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Biomechanical responses of individuals with transtibial amputation stepping on a coronally uneven and unpredictable surface

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ABSTRACT

Coronally uneven surfaces are prevalent in natural and man-made terrain, such as holes or bumps in the ground, curbs, sidewalks, and driveways. These surfaces can be challenging to navigate, especially for individuals with lower limb amputations. This study examined the biomechanical response of individuals with unilateral transtibial amputation (TTA) taking a step on a coronally uneven surface while wearing their clinically prescribed prosthesis, compared to individuals without mobility impairments (controls). An instrumented walkway was used with the middle force plate positioned either flush or rotated $\pm$ 15° in the coronal plane and concealed (blinded). TTAs used greater hip abduction compared to controls across all conditions, but especially during blinded inversion. The recovery step width of TTAs was wider after blinded eversion and narrower after blinded inversion, but unchanged for controls. These results suggest TTAs may have decreased balance control on unexpected, uneven surfaces. Additionally, TTAs generated less positive prosthetic ankle joint work during blinded inversion and eversion, and less negative coronal hip joint work during blinded inversion compared to controls. These biomechanical responses could lead to increased energy expenditure on uneven terrain. Surface condition had no effect on the vertical center of mass for either group of participants. Finally, the TTAs and the control group generated similar vertical GRF impulses, suggesting the TTAs had sufficient body support despite differences in surface conditions. These results are important to consider for future prosthetic foot designs and rehabilitation strategies.

1. Introduction

Individuals with lower limb amputations generally have greater difficulty maintaining balance, especially when walking on uneven surfaces. Seventy-three percent of outdoor falls are due to environmental factors such as uneven surfaces on sidewalks, curbs, and streets (Li et al., 2006). A fall outdoors is three times more likely to result in an injury (Kelsey et al., 2012), which may contribute to the fear of falling and reduced activity levels (Miller et al., 2001; Miller and Deathe, 2011). Thus, maintaining balance when walking on uneven surfaces is important for participation in community activities.

Walking on uneven surfaces requires several biomechanical responses by the lower limbs to conform to the surface and maintain balance. Individuals without mobility impairments can navigate uneven surfaces by reducing step width after a disturbance (Dixon and Pearsall, 2010; Yeates et al., 2016), increasing knee and hip flexion (Gates et al., 2012), and increasing ankle and hip power (Panizzolo et al., 2017; Segal et al., 2018). Individuals with lower limb amputations likely have a harder time adapting to uneven surfaces due to a lack of ankle–foot muscle actuation and the passive characteristics of their prosthetic foot. Most currently prescribed multiaxial prosthetic feet for individuals with transtibial amputation (TTA) are passive energy-storing-and-returning feet. These types of feet aim to improve the performance and activity level of individuals with TTA by attempting to mimic the shock absorption and propulsion of the intact foot (Hafner et al., 2002). However, these prosthetic feet can only return as much energy as they absorb, and act passively in all axes of deformation, which may not be desirable over certain uneven surfaces. Different types of uneven

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surfaces may even further decrease the propulsive output of these prosthetic feet. Understanding how individuals with TTA respond to uneven surfaces while wearing their currently prescribed prosthetic feet, compared to individuals without mobility impairments, can provide insight into possible prosthetic or therapeutic interventions that minimize needed biomechanical responses.

Cross-slopes can be challenging even for individuals without mobility impairments. For example, adult males taking several steps on a 10% grade cross-slope walked with increased hip adduction on the up-slope limb, and decreased hip adduction on the down-slope limb (Dixon and Pearsall, 2010). During prolonged cross-slope walking, individuals with TTA have been shown to increase their prosthetic knee and hip flexion during swing, and decrease their contralateral knee flexion during stance (Villa et al., 2017). When stepping on a coronally uneven and unpredictable surface, individuals with TTA wearing a novel, coronally clutching ankle (i.e., conformal inverting and everting) showed improvement in some aspects of balance, such as center of mass path regulation, but not all when compared to their prescribed prosthesis (Yeates et al., 2018). Yet rarely are individuals with TTA trained on how to walk on uneven surfaces (Matjaĉić and Burger, 2003; Sjödahl et al., 2001). Current research has yet to explore the biomechanics of individuals with TTA taking a single step on a coronally uneven surface while wearing their prescribed prosthesis. A transient, single step on a coronally uneven surface, simulating unexpected variations in natural and manufactured terrain, may present a greater challenge to balance for individuals with TTA compared to individuals without mobility impairments.

