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Amputee Locomotion: Joint Moment Adaptations to Running Speed using Running-Specific Prostheses after Unilateral Transtibial Amputation

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Abstract

Objective—The objective of this study was to investigate 3D lower extremity joint moment differences between limbs and speed influences on these differences in individuals with lower extremity amputations using running-specific prostheses.

Design—8 individuals with unilateral transtibial amputations and 8 control subjects with no amputations ran overground at three constant velocities (2.5, 3.0, and 3.5 m/s). A 2×2×3 (group × leg × speed) repeated-measures ANOVA with Bonferroni adjustments determined statistical significance.

Results—The prosthetic limb generated significantly greater peak ankle plantarflexion moments and smaller peak ankle varus, knee stance extension, knee swing flexion, knee internal rotation, hip stance flexion, hip swing flexion, hip swing extension, hip valgus, and hip external rotation moments than the intact limb. The intact limb had greater peak hip external rotation moments than control limbs, but all other peak moments were similar between these limbs. Increases in peak hip stance and knee swing flexion moments associated with speed were greater in the intact limb than the prosthetic limb.

Conclusion—Individuals with amputation relied on the intact limb more than the prosthetic limb to run at a particular speed when wearing running-specific prostheses, but the intact joints were not overloaded relative to the control limbs.

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Keywords

biomechanics; amputation; gait; transtibial

Introduction

Individuals with lower extremity amputation must demonstrate different biomechanical strategies during running to compensate for limb loss. Joint kinetics adapt during running to account for missing major joints and musculature while also integrating the function of a mechanical prosthesis¹⁻⁸. Most previous studies on running mechanics in people with amputations have used non-running-specific prostheses (non-RSPs), i.e. prostheses originally designed for walking^{2,4-8}. Several studies have also examined running mechanics with RSPs^{1,3,9-11}, but their focus tends toward maximal sprinting. Studies investigating running mechanics with RSPs at submaximal running velocities have examined ground reaction force and metabolic data^{9,10,12}, but our knowledge of joint kinetic information using RSPs at submaximal running speeds is limited. Joint moment data during running after amputation can offer insights into gross muscular effort and joint control. Sagittal plane moments have been presented^{1-3,5,6,8}, but no reports of frontal or transverse plane joint moments currently exist for this population. Frontal and transverse plane moments are important to stabilizing the limbs during running but they have also been associated with loading that may lead to injury risk¹³⁻¹⁵. Consequently, a complete description of three dimensional (3D) joint moment profiles would provide greater insights into how individuals with lower extremity amputation run and compensate for replacing an active limb with a passive prosthetic device. These gaps in knowledge are important because submaximal-speed running is common in day-to-day and sporting activities, and because joint moments affect musculoskeletal health and may play a role in injury risk^{13,15,16}.

Individuals with unilateral amputations have an inherent structural and functional asymmetry where joint moments in all planes may differ between the limbs. Runners using non-RSPs exhibit greater hip extension moments, smaller knee extension moments, and smaller prosthetic ankle plantarflexion moments in the residual limb compared to their intact limb and to able-bodied limbs during stance phase^{2,5,6,8}. One study² also reported greater residual limb hip extensor moments during swing compared to their intact limb and a control group without amputations, which suggests greater hip musculature effort to slow the swinging residual limb in preparation for foot strike. Only two studies report joint moments for people running with RSPs^{1,3}, and these studies present sagittal plane, stance phase data for three total subjects during sprinting. This literature reports inconsistencies where one sprinter with bilateral transtibial amputations generated greater RSP “ankle” plantarflexion moments and minimized knee and hip extension moments compared with a control group without amputations¹. Conversely, two sprinters with unilateral transtibial amputations generated lower RSP ankle plantarflexion moments and greater residual knee and hip extension moments compared to their intact limb joints³. However, one of these subjects also demonstrated smaller residual knee extension moments when using a different type of RSP³. The differences in joint moments observed between these studies could be due to the task

(sprinting vs submaximal running¹⁴), prosthetic components (RSP design), amputation type (bilateral vs unilateral), or a combination of these factors.

