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Spatio temporal parameters and symmetry in subjects ascending and descending a ramp, using three different prosthetic feet

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Abstract This study aimed at evaluating spatiotemporal parameters (STP) and symmetry index (SI), commonly used for evaluating amputee gait for routine clinical use, in individuals with unilateral transtibial amputations wearing energy storage and return (ESAR) feet with fixed ankles, prosthetic feet with adaptive ankles (PFAA) and prosthetic feet with microcontrolled adaptive ankles (PFAA-MC) in ramp ascent and descent. Thirteen individuals with transtibial amputations walked up and down a ramp. The STPs were measured in the amputated and intact legs and the relationship between them was quantified using the SI. The results showed that the use of PFAA-MC decreases walking speed in ramp descent ($P \leq 0.018$). However, this was the only parameter that showed a significant change. Hence, the differences in the amputees' gait pattern when using the above-mentioned prostheses may not be reflected by STP and their SI.

1. Introduction

The use of prosthetic devices aims at helping amputees to successfully return to their previous activities of daily living, including functional and leisure activities. However, in many cases, the prostheses do not fully replicate the physiological function of unimpaired limbs. In fact, in terms of lower limb unilateral transtibial amputations (UTTAs), the use of prosthesis has shown a decrement in walking speed [1] and an increment of the asymmetries between limbs in spatiotemporal parameters (STP) and kinetics of the joints [2] when compared to unimpaired people. These gait deviations may produce functional limitations in daily living.

A range of prosthetic feet may be prescribed for individuals with TTAs. One of these devices, called "Energy storage and return" (ESAR) feet can use the mechanical compression occurring at the time of loading response and midstance to store elastic energy that will be returned later during terminal stance and preswing. This mechanism has the objective of contributing some of the energy that the plantarflexors would normally provide at the time of push off [3], [4].

Other devices are manufactured with the aim of allowing for a larger ankle range of motion. One of these Prosthetic Feet with Adaptive Ankles (PFAA) is produced by Chas. A. Blatchford and Sons Ltd. (Basingstoke, UK): Echelon™, which allows for 6° and 3° of plantar and dorsiflexion respectively, relative to its neutral (standing) position [5]. More recently, prosthetic feet with microprocessor-controlled adaptive ankles (PFAA-MC) have been introduced, such as the Elan foot (Chas. A. Blatchford and Sons, Ltd., Basingstoke, UK). This foot includes sensors that can determine the incline of the terrain and the walking speed or cadence. Then, with this information, hydraulic damping settings could be adjusted to predefined values so as to facilitate an optimal foot to ground interaction for conditions such as 'ramp descent' and 'ramp ascend' [6].

ESAR and PFAA hydraulic prostheses have been studied and compared when used by the UTTA population while people walked on level ground [5], [7]–[10]. However, very few studies have investigated the effect of prostheses on other than level ground surfaces. Kristal [11] showed that the hydraulic foot decreases internal stresses in the residuum of transtibial amputation to ascend stairs. William et al [12] tested an PFAA developed by his team on incline walking for three subjects and found that their ability to adapt to the walking surface did not require active control. Agrawa et al [13] studied differences in symmetry in external work between four categories of prosthetic feet (including ESAR, and PFAA-MC) during ramp ascent and descent and found that prosthetic foot category appears to influence the symmetry more during descent walking. Struchkov and Buckley [6] determined that the use of PFAA-MC improved the gait biomechanics of ramp descent in comparison to conventional ankle–foot mechanisms (PFAA and ESAR), suggesting that it could reduce the biomechanical compensations used to walk down slopes.

For the clinical evaluation of the effect of different prosthetic feet, many tools are available, including questionnaires, STP, kinematic and kinetic parameters. A review on the biomechanical parameters used in the assessment of lower limb amputees [14] showed that the STP were the most commonly reported outcome measures. Given that the systems needed to measure STP and their symmetry index are appealing for clinical use (in terms of cost, easy to don and doff, time for measuring them and portability), they appear as a promising option for evaluations in clinical environments.

