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Walking characteristics of runners with a transfemoral or knee-disarticulation prosthesis

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Abstract

Background: Running with prostheses has become a common activity for amputees participating in sports and recreation. However, very few studies have characterized the kinematic and kinetic parameters of walking in individuals with amputation who are runners. Thus, this study attempts to elucidate the kinematics and kinetics of walking in runners with a unilateral transfemoral amputation or knee-disarticulation.

Methods: This study experimentally compares the prosthetic and intact limbs of runners with prostheses as well as compares the findings against the limbs of age-matched able-bodied individuals while walking. Fourteen runners with a unilateral transfemoral amputation or knee-disarticulation were recruited and 14 age-matched able-bodied individuals were prepared using gait database. Spatiotemporal, kinematic, and kinetic parameters of walking were analyzed using a 3-demensional motion capture system.

Results: The results showed that the peak ankle positive power at pre-swing and peak hip positive power from loading response to mid stance in the intact limb were significantly larger than that in the prosthetic limb. Moreover, to compensate for missing anatomical functions on the prosthetic limb, it appeared that the intact limb of the runners generated larger peak joint power by producing more ankle plantarflexor and hip extensor moments while walking.

Interpretation: This study demonstrated that the runners rely on their intact limb while walking. Training of hip extensor muscles of the intact limb may be beneficial for these individuals.

Keywords: Athlete; Amputee locomotion; Gait; Paralympic; Running

Introduction

The physical and psychosocial benefits of sports and recreational activities have been widely recognized for individuals with amputation, but their participation in these activities is limited (Bragaru et al., 2011). Running with a transfemoral prosthesis generally requires a tremendous personal effort and training in order to develop the necessary strength in both the amputated and intact lower limbs. Fortunately, the recent technological advances in prosthetic components have paved the way for more individuals with a transfemoral amputation to participate and engage themselves in running activity (Nolan, 2008). Walking characteristics have been widely analyzed in general population of individuals with transfemoral amputation. But studies specifically analyzing walking characteristics in runners with transfemoral amputation have been limited. One example of such study includes a comparison of energy cost with runners without amputation (Mengelkoch et al., 2017). The runners with amputation demonstrated greater energy cost and lower gait performance under various prosthetic feet conditions in comparison to runners without amputation.

Previous studies on analyses of walking have provided insights on how prosthetic components, such as knees (Creylman et al., 2016; Eberly et al., 2014; Segal et al., 2006), sockets (Traballesi et al., 2011) and feet (Graham et al., 2007) may affect the walking ability of individuals with transfemoral amputation.

The walking ability of individuals with transfemoral amputation may also be influenced by their physical characteristics, such as functional levels (Sions et al., 2018) and residual limb length (Bell et al., 2013). In general, highly functional individuals could achieve comparable walking speed to able-bodied individuals. However, overall reductions in biomechanical and functional performances are often reported (Jarvis et al., 2017; Mahon et al., 2017).

Asymmetry in kinematics and kinetics of walking has been commonly observed in individuals with unilateral transfemoral amputation (Kaufman et al., 2012). Studies investigating maximal sprinting in runners with transfemoral amputation have also shown that peak ground reaction force in the intact leg was greater than those of the prosthetic leg while running (Makimoto et al., 2017; Sano et al., 2017). Asymmetric loading patterns during gait may potentially lead to the amputees experiencing lower back pain (Shojaei et al., 2016) or osteoarthritis of the knee in the intact limb (Russell Esposito et al., 2015). These loading patterns may be attributed to the fact that individuals with transfemoral amputation compensate for the missing functions of the amputated limb, such as active knee extension and ankle plantarflexion, by exerting more effort with the intact limb (Schaarschmidt et al., 2012). The runners with amputation are highly active and expected to be functionally and physically fit. And they may present unique walking characteristics, such as generating greater amount of lower-limb joint moments and power in comparison to general population of individuals with transfemoral amputation. Therefore, analyzing their walking characteristics may provide novel insights on walking in individuals with amputation.

The aim of this study is to characterize the kinematics and kinetics of walking in runners with transfemoral or knee-disarticulation amputation. We hypothesize that the intact limb of the runners with amputation generates more joint power at ankle, knee and hip when compared to the prosthetic limb as well as the limbs of age-matched able-bodied individuals while walking. In addition, we attempt to compare our results with existing literature to identify potential differences in gait between our unique group of runners with transfemoral amputation and general population of individuals with transfemoral amputation.

Methods

Study participants

Fourteen individuals with a unilateral amputation (12 individuals with a transfemoral amputation and 2 individuals with a knee-disarticulation; 4 females and 10 males) participated in this study (Table 1). Their mean (SD) age was 32.6 (10.2) years, their mean (SD) body height was 1.65 (0.10) m, and mean (SD) body mass was 60.0 (10.6) kg. The individual data on age, body height, and body mass were excluded from Table 1 in order to avoid identification of the participants. Eight participants had amputations due to trauma; five participants had amputations due to cancer, and one participant had amputation due to infection. Of the 14 individuals, eight individuals had amputations on the left side, while six individuals had amputations on the right side. The mean (SD) time since amputation was 14.1 (9.1) years. Twelve individuals with transfemoral amputation used an ischial containment socket, while two individuals with knee-disarticulation used a distal end-bearing socket. Clinical observations showed that, from the individuals with transfemoral amputation, four individuals had a long residual limb, five individuals had a middle residual limb, and three individuals had a short residual limb. All the participants used a silicon liner. The inclusion criteria of this study as runners were: 1) aged over 18 years, 2) user of a unilateral transfemoral or knee-disarticulation prosthesis, 3) active, exercising regularly (i.e., at least once a week) and able to run using a running-specific prosthesis at a speed of more than 4.0 m/s, and 4) no confounding neurological or orthopedic issues. The types of prosthetic knees and prosthetic feet used by the participants for walking and running are listed in Table 1. In addition, we utilized AIST (Advanced Industrial Science and Technology) Gait Database to prepare 14 age- and walking speed-matched able-bodied individuals as the control group (Kobayashi et al., 2015). The mean (SD) age of the control group was 32.5 (9.5) years, the mean (SD) body height was 1.69 (0.09) m, and the mean (SD) body mass was 66.9 (12.5) kg. The study was approved by the Institutional Review Board, and written informed consent was obtained from all subjects and the guardian of one subject prior to the experiment.