In addition to moment asymmetries induced by the structural and functional differences between the residual and intact limbs, running speed also influences lower extremity joint and limb kinetics. For example, ground reaction force (GRF) magnitudes^{2,9,12} and peak sagittal plane joint moments² both increase with speed during running after amputation. How speed may affect the differences in joint moments between the residual and intact limbs has not been explicitly studied during running after amputation. Visually inspecting the existing literature, however, does show subjective changes in kinetic differences between limbs. Residual and intact limb differences in average vertical GRFs appear to increase with running speed⁹, while peak joint moment differences between the limbs appear to reduce or stay the same with increases in running speed². An objective analysis of how running speed affects the differences in moment generation between the limbs can provide insights into limb control, musculoskeletal health, and injury risk.

This study aimed to investigate how 3D lower extremity joint moments change when running with RSPs under different speed constraints. Based on existing literature, it was hypothesized that (1) individuals with lower extremity amputation would demonstrate smaller peak ankle joint moments and greater peak knee and hip joint moments in the residual limb compared to the intact and control limbs throughout stance and swing phase, and (2) greater running speed would be associated with smaller differences between the residual and intact limb peak joint moments.

Methods

Subjects

Eight male subjects with unilateral transtibial amputation (mean age = 32.0 ± 10.2 years, height = 1.80 ± 0.07 m, mass = 82.3 ± 13.0 kg; see Table 1) and eight healthy male control subjects (mean age = 29.0 ± 6.9 years, height = 1.84 ± 0.05 m, mass = 79.3 ± 7.9 kg) participated. Subjects ran in their own prescribed RSPs to reduce variability due to using a new prosthetic design and to ensure proper alignment. Subjects with amputation had at least 3 months of running experience (range: 3-256 months) and amputations were either congenital (N=1) or traumatic (N=7). This study was conducted in accordance with the Declaration of the World Medical Association. Prior to participating, all subjects gave informed written consent approved by the University of Maryland Institutional Review Board. This study conforms to all STROBE guidelines and reports the required information accordingly (see Supplementary Checklist). Subjects were excluded if they had comorbidities on the intact limb affecting gait.

Material Properties and Anthropometrics

Subject masses were measured using a force platform. Each subject's height and body weight were measured, and anthropometric measurements from marker positions estimated the mass, center of mass, and moments of inertia of intact limb segments^{17,18}. Since subjects with amputation were missing one foot and part of their shank, an adjusted body mass

(ABM)¹⁹ was used as an input to anthropometric regression equations that accounted for the missing body segments.

$$ABM = \frac{MBM - m_p - m_{res}}{1 - c} \quad [1]$$

where MBM is measured body mass while wearing the prosthesis, m_p is the prosthesis mass including the socket, m_{res} is the estimated residual limb mass, and c (0.057) is the percent of ABM accounted for by the intact shank and foot¹⁸.

For subjects with amputation, residual limb length and circumferences at the knee joint and distal end of the limb were measured using a measuring tape. Residual limb inertial properties were then estimated as a frustum of a right circular cone^{17,20}. Residual limb mass was estimated from the calculated geometric volume assuming a uniform 1.10 g/cm³ tissue density^{20,21}. The prosthetic socket and RSP were treated as one unit and weighed with a laboratory scale. The socket+RSP unit's center of mass position was calculated using a reaction board²², and the moment of inertia of the socket+RSP unit was calculated from the period of oscillation measured with a trifilar pendulum^{23–26}. Lower limb segment inertial properties were then calculated by combining the residual limb and prosthetic components. The RSP keels could not be disconnected from the sockets for each subject, so inertial properties for each prosthetic keel were estimated using data reported for the RSP model and stiffness category that most closely matched those used by the subjects²⁵. The socket's inertial properties were then estimated by subtracting the inertial properties of the RSP keel from the total socket+RSP segment. Subsegments within the RSP keels were defined via marker placements and the inertial properties of each subsegment were estimated by assuming each segment as a rigid trapezoidal cuboid²⁷.

