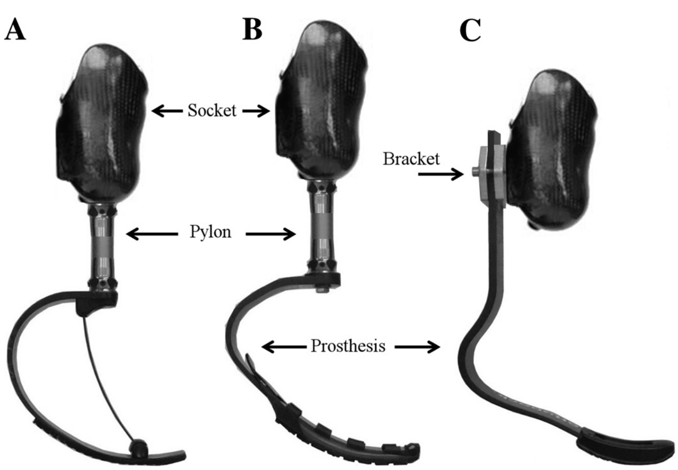

Fig. 2. From *left* to *right*: *A*, the Freedom Innovations Catapult FX6 prosthesis (C-shaped) at a representative recommended height; *B*, the Össur Flex-Run prosthesis (C-shaped) at a representative height of +2 cm; and *C*, the Ottobock 1E90 Sprinter prosthesis (J-shaped) at a representative height of −2 cm. The C-shaped prostheses are connected to sockets via aluminum pylons, and the J-shaped prostheses are connected to sockets via custom aluminum brackets.

We sought to determine how the use of RSPs with different stiffness values and heights affect the metabolic cost of running for athletes with bilateral transtibial amputations. Since reduced prosthetic stiffness may enhance mechanical energy conservation and improve the effective mechanical advantage of the stance leg, we hypothesized that using RSPs with a lower stiffness than manufacturer recommended would decrease the metabolic cost of running. Given the lack of previous data, we tested the null hypothesis that altering prosthetic height would not affect the metabolic cost of running. Based on several studies (29, 30, 39), we hypothesized that residual limb stiffness (comprising knee and hip joints) would be inversely associated with prosthetic stiffness such that athletes would maintain overall leg stiffness across different prosthetic stiffness configurations.

Finally, the metabolic cost of running is often associated with biomechanical variables such as vertical (41, 43, 56) and horizontal GRF magnitude (9, 24), ground contact time (41, 43), stride frequency (23, 35), and leg stiffness (23, 26, 35). For those reasons, we sought to quantify how the metabolic cost of running relates to these biomechanical variables in athletes with bilateral transtibial amputations.

METHODS

Subjects. Five male athletes with bilateral transtibial amputations participated (Table 1). Each athlete had over one year of experience using RSPs, which included track and field races. The protocol was approved by the Colorado Multiple Institutional Review Board and the U.S. Army Medical Research and Materiel Command (USAMRMC) Office of Research Protections, Human Research Protection Office, and before participation each athlete gave informed written consent in accordance with our protocol.

Protocol. Initially, each participant completed a fitting and accommodation session. During this session, we collected anthropometric measurements to determine the tallest height that each participant could use to compete in track and field races according to the IPC guidelines (37a). Next, a certified prosthetist fit each participant with three different prosthetic models (Freedom Innovations Catapult FX6, Irvine, CA; Össur Flex-Run, Reykjavik, Iceland; and Ottobock 1E90 Sprinter, Duderstadt, Germany) at the manufacturer's recommended stiffness category and ± 1 stiffness categories and at leg lengths that produced the IPC maximum competition height and ± 2 cm. Prosthetic stiffness categories are recommended to athletes based on user

Table 1. *Participant characteristics: age, mass, average standing metabolic power, cause of amputations, primary event(s), standing height, and leg length*

Participants	Age, yr	Mass, kg	Standing Metabolic Power, W/kg	Cause of Amputations	Primary Event	Max IPC Height, m	Max IPC Leg Length, m	Catapult Leg Length, m	Flex-Run Leg Length, m	1E90 Sprinter Leg Length, m
1	25	69.8	1.6	Congenital	100/200 m	1.80	0.97	1.12	1.12	0.97
2	23	76.1	1.5	Congenital	Long jump	1.88	1.07	1.07	1.07	1.04
3	18	73.2	1.7	Congenital	100/200 m	1.87	1.05	1.05	1.05	1.05
4	31	69.6	1.3	Traumatic	400 m	1.90	1.10	1.10	1.10	1.10
5	27	68.9	1.5	Infection	5,000 m	1.87	1.06	1.06	1.06	1.06
Average	24.8	71.5	1.5			1.86	1.05	1.08	1.08	1.04
SD	4.8	3.0	0.2			0.04	0.05	0.03	0.03	0.04

