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Gait characteristics of individuals with transtibial amputations walking on a destabilizing rock surface ☆

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Abstract

Individuals with transtibial amputation (TTA) have a high incidence of falls during walking. Environmental factors, such as uneven ground, often play a contributing role in these falls. The purpose of this study was to quantify the adaptations TTA made when walking on a destabilizing loose rock surface. In this study, 13 young TTA walked over a rock surface and level ground level ground at four controlled speeds. Subjects successfully traversed the rock surface by adopting a conservative gait characterized by shorter and wider steps. They also took shorter steps with their prosthetic limbs and exhibited greater variability in foot placement when stepping onto their intact limb. Between-limb differences in step length and width variability increased at faster walking speeds. TTA increased hip and knee flexion during initial stance, which contributed lowering the whole-body center of mass. TTA also increased hip and knee flexion during swing, enabling them to significantly increase their toe clearance on the rock surface compared to level ground. Toe clearance on the prosthetic side was aided by increased ipsilateral hip flexion. The results suggest that TTA were able to adapt their gait to overcome the challenge imposed by the rock surface. These adaptations were asymmetric and initiated proximally.

Keywords

transtibial amputation; uneven or irregular terrain; kinematics; falls; walking

1. Introduction

The incidence of falls and fear of falling are common in individuals with lower limb amputations [1, 2]. Environmental barriers, such as uneven ground or inclines, can increase the frequency of falling [3]. Walking on uneven surfaces, in particular, may be difficult for

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persons with lower limb amputations due to limited ankle joint mobility on the prosthetic side, and lack distal muscles and sensory feedback from the lower limb [4]. To mitigate fall risk, it is important to determine how individuals successfully negotiate these challenging surfaces.

Individuals with and without lower limb amputation differed in the way they respond to challenging surfaces like stairs, ramps, or uneven ground. Persons with transtibial amputations (TTA) adopted a conservative, slower gait when walking on challenging surfaces such as uneven ground [5], stairs [6], or inclines [7]. They also spent more time in double support when walking on uneven ground [8], up [8] or down an incline [7], or when ascending [6, 8] or descending stairs [6]. Due to limited range of motion at the ankle, TTA make more proximal kinematic adaptations when negotiating stairs [9, 10]. Additionally, TTA may make different adaptations on their prosthetic and intact limbs, including stance phase asymmetry during stair ambulation [7, 9].

Many surfaces encountered in everyday life have varying surface characteristics. Adapting to changes in surface characteristics requires that subjects continually adapt their movement patterns. Previously, we studied healthy young individuals without amputations walking on a destabilizing rock surface that was uneven, unpredictable and moveable [11]. This surface was challenging because it could elicit a trip if patients did not walk with sufficient toe clearance, or a slip if patients did not adapt to the movement of the rocks underneath the foot. Able-bodied individuals adapted their gait in specific ways to decrease their risk of falling when walking on the rock surface at controlled speeds [11]. Those subjects increased hip and knee flexion to lower their center of mass (COM) and exhibited a more flexed posture with a flatter foot at initial contact [11]. They also increased ankle dorsiflexion and hip and knee flexion during swing to increase toe clearance on the rock surface [11]. Similar adjustments may not be possible in TTA since they cannot use proprioceptive information or distal muscles to make adjustments during stance on their prosthetic limb. They may therefore rely more on visual input and predictive strategies to appropriately maintain their equilibrium while walking [4].

The purpose of this study was to quantify how TTA adapt their gait when walking on a destabilizing rock surface. Similar to able-bodied subjects tested previously [11], we hypothesized that the TTA would modify their gait kinematics when walking on the rock surface to decrease their foot contact angles, increase their overall flexion during early stance and swing, and increase their toe clearance. However, since TTA cannot actively adjust their ankle angle, we hypothesized that the greatest kinematic changes would be seen at the hip and knee joints on the prosthetic side. We also hypothesized that TTA would increase their step width to widen their base of support and to lower their center of mass to enhance stability. Finally, we hypothesized that kinematic changes would increase with increasing walking speed.

2. Methods

2.1 Subjects

Thirteen young adults with unilateral traumatic transtibial amputations participated (Table 1) in this institutionally approved study after providing their written informed consent. Prior to testing, all participants were screened to ensure their intact limb was free of orthopedic and neurological disorders. All patients were able to complete all activities in the Locomotor Capabilities Index-5 without assistive devices [12]. All patients wore single-axis energy storing prosthetic feet which were fit by a certified prosthetist. They also wore their own running shoes during all data collection.

