Gait & Posture 73 (2019) 8–13

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

Joint power distribution does not change within the contralateral limb one year after unilateral limb loss

Courtney M. Butowicz[a], Christopher L. Dearth[a,b,c], Brad D. Hendershot[a,b,d,*]

[a] Research and Development Section, Department of Rehabilitation, Walter Reed National Military Medical Center, Bethesda, MD, USA
[b] DoD-VA Extremity Trauma and Amputation Center of Excellence, USA
[c] Department of Surgery, Uniformed Services University of the Health Sciences & Walter Reed National Military Medical Center, Bethesda, MD, USA
[d] Department of Rehabilitation Medicine, Uniformed Services University of the Health Sciences, Bethesda, MD, USA

ARTICLE INFO

Keywords:
Biomechanics
Gait analysis
Longitudinal
Kinetics

ABSTRACT

Background: To assist with forward progression during gait, persons with unilateral lower-limb amputation typically perform more work within the unaffected versus affected limb. However, prior cross-sectional (> 2years post-amputation) studies cannot necessarily elucidate the origin or evolution of these compensatory mechanics.

Research question: Do lower limb joint kinetics change during the initial stages of independent ambulation among persons with lower-limb amputation?

Methods: Nine males with unilateral lower-limb amputation (6 transtibial; 3 transfemoral) completed instrumented gait analyses (speed = 1.2 m/s) at 2 and 12-months post-independent ambulation. Within the unaffected limb, sagittal and frontal plane total positive and negative work, peak power, average positive power, and percent contribution of each joint were compared between time points using paired t-tests.

Results: No differences existed between time points in total positive or negative work, at any joint ($p > 0.038$) in either plane. Similarly, there were no differences in percent contribution by each joint to total average power by sagittal ($p > 0.15$) or frontal ($p > 0.32$) planes.

Significance: Persons with unilateral lower-limb amputation do not alter power distribution among joints within the unaffected limb during initial independent ambulation. However, compared to previous cross-sectional reports, smaller peak powers in the unaffected hip and knee here suggest mechanical work increases with time since amputation. Future research should longitudinally monitor segment mechanics to determine when deleterious strategies develop, as these have implications for joint degeneration and pain.

1. Introduction

By the year 2050, an estimated 3.5 million individuals in the United States alone will be living with an amputation [1]. There is also a growing cohort of approximately 2100 US Service Members living with amputations resulting from traumatic injuries sustained during recent conflicts and other non-combat activities [2]. Among these individuals, particularly those with a lower-limb amputation (LLA), there is a considerably high risk for developing deleterious musculoskeletal conditions (i.e., low back pain, knee osteoarthritis) secondary to amputation [3,4]. This is especially concerning for Service Members with LLA, who are relatively young at the time of injury, and thus will likely experience decades of pain and functional limitations. Therefore, identifying risk factors for these conditions early in the rehabilitation process (e.g., increased joint loading or increased muscular work during gait) can serve as an indicator of aberrant motor strategies. Identifying these strategies early allows for the implementation of interventions that can mitigate or prevent deleterious effects of repeated exposure to altered mechanics over time.

From prior cross-sectional studies, persons with unilateral LLA regularly performed activities of daily living with both preferential use of the unaffected limb and large proximal motions [5–8]. During gait specifically, these altered movement strategies are characterized by greater unaffected limb loading (including joint moments and powers), and increased spinal loads [9–11]. Compared to uninjured populations, persons with LLA performed significantly more (sagittal plane) concentric mechanical work in the unaffected hip and knee joints during the stance phase of gait, this phenomenon exacerbated at faster walking speeds, which is thought to be a compensatory strategy to aid forward progression and account for absent and altered musculature in the

* Corresponding author at: 4954 N. Palmer Road, America Building (19), Room B-320, Bethesda, MD, 20889, USA.
E-mail address: bradford.d.hendershot2.civ@mail.mil (B.D. Hendershot).

https://doi.org/10.1016/j.gaitpost.2019.06.011
Received 19 February 2019; Received in revised form 11 June 2019; Accepted 18 June 2019
0966-6362/ Published by Elsevier B.V.

affected limb [12]. Moreover, frontal plane hip power during single-limb support is considered the primary mechanism of maintaining dynamic stability, and is reduced in the unaffected limb of persons with LLA, thereby increasing the reliance on other joints or segments (i.e., trunk) to maintain mediolateral stability [13]. Together, these cross-sectional reports suggest that, over time, persons with LLA develop and utilize compensatory strategies which includes increased work within the unaffected limb and trunk. However, the origin and evolution of such changes remain largely unknown, considering the time elapsed since amputation in prior studies is at minimum two years [11,14], and in some cases greater than seven years [6,8,9].

