Performance of conventional and X2® prosthetic knees during slope descent

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ABSTRACT

Background: Individuals with transfemoral amputation often have difficulty descending sloped surfaces due to increased lower extremity range of motion and torque requirements. The X2®, a new microprocessor controlled prosthetic knee, claims to improve gait over sloped terrain. The aim of this study was to evaluate how experienced prosthesis users descended a sloped surface using the X2®, compared to a conventional knee, either mechanical (MECH) or microprocessor (MP).

Methods: Descent technique and biomechanics were assessed in 21 service members with unilateral transfemoral amputation as they descended an instrumented 10° slope at a self-selected walking velocity.

Findings: Use of the X2® in the MECH group resulted in greater hill assessment scores (8.5 to 11.0, $P=0.026$), due primarily to decreased reliance on handrail use. Use of the X2® in the MP group increased prosthetic knee flexion to a median of 6.4° at initial contact ($P=0.002$) and 73.7° in swing ($P=0.005$), contributing to longer prosthetic limb steps ($P=0.024$) and increased self-selected velocity ($P=0.041$). Additionally, use of the X2® in the MP group increased prosthetic limb impact peaks (11.6N/kg, $P=0.004$), improving impact peak symmetry to -1.3% ($P=0.004$).

Interpretation: Decreased reliance on handrail use as MECH users descended in the X2® indicate improved function and perhaps greater confidence in the device. Additional biomechanical improvements for existing MP users suggest potential longer-term benefits with regard to intact limb health and overuse injuries.

Keywords: Transfemoral amputation; Biomechanics; slope descent; gait analysis; ambulation

Abbreviations

TFA	Transfemoral amputation
X2®	Otto Bock X2® microprocessor controlled knee
MECH	Mechanically controlled conventional prosthetic knee
MP	Microprocessor controlled conventional prosthetic knee

1. INTRODUCTION

Lower extremity amputation can limit an individual's ability to perform a range of common daily activities (Hafner, et al., 2007). Slope walking, in particular, is often negatively affected by the loss of knee and/or ankle function (Gailey, et al., 2004). Slopes are commonly encountered in everyday walking, either in moderate gradients (i.e., $2.9 - 7.3^\circ$) as replacement for stairs per the Americans with Disabilities Act guidelines (Justice, 2010), or of more substantial gradations (i.e., greater than 7.3°) during other tasks or recreational activities. Relative to overground walking, studies of able-bodied participants navigating slopes demonstrate the importance of alternative motor control strategies, namely via increased knee flexion, for controlling vertical displacement of the center of mass (Lay, et al., 2006a; Lay, et al., 2006b; McIntosh, et al., 2006; Kuster, et al., 1995; Prentice et al., 2003). However, to account for altered prosthetic limb control, motion, proprioception and power generation, individuals with unilateral lower extremity amputations must adopt compensatory strategies to successfully descend slopes (Vickers, et al., 2008); for example, taking short but equal strides, side stepping, or using a zigzag method (Gailey, et al., 2004). As a result of these compensatory strategies, those with lower extremity amputations walk slower with asymmetrical temporal-spatial measures and impact forces (Vrieling et al., 2008).

Prosthetic device manufacturers continue to develop and release new prosthetic knee components in an effort to more effectively replicate the functionality of an intact knee. Few studies have, however, sought to determine the effect of these devices while walking on slopes (Bellmann, et al., 2010; Highsmith, et al., 2012; Lura, et al., 2015; Vrieling et al., 2008). As compared to mechanically (MECH) controlled devices, the use of stance flexion resistance in microprocessor (MP) controlled devices improve intact limb shock absorption in loading response (Vrieling et al., 2008). The X2® knee (Otto Bock Healthcare, GmbH, Duderstadt, Germany) is a device specifically developed for injured U.S. military service members and is similar to the commercially available Genium™ knee (Otto Bock Healthcare, GmbH, Duderstadt, Germany) with respect to control software and sensor hardware. The X2® uses variable flexion/extension

resistance which enables users to load the knee while flexed without collapse. Furthermore, it provides new functions and ambulation modes that utilize novel algorithms and sensors that are thought to provide a more accurate recoding of movement and might improve dynamic stability control (Bellmann, et al., 2012). Improved device control could produce more biomimetic timing, reducing the need for alternate walking strategies on slopes (Bellmann, et al., 2010; Vrieling et al., 2008). While the Genium™ knee has been reported to have greater prosthetic knee flexion in descent than MP devices (Lura, et al., 2015), the effects on descent technique and associated biomechanical changes are relatively unknown.