2.2 Experimental Protocol

Patients performed the same protocol as a group of healthy able-bodied subjects previously described in [11]. Subjects walked across level ground and over a destabilizing rock surface at four controlled speeds. The level ground was a 5-m level walkway and rock surface was a 4.2-m long by 1.2-m wide by 10-cm deep pit filled with loose river rocks from a major hardware store. A photograph of the rocks with scale is provided as supplemental material. Each capture area was preceded by a 4.6-m level walkway which allowed subjects to reach steady speed prior to data collection [13]. Subjects were instructed not to look down, unless they "felt that they were about to lose their balance and fall."

Walking speeds were normalized to each subject's leg length according to $\mathit{Walking\ Speed}=\sqrt{Fn \cdot g \cdot l}$, where Fn is the Froude Number, g is the gravitational constant and l is leg length [14]. Subjects walked at Fn = 0.06, 0.10, 0.16, and 0.23 Prior to collecting these set speeds, subjects were able to walk across the level ground and rock surface at their self-selected speeds to acclimate to the task. These speeds were recorded, and the average of five trials was noted. The order of testing was randomized such that each speed was performed first on the level ground and then rock surface. At each speed, only trials what were within ±10% of the target speed were accepted. A total of five left and five right strides were collected for each subject, at each speed, on each surface. Subject #10 did not complete speed four over the rock surface due to apprehension.

2.3 Data Analysis

Kinematic data were collected at 120 Hz using a 20-camera infrared motion capture system (Motion Analysis, Santa Rosa, CA). 55 reflective makers were used to track whole body kinematics [11]. The locations of 20 bony landmarks in relation to marker clusters were found by manual palpation and recorded using a digitizing pointer (C-Motion, Inc., Germantown, MD). Kinematic data were low-pass filtered using a 4^{th} order low-pass Butterworth filter with a cut-off frequency of 6 Hz. Heel strikes were determined using a velocity-based detection algorithm [15] and then verified by visual inspection. Step length and step width were defined as the distance between the right and left heel markers at heel strike in the anterior-posterior and medial-lateral directions respectively. Step time was the time elapsed between subsequent right and left heel contacts. The standard deviations of step time, step length, and step width for each limb and each condition, computed across all five cycles, represented the within-subject variability. Foot angle at initial contact was defined as the angle between a line connecting the heel and toe markers and horizontal [16]. The initial position of the foot was subtracted such that the foot angle was zero during quiet stance. Minimum toe clearance was the vertical distance between the first metatarsal marker at its lowest point in stance and its lowest point in mid-swing [16]. Minimum toe clearance at mid-swing indicates the potential risk of tripping [16, 17], where a lower or more variable toe clearance would indicate a greater likelihood of tripping [16, 18].

Segmental markers and landmarks were used to create a 15 segment whole-body model using Visual3D (C-Motion, Germantown, MD). Local coordinate systems for the segments were defined using ISB recommendations [19, 20]. Segmental masses were assigned based on the anthropometric data of Dempster [21]. COM was calculated as the weighted average of segmental COMs. Vertical COM displacement was normalized to each subject's standing height for comparison across subjects. To ensure changes across conditions were not merely due to displacement of the rocks, the vertical height of the ankle joint center during stance was subtracted from the COM height [11, 22]. Angular motion of the ankle, knee, and hip were defined using Euler rotations according to accepted recommendations [19, 20]. Joint angles and COMheight were time normalized to 0 to 100% of the gait cycle.

Separate three-factor, within-subjects ANOVAs were conducted for each dependent measure to test for differences between walking surfaces (level ground, rock surface), speeds (speed 1–4), and limbs (Intact, Prosthetic). Statistical analyses were performed using SPSS 16 (SPSS Inc, Chicago), with a level of significance of $p \leq 0.05$ for all comparisons. Significant interaction effects were explored using the Estimated Marginal Means with a Bonferroni correction for multiple comparisons. P-values provided in the text are denoted 'Spd' for walking speed effects, 'Sur' for walking surface effects, and 'L' for limb effects.

3. Results

Subjects walked with a self-selected speed of 1.09 ± 0.13 m/s on the rock surface compared to a speed of 1.27 ± 0.14 m/s on level ground. Given the subject anthropometrics, the four controlled speeds (speed 1 – 4) were approximately 0.73, 0.97, 1.21, and 1.45 m/s. Thus, the speed 3 most closely approximated the subject's self-selected speed across surfaces.