The maximum standing height and corresponding leg lengths allowed in track and field races sanctioned by the International Paralympic Committee (IPC; Ref. 37a). The resulting Catapult, Flex-Run, and 1E90 Sprinter prosthesis leg lengths represent the closest attainable maximum IPC-regulated leg lengths from each participant and prosthetic model combination (37a). Leg lengths were measured from the greater trochanters to the most distal locations of the unloaded prostheses.

body mass with larger athletes recommended numerically greater stiffness categories (30a, 51a, 51b). The Catapult and Flex-Run prostheses are shaped like a "C" and attach distally to the sockets that encompass the residual limbs, via connective aluminum pylons (Fig. 2). The 1E90 Sprinter prostheses are shaped like a "J" and mount to the posterior wall of each socket (Fig. 2). After establishing the heights for J-shaped RSPs, they are typically bolted directly to the sockets. Instead, we constructed custom aluminum brackets that were bolted to the sockets, thus allowing us to preserve the RSPs, secure them to the sockets, and alter height between trials (Fig. 2). Sockets are carbon-fiber or fiber-glass (check sockets) negative composites of a residual limb and are secured to the limb via suction or locking mechanisms.

Because of the combined lengths of the participant's residual legs and the heights of prosthetic components, we were unable to match the maximum IPC competition height for some participants with certain prosthetic models. The build height of C-shaped RSPs limit the minimum participant height (Fig. 2). For example, the minimum height of the Flex-Run prosthesis is 277 mm before adding the components necessary for socket attachment (51a). Thus, if the maximum IPC competition height for a participant is less than the length from the top of their head to the end of their residual limb plus 277 mm, they will exceed the maximum IPC height with the respective prosthetic model under all conditions. Also, the maximum achievable height was limited while using J-shaped RSPs. The 1E90 Sprinter prostheses could not exceed their build height; consequently, a participant with short residual limbs was unable reach the maximum IPC height using the 1E90 Sprinter prostheses (Table 1), whereas the C-shaped RSPs could be made as tall as necessary through the use of connective pylons. For these cases, we set prosthetic height as close as feasible to the maximum IPC competition height. If the closest achievable height was taller than the maximum IPC competition height, ensuing prosthetic height alterations were +2 and +4 cm. If the closest achievable height was shorter than the maximum IPC competition height, ensuing prosthetic height alterations were −2 and −4 cm (Table 1).

After being fit with different prosthetic configurations, participants ran on a treadmill at self-selected speeds until both the prosthetist and participant were satisfied. Generally, athletes were accommodated to each prosthetic model at the recommended stiffness category and height. When using C-shaped RSPs, athletes also ran at additional heights (i.e., ± 2 cm) to determine proper alignment with taller/shorter pylons. When using J-shaped RSPs, the components and alignment were the same for each height; thus athletes were not typically accommodated to additional heights. The accommodation sessions lasted approximately 6–7 h per participant. All participants used their personal competition sockets for the trials with the respective prosthetic shape (4 used J-shaped and 1 used C-shaped RSPs). The 4 athletes who competed with J-shaped RSPs used their everyday walking sockets for the C-shaped RSP trials. For the athlete who competed with C-shaped RSPs, a prosthetist fabricated custom check sockets that replicated the participant's competition sockets (suspension, internal dimensions, et cetera) for the J-shaped RSP trials.

On subsequent days, participants performed a 5-min standing trial (using their personal walking prostheses) and up to 6 5-min running trials per session with at least 5 min of rest between trials. The combination of the rest periods and the moderate-intensity running trials adequately prevented any potential effects of fatigue. For example, previous studies reported that subjects who run at a moderate intensity for trial lengths up to 7 min display no signs of fatigue (31, 32).

Participants ran on a 3-dimensional 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, as indicated by a respiratory exchange ratio >1.0, running speed was set to 2.5 m/s for all of their respective trials. Each participant ran using 15 different prosthetic model, stiffness category, and height combinations. Initially, participants ran using each prosthetic model at 3 stiffness categories (recommended and ± 1) and the maximum competition height. The stiffness category for each prosthetic model that elicited the lowest net metabolic cost of transport (CoT in joules per kilogram per meter) was deemed optimal. Subsequently, participants ran using the optimal stiffness category of each prosthetic model at 2 additional heights (e.g., ± 2 cm). We randomized the trial order beginning with the 9 prosthetic model and stiffness category combinations at the maximum IPC height. Once a participant completed trials in all 3 stiffness categories with a prosthetic model, the altered height trials for the respective model at the optimal stiffness category were randomly inserted into the trial order. Data were collected over 3–5 sessions, and all participants completed the protocol within 9 days following the accommodation session.

Metabolic cost of transport. We instructed participants to fast for at least 3 h before testing. We measured their rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) using open-circuit expired gas analysis (TrueOne 2400; Parvo Medics, Sandy, UT) throughout each trial and averaged these rates during the last 2 min of each trial to calculate steady-state metabolic rates. We used the Brockway equation (16) to convert the average $\dot{V}O_2$ and $\dot{V}CO_2$ into metabolic power. Then, 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 the participant for each prosthetic condition. Participant mass included running gear. Finally, to compare 3.0 and 2.5 m/s trials, we divided net metabolic power by running velocity to calculate the net metabolic cost of transport (CoT) in joules per kilogram per meter. We tested each participant at the same time of day for all of their respective sessions.