Gait analysis

A 3D motion capture system (MX-T 160, Vicon, Oxford Metrics, UK) equipped with 15 infrared cameras and nine force plates (BP400600-1000, AMTI, USA) was used to measure spatiotemporal, kinematic, and kinetic parameters of gait. The sampling frequency of the cameras was 200 Hz, while that of the force plates was 2000 Hz. For control group, 3D positional data and ground reaction forces were collected at 200 Hz and 1000 Hz, respectively (Kobayashi et al., 2015). Reflective markers (14 mm) were placed on each participant based on the Helen-Hayes marker set (Kadaba et al., 1990). For data collection, the participants were instructed to walk upright at self-selected walking speed on a 10 m walkway for 5 times.

Data were recorded and synchronized using Vicon Nexus software (Vicon Motion Systems, Oxford, UK) and post-processed using Visual3D (CMotion, Germantown, MD, USA). The gait data were filtered using a fourth-order Butterworth low-pass filter with a cut-off frequency of 10 Hz for the marker trajectories and 100 Hz for the force data. For the control group, 3D positional data and ground reaction forces were filtered using a fourth-order Butterworth low-pass filter with a cut-off frequency of 10 Hz and 50 Hz, respectively (Kobayashi et al., 2015). Because ground reaction force data of the control group in AIST Gait Database were collected at the sample frequency of 1000 Hz, their cut-off frequency was set at 50 Hz. The spatiotemporal and kinematic parameters of gait from a prosthetic limb of the amputees, an intact lower-limb of the amputees, and a right lower-limb of the able-bodied individuals were calculated with marker data, while kinetic parameters of gait were calculated with marker and ground reaction force data using inverse dynamics (Winter, 2005). The parameters were averaged over five trials. The kinetic data were normalized to the body mass of the participants. Subsequently, the mean of the prosthetic limb of the amputee, the intact limb of the amputee, and the right lower-limb of the able-bodied individuals were plotted for the ankle, knee, and hip joint angle, as well as moment and power, which were normalized to a gait cycle in time. Ankle dorsiflexion, knee flexion, and hip flexion were defined as positive for the joint angles, while ankle plantarflexor, knee extensor, and hip extensor were defined positive for the internal joint moments.

To determine the spatiotemporal parameters of gait, the following parameters were extracted: walking speed (m/s); gait cycle time (s); cadence (steps/min); stride length (m); step length (m); step width (cm); stance time (%); and double support time (%).

Kinematic and kinetic parameters of the ankle, knee and hip joints were analyzed based on the hypothesis of this study. Phases of the gait cycle were defined according to Perry (Perry and Burnfield, 2010). As regards the kinematic parameters of gait, the following parameters were extracted: peak ankle plantarflexion angle at loading response [AA1] (°); peak ankle dorsiflexion angle from terminal stance to pre-swing [AA2] (°); peak ankle plantarflexion angle from pre-swing to initial swing [AA3] (°); peak knee flexion from loading response to mid stance [KA1] (°); peak knee flexion from initial swing to mid swing [KA2] (°); and peak hip extension angle from terminal stance to pre-swing [HA1] (°).

Finally, for the kinetic parameters of gait, the following parameters were extracted: peak ankle dorsiflexor moment at loading response [AM1] (Nm/kg); peak ankle plantarflexor moment from terminal stance to pre-swing [AM2] (Nm/kg); peak ankle negative power at terminal stance [AP1] (W/kg); peak ankle positive power at pre-swing [AP2] (W/kg); peak knee extensor moment from loading response to mid stance [KM1] (Nm/kg); peak knee flexor moment at terminal stance [KM2] (Nm/kg); peak knee extensor moment at pre-swing [KM3] (Nm/kg); peak knee negative power at loading response [KP1] (W/kg); peak knee positive power at mid stance [KP2] (W/kg); peak knee negative power at pre-swing [KP3] (W/kg); peak knee negative power from mid swing to terminal-swing [KP4] (W/kg); peak hip extensor moment from initial contact to loading response [HM1]

(Nm/kg); peak hip flexor moment from terminal stance to pre-swing [HM2] (Nm/kg); peak hip positive power from loading response to mid stance [HP1] (W/kg); peak hip negative power at terminal stance [HP2] (W/kg); and peak hip positive power at pre-swing [HP3] (W/kg) (Winter, 1987).

Statistical analysis

Statistical analyses were performed using SPSS v.22.0 (IBM Corp. Armonk, USA). Normality of data distribution was assessed using the Shapiro–Wilk and the Kolmogorov–Smirnov test. For walking speed, gait cycle time, cadence, and stride length, the statistical comparisons between amputees and able-bodied individuals were performed using t-test. For KP1 and KP2, statistical comparisons between the intact limb of the amputees and able-bodied individuals were performed using t-test. For the parameters that did not demonstrate normality (i.e., double support time, AP2, KA1, KM1, KM3 and KP4), statistical comparisons between the intact limb of the amputee, the prosthetic limb of the amputee, and able-bodied individual were performed using the Kruskal–Wallis test. As regards other parameters that demonstrated normality, one-way ANOVA was performed for the statistical comparisons. The significance level was set at P-value < 0.05 for each analysis. Post-hoc testing for multiple comparisons was performed using t-test for parametric parameters and Mann–Whitney U test for non-parametric parameters using Bonferroni correction of the *P*-value.

Results

Spatiotemporal parameters

As shown in Table 2, the statistical analysis showed no significant differences between the amputees and the able-bodied individuals with the exception of step width ($P < 0.001$) by t-test and stance time ($P < 0.001$) by one-way ANOVA.