3.1 Temporal-Spatial Parameters

As expected, step length increased and step time decreased at faster walking speeds ($p_{Spd} < 0.001$; Fig. 1A and Table 2). TTA adapted to the rock surface by taking shorter ($p_{Sur} = 0.014$), wider steps ($p_{Sur} = 0.029$). There were significant interaction effects for step length ($p_{Spd \times L} < 0.001$, $p_{L \times Sur} = 0.01$). Subjects walked with shorter steps on their intact limb when walking on the rock surface at speeds 3 and 4 and shorter steps on their prosthetic limb on level ground at speed 1.

Step time, length, and width variabilities were greater on the rock surface compared to level ground ($p_{Sur} < 0.001$; Fig. 1B and Table 2). Step time variability decreased with speed ($p_{Spd} < 0.001$). There was a significant limb effect for step width variability ($p_L = 0.002$). TTA had greater step width variability when stepping from the prosthetic limb to the intact limb. There was also a speed × limb interaction effect for step width variability ($p_{Spd \times L} = 0.021$). The difference between limbs was significant at speeds 1–3, but not speed 4.

3.2 Kinematics

Initial Contact—TTA made qualitatively similar adjustments to those made by a group of healthy able-bodied subjects previously studied [11] when walking on both surfaces (Fig. 2A & 3A). Foot contact angle was lower when subjects walked on rock surface ($p_{Sur} < 0.001$; Fig. 2B, Table 2) and at slower speeds ($p_{Spd} < 0.001$). Foot contact angle was also lower on the intact than the prosthetic limb ($p_L = 0.001$). Between-limb differences in foot contact angle were greater on the rock surface than level ground ($p_{L \times Sur} < 0.001$).

Subjects had a significantly lower COM at heel contact when walking on rock surface ($p_{Sur} < 0.001$) than level ground (Fig. 3D & E; Table 2). COM height also decreased with increasing speed ($p_{Spd} = 0.001$). Subjects contacted the ground with a lower COM when stepping onto the prosthetic limb, except at speed 1 ($p_{Spd \times L} = 0.037$). Additionally, the range of motion of the COM height during stance increased on the rock surface ($p_{Sur} < 0.001$) and at faster speeds ($p_{Spd} < 0.001$; Fig. 3F).

Early Stance—TTA exhibited increased hip and knee flexion during early stance when walking on the rock surface ($p_{Sur} < 0.001$; Fig. 3A & B; Table 2). Hip flexion increased with walking speed ($p_{Spd} < 0.001$). Knee flexion was greater on the intact limb for speeds 2–4 ($p_L = 0.038$; $p_{Spd \times L} < 0.001$). Between-limb differences in peak knee flexion were two times greater on the rock surface ($p_{L \times Sur} < 0.001$). Ankle plantarflexion during early stance was greater on level ground ($p_{Sur} < 0.001$) and on the intact limb ($p_L = 0.001$).

Late Stance—At terminal stance, the TTA reached the same posture when walking on the rock surface as level ground (Fig. 3A; Table 2). There were no main effects for walking surface in peak ankle, knee or hip angles. There were significant speed, limb, and all interaction effects for hip extension. Peak hip extension increased (ie. the hip angle became more negative) with increasing speed and was greater on the intact limb (See Supplemental Material).

Swing—Peak hip and knee flexion during swing were greater when subjects walked on the rock surface ($p_{Sur} \leq 0.001$; Fig. 3A & C; Table 2). The amount of hip and knee flexion increased with walking speed ($p_{Spd} < 0.001$). Hip flexion was also greater on the intact limb ($p_L = 0.017$). There was greater prosthetic limb ankle dorsiflexion on the rock surface ($p_{Sur} < 0.001$; $p_{L\times Sur} < 0.001$). Between-limb differences were significant at speed 1 only ($p_{Spd} = 0.001$, $p_{Spd\times L} < 0.001$).

Increased swing phase hip and knee flexion on the rock surface contributed to increased toe clearance ($p_{Sur}<0.001$; Fig. 3A–C). Toe clearance increased with speed on the rock surface ($p_{Spd} < 0.001$; $p_{Spd\times Sur} < 0.001$). Toe clearance was also greater on the intact limb ($p_L < 0.001$). Between-limb differences were greater on the rock surface than level ground ($p_{L\times Sur} > 0.001$).

4. Discussion

In contrast with unimpaired subjects [11], TTA adapted to the rock surface by taking shorter, wider steps (Fig. 1A). These changes may have been made to increase the lateral base of support due to perceived instability. Similar changes were shown in other groups with compromised balance. Elderly subjects slowed down, took wider steps and increased step time when walking on an uneven surface while younger subjects did not [23]. TTA developed an asymmetry in SL when walking across the rock surface at faster speeds, wherein they took shorter steps with their intact limb. This asymmetry is not uncommon in TTA [24], and may reflect a desire to spend less time supported by their prosthetic limb while still maintaining the required speed.