The objective of this research was to quantify the biomechanical response of individuals with a unilateral TTA wearing their clinically prescribed prosthesis versus individuals without mobility impairments (controls), during and after a single step on a coronally uneven and unexpected surface. A single step was chosen to allow observation of the biomechanical response without further disturbances during subsequent recovery steps. A novel experimental design was used to produce three different terrain conditions: unblinded flush (flat surface), 15° blinded inversion, and 15° blinded eversion. The 15° coronal angle was chosen to produce an observable effect from the disturbance without causing injury to the participants (Yeates et al., 2016). Lower limb joint angles and vertical center of mass (COM) were calculated to give insight into lower limb joint strategies and biomechanical responses to uneven terrain. Joint work was calculated to give insight into energy expenditure and the effects of the prosthetic foot on propulsion over uneven terrain. Finally, recovery step width and vertical ground reaction force (GRF) impulse were calculated to give insight into balance and body support for individuals with TTA compared to controls. These findings may contribute to current rehabilitation practices for individuals with TTA and to future prosthesis designs.

2. Methods

2.1. Participants

Completing the protocol were eight individuals with unilateral TTA (7 male, age: 49 ± 14.6 years, height: 1.78 ± 0.04 m, mass: 84.4 ± 6.6 kg, etiology: 7 traumatic, 1 diabetic neuropathy) walking with their clinically prescribed prosthesis, including an energy storing and returning prosthetic foot, and eight age and gender matched controls (7 male, age: 47 ± 18.9 years, height: 1.77 ± 0.10 m, mass: 81.4 ± 11.0 kg) with no self-reported musculoskeletal or gait disorders. All participants were free of contractures. Each subject provided informed consent approved by the governing Institutional Review Board.

2.2. Instrumentation

A custom instrumented raised walkway was created (Yeates et al., 2016) with five embedded force plates to capture GRFs (in sequential order: two AMTI force plates, BP400600, Watertown, MA, then one portable Kistler force plate, 9286AA, Winterthur, Switzerland, followed by two more AMTI force plates. See Fig. 1). The middle disturbance force plate could be rotated and rigidly positioned either flush with the walkway or at ± 15° in the coronal plane, creating an inverted or everted cross-slope disturbance (Fig. 2). Subjects could be blinded to the position of the middle disturbance force plate by concealing it with a 0.5 mm opaque latex cover (Fig. 2B, 2C), which had negligible stiffness to minimize anticipatory strategies prior to initial contact with the uneven step. The walkway had handrails along each side for safety but were never used (Fig. 1).

A 12-camera motion capture system (Vicon Motion Systems, Oxford, GBR) recorded marker trajectories at 120 Hz and force plate data at 1200 Hz. All subjects wore study provided, tight fitting spandex shorts and shirt, and were fit with a standardized walking shoe (model M577, New Balance Inc., Boston, MA). The same researcher placed reflective tracking markers on all subjects using Vicon's standard Plug-in-Gait marker set, with additional markers placed bilaterally on the medial knee epicondyle, medial malleolus, tibial tuberosity, fibular head, and first and fifth metatarsal heads. Clusters of four markers were also placed bilaterally on the upper arms and thighs. The markers on the prosthetic limb mirrored their intact limb.

2.3. Protocol

Individuals with TTA stepped on the middle disturbance plate with their prosthetic limb, while the controls stepped on it with their dominant limb, determined by which foot they would use to kick a ball. Three different surface conditions were tested: unblinded flush, blinded inversion, and blinded eversion. Subjects walked across the unconcealed walkway in each condition to establish familiarity and proper foot positioning. Only trials with single foot contact on the disturbance plate and subsequent recovery plate were included in the analysis. Subjects walked at their self-selected speed for a minimum of four repeated trials per condition. Unblinded flush trials were performed first, then blinded inversion and eversion trials were performed in randomized order. In between each blinded trial, subjects waited in a separate room while researchers switched the disturbance condition and concealed it with the latex cover.