Experimental Procedures

Retroreflective markers were placed bilaterally over the anterior and posterior iliac spines, heel, 3rd and 5th metatarsal heads, and tip of the toe on the shoe. Marker clusters were placed bilaterally on the lateral thigh and shank segments. An additional static trial included markers placed on the lateral and medial femoral condyles and the lateral and medial malleoli. On the amputated limb, the shank cluster was placed laterally on the socket and a marker was placed at the distal tip of the socket to define the long axis of the residual shank segment. Three markers were placed on the prosthesis keel including the most proximal end (“Head”), the most distal end (“Toe”), and laterally on the most acute point of the prosthesis curvature, which defined the prosthetic limb “ankle” joint (Figure 1). Although RSPs do not have an anatomical ankle joint, the term “ankle” was used here to remain consistent with past studies.

Subjects ran overground around a 100m long track at three running speeds (2.5 m/s, 3.0 m/s, and 3.5 m/s). The testing setup (see Hobara et al. for experimental setup figure²⁸) included a 10-camera motion capture system (Vicon, Centennial, CO) that captured 3D marker positional data at 200 Hz for kinematic analysis, and ten force platforms (Kistler, Amherst, NY) embedded in the track in series that collected GRFs at 1000 Hz. Subjects completed at

least five successful trials for each leg at each running speed where they ran within ± 0.2 m/s of the target speed and stepped within the force platform boundaries. Predetermined running speeds were governed using concurrent biofeedback. Six sets of laser sensors evenly distributed around the track instantaneously calculated average speed over a track section when the subject ran past the sensors. Verbal feedback was given to subjects indicating whether they were running at the desired speed and trials that were not within 0.2 m/s of the target speed were excluded. Running speed ordering was randomized. Subjects rested for as long as needed between speed conditions to reduce fatigue with a minimum rest of five minutes between conditions.

Raw marker data were filtered using a 4th order, zero lag low pass Butterworth filter with a 6 Hz cutoff frequency while raw force data were similarly filtered with a 30 Hz cutoff frequency^{10,29}. Standard inverse dynamics calculations³⁰ estimated lower extremity joint moments using a 6-degree-of-freedom kinematic model via Visual3D (C-Motion, Germantown, MD) software. The pelvic, thigh, and shank segment coordinate systems were used as the reference frames for the hip, knee, and ankle moments, respectively. The RSP “ankle” moments were calculated using the reference coordinate system for the segment defined by the RSP Head and acute point markers.

Statistical Analysis

Peak ankle, knee, and hip moments in each plane of motion during stance and swing phase were compared between intact and residual limbs of subjects with amputation and between Amputation and Control groups at each running speed. Statistical comparisons were performed in SPSS 19.0 (SPSS Inc.). A $2 \times 2 \times 3$ three factor repeated measures analysis of variance (ANOVA) identified statistical differences between the dependent variables using Group (Amputation and Control), Leg (prosthetic/intact and left/right), and Speed (2.5 m/s, 3.0 m/s, 3.5 m/s) as independent variables (IVs). Group was treated as a between-subjects variable while Leg and Speed were treated as within-subjects variables. When significant differences were identified from the full factorial model, two-way ANOVAs and pair-wise comparisons with Bonferroni adjustments for multiple comparisons were used to determine which conditions significantly differed from each other. Independent t-tests determined differences between Groups in age, height, and mass. Significance for all statistical tests were set at $\alpha = 0.05$.

Results

Demographical data and inertial properties measured for each subject with amputation are presented in Table 1. No differences in joint moments were observed between the left and right control limbs for any variable ($P > 0.05$). Consequently, these data were averaged to generate a representative control limb for clearer presentation in tables and figures. However, all statistical outcomes were based on the balanced statistical design that included both control limbs. No differences existed between groups for age ($P = 0.503$), height ($P = 0.223$), or mass ($P = 0.590$). No limb length differences existed between the Amputation and Control groups ($P = 0.619$), but prosthetic limbs were longer than the intact limbs

($P<0.01$). Figure 2 shows joint moment profiles at each running speed for the hip, knee, and ankle at each plane of motion normalized to the gait cycle.