Longitudinal evaluation of gait mechanics (e.g., joint powers and work), particularly during the initial stages (i.e., first year) of independent ambulation, can identify when such compensatory strategies originate, and therefore provide insight into the time points at which targeted mitigation intervention(s) is most critical. The purpose of this study was to evaluate joint powers and mechanical work in the unaffected limb of persons with unilateral LLA during the first year of independent ambulation. It was hypothesized that the overall (i.e., summed magnitude of all joints) joint power would not change over time, rather a redistribution of power and work among joints within the unaffected limb [12]. Specifically, there would be a shift in power and work contributions from the more distal structures (i.e., ankle) to the more proximal structures (i.e., hip) between early and late stages of independent ambulation, as individuals accommodate to walking with a prosthesis.

2. Methods

2.1. Participants

An *a priori* power analysis determined that 8–10 subjects were required to detect differences in hip and knee power with effect sizes of at least d = 0.8 as statistically significant ($\alpha = 0.05$, $\beta = 0.20$). This effect size was calculated using data from Silverman et al [12], comparing the hip power from the intact limb of persons with LLA to non-LLA controls. Data from nine males with traumatic unilateral LLA (six transtibial, three transfemoral) receiving standard-of-care rehabilitation at Walter Reed Army Medical Center were retrospectively analyzed (Table 1). Participants were excluded if they presented with: pain during activities of daily living (> 4 out of 10), bilateral lower limb amputations, history of traumatic brain injury, or the inability to walk without the use of an assistive device (other than the prosthesis). All participants provided written informed consent to procedures approved by the Walter Reed Army Medical Center Institutional Review Board.

2.2. Protocol

Gait analyses were repeated at two time points, 2- and 12-months, following initial independent ambulation after LLA (defined relative to the ability to first walk 50-ft without the use of an assistive device). These specific time points were intended to allow for initial learning effects of walking with a prosthesis and minimize any influence of residual limb pain during initial ambulation [15]. At both time points, participants wore shorts, their normal athletic shoes, prescribed energy storage and return ankle prostheses, and unpowered microprocessor knee prostheses (transfemoral LLA only). Participants walked along a 15 m level walkway at a forced speed, which was calculated using Froude number (0.4), a dimensionless speed based on lower limb length. Of note, this equated to 1.2 (± 5%) m/s for all nine participants [16]. Gait speed was enforced using an auditory tone ("beep"), signaling when speed fell within a ± 5% range of the target speed. Full-body kinematics were obtained by tracking (120 Hz) locations of 51 surface-markers (6DoF marker set) using a 23-camera motion capture system (Vicon, Oxford, UK) while ground reaction forces were simultaneously sampled (1200 Hz) from six force platforms embedded within the walkway (AMTI, Watertown, MA, USA).

2.3. Dependent measures and statistical analyses

Marker trajectories and ground reaction forces were low-pass filtered at 6 Hz and 50 Hz, respectively, with a dual-pass 2nd order Butterworth filter. Joint powers (normalized to body mass) from the ankle, knee, and hip of the unaffected limb were estimated using a three-dimensional linked-segment model and iterative Newton-Euler inverse dynamics in Visual 3D (C-Motion Inc., Germantown, MD, USA). Mechanical work (positive/negative) was calculated as the total area under the joint power curves, and respectively indicate mechanical energy generation/absorption of the unaffected limb. Segment inertial and anthropometric properties of the unaffected limb were calculated according to the regression equations of Dempster [17]. Data were time-normalized to stride (100% gait cycle), with a stride defined as subsequent heel strikes of the unaffected limb. To obtain information more closely related to the mechanical power during stance specifically, joint positive work values were divided by step time to calculate average positive mechanical joint power [18]. As positive power is required to accelerate the center of mass and the lower limbs during gait, the total average positive power output, and the contributions of each joint to the total output, illustrate the average (as opposed to instantaneous) power generated by the unaffected limb and the distribution among the joints during the stance phase of gait [18,19]. These values were then summed for each joint to determine power as a percentage of the total average power. Total positive and negative work and average positive mechanical joint power were compared between time points using paired t-tests. Temporal-spatial aspects of gait were included in the analysis to determine if they should be considered as covariates, as they influence joint kinetics. All statistical analyses were performed using SPSS (version 24; IBM Corp., Armonk, NY), with significance set at $p < 0.05$; though Bonferroni corrections were applied given multiple comparisons ($p < 0.008$ for work at each joint per plane and $p < 0.017$ for average positive mechanical joint power per plane). All values are reported as means ± standard deviations.