2.4. Data analysis

The raw data were filtered using a digital, fourth order, low-pass Butterworth filter with cutoff frequencies of 25 Hz for kinetics and 6 Hz for kinematics. A 15-segment whole body model was created in Visual 3D (C-Motion, Germantown, USA. See additional details in (Yeates

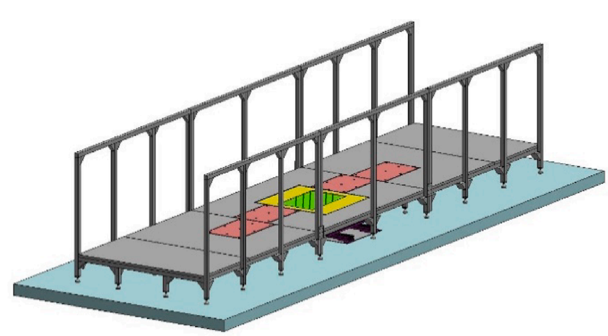

Fig. 1. CAD drawing made using SolidWorks (Dassault Systèmes SolidWorks Corporation, Waltham, MA) of the custom instrumented raised walkway (6 m × 1.5 m). Embedded AMTI force plates shown in red, middle disturbance force plate that could be rotated shown in green stripes, and walking platform and handrails shown in grey. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

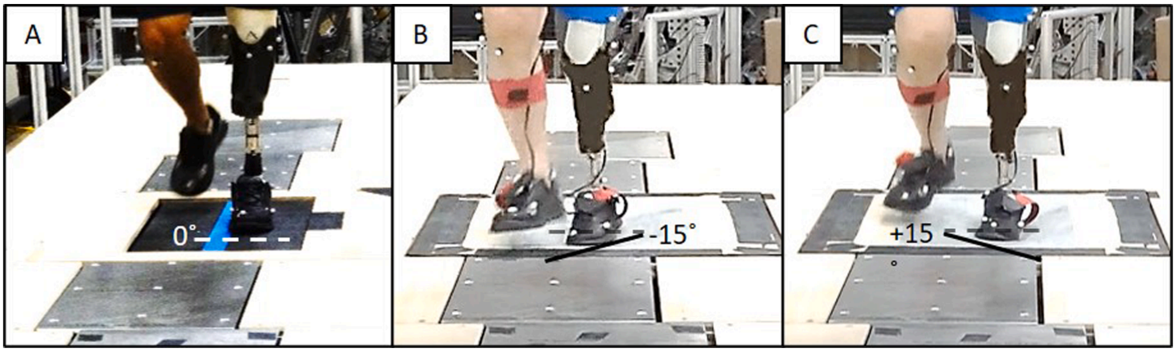

Fig. 2. Individuals with TTA stepping on the middle disturbance force plate during: (A) the unblinded flush condition, (B) the blinded eversion condition, and (C) the blinded inversion condition.

et al., 2016)), and used to calculate coronal and sagittal hip angles, sagittal knee angle, recovery step width, vertical COM position, positive and negative ankle, knee, and hip joint work, and the vertical GRF impulse. All variables were time normalized to percent gait cycle (heel strike on the middle disturbance plate to heel strike on the subsequent recovery plate). Hip and knee angles were calculated using inverse kinematics and zeroed based on standing static trial joint angles to create a steady state standing posture equivalent for all subjects. Recovery step width was calculated by taking the difference in mediolateral (x-axis) heel marker position between the heel strike on the middle disturbance plate and the opposite limb heel strike on the recovery plate. Vertical COM was normalized to each subject's body height and is reported as a percentage. Ankle power was calculated using the unified deformable segment method (Takahashi et al., 2012). This method includes power contributions from the deforming structures of the prosthesis, enabling a better direct comparison between the variable prosthetic structural components and the anatomical properties of an intact ankle–foot system (Takahashi et al., 2015; Zelik and Honert, 2018). Knee and hip powers were calculated using standard inverse dynamics techniques (Winter, 1991). Further data analysis was performed using Matlab software (The MathWorks Inc., Natick, MA). The total area of the power curves below zero (negative work) and above zero (positive work) were calculated during stance for the ankle, knee, and hip joints. Coronal and sagittal hip angles, sagittal knee angle, and vertical COM were measured at midstance on the disturbance plate. Midstance was calculated as the time when the anterior/posterior GRF crossed zero and was chosen because at this moment in the gait cycle, the subjects were assumed to be fully weight bearing, conformed to the surface, and ready to make a response with their opposite limb. GRFs were normalized by the subject's body mass. Finally, vertical GRF impulses were calculated as the time-integral of the vertical GRF across stance.