The aim of the current study was to evaluate how experienced prosthesis users descended a sloped surface while using the X2®, compared to a conventional knee, either mechanical (MECH) or microprocessor (MP). Self-selected descent techniques were compared and between-device biomechanical differences were examined. We hypothesized that compared to conventional knees (either MECH or MP), the X2® would allow the user to have more normalized gait patterns and biomechanics during slope descent, demonstrated by an increase in hill assessment index (Highsmith, et al., 2012), normalized sagittal mechanics (e.g. knee and hip flexion/extension), temporal-spatial, and loading parameters (e.g. peak and symmetry of step length, stance time, and limb loading). Overall, we hypothesized that the X2® should improve functionality during sloped walking for those with unilateral transfemoral amputation (TFA).

2. METHODS

2.1 Participants: Twenty-one young (18-40 years old) and active (K-level 4) individuals with unilateral traumatic TFA participated in the study at one of two testing sites (Table 1). All participants were more than two years post-amputation and had been using a MECH or MP device for a minimum of three months. Independent ambulation on stairs and slopes was required for inclusion into the study and was determined by the on-site, treating physical therapist. Participants did not have upper-extremity

amputations or contralateral injuries that prevented full sagittal plane lower extremity motion or strength. Participants did not have brain injuries resulting in functional impairment, were not receiving treatment for depression, and did not have uncorrected visual impairment. No participant had pain or discomfort which interfered with performance of functional activities. Prior to participation, all volunteers provided written informed consent as approved by onsite institutional review boards.

[Insert Table 1 about here]

2.2 Data Collection: Participants the same motion capture laboratory for two separate biomechanical assessments of slope descent, first wearing a conventional MECH or MP knee, previously prescribed by the participants' prosthetist, and then wearing the X2® (Table 1). Participants completed initial conventional device testing and were then fit with the X2® by a certified prosthetist at each site. After obtaining the X2®, participants were instructed by a physical therapist to ensure safe performance of advanced activities such as slope descent. The acclimation period for participants using the X2® as a primary device was a minimum of approximately 3 months (median (range) of 112 (86, 253) days) as determined by the clinical staff before returning to the same laboratory for reevaluation. The time period differed due to participant schedule availability.

It has been suggested that sloped gait requires a motor control strategy that differs from level walking but there is a paucity of information on the exact slope range where such control strategies shift. When taking one step up a ramp, participants have been seen to transition to an upslope motor control strategy (as evidenced by changes in limb positioning and sagittal trunk mechanics), when the slope is between 6-9° (Prentice et al., 2003). More recently, upslope transition has been suggested to occur at a lower (5.7° = 10% grade) slope when participants take multiple steps up a ramp (Lay et al., 2006a). This investigation assumed that utilization of a 10° slope was steep enough to trigger use of downslope motor control strategies. At least three trials were collected in each session as participants walked at a self-selected

velocity down an inclined walkway (7 m long, 1.22 m wide; AMTI, Inc., Watertown, MA, USA) set to a slope of 10 °. Handrail use was permitted, but participants were asked to only use assistance if needed. A qualitative evaluation of the self-selected method of hill descent was conducted by one rater at each site using the hill assessment index (Highsmith, et al., 2012). Both laboratories used an identical 6 degree of freedom cluster based full body marker set (Wilken, et al., 2012). In lieu of attaching the marker clusters used for able-bodied individuals onto the prosthesis, markers were placed on rigid areas of the prosthesis in a comparable location and pattern. A motion capture system consisting of at least 10 cameras (Vicon Inc. Oxford, UK or Motion Analysis Corp., Santa Rosa, CA, USA) was used to track markers at a frame rate of 120Hz. Ground reaction forces were simultaneously collected at 1200Hz from two force plates (AMTI, Inc., Watertown, MA, USA) embedded (front to back) in the middle of the walkway such that participants completed at least 3 steps on the ramp before hitting the force plates.