Interlimb asymmetry has been demonstrated for the stance time of UTTA, normally being the stance time longer on the intact limb [15], [16]. This could partly be due to subjects having more confidence in their intact leg. However, an increased stance phase and consequently loading period may lead to secondary pathologies in the intact limb [14]. Therefore achieving gait symmetry is one of the main concerns in UTTA [17], [18].

When using different prosthesis, the evaluation of STP and symmetry in the literature focused on the comparison between ESAR and PFAA, on level ground walking. Sedki and Moore [19] reported that patients using PFAA like Echelon foot have reported higher satisfaction levels in many facets of prosthetic use (during the activities of daily living). Moore [8] showed a statistically significant reduction in asymmetry of stance phase duration of level ground walking, when using hydraulic PFAAs prostheses when compared with ESARs. The author suggested that this could be one of the possible reasons for the increase in reported levels of satisfaction. Ko et al [20] evaluated ankle angle and external work symmetry on different terrains (level ground, 7° slope, 15 cm high stairs) and found an overall increase in symmetry in PFAA and PFAA-MC. The authors then discussed that this improvement in symmetry when using adaptive ankle feet may influence spatiotemporal parameters. A previous study [21] investigated the STP and symmetry index for amputees walking on level ground using different prosthesis. The results of the study did not show statistically significant differences for the conditions evaluated. However, the evaluation of STP on incline walking when using ESAR, PFAA and PFAA-MC is still pending. Hence, this paper aims at evaluating STP and their symmetry in subjects ascending and descending a 5° incline, using three different prosthetic feet (ESAR, PFAA, PFAA-MC).

2. Methods

2.1 Participants

Fourteen participants were involved in this study. The data of one of them was discarded due to loss of markers during the trials. Data from thirteen physically active, unilateral transtibial amputees (mean (SD)) age 38.23 (13.2) years, mass 75.1 (15.4) kg, height 1.76 (0.07) m), K2-K4 ambulatory level [22] was analysed. The participants had used their current foot for at least four months (10.8 (13.05) years, range 0.3-47 years). At the time of testing, six participants were using an ESAR foot, five used PFAA, one used PFAA-MC and one used PFAA and PFAA-MC. Each participant gave written informed consent prior to their involvement. The local ethics committee approval was obtained for the protocol.

2.2 Protocol and prosthetic intervention

2.2.1 The Ramp

The ramp (5 ° incline, 6 m long and 1m wide) was custom made and it incorporated a raised surface at its upper end to provide a stable area for resting and turning.

2.2.2 Prosthetic conditions

Subjects walked up and down the ramp using three different prostheses: one ESAR (Esprit), one PFAA (Echelon) and one PFAA-MC (Elan) from the Endolite family ((Chas. A. Blatchford and Sons Ltd., Basingstoke, UK).

The order of the prostheses used in the experiment depended on the type of prosthesis the participant used habitually (either ESAR or PFAA). If the participant habitually used an ESAR prosthesis, then they would start with a PFAA. And vice versa if they normally used a PFAA. In this way, a minimum familiarization time with the nonhabitual prosthesis, was ensured.

An experienced prosthetist was in charge of fitting the prosthesis and ensured the best possible alignment and setting adjustments for each prosthetic foot. The prosthetist also ensured that each foot was attached to the shank pylon with as close to the same alignment and set-up as possible. All the other components of the prosthesis (socket, suspension, shank pylon and its alignment) were not modified during the study. The settings which control the rates of articulation within the hydraulic feet (damping) were adjusted by the prosthetist until deemed to provide optimal function at self-selected, comfortable walking speed.

For the PFAA-MC device to detect the incline and swtich to a ramp mode a minimum of two steps are required. With the objective of ensuring that the prosthesis was in the appropriate ramp mode (either ascent or descent), it was switched remotely using a Bluetooth connection before the person started walking [6].

2.3 Data acquisition and processing

Participants walked up and down the ramp at self selected speed [23], using their own comfortable shoes. Kinematic data of six ramp ascent trials and six ramp descent trials were collected at 200 Hz. A motion capture system (Qualisys ProReflex, Göteborg, Sweden) with eleven cameras was used for data collection.