Kinematic and kinetic parameters of the ankle

Figures 1 illustrate the plots for mean ankle, knee and hip angle, moment, and power of the amputees (prosthetic and intact limb) and able-bodied individuals, respectively, with standard deviations of the able-bodied limb. The mean, standard deviations, and 95% confidence intervals of the ankle, knee and hip angle, moment, and power parameters as well as the statistical outcomes are summarized in Table 2. One-way ANOVA demonstrated significant differences in peak ankle plantarflexion angle from pre-swing to initial swing [AA3] ($P < 0.001$) and peak ankle plantarflexor moment from terminal stance to pre-swing [AM2] ($P < 0.001$). Kruskal-Wallis test demonstrated significant differences in peak ankle positive power at pre-swing [AP2] ($P < 0.001$). The results of the post-hoc tests for ankle parameters, which exhibited significant differences, are shown in Figure 2. The intact limb demonstrated significant differences in AA3 ($P < 0.001$), AM2 ($P < 0.001$), and AP2 ($P < 0.001$) when compared with the prosthetic limb. However, the intact limb demonstrated significant differences in AA3 ($P = 0.004$) when compared to the able-bodied limb. The prosthetic limb demonstrated significant differences in AA3 ($P < 0.001$), AM2 ($P = 0.002$), and AP2 ($P = 0.001$) in comparison to the able-bodied limb.

Kinematic and kinetic parameters of the knee joint

As shown in Table 2, the intact limb demonstrated the significant differences in peak knee negative power at loading response [KP1] ($P = 0.004$) and peak knee positive power at mid stance [KP2] ($P = 0.040$), when compared to the able-bodied limb. One-way ANOVA showed significant differences in peak knee extensor moment in peak knee negative power at pre-swing [KP3] ($P < 0.001$). Kruskal-Wallis test demonstrated significant differences in peak knee flexion from loading response to mid stance [KA1] ($P < 0.001$), peak knee extensor moment from loading response to mid stance [KM1] ($P = 0.002$), and peak knee negative power from mid swing to terminal-swing [KP4] ($P < 0.001$). The results of the post-hoc tests for the knee parameters that demonstrated significant differences are shown in Figure 2. In comparison to the prosthetic limb, the intact limb demonstrated significant differences in KA1 ($P < 0.001$) and KP3 ($P < 0.001$). Moreover, in comparison to the able-bodied limb, the prosthetic limb demonstrated significant differences in KA1 ($P < 0.001$), KM1 ($P = 0.001$) and KP3 ($P = 0.012$).

Kinematic and kinetic parameters of the hip joint

One-way ANOVA demonstrated significant differences in peak hip extensor moment from initial contact to loading response [HM1] ($P < 0.001$), peak hip flexor moment from terminal stance to pre-swing [HM2] ($P < 0.001$), peak hip positive power from loading response to mid stance [HP1] ($P < 0.001$), peak hip negative power at terminal stance [HP2] ($P = 0.020$), and peak hip positive power at pre-swing [HP3] ($P = 0.042$). The results of the post-hoc tests for the hip parameters, which exhibited significant differences, are shown in Figure 2. When compared to the prosthetic limb, the intact limb demonstrated significant differences in HM1 ($P < 0.001$), HM2 ($P < 0.001$), HP1 ($P < 0.001$), HP2 ($P = 0.016$) and HP3 ($P = 0.037$); moreover, it exhibited significant differences in HM1 ($P < 0.001$) and HP1 ($P = 0.003$) when compared with the able-bodied limb. The prosthetic limb demonstrated significant differences in HM2 ($P = 0.006$) in comparison to the able-bodied limb.

Discussion

In this study, we characterized the kinematics and kinetics of walking in individuals with transfemoral or knee-disarticulation amputation who are runners by comparing the intact and prosthetic limb of amputees as well as the limbs of age-matched able-bodied individuals. The findings suggested that the peak hip positive power from loading response to mid stance [HP1] of the intact limb was significantly larger than that for the prosthetic limb and the able-bodied limb (Table 2, Figure 2). This finding supports with our hypothesis that the intact limb of the runners with transfemoral or knee-disarticulation amputation generates more joint power when compared to the prosthetic limb and the limbs of age-matched able-bodied individuals at particular phases of the gait cycle while walking. While individuals with transfemoral amputation generally rely on their intact limb to maintain stability while walking (Kendell et al., 2016), the study showed that amputee runners were no exception. The findings suggest that to compensate for the anatomical functions on the prosthetic limb, the individuals rely on the intact limb by generating more joint power at the hip joint in stance while walking.

This study also found that the peak ankle plantarflexor moment from terminal stance to pre-swing [AM2] of the intact limb was significantly larger than the prosthetic limb, while the peak hip extensor moment from initial contact to loading response [HM1] of the intact limb was significantly larger than the prosthetic limb and the able-bodied limb (Table 2, Figure 2). As the joint power is a dot product of joint moment and joint angular velocity, the significantly larger ankle and hip positive power may be attributed to significantly larger ankle plantarflexor and hip extensor moments in the intact limb because ankle and hip joints moment and power appeared to demonstrate similar trends in their plots (Winter, 2005). Thus, ankle plantarflexors and hip extensors in the intact limb can be regarded as key muscles that compensate for the missing anatomical muscles and joint functions in the amputated limb during walking.

The peak hip flexor moment from terminal stance to pre-swing [HM2] was significantly larger in the prosthetic limb than that in the intact limb and the able-bodied limb. Moreover, the peak hip positive power at pre-swing [HP3] in the prosthetic limb was significantly larger than that in the intact limb (Table 2, Figure 2). This suggests that hip flexors in the prosthetic limb play an important role in the swing movement while walking. Based on the anthropometric data, an average mass of foot and shank of biological leg of the participants with amputation was 3.66 kg (0.061 x 60 kg)(Winter, 2005). The mass of the transfemoral prosthesis is generally around 3 to 4 kg (Meikle et al., 2003). Thus it appears that the hip flexors in the prosthetic limb require greater efforts despite mass of the prosthetic leg and biological limbs are close. This might be attributed to required efforts by the hip flexors to secure sufficient toe clearance during swing.