Ankle Moments

Sagittal plane peak ankle moments are presented in Figure 3. Group, leg, and speed main effects existed for peak ankle plantarflexion moments ($P<0.05$). The Amputation group had greater peak plantarflexion moments than the Control group due to the prosthetic “ankle” generating greater peak plantarflexion moments than all other limbs. Both legs of both groups increased peak plantarflexion moment with speed ($P<0.01$). A leg-by-group interaction ($P<0.01$) indicated peak plantarflexion moments in prosthetic and intact limbs were more sensitive to speed (i.e. greater increases) than Control limbs. Peak ankle stance varus moments had leg ($P<0.01$) main effects and leg-by-group ($P<0.01$) interaction effects. The prosthetic limb had lower peak varus moments than the intact and control limbs ($P<0.01$), and the interaction effect was due to prosthetic and intact limbs having greater differences than Control limbs. No significant differences existed for peak ankle internal/external rotation moments ($P>0.100$).

Knee Moments

Sagittal plane peak knee moments are presented in Figures 3–4. Leg and speed main effects existed for stance phase peak knee extension and swing phase peak knee flexion moments ($P<0.01$). The prosthetic limb knee had lower stance peak knee extension and swing peak knee flexion moments than the intact and control limbs ($P<0.05$). Peak knee extension moments in stance increased with speed for Amputation group limbs ($P<0.01$), but not for Control limbs. Peak knee flexion moments in swing increased with speed ($P<0.01$). Leg-by-group interactions for both of these variables ($P=0.01$) indicated greater differences between the prosthetic and intact limbs than between the Control limbs. A leg-by-speed interaction existed for Amputation group swing phase peak knee flexion moments ($P<0.01$) where the intact knee moments increased with speed at a greater rate than the prosthetic limb knee moments.

The intact limb peak valgus moment increased with speed ($P<0.01$). Peak knee internal rotation moments had a leg main effect ($P<0.05$) where the prosthetic limb had lower peak internal rotation moments than the intact limb.

Hip Moments

Sagittal plane peak hip moments are presented in Figures 3–4. Leg main effects were evident for peak hip stance and swing extension moments ($P<0.01$). The residual limb had lower peak stance hip extension moments than the intact and control limbs ($P<0.01$) at all velocities. The prosthetic limb had lower peak hip extension moments in swing than the intact limb at each speed ($P<0.01$) and was lower than the control limb at 3.0 m/s ($P<0.05$) but did not reach significance at 2.5 m/s ($P=0.053$) or 3.5 m/s ($P=0.053$). All four limbs increased peak stance and swing flexion and extension moments with speed ($P<0.01$). The Control group had greater peak hip moments than the Amputation group ($P<0.05$). Leg-by-group interactions ($P<0.01$) for hip peak stance and swing extension moments indicated the Amputation group had greater differences between the limbs than the control group. A leg-

by-speed interaction existed for peak hip stance flexion moment ($P<0.05$) where the intact limb moment increased at a greater rate with speed than did the prosthetic limb moment.

Peak stance phase hip valgus and external rotation moments had speed and leg main effects ($P<0.01$) and leg-by-group interaction effects ($P<0.01$). The control limb peak valgus and external rotation moments increased with speed ($P<0.05$), while the intact limb peak valgus moment and prosthetic limb peak external rotation moment increased with speed ($P<0.05$). The prosthetic limb generated lower peak valgus and external rotation moments than the intact and control limbs ($P<0.05$). The intact limb had greater peak external rotation moments than the control limbs ($P<0.05$). Leg-by-group interactions were caused by greater differences between the prosthetic and intact limbs than the control limbs for both of these parameters.