Reflective markers were placed on the lower limbs of the participants and on equivalent positions on the prosthetic limb. Two additional markers were placed on the toe and at equivalent position on the prosthesis (see Fig.1).

Gait events were estimated from kinematic data [24]. Spatio-temporal parameters (speed, step length, cycle time, step time, stance time and swing time) were computed using the software Visual3D (Version 6.01.08, C-Motion, Germantown, MD). The length of the step was expressed as a percentage of the participants height while the temporal parameters were normalized to the Cycle Time [25]. Symmetry Index (SI) for all the parameters was defined as [17] Eq. (1):

$$S.I. = \frac{\min(PR, PL)}{\max(PR, PL)} * 100 \quad (1)$$

Where PR and PL represent the gait parameter value for the right and left limb, respectively [17]. For each parameter and each prosthesis, a mean values was calculated for each participant.

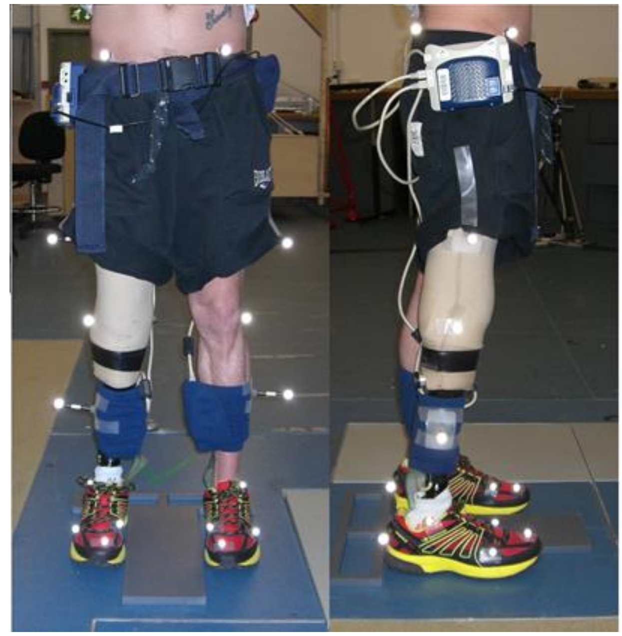

Fig. 1. Marker set used in the study.

2.4 Statistical analyses

As statistical descriptors, the median and the first and third quartiles (q1 and q3) of each outcome measure was used. Tthen a non-parametric analysis was carried out using the Friedman test. In case that the results showed a statistically significant difference, then a Dunn–Bonferroni post hoc test with adjustment for multiple comparisons [26] was performed. Statistical analysis was done using SPSS (23.0.0.0, IBM, Armonk New York U.S.A.). The alpha level set at 0.05. The effect size was calculated using the Kendall's W coefficient [27]. Values of W was interpreted as follows: W<0.11, very weak; 0.11≤W≤0.30, weak; 0.31≤ W ≤0.50, moderate; 0.51≤ W ≤0.70, strong; and W>0.71, very strong effect [28], [29].

3. Results and Discussion

3.1 Spatio Temporal Parameters (STP)

Table 1 shows the results of the STP for the sound and prosthetic limbs, for descending the ramp and for the three different prostheses. From the table it is possible to see that walking speed was faster when using ESAR, showing a difference of 0.15 m/s between ESAR and PFAA-MC, which is approximately a variation of 11.7%.

Statistically significant difference between the groups with a moderate size effect was found.

The Dunn's Post test indicated a statistically significant difference between ESAR and PFAA-MC ($p < 0.05$). PFAA-MC presented the minimum descending velocity 1.21 (1.09 1.31) m/s and ESAR showed the maximum 1.36 (1.12 1.4) m/s (see Fig. 2). Given that the working principle for PFAA-MC is to reduce the velocity of the first rocker and hence the velocity of the advancement of the tibia during ramp descent, a slower speed in descending the ramp could be expected, and could imply a better control of the speed of walking. It is also in accordance with the results found by Struchkov et al [6] who found that shank angular velocity during single-support was significantly lower for the PFAA-MC foot than either the PFAA or ESAR, with no significant differences between the later.