The results also showed that the peak ankle plantarflexion angle from pre-swing to initial swing [AA3] in the intact limb was significantly larger than that in the able-bodied limb (Table 2, Figure 2). This may be attributed to the common strategy adopted by individuals with transfemoral amputation known as vaulting. Vaulting is defined as plantarflexion of the intact ankle during single-limb support phase to assist toe clearance of the prosthetic limb (Drevelle et al., 2014). Another explanation could be that the intact ankle needed to produce larger plantarflexion moments and power generation for push-off in order to maintain a desired speed. Figure 1 shows that the slope of ankle moment (i.e., increase in plantarflexor moment) has a plateau in mid-stance in the intact limb, while the slope for the prosthetic limb and able-bodied limb is relatively constant. This can be attributed to the limited amount of dorsiflexion, which would give a sensation of walking uphill in the intact limb while walking (Bowker et al., 2002).

The results of this study also showed that the peak knee negative power at pre-swing [KP3] in the prosthetic limb was significantly larger than that in the intact limb and the able-bodied limb (Table 2, Figure 2). This demonstrates that, while the knee is flexing, the dampers of the prosthetic knee absorb significantly more energy when compared to anatomical knees from pre-swing to initial swing. The peak hip negative power at terminal stance [HP2] in the prosthetic limb was also significantly larger than that in the intact limb (Table 2, Figure 2). This finding could be explained by the significantly larger hip flexor moment from terminal stance to pre-swing [HM2] during the extension of the hip joint in the prosthetic limb. This may be attributed to eccentric contraction of the hip flexor muscles to stabilize the prosthetic limb during stance phase.

The results of this study were indirectly compared against the results of the study conducted by (Segal et al., 2006), wherein the kinematic and kinetic parameters of a general population of individuals with transfemoral amputation were studied. The comparison shows that the runners in our study demonstrated much larger peak hip positive power from loading response to mid stance [HP1] in the intact limb [2.95 (0.57) W/kg] and in the prosthetic limb [1.53 (0.68) W/kg] at a walking speed of 1.36 (0.17) m/s (Table 2). In the previous study, the HP1 in the intact limb was 0.57 (0.4) W/kg for C-Leg, and the HP1 in the prosthetic limb was 0.46 (0.1) W/kg for C-Leg at a walking speed of 1.20 – 1.30 m/s. The runners also demonstrated much larger peak hip positive power at pre-swing [HP3] in the prosthetic limb [2.28 (1.04) W/kg] when compared to the participants of the previous study [0.75 (0.3) W/kg for C-Leg]. Furthermore, the peak ankle positive power at pre-swing [AP2] in the intact limb [4.00 (1.19) W/kg] of the runners was also larger than the participants in the previous study [3.17 (0.5) W/kg for

C-Leg]. Although the gait models in the two studies were different and their statistical comparison was impossible, the walking speed that might affect these kinetic parameters was close to each other (Lelas et al., 2003). Therefore, in comparison to the general population of individuals with transfemoral amputation, the runners recruited in the present study were suggested to generate more power, particularly at the hip joint from loading response to mid-stance in the intact limb and at pre-swing in the prosthetic limb.

It is widely believed that in individuals with amputation, hip strength is weaker in the residual limb than in the intact limb (Ryser et al., 1988). However, hip strength of the residual limb in active individuals with transtibial amputation is reported to be stronger than able-bodied individuals (Nolan, 2009). A previous study also demonstrated that the training program to improve hip strength was important for enabling lower limb amputees to acquire the ability to run after amputation (Nolan, 2012). In addition, intense hip abductor strength training was reported to improve the functional performance of individuals with transfemoral amputation (Pauley et al., 2014). Therefore, the present study as well as other studies in this domain highlight the importance of hip strength for walking among active individuals with amputation.

There are certain considerations that must be acknowledged when interpreting the results of this study. First, the set of study participants (runners with amputation) was not homogeneous in terms of running ability. Their maximum running speed ranged from 4.4 m/s to 7.6 m/s and exercise frequency ranged from 1 day/week to 6 days/week. Therefore, some of the participants can be regarded as recreational runners while the others are competitive athlete runners. Second, the participants used their own prostheses for data collection, and the prosthetic knees and feet used in this study were not controlled. Moreover, their time from amputation ranged widely, from 3 years to 29 years. All being considered, this study will serve as foundation for future studies investigating the kinematics and kinetics of walking in runners with amputations.

Conclusions

This study demonstrated that the amputee runners with transfemoral or knee-disarticulation prostheses rely on their intact limb while walking. The intact limb of the amputee runners demonstrated significantly larger peak ankle positive power at pre-swing and peak hip positive power at loading response than the prosthetic limb. They appeared to generate more joint power by producing more ankle plantarflexor and hip extensor moments while walking. In addition, a comparison to a previous study suggested that runners with amputation might generate larger ankle and hip joints power than a general population of individuals with transfemoral amputation.

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Conflict of interest

None of the authors have any conflicts of interest associated with this study.

References

Bell, J.C., Wolf, E.J., Schnall, B.L., Tis, J.E., Tis, L.L., Potter, B.K., 2013. Transfemoral amputations: the effect of residual limb length and orientation on gait analysis outcome measures. J Bone Joint Surg Am 95, 408-414.

Bowker, J.H., Michael, J.W., American Academy of Orthopaedic Surgeons., 2002. Atlas of limb prosthetics : surgical, prosthetic, and rehabilitation principles, 2nd ed. Mosby Year Book, St. Louis.

Bragaru, M., Dekker, R., Geertzen, J.H., Dijkstra, P.U., 2011. Amputees and sports: a systematic review. Sports Med 41, 721-740.

Creylman, V., Knippels, I., Janssen, P., Biesbrouck, E., Lechler, K., Peeraer, L., 2016. Assessment of transfemoral amputees using a passive microprocessor-controlled knee versus an active powered microprocessor-controlled knee for level walking. Biomed Eng Online 15, 142.

Drevelle, X., Villa, C., Bonnet, X., Loiret, I., Fode, P., Pillet, H., 2014. Vaulting quantification during level walking of transfemoral amputees. Clin Biomech (Bristol, Avon) 29, 679-683.