Discussion

This study aimed to improve understanding of 3D lower extremity joint moments throughout the gait cycle for individuals running with RSPs under different speed constraints. Our first hypothesis predicted that runners with amputation would demonstrate smaller peak ankle joint moments and greater peak knee and hip joint moments in the prosthetic limb than the intact and control limbs throughout stance and swing phase at each speed. This hypothesis was only supported by smaller prosthetic ankle varus moments compared to the intact and control ankles. However, this hypothesis was rejected with the prosthetic limb generating greater peak ankle stance plantarflexion moments and smaller peak knee stance extension, knee swing flexion, knee internal rotation, hip stance and swing flexion, hip extension, hip valgus and hip external rotation moments than the intact and control limbs. In particular, smaller stance phase peak knee extension, hip extension, and hip valgus moments in the prosthetic limb support the idea that the intact limb is overloaded compared to the prosthetic limb when running with RSPs.

Smaller varus moments in the RSP compared to the intact and control ankles are indicative of the RSP primarily acting in the sagittal plane, and suggest a limited ability to control motion in the frontal plane of the residual limb. Greater plantarflexion moments in the prosthesis were due to the longer effective moment arm between the center of pressure and the most acute point of curvature on the prosthesis where the “ankle” was defined (Figure 1). RSP “ankle” plantarflexion moments are sensitive to marker placement¹ and are not directly comparable to intact limb ankle moments due to vastly different architecture and mechanical moment arms. The term “ankle” was used to describe the most acute point on the RSP in order to remain consistent with past studies. However, as RSPs do not have an ankle joint, future studies should consider using alternate joint terminology to report RSP outcomes. We recommend referring to the specific points, such as the most acute point on the RSP to clarify results and use more appropriate terminology.

The frontal plane peak knee and hip moments were similar between the intact and control limbs. The intact limb hip external rotation moments were greater than the control limbs, which may imply that these subjects are at greater risk of injury when running. Externally rotating the hip may be a compensation strategy to produce additional propulsive force or to

control the body's momentum. Targeting the hip during running training to reduce the non-sagittal plane motions could improve efficiency and reduce this potential injury risk; however, more research is needed to identify whether greater external rotation moments have negative consequences. While not significantly different due to high variability, the data suggest increased ankle joint internal rotation moments in the Amputation group compared to the Control subjects. These moments could also be an indicator of future orthopaedic issues, but more specific investigation into these variables is needed to confirm this speculation.

The lower extremity extension moments that provide resistance to limb collapse and propulsive power generation along with the frontal and transverse plane moments were lower in the prosthetic limb compared to both the intact and control limbs. Smaller prosthetic limb stance knee extension moments are consistent with previous running studies of individuals with unilateral transtibial amputation wearing non-RSPs^{2,5,6,8} and of a single subject sprinting with bilateral RSPs¹. Smaller prosthetic limb stance hip extension moments are consistent with the bilateral sprinter (compared to control subjects)¹. These results are indicative of a more extended limb throughout the gait cycle. Residual limb knee extensor strength may be reduced after amputation², so reducing the range of motion and thus the moments could minimize the muscular requirements in the affected leg. Keeping the limb more extended could increase loading rates in the residual joints, thus increasing injury risks. However, lower peak and average GRFs¹², vertical impact peaks²⁸, and vertical average loading rates of GRFs²⁸ previously reported in the prosthetic limb likely lead to lower joint loading rates. This could allow individuals with amputation to run with more extended residual limbs without increasing their injury risk in this limb. Reduced residual knee flexion loading responses during stance phase would also minimize the muscular demand from the residual knee and hip joints to control knee flexion and prevent knee collapse during weight acceptance. An overall reduced muscular demand could explain the reduced frontal and transverse plane moments. These data may also indicate an attempt to minimize the joint loading on or ease the control demands of the prosthetic limb. Whether this is due to an effort to reduce these muscular requirements or the limb's inability to generate these requirements is not clear. In either case, general strengthening of the lower extremity musculature would likely benefit runners with amputation. An alternate possibility is that the affected limb generates greater co-contractions, consequently reducing the net joint moments. Investigating muscle activities via electromyography could confirm this conjecture. No differences in peak moment values were identified between the intact and control limbs during stance phase suggesting that the joint loading on the intact limb is not greater than normal. These data also indicate that the intact limb behaves very similarly to control limbs with respect to the sagittal plane joint moment profiles. Assuming no major differences in antagonistic co-contraction, joint loading in the intact and control limbs may therefore be similar when running with RSPs. This would allow clinicians greater freedom in running training for individuals with lower extremity amputation without concern for overloading the intact limb.