No other parameter showed statistically significant differences between the prosthesis, and the results are supported by a weak W value.

Table 1. Median value, first and the third quartile (q1 q3) for Step length (SL) expressed as % of body high (BH), Step Time (StT), Stance Time (ST), Swing time (SwT), expressed in % of Gait Cycle and Speed (m/s) for Ramp Descent

Ramp Descent					
Parameter	Prosthesis			p	W
	ESAR	PFAA	PFAA-MC		
Speed (m/s)	1.36	1.24	1.21	0.018	0.31
(q1 q3)	(1.12 1.40)	(1.11 1.30)	(1.09 1.31)		
Prosthetic limb					
SL (%)	40	39	39	0.526	0.05
(q1 q3)	(35 40)	(36 42)	(35 41)		
StT (%)	50	50	50	0.595	0.04
(q1 q3)	(49 51)	(49 51)	(49 51)		
ST (%)	65	65	65	0.332	0.09
(q1 q3)	(64 67)	(64 67)	(65 67)		
SwT (%)	35	35	35	0.294	0.09
(q1 q3)	(34 37)	(33 36)	(32 36)		
Sound limb					
SL (%)	38	36	37	0.728	0.02
(q1 q3)	(34 40)	(35 39)	(33 39)		
StT (%)	51	50	51	0.226	0.11
(q1 q3)	(50 51)	(49 51)	(50 51)		
ST (%)	66	66	67	0.135	0.15
(q1 q3)	(65 68)	(66 68)	(66 68)		
SwT (%)	35	33	34	0.207	0.12
(q1 q3)	(33 35)	(33 36)	(33 34)		

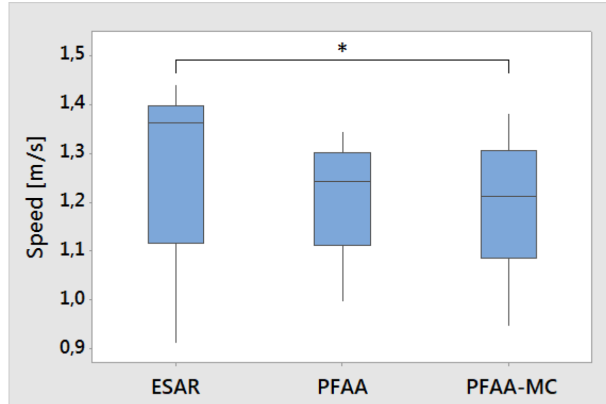

Fig. 2. Box plot of Walking Speed in descending the ramp for the three prosthesis conditions. Walking speed was the only parameter that showed a statistically significant differences between prostheses ($P<0.05$, W =0.31) and the post hoc analysis showed difference between ESAR y PFFA-MC, ($P<0.05$ shown by *)

Also the differences in the median values of the temporal parameters were within 2% of Gait Cycle (approximately 20 ms) and for Step Length, the difference was 2 % of Body Height, which implies a small variation in the parameters.

Table 2 shows the results for ascending the ramp. In this case, none of the median of the parameters showed statistically significant differences. The differences in the median values of the temporal parameters were within 1% of Gait Cycle and for Step Length, the difference was 4 % of Body Height.

Similar studies are scarce so comparison of results with others is limited. In particular, a full comparison of spatio temporal parameters between different prosthesis when walking on ramps was still pending, so a direct comparison of results is not possible.

Ko et al [20] presented qualitative results only for the prosthethic side of one person with amputation walking on ramps with different prosthesis. Their results on the prosthetic side showed a relationship between stance and swing phase of approximately 60 to 40% [20], which is similar to the results obtained in this study.

Other studies measured spatio temporal parameters in amputees walking on level ground and reached similar results [21], [30], [31]. In particular, Kovac et al [30] studied differences in STP between healthy and transtibial amputees with their current prosthetic foot. Their results showed values of walking speed of 1.3 (m/s), stance phase in the range between 65 and 67%, Swing Time between 33 and 35% and Step Length between 40 and 42, results very similar to the ones presented in this paper.