2.3 Data Analysis: A total of 21 participants (MECH: n = 8; MP: n = 13) completed both data collection sessions; however, use of handrails influenced what data were available for biomechanical analysis. As such, data from participants who self-selected to use handrails for support were not included in temporal-spatial, kinematic, or kinetic analyses (Table 1; see * denotation). Temporal-spatial, kinematic, and bodyweight-normalized kinetic parameters were calculated for 3-5 prosthetic and intact limb strides for each participant using Visual3D (C-Motion Inc., Germantown, MD, USA). Segment coordinate systems followed ISB recommendations (Wu and Cavanagh, 1995). Kinematic and ground reaction force data were filtered with a 4th order, zero lag Butterworth filter at 6 Hz and 50 Hz, respectively. All data were exported from Visual3D into MATLAB (The MathWorks, Natick, MA, USA) to identify and extract peak sagittal joint kinematics and kinetics at the ankle, knee and hip, as well as peak vertical ground reaction forces. The support moment for each limb was calculated as the sum of the hip and knee extensor moments and the ankle plantarflexor moment, and indicates the net response of the lower extremity. Metrics indicated as measured during early and late stance designate that peaks were obtained during 0-30 % and 35-75 % of the gait cycle, respectively.

This analysis utilized a symmetry index calculation to quantify between-limb differences in step length, stance time and vertical impact force values. All symmetry index calculations represented perfect symmetry with an index value of zero, percentage asymmetry was indicated by the magnitude of the symmetry index, and negative index measures indicated a larger prosthetic limb value (Davis, et al., 2004).

2.4 Statistical Analysis:

Normality of all variables was assessed using Shapiro-Wilk analysis. Participant demographics were normally distributed and compared using an independent samples t-test. Wilcoxon signed-rank tests were used to determine between-session differences by conventional knee-type using a significance value of $P \leq 0.050$ for hill assessment index, temporal-spatial, kinematic and kinetic outcome measures. Group results are displayed as median (min, max).

3. RESULTS

Demographic parameters (age, height, mass; Table 1) were not different (all $P \geq 0.111$) between conventional prosthetic knee groups. Acclimation time was not different ($P = 0.569$) between MECH and MP groups.

3.1 Self-Selected Method of Descent: Hill assessment index scores for the MECH group increased ($P = 0.026$) from 8.5 (5-11) at the initial session to 11 (10-11) with X2® use. In contrast, hill assessment index scores for the MP group were similar ($P = 0.495$) between sessions; 11 (9-11) at the initial visit and 11 (7-11) in an X2®.

3.2 Temporal-spatial:

Velocity of descent was similar ($P = 0.273$) for MECH participants utilizing the X2®. The MECH group did not change step length (all $P > 0.109$) nor average step length asymmetry (initial visit: 7.8 (2.8, 19.4) %; X2® 3.9 (-8.2, 9.4) %) between sessions. However, MP users increased ($P = 0.041$) velocity of descent and took longer ($P = 0.016$) prosthetic limb steps while using an X2®. This increase in step length for MP users did not influence ($P = 0.307$) a change in average step length asymmetry (initial visit: 6.0 (-36.0, 15.9) %; X2® 0.0 (-18.6, 7.4) %). Regardless of knee device used, no difference was observed in prosthetic or intact limb stance times between sessions (all $P \geq 0.181$). Stance time asymmetry also did not change (MECH: -10.7 (-24.8, -7.8) % and -7.9 (-13.5, -6.1) %; MP: -8.7 (-14.7, -1.8) % and -7.5 (-10.5, -2.6) %; for conventional and X2® visits respectively, all $P \geq 0.285$).

[Insert Table 2 about here]

3.3 Kinematics & Kinetics: Data from seven participants were not included in this analysis: Five participants who relied on handrail use for descent (Table 1; see * denotation) and two participants that descended without handrail assistance but had insufficient step lengths to allow the collection of kinetic data with only one foot on the plate (Table 1; see ϕ indicator). Ankle excursions for the intact and prosthetic limb were similar (all $P \geq 0.388$) between sessions in each group. Prosthetic knee flexion at initial contact ($P = 0.324$) and during swing ($P = 0.426$) were similar among the MECH group in the X2® device. However, MP users had greater prosthetic knee flexion at initial contact ($P = 0.002$) and at swing ($P = 0.005$) when utilizing the X2® device. Hip excursions on the prosthetic and intact limb were similar (all $P \geq 0.144$) between sessions.

Prosthetic and intact limb support moments did not change (all $P \geq 0.109$) for the MECH group during early or late stance. For the MP group, however, prosthetic limb support moment peaks were greater

when using the X2® in early ($P = 0.041$) and late stance ($P = 0.010$). Intact limb braking and propulsive forces did not change (all $P \geq 0.055$) for either group using an X2®.