Consistent with other studies involving uneven terrain [11, 22], TTA exhibited more variable step time, step length, and step width when walking over the rock surface compared to level ground. TTA also exhibited greater variability in step width when stepping from prosthetic onto intact limbs. This is partially explained by the fact that TTA cannot actively control the ankle to adapt when standing on their prosthetic limb. Therefore, they may use the subsequent step with their intact limb to help redirect the COM and adjust the base of support to ensure stability when supported by their prosthetic limb. Differences between the limbs decreased with increasing speed. At faster walking speeds the range of motion of the COM in the medial-lateral direction has been found to decrease [25]. Thus, the subjects may have been able to make more subtle adjustments to their foot placement since their COM was less likely to approach its outer limits.

Similar to able-bodied subjects [11], TTA walked such that they were more flat footed at initial contact on the rock surface. TTA also took slightly shorter steps on the rock surface (Fig. 1A; Table 2), with the largest decrease in step length on the intact limb at the faster speeds. Similarly, several studies have found that subjects with knowledge of a slippery surface reduce their foot contact angles and stride length [16, 26, 27]. These adjustments may reduce the required coefficient of friction (the ratio of shear to normal force) at the shoe-floor interface [26, 28]. Previous studies have shown that subjects are able to successfully reduce RCOF by more slowly rotating the foot down onto the walking surface after heel strike, adopting shorter strides, and reducing foot contact angles [26].

Additionally, decreased step length and decreased foot contact angle have been associated with decreased probability of a hazardous slip [29]. Thus, these adjustments may decrease the risk of initiating a slip or decreased the severity of the slip, if it occurred. On this particular loose rock surface, the decreased foot contact angle enabled subjects to increase the contact area between their foot and the surface thus distributing the ground reaction force across more rocks. This likely enabled the subjects to remain on the top of the surface without sinking into the rocks or pushing them forward or sideways on contact. The strategy that the TTA employed was different on their prosthetic and intact limbs. Since TTA cannot actively control ankle position of their prosthetic limb, they are forced to use knee and hip position to control foot orientation. Foot contact angle was greater on the prosthetic limb. This may suggest that patients were at increased risk of slipping or sinking into the rocks when stepping onto their prosthetic limb.

TTA exhibited increased hip and knee flexion during early stance when walking on the rock surface. In addition to affecting foot orientation at initial contact, this may have enabled them to lower their COM, similar to what able-bodied subjects did when walking on the same rock surface [11], a compliant surface [22], and a slippery surface [30]. This adaptation may enhance stability by decreasing the moment arm between the COM and ground reaction force, such that a larger horizontal force is necessary to induce a fall [22]. This flexed posture may also be advantageous as it allows the subject to adjust the lateral COM position simply by extending the stance leg.

Minimum toe clearance was four times greater on the rock surface compared to level ground (Fig. 2A, C). Toe clearance is a measure of the margin of safety during walking where decreased toe clearance leads to increased risk of tripping [16, 17]. Thus this adaptation likely reduced the subjects' tripping risk when walking on the rock surface. This finding matches previous studies in young adults without amputations. Young adults exhibited a 2- to 4-times greater toe clearance when walking over surfaces with various sized obstacles (13 – 50 mm obstacles) and when walking on a loose rock surface [11], compared to an obstacle-free level surface [16, 18, 31]. Additionally, TTA walked with 1.3 times greater toe clearance with their intact limb than their prosthetic limb. TTA are better able to respond when standing on their intact limb since they have active ankle control. They may therefore, not need as much toe clearance on their prosthetic limb. Toe clearance on the prosthetic limb was also likely limited since the prosthetic ankle could not be actively dorsiflexed to raise the height of the toe. To compensate, subjects increased ipsilateral hip flexion (Fig. 3C; Supplemental Fig 2). This compensation is supported by modeling work [32], which shows that toe clearance is most sensitive to changes in the angle of the ankle, then hip, then knee.

Limitations

The results of this study may not be generalizable to all amputee populations. These patients were young, active, and highly functional, with amputations due to limb trauma. As a result, they may be better able to respond to surface challenges than elderly or pathologic TTA. For example, patients with vascular disease often have reduced somatosensory feedback from their intact limb. This is linked to poor balance and may influence the strategy utilized to walk on irregular terrain [33]. Therefore, we might expect greater compensations and poorer performance from these individuals. It should be noted that each subject attempted one faster speed (Fr = 0.31), but not all of these high-functioning individuals were able to walk comfortably at that pace.