3. Results

3.1. Temporal-spatial characteristics

Between time points, there were no differences in stride width ($t = 0.78$, $p = 0.46$, $d = 0.14$), stride length ($t = 0.24$, $p = 0.82$, $d = 0.08$), or step time ($t = -1.34$, $p = 0.22$, $d = 0.19$) (data included as Supplementary Material). Therefore, temporal-spatial aspects of gait were not considered as covariates in analyses.

3.2. Joint work

In the sagittal plane, there were no differences between time points

Table 1
Mean (SD) participant characteristics. Level of amputation denoted as TTA: transtibial amputation, TFA: transfemoral amputation.

Level of Amputation	N	Age (yrs)	Stature (cm)	Body Mass (kg)	Time between1st prosthesis and 2-month visit (months)
TTA	6	30.8 (9.3)	178.2 (3.1)	80.9 (10.3)	3.7 (1.7)
TFA	3	28.7 (5.0)	174.7 (7.8)	92.5 (11.3)	3.0 (1.1)

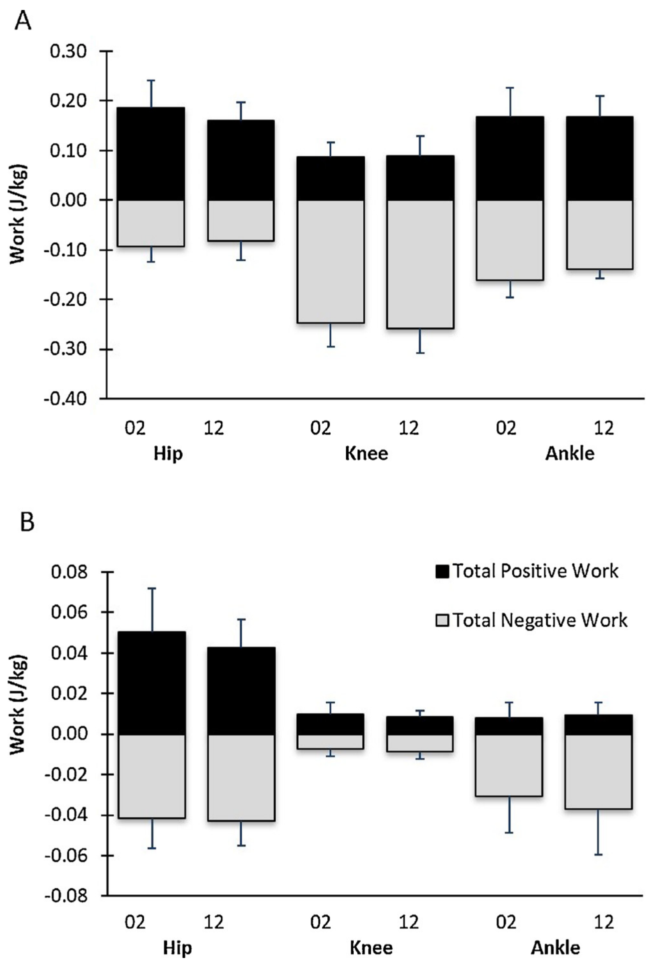

Fig. 1. Sagittal (A) and frontal (B) plane total positive and negative work at the hip, knee, and ankle at 2- and 12-months post-independent ambulation.

in total positive or negative work in the hip (Figs. 1a and 2a), knee (Figs. 1a, 2b), or ankle (Figs. 1a, 2c) ($p > 0.038$). Similarly, there were no differences ($p > 0.27$) in total positive or negative work (Fig. 1b) at any joint in the frontal plane (Figs. 2d-f).

3.3. Mean positive mechanical joint power

In general, magnitudes of instantaneous joint powers at the unaffected limb hip, knee, and ankle were similar throughout the entire stride between time points (Fig. 2). Many of these similarities were also reflected in the average positive total power in the sagittal plane ($t = 1.16$, $p = 0.28$, $d = 0.38$), average positive hip power ($t = 2.35$, $p = 0.05$, $d = 0.65$), average positive knee power ($t = 0.02$, $p = 0.98$), and average positive ankle power ($t = 0.16$, p = 0.87). In the sagittal plane, the percentage contribution of each unaffected lower limb joint to the total average power was not different between time points at the hip ($t = 1.60$, $p = 0.15$, $d = 0.41$), knee ($t = 0.08$, $p = 0.94$, $d = 0.02$), and ankle ($t = -1.24$, $p = 0.25$, $d = 0.33$) (Fig. 3).