Intact and prosthetic limb vertical impact peaks were similar for the MECH group (all $P \geq 0.109$) and impact peak symmetry did not change ($P = 0.109$) with X2® use (MECH: 16.1 (-4.1, 27.3) % and X2® 9.2 (-10.1, 13.8) %). During early stance, MP users experienced greater ($P = 0.004$) prosthetic limb vertical impact when using the X2® and a decrease ($P = 0.001$) in between-limb vertical impact peak asymmetry during early stance when wearing a conventional device compared to the X2® (MP: 12.1 (-12.3, 24.7) % and X2®: 0.0 (-18.6, 7.4) %).

[Insert Table 3 about here]

4. DISCUSSION

Conventional prosthetic knees typically do not allow those with unilateral TFA to descend a 10° slope without using compensatory gait mechanics or handrail assistance (Gailey, et al., 2004). The manufacturer claims that the X2® incorporates increased prosthetic knee flexion at prosthetic heel strike, improved stance flexion control, resistive flexion during stance, and adaptive swing phase control which should allow those with unilateral TFA to more easily handle advanced terrain such as downhill slopes ("Advanced prosthetic knee technology," 2012). MECH and MP devices have been available for years and differences between these conventional devices have been established in the literature (Bellmann, et al., 2010; Hafner, et al., 2007; Johansson, et al., 2005; Kahle, et al., 2008; Kaufman, et al., 2008; Orendurff, et al., 2006; Segal, et al., 2006; Vrieling et al., 2008). The current study was conducted to determine if use of the X2® improves overall slope descent mechanics by assessing self-selected methods of descent, and resulting temporal-spatial outcomes and joint mechanics.

While descending the slope in a MECH knee, 4 of the 8 participants relied on handrail assistance; whereas this number was decreased to only one participant requiring handrail assistance when using the X2® for slope descent. Participants using a MP device did not select to use handrail assistance. Accordingly, individuals using either MP device or the X2® rated at the top of the hill assessment index scale (11), indicating they descended independently without assistive device use while taking even steps. The median self-selected walking velocity of all 16 participants (MECH and MP) using the X2®, was 1.14 (0.88, 1.48) m/s, 0.58 m/s slower than the speed (1.72 (0.19) m/s) reported for a young (22.4 (3.0) years), able-bodied population walking down a 10° slope (McIntosh, et al., 2006). Although MP users in the X2® increased speed of descent and took longer prosthetic limb steps (Table 2), overall strides for both groups at both evaluations were less than the 1.50 (0.15) m reported for an able-bodied population (McIntosh, et al., 2006). Though we hypothesized improvements in temporal-spatial symmetry with the X2®, no changes in step length or stance time symmetry were seen.

Temporal-spatial parameters were likely influenced by differences in prosthetic knee flexion between conventional devices and the X2®. The MP group increased prosthetic knee flexion at initial contact which brought the MP user group median out of hyperextension and into flexion (Table 3). Flexion at initial contact for each conventional group in an X2® was more comparable to able-bodied individuals who were reported to land in 2.32 (5.65) ° of flexion when descending a similar (15% gradient = 8.5 °) slope (Lay, et al., 2006a). The MP group had considerably greater (11.6°) swing knee flexion while descending in the X2®. Although no differences were found between MECH and X2® prosthetic knee flexion, the swing flexion values of both knee type groups when using an X2® (Table 3) were comparable with reported values of 70.21 (3.09) ° for an able-bodied population descending a similar slope (Lay, et al., 2006a). The overall increase in prosthetic knee flexion at initial contact and swing for MP users could contribute to the gain in prosthetic limb step length. Increased prosthetic limb step lengths with the X2® suggest that MP users were able to accommodate a greater height difference with each prosthetic foot

placement (Vrieling et al., 2008), and supports the hypothesis that users would improve sagittal mechanics in slope descent with the X2®.