2.5. Statistical methods

Linear mixed effects regression was used to test for an association between outcome (the dependent variable), group (TTA vs. control) and surface condition (unblinded flush, blinded inversion or blinded eversion) by modeling a group by condition interaction. Study participant and study participant by condition interaction were modeled as random effects. The group by condition interaction enabled the estimation of specific pairwise differences: within condition differences between the TTA group compared to the control group for the blinded inversion, blinded eversion, and unblinded flush conditions separately, and within group differences between the blinded inversion or blinded eversion condition compared to the unblinded flush condition for the TTA group and control group separately. Within condition comparisons (TTA vs. control) were performed for all outcome measures, while within group comparisons (unblinded flush vs. blinded inversion and unblinded flush vs. blinded eversion) were only performed for coronal hip angle and recovery step width. We chose not to compare within group differences for any sagittal plane metrics because our study design required a small (approximately 3 cm) sagittal plane step down to obscure the blinded eversion and blinded inversion conditions while there was no such step down for the unblinded flush condition. Differences in outcome variance between the TTA group and the control group were addressed by estimating different variances for each group separately to achieve variance homogeneity across participants within each group. The Benjamini-Hochberg correction for multiple comparisons was applied to hypothesis tests for differences in biomechanical outcomes by TTA vs. control group for each surface condition separately to maintain a false discovery rate of 0.05. Standard errors are reported as they pertain to the precision of the mean estimates and are related to hypothesis testing. Statistical analyses were carried out using R 3.5.3 (R Foundation for Statistical Computing, Vienna, Austria, 2019).

3. Results

The TTA group had a less adducted coronal hip angle than the control group at midstance for all three surface conditions, ranging from a difference of -5.9° (95% confidence interval [-10.2, -1.5], $p = 0.038$) for the blinded eversion condition to -7.6° ([-12.4, -3.0], $p = 0.013$) for the blinded inversion condition (Fig. 3 and Table 1). Within group comparisons showed that the TTA group had a more abducted coronal hip angle by -3.0° ([-4.9, -1.3], $p = 0.001$) at midstance during the blinded inversion condition compared to the unblinded flush condition, but they had a similar magnitude for the blinded eversion condition compared to the unblinded flush condition (Fig. 3 and Table 2). The control group's coronal hip angles were similar across surface conditions (Table 2).

Sagittal hip angle, sagittal knee angle, and vertical COM position at midstance were similar for all surface conditions between subject groups (Table 1).

For the unblinded flush condition, the recovery step width was similar between groups. However, the TTA group used a larger recovery step width by 6.2 cm for the blinded eversion condition compared to the control group ([2.7, 9.5], $p = 0.016$) (Table 1). Within group comparisons showed that the TTA group had a 4.3 cm smaller recovery step width for the blinded inversion condition compared to the unblinded flush condition ([-6.8, -1.8], $p = 0.001$), and a 4.0 cm larger recovery step width for the blinded eversion condition compared to the unblinded flush condition ([1.2, 6.6], $p = 0.004$) (Table 2). The control group exhibited no differences between the flush condition and either of the blinded surface conditions (Table 2).

The TTA group had more negative ankle joint work than the control group during the blinded eversion condition (-0.07 J/kg difference, [-0.13, -0.02], $p = 0.038$), however the control group produced more positive ankle joint work during the blinded eversion condition (-0.11 J/kg difference, [-0.19, -0.04], $p = 0.038$) and the blinded inversion