Prosthetic stiffness. Recommended prosthetic stiffness (in kilonewtons per meter) differs between models (11). Therefore, we assessed the influence of the manufacturer's recommended prosthetic stiffness category as well as actual prosthetic stiffness (in kilonewtons per meter) on the net CoT during running (11) using established data. We calculated prosthetic stiffness from the mean peak vertical GRF measured from both legs during each trial (present study) and the force-displacement equations from Ref. 11 to estimate prosthetic displacement. Subsequently, we divided the measured peak vertical GRF magnitude by the estimated prosthetic displacement to yield stiffness.

Biomechanics. We measured vertical and anterior-posterior components of the ground reaction forces (GRFs) between *minutes 2.0* and *3.0* and *minutes 3.5* and *5.0* of each trial. We collected GRFs at 1,000 Hz, filtered them using a 4th-order low-pass Butterworth filter with a 30-Hz cutoff frequency, and then used filtered data to calculate GRF parameters, stride kinematics, and leg stiffness values from 10 consecutive strides (20 steps) with a custom MATLAB script (The MathWorks, Natick, MA). We set our GRF threshold at 1% of user body weight to detect periods of ground contact.

We calculated overall leg stiffness (k_{leg}) as the quotient of peak vertical GRF (F_{peak}) and maximum leg spring compression (ΔL) during ground contact (Fig. 1; Ref. 25):

$$k_{leg}=\frac{F_{peak}}{\Delta L}. \tag{1}$$

To calculate the maximum compression of the leg spring (ΔL), we measured initial leg length (L_0) as the distance from the greater trochanter to the distal end of the unloaded RSP (33, 46). Next, we used initial leg lengths to calculate θ, which is the angle of the leg spring at initial ground contact relative to vertical (Fig. 1), using *Eq. 2*.

$$\theta = \sin^{-1}\left(\frac{vt_c}{2L_0}\right). \tag{2}$$

Because the spring-mass model assumes step symmetry about the vertical axis (15, 25, 47), θ equals half of the angle swept by the stance leg, as determined from running velocity (v), ground contact time (t_c), and initial leg length (L_0). The maximum stance leg spring compression (ΔL) was calculated using *Eq. 3*:

$$\Delta L = \Delta y + L_0\left(1 - \cos\theta\right), \tag{3}$$

which incorporates 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 (19). 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 (19).

Since biological legs and RSPs have relatively linear force-displacement profiles (11, 25), we modeled overall leg stiffness (k_{leg}) as two in-series springs (Fig. 1). We used previously established measurements of prosthetic stiffness (k_{RSP}; Ref. 11) to estimate residual limb stiffness (k_{res}) using *Eq. 4*.

$$\frac{1}{k_{leg}}=\frac{1}{k_{res}}+\frac{1}{k_{RSP}}. \tag{4}$$

Because of the potential association between the mechanical energy delivered by the RSPs and the metabolic cost of running, we calculated mechanical power return from the RSPs for each step ($\dot{P}_{RSP}$):

$$\dot{P}_{RSP} = \frac{k_{RSP}(\Delta d)^2\left(1 - Hst_{RSP}/100\right)}{2t_{step}}, \tag{5}$$

determined by prosthetic stiffness (k_{RSP}), peak prosthetic displacement (Δd), percentage prosthetic hysteresis (Hst_{RSP}; Ref. 11), and step time (t_{step}). To relate prosthetic mechanical energy return to metabolic cost of transport (in joules per kilogram per meter), we divided the energy return averaged per step by user body mass (m) and running velocity (v) to calculate mechanical energy return ($\dot{E}_{RSP}$) in joules per kilogram per meter:

$$\dot{E}_{RSP} = \frac{\dot{P}_{RSP}}{mv}. \tag{6}$$

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 with actual prosthetic stiffness (in kilonewtons per meter) instead of stiffness category to evaluate the effects of using different prosthetic models, stiffness, and heights on net CoT.

Three of our participants ran at 3.0 m/s, and two ran at 2.5 m/s. Accordingly, we used linear mixed models to control for speed while independently testing the associations of the predetermined GRF parameters (stance average vertical GRF, peak vertical GRF, and peak horizontal braking and propulsive GRFs), stride kinematics (ground contact time and stride frequency), and leg stiffness on the net CoT. To evaluate the influence of prosthetic mechanical energy return on net CoT, in addition to the relationships between leg stiffness, prosthetic stiffness, and residual limb stiffness, we performed simple linear regressions. We performed paired two-tailed *t*-tests to compare each biomechanical variable from *minutes 2.0* to *3.0* to the respective variable from *minutes 3.5* to *5.0* to ensure participants achieved a biomechanical steady-state. We reported the fixed effect (β) from each statistically significant association (dependent variable = β independent variable + intercept). When appropriate, we implemented a Bonferroni correction and tested for potential interaction effects across all statistical comparisons. We set the level of significance at $\alpha = 0.05$ and performed statistical analyses using RStudio software (Boston, MA).