Though participants used the same prosthetic foot for both sessions, changes in prosthetic limb support could indicate changes in how participants load the prosthetic foot for increased energy return, or use the knee or hip. In both sessions, each group had intact limb support moments in early stance (Table 3) that exceeded values (MECH initial visit: 93.7% greater; MECH X2® 98.2% greater; MP initial visit: 34.2% greater; MP X2® 47.8% greater) of 1.11 (0.33) N•m/kg reported for able-bodied participants descending a similar slope (Lay, et al., 2006a). Similarly, intact limb support in late stance (Table 3) for both user groups exceeded values (MECH initial visit: 5.9% greater; MECH X2® 28.8% greater; MP initial visit: 15.0% greater; MP X2® 3.9% greater) of 1.53 (0.23) N•m/kg reported for able-bodied participants descending a similar slope (Lay, et al., 2006a). With X2® use, the MP users increased prosthetic limb support moments in late stance (Table 3), to a value that is similar to able-bodied average of 1.53 (0.23) N•m/kg (Lay et al., 2006a). Exceeding able-bodied averages could indicate a heightened use of the intact limb for support in descent.

To control the body's center of mass, downslope walking is characterized by greater peak braking vs. propulsive forces (Lay et al. 2006). Braking forces were not seen to change as either group transitioned to an X2®. Interestingly, intact limb braking forces for the MP group were greater (initial visit: 27.8% greater, X2® visit 18.1% greater) than respective averages of -1.55 N/kg obtained from an able-bodied population (Lay, et al., 2006a), indicating that the intact limb was relied upon more for control in descent. Between-limb vertical impact force asymmetry was apparent as users descended with conventional devices. The intact limb vertical impact forces among both groups, at both sessions, were larger than the reported average of 10.4 N/kg for an able-bodied population (Lay, et al., 2006a). This smoother transition may be due to use of resistive flexion control in the X2® during late stance. Increased prosthetic limb vertical impact peaks for MP users resulted in reduced impact force asymmetry and could signify that the X2®

allows for more normative loading methodology throughout weight acceptance, despite loading peaks remaining elevated relative to normative values. Increasing prosthetic limb utilization and more normative loading methodology throughout weight acceptance could reduce intact limb reliance and, in turn, lessen the need for compensatory gait mechanics in descent.

5. LIMITATIONS

Multiple factors may have influenced the study results, to include subject enrollment and testing across two sites, which has the potential to affect data consistency and therefore the ability to detect differences. The study team did, however, use consistent methods and the effects of multi-site collection were minimized by using a repeated measures design with all testing for an individual taking place at a single site. The study was also conducted in parallel with clinical care, and therefore participants were initially assessed in their clinically prescribed device that was fit by their personal prosthetist. Although the clinical rationale for prescribing the initial device was not determined, the experimental approach does allow between device comparisons that replicate a clinical transition from a MECH or MP device to a new system such as the X2®. As previously mentioned, the participants were allowed to use the handrails for assistance, although the rails were not instrumented. Thus, sample sizes were small for some variables due to the fact that participants were allowed to use the un-instrumented handrails for assistance. Finally, participants were young, active, and highly functional with traumatic amputations and therefore the results of this study may not be generalizable to all populations of people with amputation. Follow-up studies should control for the above noted variables and investigate how the X2 functions over slopes for people with TFA who are at a lower activity level.

6. CONCLUSION

The aim of the current study was to determine if use of the X2® improves overall slope descent mechanics by assessing self-selected technique of descent, and subsequent changes in temporal-spatial outcomes and joint mechanics. Although this analysis finds normalization of some temporal-spatial outcomes and joint mechanics were likely achieved due to the use of stance flexion resistance with the X2® device allowing for improved control lowering the body when both leading and trailing, some values continued to deviate from those of able-bodied individuals. Nevertheless, decreased reliance on handrail use as MECH users descended in the X2® suggest improved function and perhaps greater confidence in the device possibly reducing the risk of falling. Furthermore, overall reductions in intact limb loading and more symmetric loading at impact could indicate more normative loading patterns and a possible reduction of intact limb overuse during downslope walking. Reducing compensatory gait strategies during slope descent, perhaps through use of the X2®, could thus play a role in mitigating longer-term overuse injuries commonly associated with TFA.

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TABLE CAPTIONS

Table 1. Participant demographics, conventional knee type, and X2® acclimation time.

Table 2. Temporal-spatial results for participants that descended without handrail assistance. Median (range) data are presented for each session and organized by conventional knee type group.

Table 3. Kinematic and kinetic peak values for participants that descended without handrail use. Median (range) data are presented for each session and organized by conventional knee type group.

Table 1. Participant demographics, conventional knee type, and X2® acclimation time.