Eberly, V.J., Mulroy, S.J., Gronley, J.K., Perry, J., Yule, W.J., Burnfield, J.M., 2014. Impact of a stance phase microprocessor-controlled knee prosthesis on level walking in lower functioning individuals with a transfemoral amputation. Prosthet Orthot Int 38, 447-455.

Graham, L.E., Datta, D., Heller, B., Howitt, J., Pros, D., 2007. A comparative study of conventional and energy-storing prosthetic feet in high-functioning transfemoral amputees. Arch Phys Med Rehabil 88, 801-806.

Jarvis, H.L., Bennett, A.N., Twiste, M., Phillip, R.D., Etherington, J., Baker, R., 2017. Temporal Spatial and Metabolic Measures of Walking in Highly Functional Individuals With Lower Limb Amputations. Arch Phys Med Rehabil 98, 1389-1399.

Kadaba, M.P., Ramakrishnan, H.K., Wootten, M.E., 1990. Measurement of lower extremity kinematics during level walking. Journal of orthopaedic research : official publication of the Orthopaedic Research Society 8, 383-392.

Kaufman, K.R., Frittoli, S., Frigo, C.A., 2012. Gait asymmetry of transfemoral amputees using mechanical and microprocessor-controlled prosthetic knees. Clin Biomech (Bristol, Avon) 27, 460-465.

Kendell, C., Lemaire, E.D., Kofman, J., Dudek, N., 2016. Gait adaptations of transfemoral prosthesis users across multiple walking tasks. Prosthet Orthot Int 40, 89-95.

Lelas, J.L., Merriman, G.J., Riley, P.O., Kerrigan, D.C., 2003. Predicting peak kinematic and kinetic parameters from gait speed. Gait Posture 17, 106-112.

Mahon, C.E., Pruziner, A.L., Hendershot, B.D., Wolf, E.J., Darter, B.J., Foreman, K.B., Webster, J.B., 2017. Gait and Functional Outcomes for Young, Active Males With Traumatic Unilateral Transfemoral Limb Loss. Mil Med 182, e1913-e1923.

Makimoto, A., Sano, Y., Hashizume, S., Murai, A., Kobayashi, Y., Takemura, H., Hobara, H., 2017. Ground Reaction Forces During Sprinting in Unilateral Transfemoral Amputees. J Appl Biomech 33, 406-409.

Meikle, B., Boulias, C., Pauley, T., Devlin, M., 2003. Does increased prosthetic weight affect gait speed and patient preference in dysvascular transfemoral amputees? Arch Phys Med Rehabil 84, 1657-1661.

Mengelkoch, L.J., Kahle, J.T., Highsmith, M.J., 2017. Energy costs and performance of transfemoral amputees and non-amputees during walking and running: A pilot study.

Prosthet Orthot Int 41, 484-491.
Nolan, L., 2008. Carbon fibre prostheses and running in amputees: a review. Foot Ankle Surg 14, 125-129.
Nolan, L., 2009. Lower limb strength in sports-active transtibial amputees. Prosthet Orthot Int 33, 230-241.
Nolan, L., 2012. A training programme to improve hip strength in persons with lower limb amputation. Journal of rehabilitation medicine 44, 241-248.
Pauley, T., Devlin, M., Madan-Sharma, P., 2014. A single-blind, cross-over trial of hip abductor strength training to improve Timed Up & Go performance in patients with unilateral, transfemoral amputation. Journal of rehabilitation medicine 46, 264-270.
Perry, J., Burnfield, J.M., 2010. Gait analysis : normal and pathological function, 2nd ed. SLACK, Thorofare, NJ.
Russell Esposito, E., Aldridge Whitehead, J.M., Wilken, J.M., 2015. Sound limb loading in individuals with unilateral transfemoral amputation across a range of walking velocities. Clin Biomech (Bristol, Avon) 30, 1049-1055.
Ryser, D.K., Erickson, R.P., Cahalan, T., 1988. Isometric and isokinetic hip abductor strength in persons with above-knee amputations. Arch Phys Med Rehabil 69, 840-845.
Sano, Y., Makimoto, A., Hashizume, S., Murai, A., Kobayashi, Y., Takemura, H., Hobara, H., 2017. Leg stiffness during sprinting in transfemoral amputees with running-specific prosthesis. Gait Posture 56, 65-67.
Schaarschmidt, M., Lipfert, S.W., Meier-Gratz, C., Scholle, H.C., Seyfarth, A., 2012. Functional gait asymmetry of unilateral transfemoral amputees. Human movement science 31, 907-917.
Segal, A.D., Orendurff, M.S., Klute, G.K., McDowell, M.L., Pecoraro, J.A., Shofer, J., Czerniecki, J.M., 2006. Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg and Mauch SNS prosthetic knees. J Rehabil Res Dev 43, 857-870.
Shojaei, I., Hendershot, B.D., Wolf, E.J., Bazrgari, B., 2016. Persons with unilateral transfemoral amputation experience larger spinal loads during level-ground walking compared to able-bodied individuals. Clin Biomech (Bristol, Avon) 32, 157-163.
Sions, J.M., Beisheim, E.H., Manal, T.J., Smith, S.C., Horne, J.R., Sarlo, F.B., 2018. Differences in Physical Performance Measures Among Patients With Unilateral Lower-Limb Amputations Classified as Functional Level K3 Versus K4. Arch Phys Med

Rehabil 99, 1333-1341.

Traballesi, M., Delussu, A.S., Averna, T., Pellegrini, R., Paradisi, F., Brunelli, S., 2011. Energy cost of walking in transfemoral amputees: Comparison between Marlo Anatomical Socket and Ischial Containment Socket. Gait Posture 34, 270-274.

Winter, D.A., 1987. The biomechanics and motor control of human gait, 1st ed. ed. University of Waterloo Press, Waterloo, Ontario, Canada.

Winter, D.A., 2005. Biomechanics and motor control of human movement, 3rd ed ed. Wiley, Hoboken, New Jersey.