RESULTS

While controlling for covariates, use of different prosthetic stiffness (category and in kilonewtons per meter; $P \leq 0.008$; Fig. 3), but not height ($P \geq 0.089$; Fig. 4), affected the net CoT of athletes with bilateral transtibial amputations. Each integer reduction in stiffness category decreased the average net CoT by 3.7% ($\beta = 0.135$; $P < 0.001$). Actual prosthetic stiffness values ranged from 19.3 to 29.6 kN/m and averaged 22.9 ± 2.3 kN/m (±SD). Overall, every 1 kN/m reduction in prosthetic stiffness decreased net CoT by 1.3% ($\beta = 0.036$; $P = 0.008$; Fig. 3).

The metabolic cost of running was associated with the equipped prosthetic model. The influence of prosthetic model on net CoT was largely the same when controlling for either prosthetic stiffness category or actual stiffness (in kilonewtons per meter), thus unless otherwise specified, we will interpret prosthetic model effects while controlling for actual prosthetic stiffness (in kilonewtons per meter). Within our prosthetic stiffness range, when athletes with bilateral transtibial amputations used the 1E90 Sprinter prostheses, their net CoT was 4.3–4.7% lower compared with using Catapult prostheses ($\beta = -0.176$; $P = 0.009$). The net CoT was similar when athletes used the Flex-Run vs. 1E90 Sprinter prostheses ($P = 0.597$). When controlling for stiffness category, the use of Flex-Run prostheses elicited similar net CoT values compared with the use of Catapult prostheses ($P = 0.138$), whereas while controlling for actual prosthetic stiffness (in kilonewtons per

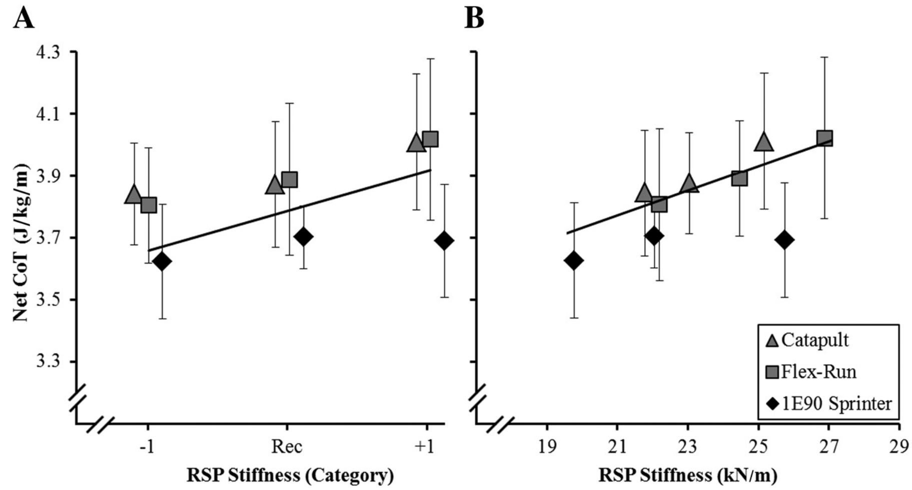

Fig. 3. *A*: mean (±SE) net metabolic cost of transport (CoT) as a function of using different models of running-specific prostheses (RSPs) with different stiffness categories (Cat). Symbols are offset for clarity. Regression equation: net CoT = 0.129 ΔCat + 3.786. Rec, recommended stiffness. *B*: mean (±SE) net metabolic cost of transport (CoT) as a function of actual prosthetic stiffness (in kilonewtons per meter) across prosthetic models. Regression equation: net CoT = 0.036 ΔkN/m + 2.931. Triangles represent use of the C-shaped Catapult, squares represent use of the C-shaped Flex-Run, and diamonds represent use of the J-shaped 1E90 Sprinter prostheses.

meter), the use of Flex-Run prostheses reduced net CoT 4.4–4.9% compared with the use of Catapult prostheses ($\beta = -0.139$; $P = 0.044$), highlighting the dissimilarity in manufacturer-recommended stiffness values (Fig. 3). There were no significant interaction effects between prosthetic model, stiffness, and/or height on net CoT ($P \geq 0.230$). Additionally, there was an extremely weak but significant correlation between the RSP mechanical energy return and the elicited net CoT ($P = 0.042$; $r^2 = 0.055$; net CoT = −0.660 RSP mechanical energy return + 4.360; Fig. 5).