Participant	Collection Site	Age (years)	Mass (kg)	Height (m)	Conventional prosthetic knee		Acclimation time (days)
P01	BAMC	25	72.7	1.68	Otto Bock C-Leg®	Microprocessor	86
P02	BAMC	32	103.2	1.85	Otto Bock C-Leg®	Microprocessor	94
P03*	BAMC	36	93.8	1.78	Össur Total Knee ®	Mechanical	129
P04	BAMC	28	78.8	1.75	Otto Bock C-Leg®	Microprocessor	112
P05*	BAMC	39	104.0	1.83	Otto Bock C-Leg®	Microprocessor	105
P06	BAMC	27	82.8	1.85	Össur Total Knee ®	Mechanical	217
P07 ^φ	BAMC	33	83.7	1.75	Otto Bock C-Leg®	Microprocessor	103
P08	BAMC	36	93.1	1.83	Otto Bock C-Leg®	Microprocessor	120
P09	BAMC	23	80.1	1.88	Otto Bock C-Leg®	Microprocessor	122
P10	BAMC	26	74.1	1.80	Otto Bock C-Leg®	Microprocessor	90
P11	BAMC	25	81.9	1.68	Otto Bock C-Leg®	Microprocessor	112
P12	WRAMC	33	90.0	1.74	Otto Bock C-Leg®	Microprocessor	163
P13	WRAMC	38	85.6	1.82	Otto Bock C-Leg®	Microprocessor	188
P14	WRAMC	34	107.0	1.83	Otto Bock C-Leg®	Microprocessor	182
P15 ^φ	WRAMC	40	85.2	1.79	Össur Mauch ®	Mechanical	102
P16*	WRAMC	33	91.7	1.91	Össur Mauch ®	Mechanical	110
P17	WRAMC	39	84.0	1.87	Össur Mauch ®	Mechanical	98
P18*	WRAMC	28	70.0	1.77	Össur Mauch ®	Mechanical	253
P19*	WRAMC	33	104.9	1.86	Össur Mauch ®	Mechanical	91
P20	WRAMC	38	78.9	1.74	Össur Rheo Knee ®	Microprocessor	113
P21	WRAMC	40	99.9	1.78	Össur Mauch ®	Mechanical	92

Data available for each subject are indicated by symbol: * denotes that data was not used in biomechanical analysis (participants used hand rail assistance); ^φ indicates participants without kinetic data. Abbreviations: BAMC, Brooke Army Medical Center; WRAMC, Walter Reed Army Medical Center

Table 2. Temporal-spatial results for participants that descended without handrail assistance. Median (range) data are presented for each session and organized by conventional knee type group.

		Intact Limb						Prosthetic Limb					
		Initial visit			X2® visit			Initial visit			X2® visit		
		Media n	min	ma x	Media n	min	ma x	Media n	min	ma x	Media n	min	ma x
Mechanical (n = 4)													
Velocity	(m/s)	0.98	0.59	1.17	1.06	0.88	1.36						
Step Length	(m)	0.55	0.28	0.56	0.63	0.53	0.73	0.62	0.32	0.83	0.63	0.61	0.76
Stance Time	(s)	0.78	0.76	0.83	0.83	0.63	0.88	0.62	0.50	0.65	0.67	0.55	0.77
Microprocess or (n = 12)													
Velocity	(m/s)	1.04	0.77	1.36	1.14	0.93	1.48						
Step Length	(m)	0.57	0.45	0.71	0.63	0.48	0.74	0.65	0.24	0.79	0.76	0.69	0.48
Stance Time	(s)	0.78	0.67	0.94	0.75	0.68	0.83	0.66	0.56	0.85	0.64	0.57	0.74

‡P < 0.05 indicates a significant within-limb within-group difference between visits.

Table 3. Kinematic and kinetic peak values for participants that descended without handrail use. Median (range) data are presented for each session and organized by conventional knee type group.