The unaffected limb hip and ankle were the dominant sources of power output at both the 2-month (42% and 37%, respectively) and 12-month (39% and 40%, respectively) time points. Similarly, there were no differences in frontal plane average total positive power ($t = 1.03$, $p = 0.33$, $d = 0.63$), average positive hip power ($t = 1.36$, $p = 0.21$, $d = 0.79$), average positive knee power ($t = 1.30$, $p = 0.23$, $d = 0.38$), and average positive ankle power ($t = -0.33$, $p = 0.75$, $d = 0.20$). At both time points, the hip provided the largest contribution (≈ 73% and 70%, respectively) to frontal plane total positive power output, while the knee and ankle contributed noticeably less (knee ≈ 15% at both time points; ankle ≈ 12% and 15%, respectively). There were no differences in hip ($p = 0.32$, $d = 0.32$), knee ($p = 0.77$, $d = 0.07$), or ankle ($p = 0.32$, $d = 0.24$) percent contributions to total positive power output over time in the frontal plane (Fig. 4).

4. Discussion

This study compared joint powers and work within the unaffected limb of persons with unilateral LLA within the first year of independent ambulation post-amputation. There were no changes in instantaneous power or total work performed in the unaffected limb, nor a proximally-oriented redistribution of power, which is contrary to the study hypothesis. Furthermore, there were no longitudinal changes in the temporal-spatial characteristics of gait Table 1.

4.1. Instantaneous power and joint work

The shapes of the instantaneous joint power curves (sagittal plane), and the corresponding positive and negative joint work values, were similar at both time points (Figs. 2 and 3). The minimal differences in instantaneous sagittal plane peak powers between time points were less than published MDCs (Table 2 in Supplement) for uninjured adults walking at the same speed [20], further supporting the conclusion of no change over time. While previous cross-sectional comparisons suggest persons with LLA, compared to without LLA, favor use of the unaffected limb during gait [6,7,9]; here, instantaneous peak powers (sagittal plane) of the unaffected limb joints, regardless of level of amputation (Fig. 2), were smaller compared to previous cross-sectional reports of LLA joint powers [21,22]. The time since amputation in previous studies ranges from 1 to 23 years, potentially illustrating that persons with LLA increase the mechanical work done by the unaffected limb joints as time since amputation increases. Interestingly, the peak power generated at the unaffected ankle during push-off in the current cohort of persons with LLA was larger (2.5 vs. 1.3 W/kg) than a previously reported cohort walking at 1.2 m/s; however, the peak knee and hip powers of the current cohort were similar to the uninjured controls used in that study [9]. This is in contrast to prior cross-sectional studies, suggesting that the work performed at these joints may not elucidate mechanical factors associated with increased prevalence of hip and knee osteoarthritis among persons with LLA. This may be a compensatory strategy to account for decreased affected limb power (secondary to muscle atrophy) over time. Therefore, no changes in instantaneous power, total positive, and negative work were demonstrated between time points in the unaffected limb. It is possible that there were changes in affected limb gait mechanics between time points; however, this was not investigated as this was not an objective of the current study.

Features of gait mechanics within the sagittal plane are often of interest considering forward progression is the primary objective of bipedal gait; however, aberrant frontal plane mechanics are associated with postural and balance impairments and increased joint loading that can lead to falls and joint degeneration [11]. The instantaneous frontal plane joint powers, and corresponding total negative and positive work performed at each joint, were similar at both 2-months and 12-months (Fig. 2). While there are no published (to date) MDCs for frontal plane lower limb joint powers, the effect sizes of the differences in positive and negative work were considered small ($d < 0.5$), supporting the lack of statistical significance ($p > 0.27$). Similar to Sadeghi et al (2001), here, eccentric contraction of the hip abductors during midstance produces an absorptive energy, providing a mechanism to maintain stability during single-limb stance. [13] However, whereas Sadeghi (2001) observed a concurrent increase in hip extensor power and decrease in hip abductor power on the unaffected limb, the current data demonstrate persons with LLA, regardless of level of amputation, in the initial year of rehabilitation concurrently increase hip abductor and extensor powers (Fig. 2). An increased reliance on the hip to both propel and stabilize the trunk during single-limb stance signifies the