There were no differences between any tested biomechanical parameters from *minutes 2.0* to *3.0* compared with *minutes 3.5* to *5.0* ($P \geq 0.430$). Consequently, we only report biomechanical data collected between *minutes 3.5* to *5.0* of each trial. Residual limb stiffness values ranged from 18.7 to 82.8 kN/m and averaged 42.5 ± 15.1 kN/m (±SD; Fig. 6). There was a moderate positive association between prosthetic stiffness (in kilonewtons per meter) and leg stiffness ($P < 0.001$; $r^2 = 0.437$; leg stiffness = 0.703 prosthetic stiffness − 1.623; Fig. 6) and a strong positive association between residual limb stiffness and leg stiffness ($P < 0.001$; $r^2 = 0.825$; leg stiffness = 0.149 residual limb stiffness + 8.159). There was a weak yet statistically significant, positive association between prosthetic stiffness (in kilonewtons per meter) and residual limb stiffness ($P = 0.003$; $r^2 = 0.115$; residual limb stiffness = 2.186 prosthetic stiffness − 7.704; Fig. 6).

Net CoT was associated with peak braking horizontal GRF, stride frequency, and leg stiffness. Independently, every 0.1× body weight decrease in peak braking horizontal GRF was related to a 6.4% reduced net CoT (net CoT = 2.789 peak braking GRF + 4.354; $P = 0.001$), every 0.1-Hz decrease in stride frequency was related to an 8.3% reduced net CoT (net CoT = 0.911 stride frequency + 1.099; $P < 0.001$), and each 1 kN/m decrease in leg stiffness was associated with a 1.8% reduced net CoT (net CoT = 0.053 leg stiffness + 2.991; $P = 0.009$). Stance average vertical GRF ($P = 0.592$), peak vertical GRF ($P = 0.723$), peak propulsive horizontal GRF ($P = 0.063$), and ground contact time ($P = 0.116$) were not associated with net CoT.

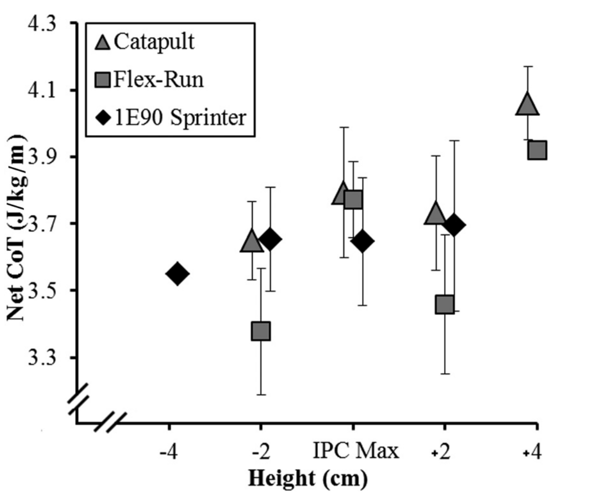

Fig. 4. Mean (±SE) net metabolic cost of transport (CoT) as a function of using different models of running-specific prostheses (RSPs) at different heights (in centimeters) using the stiffness category that produced the lowest net CoT. Symbols are offset for clarity. IPC Max indicates the prosthetic height for each participant that elicits the maximum competition height based on the International Paralympic Committee guidelines (37a), and deviations indicate heights of ± 2 and ± 4 cm. Triangles represent use of the C-shaped Catapult, squares represent use of the C-shaped Flex-Run, and diamonds represent use of the J-shaped 1E90 Sprinter prostheses.

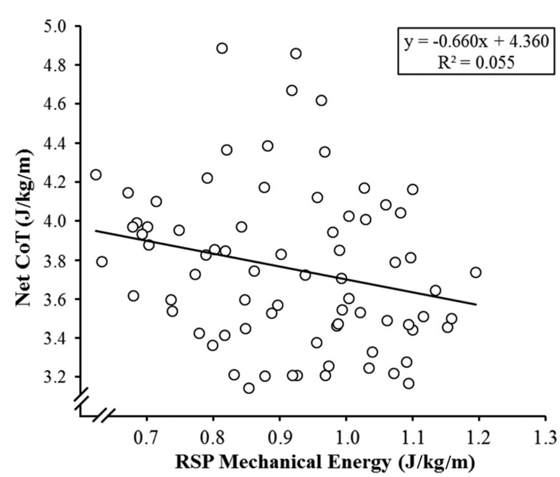

Fig. 5. The net metabolic cost of transport (CoT) as a function of running-specific prosthesis (RSP) mechanical energy return for each running trial. Increased prosthetic mechanical energy return lowered net CoT ($P = 0.042$).