Captions for Tables

Table 1. Demographic data of the runners with amputation.

Table 2. Spatiotemporal, kinematic and kinetic parameters of gait in the intact limb, the prosthetic limb, and the able-bodied limb and outcome of statistical analyses.

Captions for Figures

Figure 1. Comparison of the kinematics and kinetics of the intact limb (blue and solid), the prosthetic limb (red and dash) and the able-bodied limb (green and dot). Plot of their mean and standard deviation of the able-bodied limb in the angle, moment and power for ankle, knee and hip joints as percentage of a gait cycle, where 0% is initial contact and 100% is subsequent initial contact.

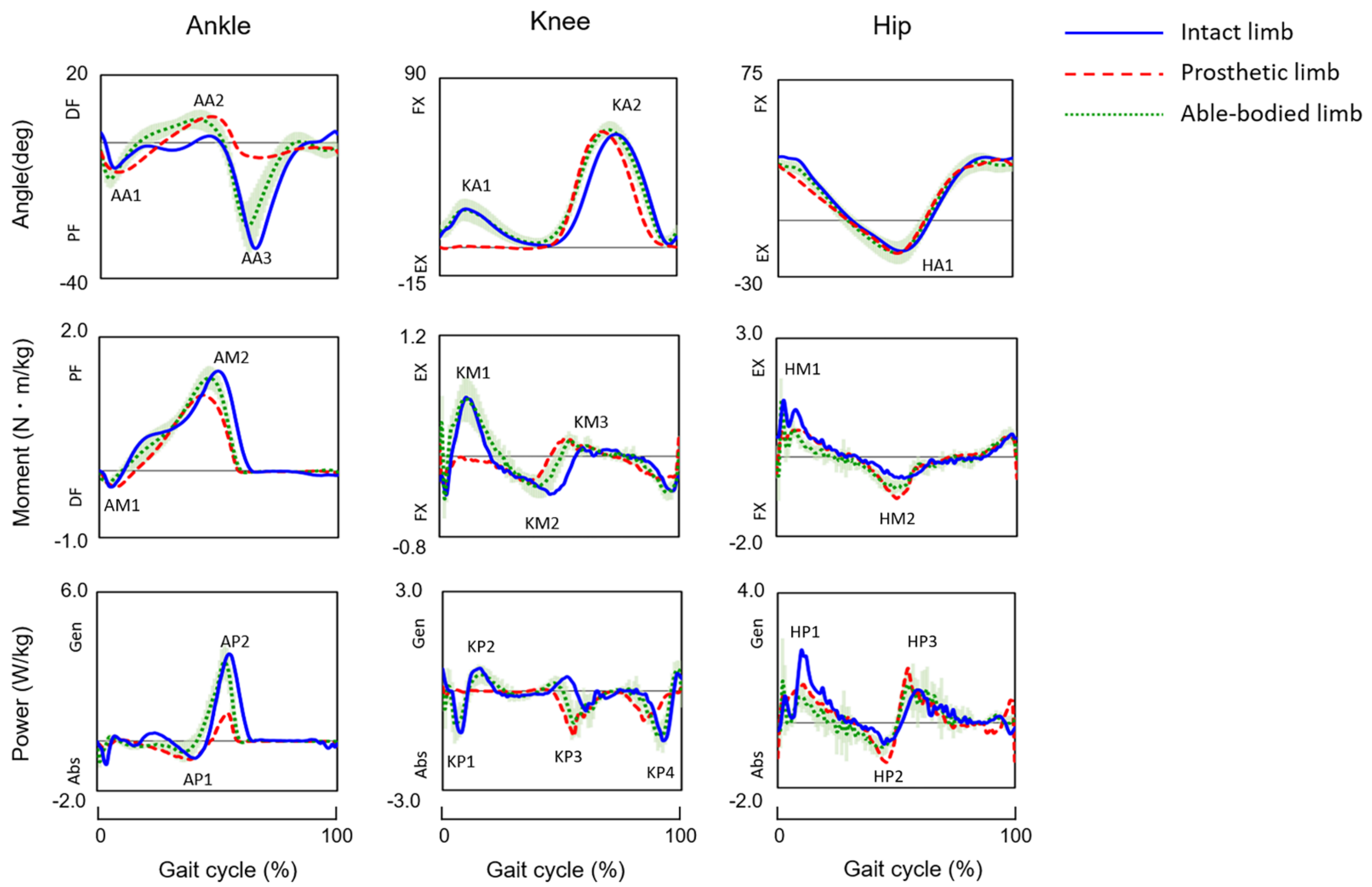
Abbreviations for the kinematic parameters: AA1, peak ankle plantarflexion angle at loading response; AA2, peak ankle dorsiflexion angle from terminal stance to pre-swing; AA3, peak ankle plantarflexion angle from pre-swing to initial swing, KA1, peak knee flexion from loading response to mid stance; KA2, peak knee flexion from initial swing to mid swing; HA1, peak hip extension angle from terminal stance to pre-swing.

Abbreviations for the kinetic parameters: AM1, peak ankle dorsiflexor moment at loading response; AM2, peak ankle plantarflexor moment from terminal stance to pre-swing; AP1, peak ankle negative power at terminal stance; AP2, peak ankle positive power at pre-swing; KM1, peak knee extensor moment from loading response to mid stance; KM2, peak knee flexor moment at terminal stance; KM3, peak knee extensor moment at pre-swing; KP1, peak knee negative power at loading response; KP2, peak knee positive power at mid stance; KP3, peak knee negative power at pre-swing; KP4, peak knee negative power from mid swing to terminal-swing; HM1, peak hip extensor moment from initial contact to loading response; HM2, peak hip flexor moment from terminal stance to pre-swing; HP1, peak hip positive power from loading response to mid stance; HP2, peak hip negative power at terminal stance; HP3, peak hip positive power at pre-swing.