Also, and as reported before [15], [16], the stance phase on the prosthetic side is slightly shorter than on sound side, contributing to a more asymmetrical gait.

Table 2. Median value, first quartile and the third quartile (q1 q3) for Step length (SL) expressed as % of body high (BH), Step Time (StT), Stance Time (ST), Swing time (SwT), expressed in % of Gait Cycle and Speed (m/s) for Ramp Accent.

Ramp Ascent					
Parameter	Prosthesis			p	W
	ESAR	PFAA	PFAA-MC		
Speed (m/s)	1.31	1.22	1.25	0.05	0.23
(q1 q3)	(1.13 1.40)	(1.11 1.34)	(1.09 1.36)		
Prosthetic limb					
SL (%)	44	40	40	0.146	0.15
(q1 q3)	(37 46)	(37 46)	(36 45)		
StT (%)	50	50	50	0.900	0.01
(q1 q3)	(49 51)	(49 52)	(49 51)		
ST (%)	66	66	66	0.690	0.03
(q1 q3)	(64 69)	(64 68)	(65 69)		
SwT (%)	34	34	34	0.657	0.03
(q1 q3)	(32 36)	(32 35)	(32 35)		
Sound limb					
SL (%)	39	39	40	0.746	0.02
(q1 q3)	(37 43)	(37 41)	(36 41)		
StT (%)	50	50	50	0.227	0.11
(q1 q3)	(49 51)	(49 51)	(49 51)		
ST (%)	67	68	67	0.469	0.06
(q1 q3)	(66 68)	(66 69)	(66 69)		
SwT (%)	32	32	33	0.616	0.04
(q1 q3)	(32 34)	(31 34)	(30 34)		

3.2 Symmetry Index (SI)

Table 3 shows the results of the median, q1 and q3, statistical p values and sample effect size (W) of the symmetry index for each STP. Results showed median values of SI was at least 90% for all conditions (ascent or descent, prostheses and all STP).

Similar results have been reported in the literature. For example, symmetry values greater than 88% have been reported in UTTA [17], [21], [32], [33], while results greater than 94% were found for unimpaired subjects [33], [34].

The differences between prosthesis in SI were small (around 2%) and the results from the statistical test showed no signficiant difference between them. CT and ST showed a statistically significance differences when comparing the three prostheses however when performing the post hoc analysis, the comparison between the pairs reported non-significant differences. For CT and ST, the difference between each pair was for all cases $P > 0.05$. This effect of finding a statistically significant difference when performing the Friedman test, but not finding significant differences between the pairs could due to the Bonferroni correction, in which case the results are considered not statistically different [35].

The lack of statistical significance may be explained by a number of reasons, some of which are limitations of the study. In general, the effect size was weak and in many cases, it was very weak. This could have been influenced by different variables. A larger sample may be needed to observe differences. Also, the functional capabilities of participants of this study ranged from the ability to walk independently (K2) to the ability to participate in professional triathlons (K4). This variability in the mobility ability of the participants may disguise real differences in walking patterns. It should also be considered that the action of prostheses on the gait of transtibial amputees is quantitatively reflected differently according to the parameter studied.

Table 3 Median values, first quartile and the third quartile (q1 q3) of SI for Step length (SL), Step Time (StT), Stance Time (ST), Swing time (SwT) and Cycle Time (CT).