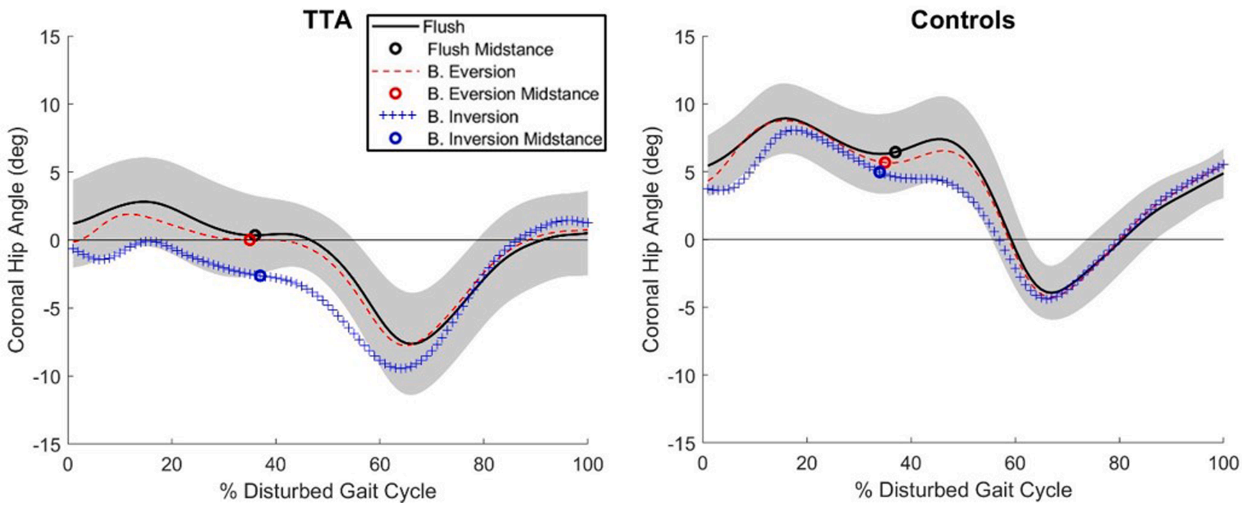

Fig. 3. Mean coronal hip angle plotted across the disturbed gait cycle (heel strike to heel strike) for each surface condition: unblinded flush with standard deviation shown in grey, blinded (B.) eversion, blinded (B.) inversion. Circles represent the average time of midstance during the disturbed gait cycle for each surface condition. Adduction is positive and abduction is negative.

condition (-0.10 J/kg difference, [-0.17, −0.04], $p = 0.01$) (Table 1).

The control group produced more positive sagittal knee joint work during the blinded eversion condition (-0.05 J/kg difference, [-0.07, −0.01], $p = 0.038$), and had more negative coronal hip joint work during the blinded inversion condition (0.04 J/kg difference, [0.02, 0.05], $p = 0.01$) compared to the TTA group (Table 1). Lastly, the vertical GRF impulses were similar for all surface conditions between subject groups (Table 1).

4. Discussion

The purpose of this study was to quantify the biomechanical response of individuals with TTA stepping on a coronally uneven and unexpected surface while wearing their prescribed prosthesis compared to able-bodied controls. For individuals with TTA, the introduction of a sudden cross-slope causes a different biomechanical response than continuous cross-slope walking. For example, the TTA group did not lower their vertical COM or increase sagittal knee or hip flexion, which has been shown in previous studies of prolonged walking on uneven surfaces (Gates et al., 2012).

The TTA group had less hip adduction on all surface conditions compared to the control group, especially during the blinded inversion condition. Surface condition did not significantly affect the control group’s coronal hip angle, however the TTA group abducted their hip three degrees more during the blinded inversion condition compared to the unblinded flush condition. This is consistent with previous findings demonstrating premature activation of hip abductors during unexpected inversion of the ankle for subjects with hypermobility (Beckman and Buchanan, 1995), which suggests this could be a biomechanical response to ankle inversion for subjects with altered ankle mobility.

The blinded surface conditions affected the recovery step width of the TTA group more than the control group. The TTA group’s recovery step was swayed in the direction of the disturbance, shown by a wider step taken after the blinded eversion condition and a narrower step after the blinded inversion condition compared to the unblinded flush condition. Both groups had similar step widths during undisturbed walking; however, the TTA group took a wider step during the blinded eversion condition compared to the control group. Due to the passive nature of the ankle–foot prosthesis, individuals with TTA may not have as much control of their COM and balance after stepping on an unexpected everting cross-slope, which resulted in an increased recovery step width. The control group accommodated the unexpected coronally uneven step without the use of a stepping strategy, showing no change in step width across conditions. This is in contrast with previously reported results by Dixon and Pearsall, who found controls used a narrower step width when continuously walking on a cross-slope compared to level ground walking (Dixon and Pearsall, 2010). The difference may be due to foot clearance expectations during steady-state walking that might not occur in response to a single step on uneven terrain.