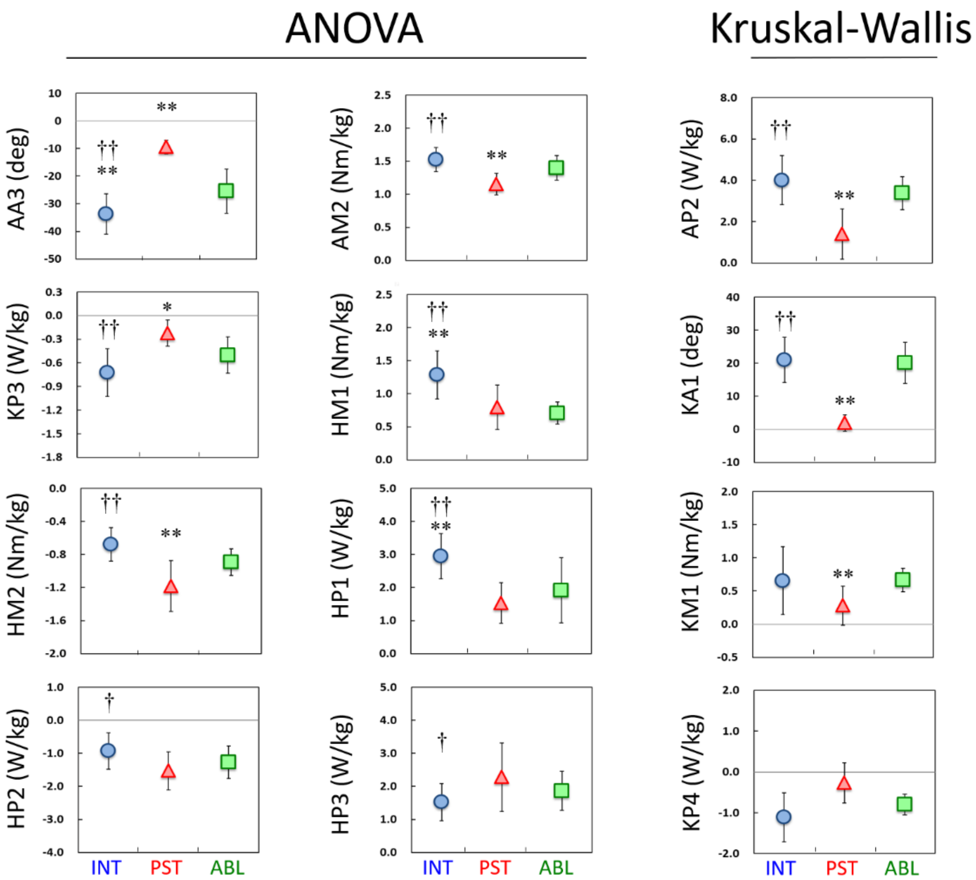
Figure 2. Post-hoc analyses of the kinematic and kinetic parameters of gait in the intact limb (blue, circle), the prosthetic limb (red, triangle) and the able-bodied limb (green, square), which demonstrated significant differences by ANOVA or Kruskal-Wallis test.

Participant	Gender	Amputated side	Etiology	Time since amputation (years)	Residual limb length	Maximum running speed (m/s)	Exercise frequency (days/week)	Prosthetic knee	Prosthetic foot
1	M	L	Trauma	3	Long TF	5.1	1	3R95	Variflex
2	M	L	Cancer	4	Long TF	5.7	1	3R106	Triton
3	M	L	Trauma	29	KD	6.0	6	4-bar intelligent knee	Lapoc J-foot
4	M	R	Trauma	28	Short TF	4.5	2	3R80	Triton
5	F	R	Infection	15	Long TF	5.6	4	4-bar intelligent knee	Total concept
6	M	L	Cancer	18	Short TF	5.3	3	3R80	Variflex
7	M	L	Trauma	16	Middle TF	5.9	4	3R80	Variflex
8	F	R	Cancer	8	Short TF	4.4	5	3R106	1H38
9	M	L	Trauma	16	Long TF	7.1	5	3R95	Lapoc J-foot
10	F	R	Trauma	7	Middle TF	5.6	5	Total knee	Total concept
11	F	R	Trauma	4	Middle TF	5.9	5	Total knee	Variflex xc
12	M	L	Cancer	24	KD	7.6	6	Genium Xtreme	1D35
13	M	R	Trauma	5	Middle TF	6.8	5	3R80	Reflex rotate
14	M	L	Cancer	21	Middle TF	5.4	2	3R95	Variflex