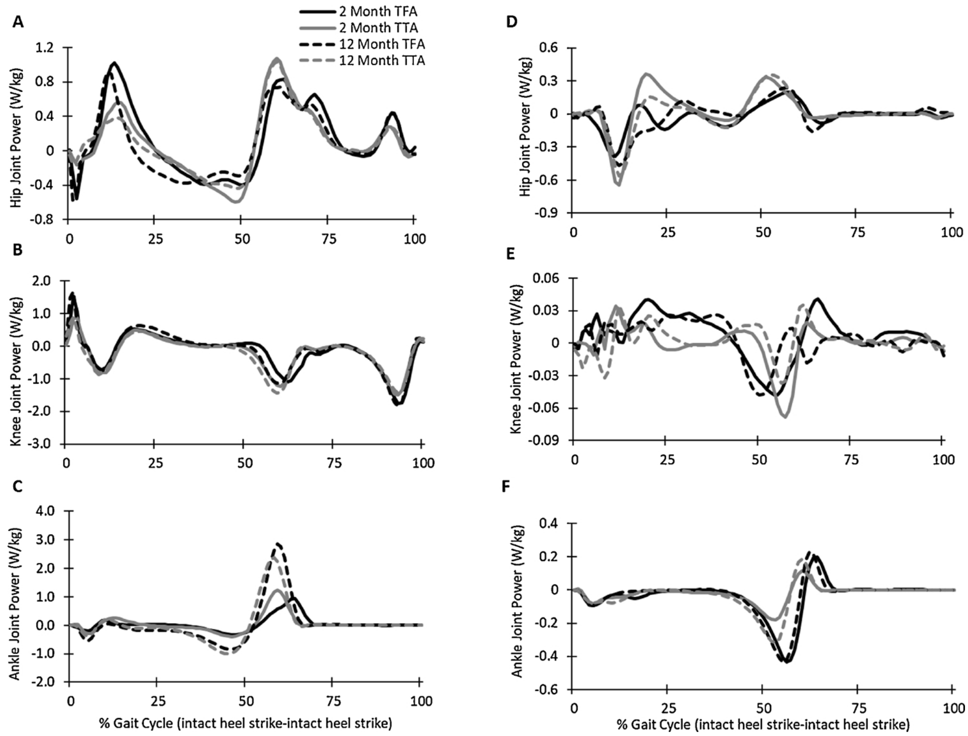

Fig. 2. Ensemble means of instantaneous powers of the: (A) hip, (B) knee, (C) ankle in the sagittal plane and ensemble averages of instantaneous powers of the: (D) hip, (E) knee, (F) ankle in the frontal plane. TTA: persons with transtibial lower limb amputation; TFA: persons with transfemoral lower limb amputation.

importance of bilateral hip extensor and abductor strengthening exercises within the first year of rehabilitation.

4.2. Average positive mechanical joint power

Persons with LLA compensate for altered or absent affected limb musculature by increasing power generation within the unaffected limb [9,13,21,23]. In both planes, the total average positive power output (i.e., sum of all three unaffected limb joints) minimally decreased during the first year of independent ambulation, possibly the result of a decreased reliance on the unaffected limb, and subsequent increased utilization of the affected limb, to generate the power required for forward progression of the center of mass. Considering it is likely there are multiple factors influencing accommodation to walking with a prosthesis, these results are perhaps not surprising given persons with transtibial LLA are able to increase positive work performed by an energy storage and return prosthesis over the course of three weeks [24]. Consistent with previous work in uninjured populations [18], the hip and ankle were the primary positive power sources, which is also consistent with forward-dynamic simulations that adduce the hip extensors and ankle plantar-flexors are driving positive muscle work during stance [18]. Contrary to our hypothesis, the relative contributions of the (sagittal) hip and ankle did not change over time, although, the moderate to large effect size (d = 0.79) of the difference between the percent contribution of the hip to the total average positive power output suggests a shift to a more proximal strategy may be occurring. This shift may not be completely surprising considering persons with LLA demonstrate greater hip extensor power generation compared to those without LLA during stance, posited as a compensatory strategy secondary to limited prosthetic ankle push-off [13]. Furthermore, the current findings suggest that persons with LLA utilize the knee to generate about 50% more power (i.e., positive work) than uninjured individuals walking at similar speeds (21% here compared to 14% as reported by Farris and Sawicki 2011) (Fig. 3). The lack of prosthetic