5. Conclusion

This study showed that young, active, TTA respond to the additional challenges of modified speed and surface to minimize their risk of falling. Specifically, they were able to alter their foot contact angle when walking on the loose rock surface to reduce their risk of slipping or moving the rocks. Subjects also increased toe clearance by increasing hip and knee flexion on the prosthetic limb and hip, knee and ankle dorsiflexion on their intact limb to reduce their risk of tripping. Additionally, subjects likely increased their stability during stance by increasing their step width and increasing lower extremity flexion to lower their COM height.

* 13 Patients with transtibial amputations walked on level ground and a loose rock surface.
* Patients took shorter, wider steps on the rock surface.
* Patients increased hip and knee flexion, to increase toe clearance when walking on the rocks.
* Patients made adaptations on the rocks that were asymmetric, and initiated proximally.

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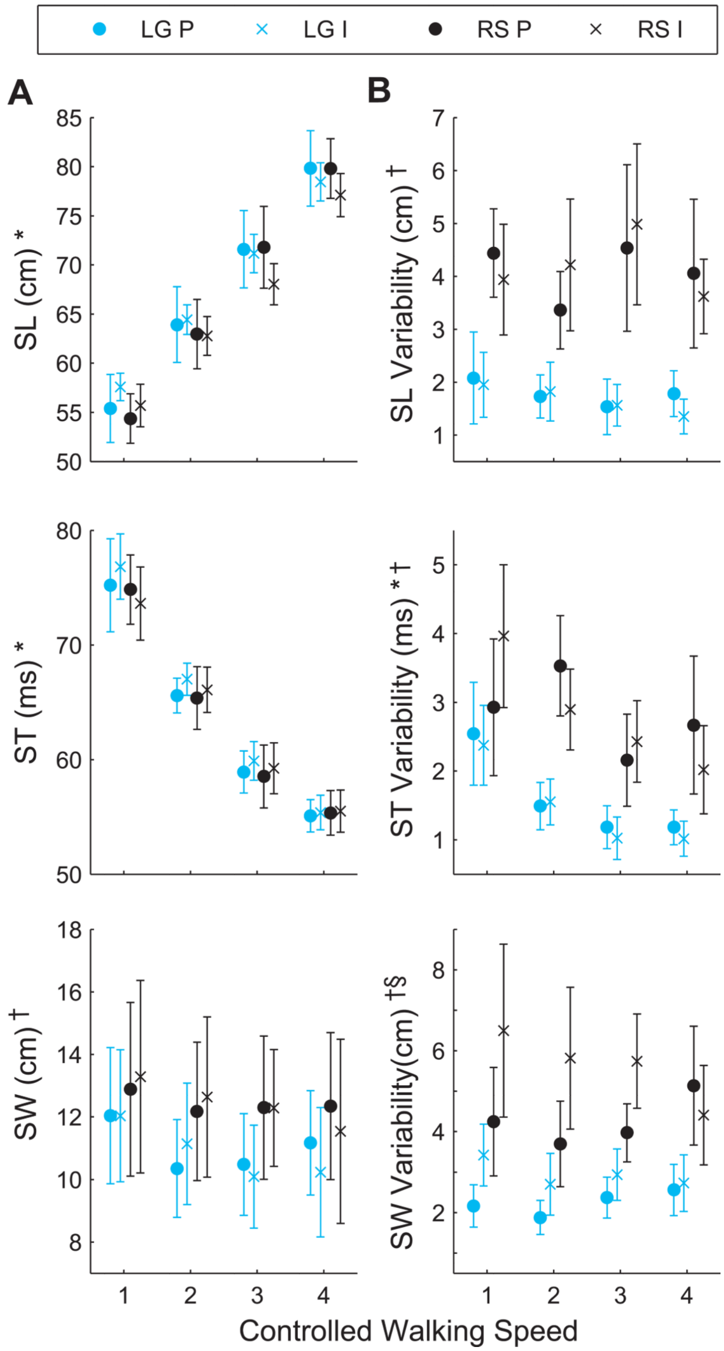

Figure 1.