The TTA group had more negative ankle joint work during the blinded eversion condition than the control group, however they did not produce as much positive ankle joint work as the control group during either the blinded eversion or the blinded inversion conditions. This suggests that currently prescribed prosthetic feet have the capability to store just as much energy as the human ankle–foot system during a step on a coronally uneven surface but are not able to generate additional positive joint work to assist with propulsion. A similar trend was found for individuals with TTA walking uphill, where their prosthetic limb absorbed nearly the same amount of energy as their intact limb but did not return as much energy (Childers and Takahashi, 2018). This indicates that individuals with TTA may have to work harder or compensate in other ways to keep a forward progression, especially on uneven surfaces. Future prosthetic designs should consider these findings and aim to generate additional net positive joint work while walking over different types of terrain. For example, performance of powered prostheses could be improved by generating increased net positive joint work when a step on coronally uneven terrain is detected.

In addition to producing more positive ankle joint work during the blinded eversion condition, the control group also produced more positive sagittal knee joint work compared to the TTA group. The lower positive sagittal knee joint work by the TTA group suggests a biomechanical response that reduces energy expenditure to the extent possible when encountering everting terrain. Prosthetic designs that produce more positive ankle joint work may alter this response and affect overall energy expenditure.

The TTA group did not perform as much coronal hip joint work as the controls during blinded inversion despite their greater coronal hip angle in abduction. Hip abductor work has been shown to be one of the major stabilizing elements during single limb support (Sadeghi et al., 2001). The TTA group may have weaker hip abductors than the control group, especially during blinded inversion where these muscles are needed for balance control. Rehabilitation therapies that strengthen hip abductor output may improve the ability of TTAs to walk on inverted surfaces. Future studies could include electromyography to aid in understanding the timing of TTA muscle activity while walking over unexpected uneven terrain.

Table 1
Mean (SE) outcome metrics and within condition comparisons between the TTA group and the control group for the unblinded flush, blinded eversion, and blinded inversion conditions separately. Bolded *p* values indicate significance. Abbreviations: (SE) standard error, (CI) confidence interval, (B) blinded, (TTA) individuals with transtibial amputation, (C) controls, (COM) center of mass, (deg) degrees, (cm) centimeters, (%BH) percent body height, (J/kg) joules per kilogram, (Ns/kg) Newton seconds per kilogram.