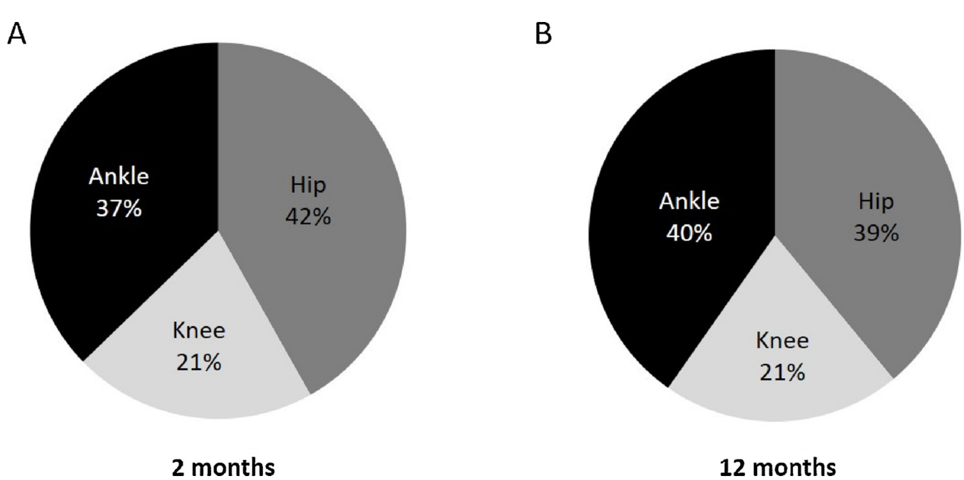

Fig. 3. Percent contribution of hip, knee, ankle to sagittal plane average positive power at (A) 2 and (B) 12-months post independent ambulation.

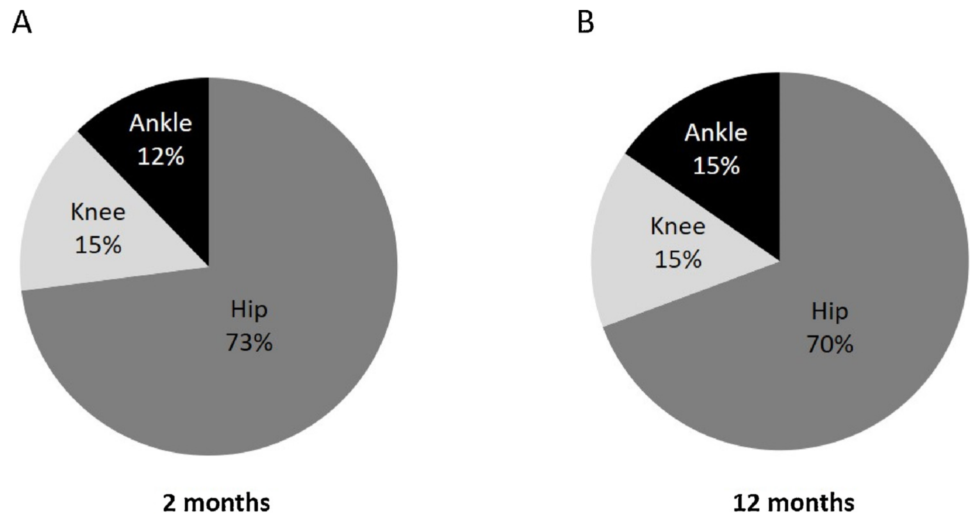

Fig. 4. Percent contribution of hip, knee, ankle to frontal plane average positive power at (A) 2 and (B) 12-months post independent ambulation.

ankle power leads to increased mechanical power demands at more proximal joints (i.e., knee and hip), which propagates to increased positive work demands on the unaffected limb. Thus, the lack of ankle power generation at both the affected ankle and prosthesis impacts workload of unaffected limb proximal musculature and should be considered a goal of early stage rehabilitation interventions.

Cross-sectional studies have previously associated altered or aberrant (unaffected) limb mechanics with joint degeneration and musculoskeletal pain conditions secondary to unilateral LLA [25,26]. The increased risk of developing knee osteoarthritis within this population is posited as the result of increased mechanical stress due to greater demands or loading of the unaffected limb during gait [3,25]. Features of the knee moment (i.e., peak or loading rate) are considered proxies for joint loading during gait, with larger features characteristic of knee osteoarthritis severity and progression; however, recent evidence suggest changes in knee flexion moment may precede onset of the condition, while changes in knee adduction moment may not occur until pain is present [27,28]. Here, the similarities in instantaneous power at the knee between two and twelve months may reflect similarities in knee moments (i.e., based on the power calculation used herein), which is consistent with a lack of longitudinal changes in knee moments between zero and six months post-independent ambulation among persons with LLA [28]. Thus, it appears that no longitudinal changes in knee joint work occur within the first year of walking with a prosthesis; however, it is possible that proximal changes (i.e., increased trunk motion) are occurring concurrently, thereby influencing unaffected limb joint dynamics [29,30]. Increased frontal plane trunk motion will decrease medial compartment knee joint loading, although this likely increases spinal loading and increases the risk for low back pain [29]. Therefore, alterations in both unaffected limb, trunk, and pelvis should be monitored regularly to determine if maladaptive strategies have been adopted, allowing for preventative or prophylactic intervention prior to joint degeneration.