Temporal-Spatial Parameters. Data is shown for the rock surface (*RS*; black) and level ground (*LG*; cyan) at four, evenly-spaced, controlled speeds. Data for the prosthetic limb (*P*) is given by '○' while data for the intact limb (*I*) is given by '×'. **A)** The average and **B)** within-subject variability of step length (*SL*), step time (*ST*), and step width (*SW*) across subjects are shown for each condition. Error bars represent ± 95% confidence intervals about the mean. *Statistically significant main effects for walking speed ($p < 0.05$), =Statistically significant main effects for walking surface ($p < 0.05$), §Statistically significant main effects for limb ($p < 0.05$)

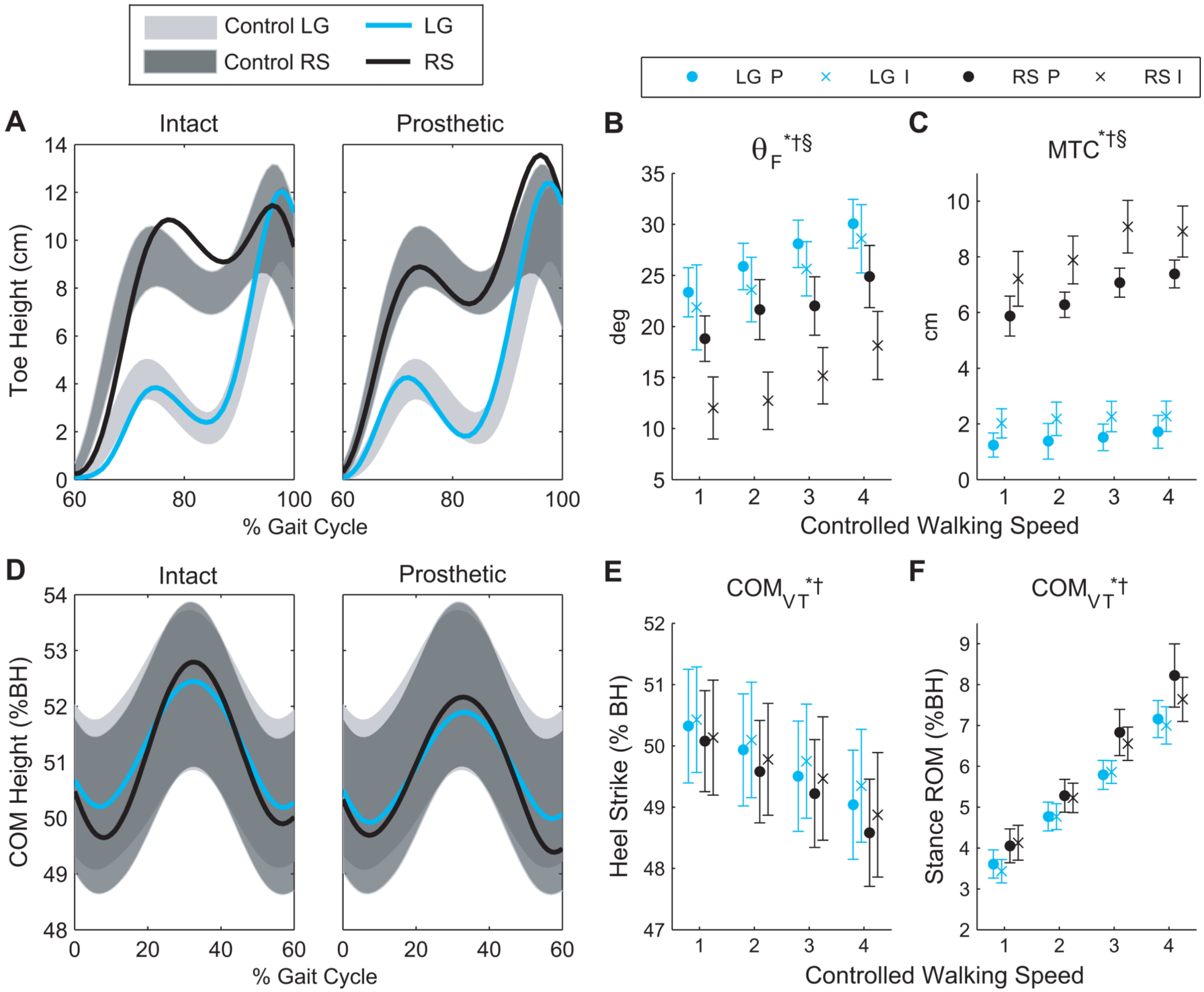

Figure 2.