Metrics	Units	Mean (SE)						P value, [95% CI]		
		Flush		B. Eversion		B. Inversion		Flush	B. Eversion	B. Inversion
		TTA	Control	TTA	Control	TTA	Control	TTA-C	TTA-C	TTA-C
Coronal Hip Angle	deg	0.5 (1.0)	6.5 (1.0)	0.0 (1.3)	5.9 (1.3)	−2.5 (1.4)	5.1 (1.4)	**0.030, [-9.3, −2.5]**	**0.038, [-10.2, −1.5]**	**0.013, [-12.4, −3.0]**
Sagittal Hip Angle	deg	−4.4 (1.4)	−4.8 (1.3)	−3.4 (1.4)	−3.6 (1.3)	−3.7 (1.6)	−3.3 (1.5)	1, [-4.3, 5.1]	1, [-4.5, 4.9]	1, [-5.8, 4.9]
Sagittal Knee Angle	deg	6.8 (1.4)	3.1 (1.4)	7.0 (1.7)	3.7 (1.7)	9.7 (1.5)	3.5 (1.5)	0.51, [-1.1, 8.4]	0.53, [-2.6, 9.1]	0.096, [0.8, 11.5]
Recovery Step Width	cm	10.5 (1.0)	10.4 (1.0)	14.5 (0.9)	8.3 (0.9)	6.2 (1.0)	8.7 (0.9)	1, [-3.7, 3.9]	**0.016, [2.7, 9.5]**	0.38, [-6.0, 1.0]
Vertical COM	%BH	72.6 (0.3)	72.8 (0.3)	72.1 (0.3)	72.0 (0.3)	71.4 (0.3)	71.8 (0.2)	1, [-1.1, 0.7]	0.94, [-0.7, 1.0]	0.76, [-1.2, 0.5]
Negative Ankle Work	J/kg	−0.21 (0.02)	−0.17 (0.02)	−0.23 (0.02)	−0.16 (0.02)	−0.21 (0.02)	−0.17 (0.02)	0.66, [-0.10, 0.03]	**0.038, [-0.13, −0.02]**	0.27, [-0.10, 0.01]
Positive Ankle Work	J/kg	0.22 (0.02)	0.28 (0.02)	0.20 (0.02)	0.31 (0.02)	0.19 (0.02)	0.29 (0.02)	0.45, [-0.13, 0.01]	**0.038, [-0.19, −0.04]**	**0.01, [-0.17, −0.04]**
Negative Sagittal Knee Work	J/kg	−0.16 (0.01)	−0.14 (0.01)	−0.13 (0.02)	−0.15 (0.02)	−0.13 (0.01)	−0.13 (0.01)	1, [-0.07, 0.04]	0.65, [-0.03, 0.09]	1, [-0.06, 0.05]
Positive Sagittal Knee Work	J/kg	0.03 (0.09)	0.06 (0.01)	0.03 (0.01)	0.08 (0.01)	0.04 (0.01)	0.07 (0.01)	0.45, [-0.07, 0.00]	**0.038, [-0.07, −0.01]**	0.20, [-0.06, 0.00]
Negative Sagittal Hip Work	J/kg	−0.15 (0.02)	−0.17 (0.02)	−0.13 (0.02)	−0.15 (0.02)	−0.18 (0.02)	−0.18 (0.02)	1, [-0.05, 0.09]	0.89, [-0.05, 0.09]	1, [-0.08, 0.08]
Positive Sagittal Hip Work	J/kg	0.14 (0.01)	0.12 (0.01)	0.120 (0.01)	0.10 (0.01)	0.11 (0.02)	0.12 (0.02)	1, [-0.03, 0.05]	0.85, [-0.03, 0.07]	1, [-0.06, 0.05]
Negative Coronal Knee Work	J/kg	−0.02 (0.01)	−0.02 (0.01)	−0.01 (0.04)	−0.01 (0.00)	−0.05 (0.01)	−0.03 (0.01)	1, [-0.02, 0.02]	0.94, [-0.01, 0.02]	0.41, [-0.05, 0.01]
Positive Coronal Knee Work	J/kg	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.01 (0.00)	0.02 (0.01)	0.02 (0.01)	1, [-0.01, 0.01]	0.53, [-0.01, 0.00]	1, [-0.02, 0.02]
Negative Coronal Hip Work	J/kg	−0.02 (0.01)	−0.04 (0.01)	−0.03 (0.01)	−0.04 (0.01)	−0.02 (0.01)	−0.06 (0.01)	0.51, [-0.01, 0.05]	0.53, [-0.01, 0.04]	**0.01, [0.02, 0.05]**
Positive Coronal Hip Work	J/kg	0.08 (0.01)	0.10 (0.01)	0.07 (0.01)	0.09 (0.01)	0.08 (0.01)	0.11 (0.01)	0.94, [-0.05, 0.02]	0.53, [-0.07, 0.02]	0.32, [-0.06, 0.01]
Vertical GRF Impulse	Ns/kg	5.35 (0.16)	5.74 (0.16)	5.34 (0.20)	5.69 (0.20)	5.26 (0.21)	5.66 (0.20)	0.51, [-0.95, 0.16]	0.53, [-1.02, 0.32]	0.45, [-1.10, 0.30]

Table 2
Mean (SE) for Coronal Hip Angle and Recovery Step Width and within group comparisons between unblinded flush, blinded eversion, and blinded inversion conditions for the TTA group and control group separately. Bolded *p* values indicate significance. Abbreviations: (SE) standard error, (CI) confidence interval, (B) blinded, (F) flush, (BE) blinded eversion, (BI) blinded inversion, (TTA) individuals with transtibial amputation, (deg) degrees, (cm) centimeters.