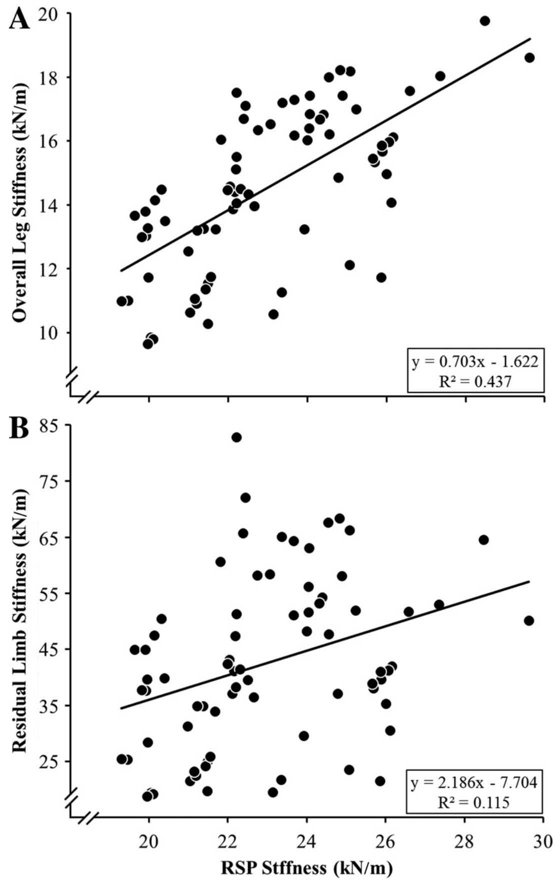

Fig. 6. Overall leg stiffness compared with running-specific prosthesis (RSP) stiffness ($P < 0.001$; *A*) and residual limb stiffness compared with prosthetic stiffness ($P = 0.003$; *B*). There was a positive association between both overall leg stiffness and residual limb stiffness compared with prosthetic stiffness.

DISCUSSION

We accept our initial hypothesis based on our findings that athletes with bilateral transtibial amputations consume less metabolic energy while running with RSPs that are less stiff than manufacturer recommended. Since prosthetic stiffness category recommendations are based on user body mass, we ran a linear mixed model with prosthetic stiffness (in kilonewtons per meter) normalized to each corresponding participant's body mass. Every 0.1 $kN \cdot m^{-1} \cdot kg^{-1}$ decrease in prosthetic stiffness (while controlling for prosthetic model) reduced net CoT by 9.2% ($\beta = 2.499$; $P = 0.012$), further supporting the notion that the use of less stiff RSPs reduces the metabolic cost of running for athletes with bilateral transtibial amputations. The decreased metabolic cost while using less stiff RSPs is likely related to improved biomechanics. Overall, the use of less stiff RSPs lowered peak braking GRF, stride frequency, and leg stiffness ($P \leq 0.022$). Moreover, while considering prosthetic models, further linear mixed-model analyses revealed that for every 1 kN/m prosthetic stiffness reduction, net CoT decreased while using the Catapult ($\beta = 0.085$; $P < 0.001$) and Flex-Run ($\beta = 0.084$; $P < 0.001$) but not the 1E90 Sprinter ($P = 0.258$) prostheses (Fig. 3). Therefore, the effects of prosthetic stiffness on net CoT depend on the prosthetic model. Future studies should investigate whether use of C-shaped RSPs that are more than one stiffness category lower than the manufacturer recommended optimize net CoT and whether net CoT remains independent of the 1E90 Sprinter or J-shaped prosthetic stiffness across a wider range of stiffness values.

In addition to improved biomechanics, the metabolic cost of running when using the J-shaped 1E90 Sprinter prostheses compared with the C-shaped RSPs may be due to better sagittal plane alignment, reduced mechanical energy dissipation (less hysteresis), and/or enhanced stability. The sagittal plane alignment of the 1E90 Sprinter prostheses may have elicited GRF vectors that were more aligned with the stance limb, thus mitigating muscular force requirements (13, 14). Also, J-shaped RSPs return ~1% more of the stored elastic energy (~1% less hysteresis) than C-shaped RSPs (11), thus potentially minimizing mechanical work performed by the muscles when using the 1E90 Sprinter prostheses compared with the C-shaped RSPs. Another possible explanation for the reduced metabolic cost of running with the 1E90 Sprinter prostheses compared with the C-shaped RSPs may have been owed to improved lateral stability (6–9). Arellano et al. (10) found that an athlete with bilateral transtibial amputations had greater mediolateral "foot" placement variability than nonamputees while running (10), indicating that lateral balance is compromised compared with nonamputees. Accordingly, it is possible that there is a considerable metabolic cost of maintaining lateral balance during running for athletes with bilateral transtibial amputations (6–9). Overall, 1E90 Sprinter prostheses are wider (0–2.5 cm) and thicker (0.1–0.9 cm) than the C-shaped RSPs at each segment (i.e., proximal, medial, and distal; Refs. 30a, 51a, 51b). Thus the design of the 1E90 Sprinter prostheses may have improved mediolateral stability and consequently reduced the metabolic cost of running compared with the use of C-shaped RSPs.

The improved metabolic cost of running with the J-shaped 1E90 Sprinter vs. C-shaped RSPs was despite the relatively heavy attachments used for the 1E90 Sprinter prostheses. The mass of two brackets plus 1E90 Sprinter prostheses (2,008 g) were 384 and 630 g greater than the mass of the Catapult and Flex-Run prostheses, respectively. Adding mass to the lower legs or feet of nonamputees increases the metabolic cost of running, such that 100 g added to the feet increases metabolic cost by ~1% (22, 37, 45). It is likely that the use of standard, lighter attachments for the J-shaped 1E90 Sprinter prostheses would further decrease the metabolic cost of running.