				Intact Limb			Prosthetic Limb		
				Initial visit			X2@ visit		
Mechanical (kinematic n = 4, kinetic n = 3)				Median	min	max	Median	min	max
Ankle excursion	0-100% GC	(deg)		30.6	26.8	35.0	31.6	19.0	34.9
Initial knee flexion	0% GC	(deg)	*	5.7	-2.6	15.5	3.3	0.1	14.9
Max knee swing flexion	50-100% GC	(deg)	*	70.7	69.4	71.9	72.3	62.4	92.5
Hip excursion	0-100% GC	(deg)		32.7	26.2	44.5	25.3	19.2	27.0
Max support moment	0-30% GC	(N·m/kg)		2.15	0.85	2.33	2.20	1.55	2.55
Max support moment	35-75% GC	(N·m/kg)		1.62	0.98	1.67	1.97	1.53	2.03
Max braking force	0-30% GC	(N/kg)		-2.15	-2.21	-0.72	-1.15	-2.78	-0.77
Max propulsive force	30-80% GC	(N/kg)		1.66	1.07	2.22	2.00	1.54	2.51
Max vertical force	0-30% GC	(N/kg)		14.01	13.19	20.88	10.91	10.88	17.78
Max vertical force	35-75% GC	(N/kg)		8.13	7.93	10.49	8.24	7.87	11.30
Ankle excursion	0-100% GC	(deg)		31.7	24.1	44.3	33.8	16.9	42.8
Initial knee flexion	0% GC	(deg)	*	1.6	12.8	11.0	0.1	15.1	15.3
Max knee swing flexion	50-100% GC	(deg)	*	75.5	61.3	87.1	75.8	68.9	93.2
Hip excursion	0-100% GC	(deg)		26.9	16.8	38.2	29.9	21.2	34.4

				Initial visit			X2@ visit		
Microprocess or (kinematic n = 12, kinetic n = 11)				Median	min	max	Median	min	max
Ankle excursion	0-100% GC	(deg)		19.2	10.9	18.7	19.2	10.9	18.7
Initial knee flexion	0% GC	(deg)	*	-1.1	-3.0	5.5	-1.1	-3.0	5.5
Max knee swing flexion	50-100% GC	(deg)	*	75.2	70.0	77.2	75.2	70.0	77.2
Hip excursion	0-100% GC	(deg)		24.5	20.2	32.2	24.5	20.2	32.2
Max support moment	0-30% GC	(N·m/kg)		1.09	0.16	0.56	1.09	0.16	0.56
Max support moment	35-75% GC	(N·m/kg)		1.46	0.17	0.70	1.46	0.17	0.70
Max braking force	0-30% GC	(N/kg)		-0.89	-3.62	-0.32	-0.89	-3.62	-0.32
Max propulsive force	30-80% GC	(N/kg)		0.66	0.95	2.25	0.66	0.95	2.25
Max vertical force	0-30% GC	(N/kg)		13.37	10.13	14.30	13.37	10.13	14.30
Max vertical force	35-75% GC	(N/kg)		8.80	7.54	8.55	8.80	7.54	8.55
Ankle excursion	0-100% GC	(deg)		19.2	10.2	25.6	19.2	10.2	25.6
Initial knee flexion	0% GC	(deg)	*	6.4	0.3	2.2	6.4	0.3	2.2
Max knee swing flexion	50-100% GC	(deg)	*	73.7	46.0	74.3	73.7	46.0	74.3
Hip excursion	0-100% GC	(deg)		24.6	18.1	37.9	24.6	18.1	37.9

													0			
													.	1	1.7	
Max support moment	0-30% GC	(N·m/kg)	1.49	0.84	1.79	1.64	0.78	2.56	0.80	0.29	1.08	1.01	4	7	‡	
													0			
													.	7	1.8	
Max support moment	35-75% GC	(N·m/kg)	1.76	0.76	2.19	1.59	1.34	2.11	1.32	0.13	1.84	1.60	2	5	‡	
													-			
													2			
													.	0	0.6	
Max braking force	0-30% GC	(N/kg)	-1.98	3.10	1.30	-1.83	2.83	1.05	-1.02	1.82	0.28	-1.06	4	9		
													0			
													.	1	2.3	
Max propulsive force	30-80% GC	(N/kg)	2.31	1.64	3.90	2.21	0.84	3.53	0.66	0.16	1.53	0.91	5	7		
													9			
													.	8	18.	
Max vertical force	0-30% GC	(N/kg)	13.03	9.14	17.75	11.72	10.35	15.81	10.21	5.51	15.89	11.56	6	17	‡	
													6			
													.	7	10.	
Max vertical force	35-75% GC	(N/kg)	9.08	7.10	17.75	8.39	7.13	11.50	8.26	3.70	10.47	7.54	3	37		

‡P < 0.05 indicates a significant within-limb within-group difference between visits. Knee flexion is a positive value (*); All moments and forces are normalized by body mass. Abbreviations: max, maximum; min, minimum; GC, gait cycle.