Runners using RSPs have significantly lower terminal swing phase knee flexion moments in the prosthetic limb knee than the intact and control limbs. This is most likely due to the reduced mass and inertia of the prosthetic shank/foot complex when wearing RSPs. Less

demand is placed on knee flexor musculature to slow down knee extension as the knee joint approaches full extension. The range of subjects' residual limb lengths here (0.12m-0.24m) could affect limb control by altering the inertial properties and muscular attachment points of the lower leg, and further investigations into how residual limb length may influence running kinetics and control are warranted. Hip extension moments at terminal swing were also lower in the prosthetic limb as compared to the intact limb. Again, this may reflect lower muscular demand to slow down the thigh due to the lower overall mass of the amputated limb. The intact limb hip peak swing flexion moments were lower than the control group peaks at each running speed, which correlate to reported shorter step lengths in this limb compared to controls²⁷. Generating shorter step lengths lowers the peak hip flexion moment demand since the hip would not need to flex as much. Electromyography and joint power analyses would provide valuable information on the muscle firing patterns and mechanical energy flow during running after amputation and further elucidate altered control strategies.

The second hypothesis that increased running speed is associated with greater increases in peak intact limb joint moments compared to the prosthetic limb was rejected for most parameters but supported for peak hip stance and knee swing flexion moments. Greater stance hip flexion moments indicates an increased reliance on hip flexor musculature at faster running speeds in order to progress the limb forward. Consequently, strengthening the hip flexors may be an important component of running training for the amputee population. No leg-by-speed interactions existed for the knee or hip stance extension moments, which indicates subjects with amputation did not increase their reliance on the intact limb at faster running velocities. It appears that the intact and residual limb joints increase their moment magnitudes with speed in a similar manner when supporting body weight or generating propulsion. However, the increase in peak hip stance flexion moment associated with running speed was greater in the intact limb than the prosthetic limb. This was influenced by greater intact limb hip joint accelerations at faster running speeds. Running speed also had a greater effect on the intact limb knee swing flexion moment than the prosthetic limb. These results contrast with the only previous running study to report swing phase joint moments in subjects with amputation, which reported similar swing phase knee moment profiles between the intact, residual, and able-bodied limbs². However, in that study subjects wore non-RSPs that had masses and inertial properties much closer to those of the intact limbs than the lighter RSPs in this study. Thus, the prosthetic and intact limb knee joint musculature would have similar mechanical demands to slow knee extension at terminal swing. Intact limbs have greater masses than the RSPs worn in the current study. Therefore, as limb swing speeds increased the intact limb induced an increasingly greater knee flexor demand compared to the prosthetic limb in order to slow the shank extension.

This study is limited by several factors. First, the subject pool is small (N=8) comprising of individuals with traumatic and congenital amputations and thus may limit generalization to a wider population. The range of running speeds was limited between 2.5 and 3.5 m/s. While the results provide insights into running adaptations within this speed range, it is unknown whether these data can be extrapolated to slower or faster submaximal speeds or to sprinting speeds. Therefore, future studies should examine running with RSPs across a larger range of running speeds. Furthermore, the different RSP designs used by our subjects were not

considered as a confounding factor. Different RSP designs may induce different joint moment profiles as reported for peak hip extension moments in two subjects sprinting³. Our eight subjects used three different RSP designs, and there did not appear to be any obvious consistent differences between the moment patterns of subjects using each design. However, a systematic study of subjects running with different RSP designs would provide an important clarification of this issue. The current study focuses on peak joint moments, but the timing of these peaks and the integral of the moment curves may provide further insights into adaptations to running after amputation. Future investigations into these variables would be valuable to our understanding. Individuals with congenital amputations may have different motor control patterns compared to those with traumatic amputations, although visual inspection of the data revealed no obvious difference between the one subject with congenital amputation and subjects with traumatic amputation. Clarifying any potential functional differences between individuals with congenital vs traumatic amputations would be valuable additions to our knowledge.