Parameter	Prosthesis			p	W
	ESAR	PFAA	PFAA-MC		
Descent					
SL (%)	93	92	94	0.728	0.02
(q1 q3)	(89 95)	(88 96)	(89 96)		
StT (%)	95	95	95	0.513	0.05
(q1 q3)	(94 96)	(94 96)	(94 96)		
ST (%)	97	97	96	0.296	0.09
(q1 q3)	(96 98)	(95 97)	(96 98)		
SwT (%)	94	94	94	0.414	0.07
(q1 q3)	(92 96)	(93 96)	(93 96)		
CT (%)	98	98	98	0.011	0.35
(q1 q3)	(97 98)	(97 98)	(98 99)		
Ascent					
SL (%)	90	90	91	0.375	0.08
(q1 q3)	(86 94)	(88 92)	(88 93)		
StT (%)	95	95	95	0.285	0.10
(q1 q3)	(94 96)	(93 96)	(94 97)		
ST (%)	97	96	96	0.014	0.33
(q1 q3)	(96 98)	(95 97)	(95 97)		
SwT (%)	94	93	94	0.933	0.01
(q1 q3)	(90 95)	(90 95)	(93 96)		
CT (%)	98	98	98	0.303	0.09
(q1 q3)	(98 99)	(98 98)	(97 98)		

Finally, it has been suggested that experienced prosthetic users (as the ones that participated in this study) may quickly adapt to modifications performed in their prosthesis, and this would not be reflected as a change in their pattern [36]. In fact, Varrecchia et al [37] studying STP in transfemoral amputees when using different prostheses found statistically significant differences only for step length in the sound limb.

This could imply that gait retraining may be needed in order to exploit any prosthetic modification and also to modify the gait pattern. And probably only then statistical differences between prosthesis may be found. In this study, only one of the participants used two of the three types of prosthesis evaluated. Therefore, it may be necessary to provide appropriate retraining to the subjects before the measurements.

From the results of this study and under the set conditions, the STP did not varied when using different prosthesis, except for walking speed, even when the parameters studied in this research are those commonly reported [14] and it has been shown that transtibial amputees present statistically significant changes in the parameters when compared to unimpaired people [37]. In this study, there was a speed difference of 0.15 m/s between ESAR and PFAA-MC, (which represents approximately a variation of 11.7%) for descending the ramp. Gard [36] suggested that walking speed probably provides a better indication of a person´s walking ability than any other quantitative gait measure. And this may be reflected in its ability to report changes between prostheses even when gait retraining was not provided.

Other biomechanics variables, such as joint angle, joint moments or power may reflect better the difference in gait pattern due to a change in prosthesis.

4. Conclusions

This paper aimed at evaluating STP and their SI in subjects with unilateral transtibial amputation when ascending and descending a 5° incline, using three different prosthetic feet (ESAR, PFAA, and PFAA-MC).

Walking speed showed differences between the prostheses when subjects descended the ramp. However, this was the only parameter that showed a significant change. Hence, the differences in the amputees' gait pattern when using the above-mentioned prostheses may not be reflected by STP and their SI in the conditions proposed in this study.

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References

[1] Y. Hermodson, C. Ekdahl, B. M. Persson, and G. Roxendal, “Gait in male trans-tibia1 amputees : a comparative study with healthy subjects in relation to walking speed,” pp. 68–77, 1994.

[2] G. R. B. Hurley, R. Mckenney, M. Robinson, M. Zadravec, and M. R. Pierrynowski, “The role of the contralateral limb in below-knee amputee gait,” Prosthet. Orthot. Int., vol. 14, no. 1, pp. 33–42, 1990.

[3] B. J. Hafner, “Clinical Prescription and Use of Prosthetic Foot and Ankle Mechanisms: A Review of the Literature,” JPO J. Prosthetics Orthot., vol. 17, no. Supplement, pp. S5–S11, 2005.

[4] R. J. Zmitrewicz, R. R. Neptune, J. G. Walden, W. E. Rogers, and G. W. Bosker, “The Effect of Foot and Ankle Prosthetic Components on Braking and Propulsive Impulses During Transtibial Amputee Gait,” Arch. Phys. Med. Rehabil., vol. 87, no. 10, pp. 1334–1339, 2006.

[5] A. R. De Asha, L. Johnson, R. Munjal, J. Kulkarni, and J. G. Buckley, “Attenuation of centre-of-pressure trajectory fluctuations under the prosthetic foot when using an articulating hydraulic ankle attachment compared to fixed attachment,” Clin. Biomech., vol. 28, no. 2, pp. 218–224, 2013.