4.3. Limitations

The longitudinal evaluation of gait strategies among Service Members with traumatic unilateral transtibial and transfemoral LLA is novel and allows for the generalization of the results to similar young, active populations; though caution is recommended, as the results of the current study may not be generalizable to persons with LLA due to causes other than trauma. The initial physical fitness levels and specialized and intensive rehabilitation of the individuals within this study also may be associated with the outcomes demonstrated after amputation and may not necessarily reflect outcomes in populations other than Service Members. The two and twelve month time points were chosen as the result of convenience sampling and to mitigate the potential of residual limb pain, which is most prevalent immediately after amputation, affecting gait during the first weeks of independent ambulation (note, the mean pain level reported at the two-month time point was 0.4 via visual analog scale) [15]. The general lack of differences between time points may be a function of small size and type II error; however, the moderate effect sizes suggest differences may exist. Moreover, while the study sample did not facilitate direct comparisons between levels of LLA (i.e., transtibial compared to transfemoral), considering the within-subjects design, it is not likely that inherent differences in gait mechanics between persons with transtibial and transfemoral LLA [9,14] affected the time point comparisons. Further, any influences of gait speed, and other temporal-spatial parameters, on joint powers were mitigated as participants were forced to walk at the same speed at both time points. Of note, rehabilitation protocols between time points were not necessarily standardized for all participants, yet, on average, all attended physical therapy sessions for two hours, three times per week.

5. Conclusions

The current results suggest that during the first year of ambulation, persons with unilateral LLA do not alter the distribution of power among joints within the unaffected limb. It is possible these individuals will maintain the current gait strategy until there is cause to alter it, such as the development of joint pain in the unaffected limb. Future research should nevertheless continue monitoring gait mechanics after amputation and with continued use of a prosthesis, considering the lower extremities and proximal segments (i.e., trunk and pelvis), and the collective effects of these strategies on joint health.

Acknowledgements

This work was supported, in part, by the Office of the Assistant Secretary of Defense for Health Affairs, through the Peer Reviewed Orthopaedic Research Program (award #W81XWH-14-2-0144), as well as the Military Amputee Research Program. The views expressed herein are those of the authors, and do not necessarily reflect the official policy or position of the U.S. Department of Defense, nor the U.S. government.

Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:https://doi.org/10.1016/j.gaitpost.2019.06.011.

References

[1] K. Ziegler-Graham, E.J. MacKenzie, P.L. Ephraim, T.G. Travison, R. Brookmeyer, Estimating the prevalence of limb loss in the United States: 2005 to 2050, Arch. Phys. Med. Rehabil. 89 (3) (2008) 422–429.

[2] EACE, Extremity Trauma and Amputation Center of Excellence (EACE) Registry, U.S. Army Medical Command, San Antonio, Texas, USA, 2018.

[3] R. Gailey, K. Allen, J. Castles, J. Kucharik, M. Roeder, Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use, J. Rehabil. Res. Dev. 45 (1) (2008) 15–29.

[4] S. Farrokhi, B. Mazzone, S. Eskridge, K. Shannon, O.T. Hill, Incidence of overuse musculoskeletal injuries in military service members with traumatic lower limb amputation, Arch. Phys. Med. Rehabil. 99 (2) (2018) 348–354 e341.

[5] B.D. Hendershot, M.A. Nussbaum, Altered flexion-relaxation responses exist during asymmetric trunk flexion movements among persons with unilateral lower-limb amputation, J. Electromyogr. Kinesiol. 24 (1) (2014) 120–125.

[6] C.H. Lloyd, S.J. Stanhope, I.S. Davis, T.D. Royer, Strength asymmetry and osteoarthritis risk factors in unilateral trans-tibial, amputee gait, Gait Posture 32 (3) (2010) 296–300.

[7] T.D. Royer, C.A. Wasilewski, Hip and knee frontal plane moments in persons with unilateral, trans-tibial amputation, Gait Posture 23 (3) (2006) 303–306.

[8] M.J. Highsmith, J.T. Kahle, S.L. Carey, et al., Kinetic asymmetry in transfemoral amputees while performing sit to stand and stand to sit movements, Gait Posture 34 (1) (2011) 86–91.

[9] L. Nolan, A. Lees, The functional demands on the intact limb during walking for active trans-femoral and trans-tibial amputees, Prosthet. Orthot. Int. 24 (2) (2000) 117–125.

[10] E.R. Esposito, J.M. Wilken, Biomechanical risk factors for knee osteoarthritis when using passive and powered ankle–foot prostheses, Clin. Biomech. 29 (10) (2014) 1186–1192.