Conclusions

The intact limb knee and hip joints generated greater peak moments than the prosthetic limb in all three planes of motion suggesting a greater reliance on the intact limb during running. However, with the exception of hip internal rotation moments, the intact limb generated similar peak moment values to the control limbs, which indicates that the intact limb is not overloaded when individuals with lower extremity amputation run with RSPs. Only the hip peak stance flexion moment and knee swing flexion moment increases associated with speed were greater in the intact limb than the prosthetic limb, which is most likely a result of the greater joint accelerations and inertial properties in the intact limb requiring a greater muscular demand at faster running velocities.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

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Figure 1. Marker placements on running-specific prostheses with views of the frontal plane (a) for all prostheses and sagittal planes for the (b) Flex-Run (shown) and Catapult and (c) Cheetah prostheses. L, M, P, and A refer to the lateral, medial, posterior, and anterior directions, respectively.

Figure 2.

Average internal hip, knee, and ankle joint moments normalized to body mass and the gait cycle at each tested speed for (a) sagittal, (b) frontal, and (c) transverse planes of motion. Flex/Ext, Var/Val, and In/Ex represent flexion/extension, varus/valgus, and internal/external rotational moments, respectively. Positive values indicate flexion, varus, and internal rotation moments.

Figure 3.

Peak stance phase flexion/extension moments for the hip, knee, and ankle normalized to body mass across the tested speeds. Error bars represent ± 1 standard deviation. * indicates significant differences ($P < 0.05$) between limbs.

Figure 4.

Peak swing phase hip and knee flexion/extension moments normalized to body mass across the tested speeds. Error bars represent ± 1 standard deviation. * indicates significant differences ($P < 0.05$) between limbs.

Table 1

Subject residual limb length and prosthesis inertial properties. The prostheses could not be disconnected from the sockets so all measured inertial properties were taken for the prosthesis and socket combined unit.

Subject with Amputation	RSP model	Amputated Limb	Residual Limb Length (m)	RSP mass (kg)	CM _x (m)	CM _y (m)	CM _z (m)	I _{xx} (kg·m ²)	I _{yy} (kg·m ²)	I _{zz} (kg·m ²)
1	Flex-Run	Right	0.12	1.478	-0.050	-0.0003	0.056	0.0356	0.0377	0.0041
2	Flex-Run	Left	0.14	1.328	-0.042	0.0004	0.082	0.0302	0.0317	0.0037
3	Flex-Run	Left	0.21	1.315	-0.044	0.0036	0.016	0.0310	0.0330	0.0038
4	Cheetah	Left	0.24	1.439	0.060	-0.0093	-0.086	0.0369	0.0404	0.0065
5	Cheetah	Right	0.20	1.892	0.044	-0.0045	-0.088	0.0858	0.0892	0.0104
6	Flex-Run	Left	0.21	1.770	-0.044	0.0057	0.070	0.0579	0.0598	0.0284
7	Catapult	Left	0.24	1.341	-0.036	-0.0025	0.028	0.0264	0.0284	0.0042
8	Catapult	Right	0.18	1.802	-0.032	0.0050	0.095	0.0775	0.0799	0.0067

RSP = Running-specific prosthesis

x, y, and z represent the anteroposterior, mediolateral, and vertical axes, respectively

CM_x, CM_y, CM_z = Center of mass position of the RSP+socket along the x, y, and z axes with respect to the most proximal end of the prosthesis keel

I_{xx}, I_{yy}, I_{zz} = Principal axis moments of inertia about the center of mass of the RSP+socket