Foot and center of mass motion. **A)** The position of the toe marker is shown for the uninvolved and involved limbs as subjects walked over the rock surface (*RS*) and level ground (*LG*) at four, evenly-spaced, controlled speeds. Bands represent the mean ± standard deviation of the average toe position across a group of young healthy subjects [12]. Solid lines represent the average joint angles across subjects for each condition for the intact (*I*; right panel) and prosthetic (*P*; left panel) limbs. **B)** The average foot contact angle, θ_F; **C)** The average minimum toe clearance, *MTC*; **D)** *COM* height as a percent of body height ('BH') is shown for the stance phase. **E)** The average *COM* height at heel strike. **F)** The range of motion of *COM* height across the stance phase (0–60% of the gait cycle). Error bars represent ± 95% confidence intervals about the mean. *Statistically significant main effects for walking speed ($p < 0.05$), †Statistically significant main effects for walking surface ($p < 0.05$), §Statistically significant main effects for limb ($p < 0.05$)

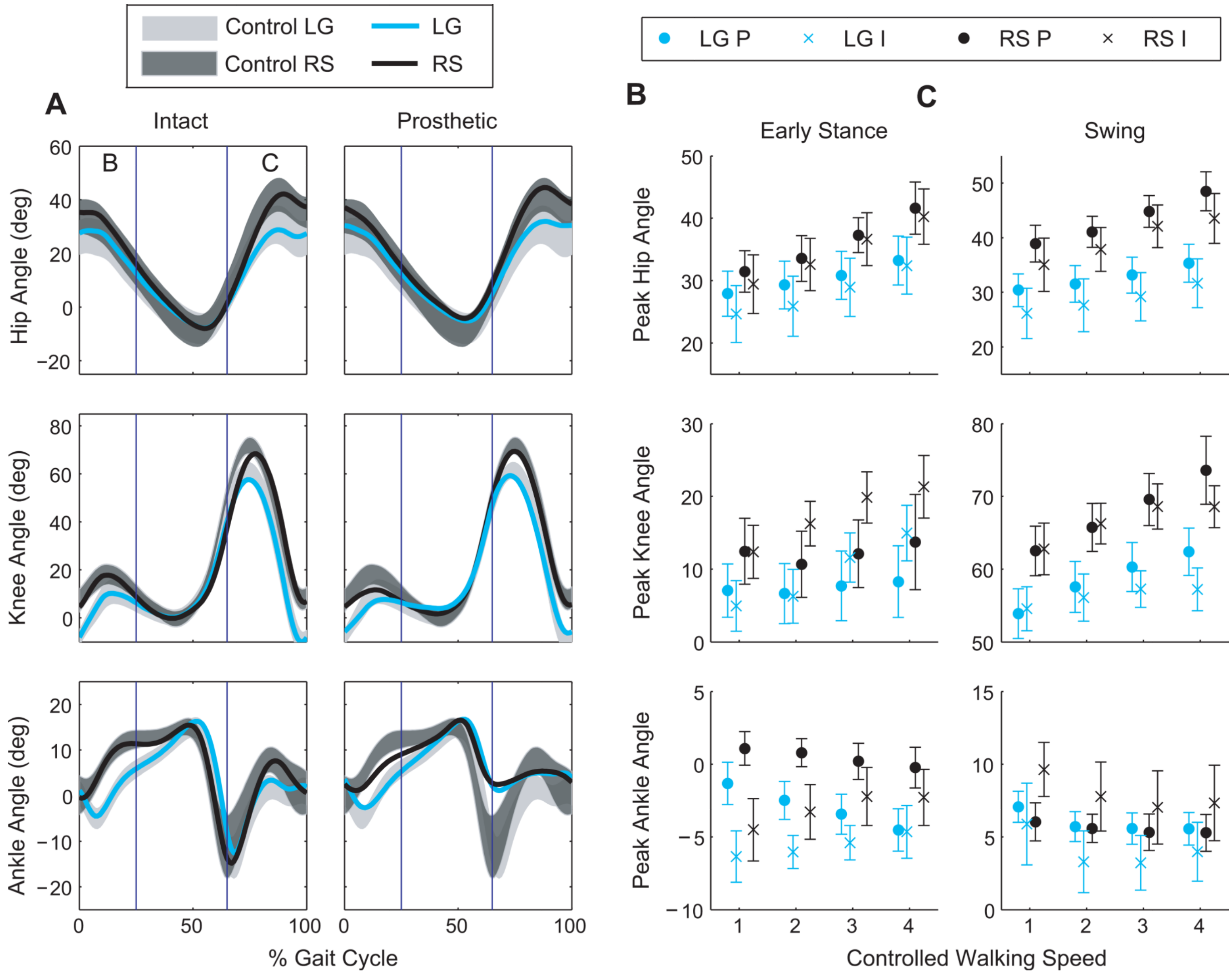

Figure 3.