Gait parameters	Intact limb		Prosthetic limb		Able-bodied limb		*t*-test	ANOVA *p*-Value	Kruskal-Wallis	Post Hoc Test *p*-Value		
	Mean (SD)	95% CI	Mean (SD)	95% CI	Mean (SD)	95% CI				INT vs. PST	INT vs. ABL	PST vs. ABL
Walking speed (m/s)	1.36 (0.17)	1.20, 1.53	–	–	1.36 (0.17)	1.20, 1.53	1.000	—	—	—	—	—
Gait cycle time (s)	1.04 (0.05)	1.00, 1.09	–	–	1.01 (0.07)	0.94, 1.09	0.215	—	—	—	—	—
Cadence (steps/min)	115 (6)	110, 121	–	–	119 (8)	111, 127	0.184	—	—	—	—	—
Stride length (m)	1.41 (0.14)	1.27, 1.55	–	–	1.36 (0.10)	1.26, 1.46	0.308	—	—	—	—	—
Step width (m)	0.11 (0.03)	0.08, 0.14	–	–	0.03 (0.02)	0.57, 0.78	< 0.001	—	—	—	—	—
Step length (m)	0.68 (0.08)	0.60, 0.76	0.73 (0.06)	0.67, 0.79	0.68 (0.05)	0.63, 0.73	—	0.060	—	—	—	—
Stance time (%)	56.47 (1.70)	54.78, 58.17	61.94 (1.08)	60.67, 63.21	58.86 (1.08)	57.78, 59.94	—	< 0.001	—	< 0.001	< 0.001	< 0.001
Double support time (%)	8.74 (1.14)	7.60, 9.88	9.95 (3.28)	6.68, 13.23	8.50 (0.79)	7.71, 9.29	—	—	0.279	—	—	—
AA1 (°)	−9.58 (4.16)	−17.74, −1.43	−9.05 (2.13)	−13.22, −4.88	−10.92 (3.17)	−17.14, −4.71	—	0.303	—	—	—	—
AA2 (°)	7.1 (4.27)	−1.27, 15.47	8.12 (2.39)	3.43, 12.82	7.45 (2.37)	2.8, 12.1	—	0.685	—	—	—	—
AA3 (°)	−33.73 (7.29)	−48.02, −19.43	−9.44 (2.46)	−14.26, −4.63	−25.43 (7.96)	−41.04, −9.83	—	< 0.001	—	< 0.001	0.004	< 0.001
AM1 (Nm/kg)	−0.27 (0.18)	−0.62, 0.07	−0.25 (0.1)	−0.44, −0.07	−0.2 (0.06)	−0.33, −0.08	—	0.302	—	—	—	—
AM2 (Nm/kg)	1.53 (0.18)	1.17, 1.89	1.15 (0.16)	0.83, 1.47	1.4 (0.19)	1.03, 1.76	—	< 0.001	—	< 0.001	0.181	0.002
AP1 (W/kg)	−0.81 (0.27)	−1.33, −0.28	−0.87 (0.33)	−1.52, −0.23	−0.64 (0.32)	−1.27, −0.01	—	0.139	—	—	—	—
AP2 (W/kg)	4 (1.19)	1.67, 6.33	1.4 (1.21)	−0.98, 3.78	3.37 (0.79)	1.82, 4.93	—	—	< 0.001	< 0.001	0.644	0.001
KA1 (°)	21.03 (6.86)	7.58, 34.48	1.89 (2.51)	−3.04, 6.82	20.14 (6.22)	7.94, 32.33	—	—	< 0.001	< 0.001	1.000	< 0.001
KA2 (°)	60.72 (6.53)	47.91, 73.52	62.52 (8.65)	45.56, 79.47	62.91 (4.5)	54.08, 71.73	—	0.662	—	—	—	—
KM1 (Nm/kg)	0.66 (0.51)	−0.35, 1.66	0.28 (0.29)	−0.3, 0.85	0.67 (0.18)	0.32, 1.01	—	—	0.002	0.051	1.000	0.001
KM2 (Nm/kg)	−0.44 (0.2)	−0.82, −0.06	−0.31 (0.14)	−0.58, −0.04	−0.34 (0.1)	−0.53, −0.15	—	0.071	—	—	—	—
KM3 (Nm/kg)	0.18 (0.12)	−0.05, 0.42	0.36 (0.37)	−0.35, 1.08	0.24 (0.1)	0.04, 0.45	—	—	0.062	—	—	—
KP1 (W/kg)	−1.31 (0.58)	−2.44, −0.18	—	—	−0.73 (0.3)	−1.32, −0.15	0.004	—	—	—	—	—
KP2 (W/kg)	1.15 (1.01)	−0.84, 3.14	—	—	0.51 (0.34)	−0.16, 1.18	0.040	—	—	—	—	—
KP3 (W/kg)	−0.72 (0.3)	−1.32, −0.13	−0.22 (0.17)	−0.55, 0.1	−0.5 (0.23)	−0.96, −0.04	—	< 0.001	—	< 0.001	0.059	0.012
KP4 (W/kg)	−1.11 (0.6)	−2.29, 0.06	−0.27 (0.49)	−1.23, 0.69	−0.8 (0.25)	−1.29, −0.3	—	—	< 0.001	0.104	0.179	1.000
HA1 (°)	−16.6 (4.13)	−24.69, −8.5	−18.15 (4.83)	−27.62, −8.68	−17.33 (6.39)	−29.85, −4.82	—	0.734	—	—	—	—
HM1 (Nm/kg)	1.28 (0.36)	0.57, 2	0.79 (0.34)	0.13, 1.46	0.71 (0.16)	0.39, 1.03	—	< 0.001	—	< 0.001	< 0.001	1.000
HM2 (Nm/kg)	−0.68 (0.2)	−1.08, −0.28	−1.18 (0.31)	−1.79, −0.58	−0.89 (0.16)	−1.21, −0.58	—	< 0.001	—	< 0.001	0.060	0.006
HP1 (W/kg)	2.95 (0.68)	1.61, 4.29	1.53 (0.61)	0.33, 2.74	1.91 (0.99)	−0.02, 3.84	—	< 0.001	—	< 0.001	0.003	0.614
HP2 (W/kg)	−0.93 (0.55)	−2, 0.14	−1.53 (0.58)	−2.66, −0.4	−1.27 (0.49)	−2.23, −0.3	—	0.020	—	0.016	0.317	0.620
HP3 (W/kg)	1.52 (0.57)	0.41, 2.63	2.28 (1.04)	0.25, 4.31	1.86 (0.59)	0.7, 3.03	—	0.042	—	0.037	0.725	0.479



Ankle
Knee
Hip
Intact limb
Prosthetic limb
Able-bodied limb
Angle(deg)
Moment (N · m/kg)
Power (W/kg)
20
-40
DF
PF
AA1
AA2
AA3
90
-15
FX
EX
KA1
KA2
75
-30
HA1
2.0
-1.0
AM1
AM2
1.2
-0.8
KM1
KM2
KM3
3.0
-2.0
HM1
HM2
6.0
-2.0
Gen
Abs
AP1
AP2
3.0
-3.0
KP1
KP2
KP3
KP4
4.0
-2.0
HP1
HP2
HP3
0
100
Gait cycle (%)



(*) Significant difference from ABL at $P < 0.05$ (†) Significant difference from PST at $P < 0.05$

(**) Significant difference from ABL at $P < 0.01$ (††) Significant difference from PST at $P < 0.01$

Highlights

- Walking characteristics of runners with a transfemoral amputation was analyzed.
- The intact limb generated more plantarflexor and hip extensor moment.
- The intact limb generated more positive ankle and hip joint power.
- Runners with a transfemoral amputation rely on the intact limb while walking.