Metric	Units	Mean (SE)						P value, [95% CI]			
		TTA			Controls			TTA		Controls	
		Flush	B. Eversion	B Inversion	Flush	B. Eversion	B Inversion	BE-F	BI-F	BE-F	BI-F
Coronal Hip Angle	deg	0.5 (1.0)	0.0 (1.3)	−2.5 (1.4)	6.5 (1.0)	5.9 (1.3)	5.1 (1.4)	0.77, [-1.9, 1.0]	**0.001, [-4.9, −1.3]**	0.61, [-2.0, 0.8]	0.18, [-3.1, 0.4]
Recovery Step Width	cm	10.5 (1.0)	14.5 (0.9)	6.2 (1.0)	10.4 (1.0)	8.3 (0.9)	8.7 (0.9)	**0.004, [1.2, 6.6]**	**0.001, [-6.8, −1.8]**	0.14, [-4.7, 0.5]	0.21, [-4.0, 0.7]

The surface conditions did not affect the vertical GRF impulses of the TTA group compared to the control group. Previous research reporting comparisons between the prosthetic limb vertical GRF impulses of individuals with unilateral transfemoral amputations versus controls report mixed findings. Kobayashi (Kobayashi et al., 2022) found the prosthetic limb vertical GRF impulses were significantly smaller than controls (and the intact limb) across a range of walking speeds whereas Zhang (Zhang et al., 2019) found the prosthetic limb and controls were similar (but the intact limb was larger) across a range of prosthetic limb alignments. In a study involving TTAs, Shell (Shell et al., 2017) showed decreasing prosthetic ankle–foot stiffness decreased prosthetic limb vertical GRF impulses. Despite differences in joint level and recovery strategies, the results presented here suggest that the TTA group and the control group generate similar vertical GRF impulses, indicating no differences in body support across surface conditions.

Limitations of this study should be considered while interpreting the results. Our unique study setup may not be generalizable for walking on the many different uneven surfaces in the built or natural environment. This study design also required a small step down to obscure the blinded eversion and blinded inversion conditions. Because of this we chose not to compare between surface conditions within each group for any sagittal plane metrics. Future studies with a similar design should add another surface condition that includes a small step down to the unblinded flush condition. Additionally, future study designs could explore the effect of multiple uneven steps on the biomechanical responses of individuals with TTA.

Another limitation was having to place the knee markers directly on the prosthetic socket instead of the skin of individuals with TTA. This could cause a slight misalignment with the true joint axis of rotation for the prosthetic limb, which could potentially lead to slightly altered kinematics at the knee and hip. To minimize these errors, we had the same trained researcher place markers on all subjects, however the use of digitized body landmarks could be explored in the future for more accurate marker placement.

This study gives insight and quantifies key biomechanical responses used by individuals with TTA when taking a single step on a coronally uneven and unexpected surface. During both blinded conditions, individuals with TTA adjusted their recovery step width significantly more than the control group and produced less positive ankle joint work. The blinded inversion condition presented the most difficulty for the TTA group, causing significantly increased hip abduction at mid-stance followed by a narrowed step width by recovery step heel strike when compared to flush. These biomechanical responses by the TTA group, when combined with less hip adduction on all surfaces, but especially during blinded inversion, and lower negative coronal hip joint work during blinded inversion when compared to controls, are suggestive of decreased balance control and a higher risk of falling when ambulating on uneven terrain.

Future prosthesis designs and rehabilitation strategies for individuals with TTA should take these results into consideration, expanding design criteria and therapies to include walking over uneven surfaces. This may help individuals with TTA maintain or increase participation in desired activities over a variety of terrain. Additionally, future work for clinical validation could explore the epidemiology of falls for individuals with TTA living in rural and urban areas to better understand frequency of steps on uneven terrain and their incidents of injury.

CRediT authorship contribution statement

Krista M. Cyr: Writing – review & editing, Writing – original draft, Methodology, Investigation, Formal analysis, Conceptualization. **Ava D. Segal:** Methodology, Investigation, Formal analysis, Writing - review & editing. **Richard R. Neptune:** Methodology, Investigation, Formal analysis, Conceptualization, Funding acquisition, Writing - review & editing. **Glenn K. Klute:** Supervision, Resources, Project administration, Methodology, Investigation, Formal analysis, Conceptualization, Funding acquisition, Writing - original draft, Writing - review & editing.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Kyle Yeates provided significant contributions in experimental design and data collection, and Jane Shofer performed the statistical analysis.

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