Kinematics. **A)** Bands represent the mean ± standard deviation of the average joint angle across a group of young healthy subjects [11] walking over level ground (*LG*) and over the rock surface (*RS*). Solid lines represent the average joint angles across subjects for each condition for the intact (I; right panel) and prosthetic (*P*; left panel) limbs. **B)** Peaks during early stance were defined as the maximum excursion between 0 and 25% of the gait cycle. For the hip and knee, this was positive or flexion, while for the ankle; this was negative, or plantarflexion. **C)** Swing phase peaks were the maximum (positive) joint angle between 65 and 100% of the gait cycle. Peaks are shown for each of the four, evenly-spaced, controlled speeds (Fr = 0.06, 0.1, 0.16, and 0.23). Error bars represent the 95% confidence intervals about the mean.

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Table 1

Subject characteristics

Subject	Gender	Age (years)	Height (m)	Mass (kg)	Leg Length (m)	Residual Limb Length (m)	Time in Prosthesis (weeks)	Involved Side
1	M	26	1.80	67.4	0.93	0.16	12	Left
2	M	32	1.85	83.7	0.95	0.19	26	Left
3	M	29	1.75	75.3	0.93	0.19	8	Left
4	M	32	1.76	77.1	0.92	0.10	21	Right
5	M	20	1.93	119.5	0.95	0.25	8	Right
6	M	27	1.77	85.7	0.90	0.14	4	Right
7	M	23	1.80	92.5	0.94	0.25	8	Left
8	M	27	1.88	96.4	0.95	0.14	12	Left
9	M	24	1.81	90.7	0.91	0.21	4	Right
10	F	32	1.65	76.7	0.84	0.15	16	Left
11	M	30	1.97	105.2	1.05	0.25	36	Right
12	M	35	1.71	93.0	0.90	0.20	52	Left
13	M	25	1.80	77.6	0.90	*	52	Right
Mean (SD)		28 (4)	1.82 (0.08)	88.7 (14.3)	0.93 (0.04)	0.19 (0.05)	20 (18)	

* Not measured

Table 2

Statistical Results. P-values are given for each dependent measure. Significant results ($p < 0.05$) are highlighted in bold font.

Dependent Measure	Speed p_{Spd}	Walking Surface p_{Sur}	Limb p_L	Speed × Limb $p_{Spd×L}$	Speed × Surface $p_{Spd×Sur}$	Limb × Surface $p_{L×Sur}$
Step Length	**< 0.001**	**0.014**	0.711	**< 0.001**	0.986	**0.010**
Step Time	**< 0.001**	0.349	0.169	0.677	0.456	0.116
Step Width	0.103	**0.029**	0.843	0.289	0.698	0.836
Step Length Variability	0.330	**< 0.001**	0.965	0.306	0.128	0.599
Step Time Variability	**< 0.001**	**< 0.001**	0.536	0.160	0.331	0.684
Step Width Variability	0.316	**< 0.001**	**0.002**	**0.021**	0.918	0.209
Peak Hip Flexion – Early Stance	**< 0.001**	**< 0.001**	0.152	0.132	0.078	**0.050**
Peak Knee Flexion – Early Stance	**< 0.001**	**< 0.001**	**0.038**	**< 0.001**	0.499	**< 0.001**
Peak Ankle Plantarflexion – Early Stance	0.964	**< 0.001**	**0.001**	0.075	**< 0.001**	0.053
Peak Hip Extension – Late Stance	**< 0.001**	0.139	**0.006**	**< 0.001**	**0.001**	**< 0.001**
Peak Knee Extension – Late Stance	0.794	0.802	0.080	**0.011**	**0.026**	0.296
Peak Ankle Dorsiflexion – Late Stance	0.485	0.777	0.517	0.357	0.112	**0.002**
Peak Hip Flexion – Swing	**< 0.001**	**< 0.001**	**0.017**	0.908	**0.042**	0.372
Peak Knee Flexion – Swing	**0.001**	**< 0.001**	0.529	0.319	0.595	0.216
Peak Ankle Dorsiflexion – Swing	**< 0.001**	**< 0.001**	0.727	**0.019**	0.432	**< 0.001**
COM Height, Range of Motion	**< 0.001**	**< 0.001**	0.451	0.224	0.123	0.177
COM Height at Heel Strike	**< 0.001**	**0.001**	0.315	**0.037**	0.868	0.915
Foot Contact Angle	**< 0.001**	**< 0.001**	**0.018**	0.170	0.190	**< 0.001**
Minimum Toe Clearance	**< 0.001**	**< 0.001**	**< 0.001**	**0.405**	**< 0.001**	**0.004**