[11] I. Shojaei, B.D. Hendershot, E.J. Wolf, B. Bazrgari, Persons with unilateral transfemoral amputation experience larger spinal loads during level-ground walking compared to able-bodied individuals, Clin Biomech (Bristol, Avon). 32 (2016) 157–163.

[12] A.K. Silverman, N.P. Fey, A. Portillo, J.G. Walden, G. Bosker, R.R. Neptune, Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds, Gait Posture 28 (4) (2008) 602–609.

[13] H. Sadeghi, P. Allard, M. Duhaime, Muscle power compensatory mechanisms in below-knee amputee gait, Am. J. Phys. Med. Rehabil. 80 (1) (2001) 25–32.

[14] B.D. Hendershot, E.J. Wolf, Three-dimensional joint reaction forces and moments at the low back during over-ground walking in persons with unilateral lower-extremity amputation, Clin. Biomech. 29 (3) (2014) 235–242.

[15] M.T. Schley, P. Wilms, S. Toepfner, et al., Painful and nonpainful phantom and stump sensations in acute traumatic amputees, J. Trauma Acute Care Surg. 65 (4) (2008) 858–864.

[16] R.M. Alexander, A. Jayes, A dynamic similarity hypothesis for the gaits of quadrupedal mammals, J. Zool. 201 (1) (1983) 135–152.

[17] W.T. Dempster, Space Requirements of the Seated Operator, Geometrical, Kinematic, and Mechanical Aspects of the Body With Special Reference to the Limbs, Michigan State Univ East Lansing, 1955.

[18] D.J. Farris, G.S. Sawicki, The mechanics and energetics of human walking and running: a joint level perspective, J. R. Soc. Interface (2011) rsif20110182.

[19] B.R. Umberger, P.E. Martin, Mechanical power and efficiency of level walking with different stride rates, J. Exp. Biol. 210 (18) (2007) 3255–3265.

[20] J.M. Wilken, K.M. Rodriguez, M. Brawner, B.J. Darter, Reliability and minimal detectible change values for gait kinematics and kinetics in healthy adults, Gait Posture 35 (2) (2012) 301–307.

[21] R.E. Seroussi, A. Gitter, J.M. Czerniecki, K. Weaver, Mechanical work adaptations of above-knee amputee ambulation, Arch. Phys. Med. Rehabil. 77 (11) (1996) 1209–1214.

[22] H.A. Underwood, C.D. Tokuno, J.J. Eng, A comparison of two prosthetic feet on the multi-joint and multi-plane kinetic gait compensations in individuals with a unilateral trans-tibial amputation, Clin. Biomech. 19 (6) (2004) 609–616.

[23] H. Bateni, S.J. Olney, Kinematic and kinetic variations of below-knee amputee gait, Jpo J. Prosthet. Orthot. 14 (1) (2002) 2–10.

[24] S.F. Ray, S.R. Wurdeman, K.Z. Takahashi, Prosthetic energy return during walking increases after 3 weeks of adaptation to a new device, J. Neuroeng. Rehabil. 15 (1) (2018) 6.

[25] D.C. Norvell, J.M. Czerniecki, G.E. Reiber, C. Maynard, J.A. Pecoraro, N.S. Weiss, The prevalence of knee pain and symptomatic knee osteoarthritis among veteran traumatic amputees and nonamputees, Arch. Phys. Med. Rehabil. 86 (3) (2005) 487–493.

[26] T. Royer, M. Koenig, Joint loading and bone mineral density in persons with unilateral, trans-tibial amputation, Clin. Biomech. 20 (10) (2005) 1119–1125.

[27] R.H. Miller, W.B. Edwards, S. Brandon, A.M. Morton, K.J. Deluzio, Why don't most runners get knee osteoarthritis? A case for per-unit-distance loads, Med. Sci. Sports Exerc. 46 (3) (2014) 572–579.

[28] R.L. Krupenevich, R.H. Miller, B.D. Hendershot, B.L. Schnall, A.L. Pruziner, Knee adduction moment peak and impulse do not change during the first six months of walking with a prosthesis, Gait Posture 63 (2018) 86–90.

[29] M.A. Hunt, M. Simic, R.S. Hinman, K.L. Bennell, T.V. Wrigley, Feasibility of a gait retraining strategy for reducing knee joint loading: increased trunk lean guided by real-time biofeedback, J. Biomech. 44 (5) (2011) 943–947.

[30] M. Hunt, T. Birmingham, D. Bryant, et al., Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis, Osteoarthr. Cartil. 16 (5) (2008) 591–599.