Numerical reductions in three biomechanical variables, peak braking GRF, stride frequency, and leg stiffness, were associated with improved net CoT. Decreased peak braking GRFs may reduce metabolic cost by mitigating the muscular force generated by the legs during running (9, 24). The potential influence of stride frequency and leg stiffness on metabolic cost is not straightforward. The metabolic cost of running for nonamputees increases when they adopt unnatural stride frequencies (23, 35). However, in the present study, participants used a self-selected stride frequency for each prosthetic configuration. Similarly, when nonamputees adopt higher or lower leg stiffness values than preferred, their metabolic cost of running increases (23, 26, 50, 54). Hypothetically, compliant leg springs decrease the metabolic cost of running compared with stiffer leg springs by prolonging ground contact time and by storing and returning more elastic energy per unit of applied force. Longer ground contact time enables athletes to produce the required vertical force on the ground with slower, more economical muscle fibers (41, 43, 52). Also, storing and returning more elastic energy during running mitigates the muscular mechanical work needed to sustain running, which also elicits more economical muscular force production (5, 20, 21, 55). However, the notion that force generated by isometric muscle contractions is more economical than continuous stretching-shortening contractions has been challenged. Holt et al. (36) reported that while generating force, frog muscles in vitro consume metabolic energy at the same rate when continuously stretching and shortening vs. operating isometrically. Moreover, stiffer leg springs generally have an improved effective mechanical advantage compared with compliant leg springs (13, 14) due to reduced ankle, knee, and hip joint flexion (27, 28). Because the greatest GRF magnitudes are approximately vertical and occur when the runner's center of mass is directly above the center of pressure of the body (Fig. 1; Refs. 15, 25, 47), reduced joint flexion theoretically decreases peak GRF-joint moments due to shorter moment arm lengths, mitigating muscular force requirements. Collectively, moderate leg stiffness seems to minimize the metabolic cost of running by optimizing the interplay of multiple biomechanical factors.

We accept our second (null) hypothesis; the metabolic cost of running was independent of prosthetic height. Since the influence of prosthetic height did not achieve statistical significance, our results generally support those of Williams and Cavanagh (60), who reported a weak relationship between the metabolic cost of running and the leg lengths of nonamputees. Athletes with long residual limbs that compete in sprint events within the T43 classification (athletes with bilateral below-knee amputations) may not able to use C-shaped RSPs because their overall height would exceed the IPC's regulated competition height (37a). However, based on the disassociation between prosthetic height and net CoT from our study, these athletes could increase their height beyond the IPC's regulated competition height without affecting their distance running performance. Further linear mixed-model analyses reveal that prosthetic height was unrelated to stride frequency ($P = 0.162$) or leg stiffness ($P = 0.914$) but was associated with peak braking GRF ($\beta = 0.005$; $P = 0.049$). For every 2-cm increase in prosthetic height, peak braking GRF magnitude increased 4.8%. Additional paired two-tailed *t*-tests revealed that prosthetic mass was similar across height alterations ($P \geq 0.352$), indicating that prosthetic mass did not statistically affect our prosthetic height results.

To our knowledge, only one study has investigated the influence of prosthetic configuration on a facet of running performance. Tominaga et al. (57) altered the sagittal plane alignment of the RSPs $\pm$ 4° for athletes with unilateral transtibial amputations and found no association between alignment and running speed during the acceleration phase of an all-out sprint (57). However, sagittal plane alignment may affect net CoT based on our previous finding that a 1° alignment change alters prosthetic stiffness 0.46–0.79 kN/m, depending on the prosthetic model (11).

Similar to Kerdok et al. (39), we found that reduced in-series stiffness with respect to the stance leg as well as increased mechanical energy return from the in-series spring were associated with a reduced metabolic cost of running. In contrast to Kerdok et al. (39), who found a strong correlation between the metabolic cost of running and the mechanical energy return of the in-series compliant surface, we found an extremely weak correlation between the metabolic cost of running and the mechanical energy returned by the in-series RSPs ($r^2 = 0.055$). Furthermore, we found that prosthetic mechanical energy return was independent of prosthetic stiffness (linear regression: $P = 0.718$) and that overall leg stiffness decreased with reduced in-series stiffness. Collectively, it appears that athletes with and without amputations both run with lower metabolic costs when in-series stiffness is reduced, yet the underlying mechanisms responsible for these changes are different.

We found that the overall leg stiffness (residual limb plus RSP in-series stiffness) of athletes with bilateral transtibial amputations is affected by changes in prosthetic stiffness. Our results coincide with those of McGowan et al. (46), suggesting that prosthetic stiffness governs overall leg stiffness. We found a positive association between residual limb stiffness (biological limb stiffness) and prosthetic stiffness (in-series stiffness; Fig. 5). Therefore, we reject our third hypothesis. Our results are in contrast to those of nonamputee runners whom adjust their biological leg stiffness with altered in-series (surface) stiffness to maintain overall leg plus in-series stiffness (29, 30, 39). Our results indicate that in-series prosthetic stiffness affects the running mechanics of athletes with bilateral transtibial amputations; consequently, traversing terrain of varying compliance likely alters their running mechanics. Biomechanically, leg stiffness is a composite of sagittal plane joint torsional stiffness and leg segment geometries (27, 28, 48). Since RSP stiffness cannot yet be modulated neurally (11) and the hip joint has a negligible influence on leg stiffness (27, 28, 48), it is possible that athletes with bilateral transtibial amputations primarily rely on knee joint mechanics to alter leg stiffness. Future studies are needed to understand the mechanisms underlying the unique leg stiffness results of athletes with bilateral transtibial amputations.

Previously, the metabolic cost of running had only been reported for two athletes with bilateral transtibial amputations (42, 59); this data set now totals seven athletes with bilateral transtibial amputations (Table 2). Selecting the most economical trial for each of our participants and the reported values in the literature, average gross CoT (ml $O_2 \cdot kg^{-1} \cdot km^{-1}$) from these seven athletes with bilateral transtibial amputations is

Table 2. *The lowest and highest elicited gross metabolic cost of transport (CoT) values for the participants in the present study (athletes 1–5) as well as those reported in the literature (athletes 6 and 7)*

Athletes with Bilateral Transtibial Amputations	Lowest Gross CoT, ml $O_2 \cdot kg^{-1} \cdot km^{-1}$	Highest Gross CoT, ml $O_2 \cdot kg^{-1} \cdot km^{-1}$
1	207.0	264.0
2	185.6	216.0
3	182.0	230.2
4	174.2	204.2
5	182.4	220.7
Average ± SD	186.2 ± 12.3	227.0 ± 22.7
6	174.9	N/A
7	216.5	N/A
Average ± SD	188.9 ± 16.3	

Athlete 6 is from Weyand et al. (59), *athlete 7* was tested in Brown et al. (17), and their individual CoT data were reported by Kram et al. (42). N/A, not applicable.

188.9 ± 16.3 ml O_2·kg $^{-1}$·km $^{-1}$ (mean ± SD). For context, Olympic-qualifying, subelite, and good nonamputee runners tested by Morgan et al. (49) elicited mean gross CoT values of 181.9 ± 9.1, 187.5 ± 9.7, and 190.5 ± 13.6 ml O_2·kg $^{-1}$·km $^{-1}$ (mean ± SD), respectively. Furthermore, our study demonstrates the importance of optimizing prosthetic model and stiffness recommendations since the least economical prosthetic configuration for each of our participants yielded average gross CoT values that were 21.9% higher than their most economical trials (227.0 ± 22.7 vs. 186.2 ± 12.3 ml O_2·kg $^{-1}$·km $^{-1}$; paired 2-tailed *t*-test; $P = 0.001$; Table 2).

We were unable to match the maximum IPC competition height for all participants and prosthetic models due to residual limb lengths and/or prosthetic component dimensions. In turn, we adopted a statistical approach that accounted for the discrepancies in participant height across trials. Also, our participants used 2 sets of sockets to complete our protocol (1 set for C-shaped RSPs and 1 set for J-shaped RSPs), thus there could have been unequal residual limb movement within the different sockets. This may have led to varying levels of muscular contraction and/or mechanical energy dissipation. Unfortunately, little is known regarding how prosthetic sockets affect athletic performance. Future studies aimed to understand the influence of socket design on the performance of athletes with lower limb amputations are warranted. Furthermore, 2 of the 5 participants were unable to complete all trials at 3.0 m/s while maintaining primarily aerobic metabolism. As a consequence, those 2 participants completed their trials at 2.5 m/s; therefore, we used net CoT because of its general independence with running speed (9, 44, 49), which we confirmed with our data set using a linear mixed-model analysis ($P = 0.572$). Even though we present the largest data set of running metabolic costs and biomechanics from athletes with bilateral transtibial amputations to date, our relatively small sample size may have falsely led us to accept null hypotheses (type II error) that would be detected with a larger participant cohort.

Conclusions. Prosthetic model and stiffness, but not height, influence the metabolic cost of running for athletes with bilateral transtibial amputations. While controlling for prosthetic stiffness (in kilonewtons per meter), using the Flex-Run and the 1E90 Sprinter prostheses yielded lower net metabolic cost of transports compared with using the Catapult prostheses. Across prosthetic models, use of RSPs that are less stiff than manufacturer recommended (e.g., numerically lower stiffness category) reduced the metabolic cost of running. The use of RSPs of different heights spanning a 4-cm range had no effect on the metabolic cost of running. Mechanically, the leg stiffness of athletes with bilateral transtibial amputations is governed by in-series prosthetic stiffness. In all, athletes with bilateral leg amputations can minimize their metabolic cost of running through the use of RSPs that are optimally designed and have lower prosthetic stiffness compared with the manufacturer-recommended.

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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

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., P.T., 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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