[6] V. Struchkov and J. G. Buckley, “Biomechanics of ramp descent in unilateral trans-tibial amputees: Comparison of a microprocessor controlled foot with conventional ankle-foot mechanisms.,” Clin. Biomech., vol. 32, pp. 164–170, Dec. 2015.

[7] L. Johnson, A. R. De Asha, R. Munjal, J. Kulkarni, and J. G. Buckley, “Toe clearance when walking in people with unilateral transtibial amputation: Effects of passive hydraulic ankle,” vol. 51, no. 3, pp. 429–438, 2014.

[8] R. Moore, “Effect on Stance Phase Timing Asymmetry in Individuals with Amputation Using Hydraulic Ankle Units,” J. Prosthetics Orthot., vol. 28, no. 1, pp. 44–48, 2016.

[9] A. R. De Asha, R. Munjal, J. Kulkarni, and J. G. Buckley, “Impact on the biomechanics of overground gait of using an ‘Echelon’ hydraulic ankle-foot device in unilateral trans-tibial and trans-femoral amputees,” Clin. Biomech., vol. 29, no. 7, pp. 728–734, 2014.

[10] C. Y. Ko et al., “Biomechanical features of level walking by transtibial amputees wearing prosthetic feet with and without adaptive ankles,” J. Mech. Sci. Technol., vol. 30, no. 6, pp. 2907–2914, 2016.

[11] A. Kristal, “Kinematics, kinetics and internal mechanical stresses of transtibial amputees walking and climbing stairs with hydraulic feet.,” in Orthopaedie + Reha-Technik Leipzig, Germany., 2012, p. Lecture 3–6.

[12] R. J. Williams, A. H. Hansen, and S. A. Gard, “Prosthetic Ankle-Foot Mechanism Capable of Automatic Adaptation to the Walking Surface,” J. Biomech. Eng., vol. 131, no. 3, p. 035002, 2009.

[13] V. Agrawal, R. S. Gailey, I. A. Gaunaurd, C. O’Toole, A. Finnieston, and R. Tolchin, “Comparison of four different categories of prosthetic feet during ramp ambulation in unilateral transtibial amputees,” Prosthet.

Orthot. Int., vol. 39, no. 5, pp. 380–389, 2015.
[14] Y. Sagawa, K. Turcot, S. Armand, A. Thevenon, N. Vuillerme, and E. Watelain, “Biomechanics and physiological parameters during gait in lower-limb amputees: A systematic review,” Gait Posture, vol. 33, no. 4, pp. 511–526, 2011.
[15] W. J. Board, G. M. Street, and C. Caspers, “A comparison of trans-tibial amputee suction and vacuum socket conditions,” Prosthet. Orthot. Int., vol. 25, no. 3, pp. 202–209, 2001.
[16] E. Isakov, H. Burger, J. Krajnik, M. Gregoric, and C. Marincek, “Influence of speed on gait parameters and on symmetry in trans-tibial amputees.,” Prosthet. Orthot. Int., vol. 20, no. 3, pp. 153–158, 1996.
[17] G. N. S. Marinakis, “Interlimb symmetry of traumatic unilateral transtibial amputees wearing two different prosthetic feet in the early rehabilitation stage,” J. Rehabil. Res. Dev., vol. 41, no. 4, p. 581, 2004.
[18] L. Nolan, “The functional demands on the intact limb during walking for active trans- femoral and trans- tibial amputees,” Prosthet. Orthot. Int., vol. 24, pp. 117–125, 2000.
[19] I. Sedki and R. Moore, “Patient evaluation of the Echelon foot using the Seattle Prosthesis Evaluation Questionnaire,” Prosthet. Orthot. Int., vol. 37, no. 3, pp. 250–254, 2013.
[20] C. Y. Ko et al., “Comparison of ankle angle adaptations of prosthetic feet with and without adaptive ankle angle during level ground, ramp, and stair ambulations of a transtibial amputee: A pilot study,” Int. J. Precis. Eng. Manuf., vol. 15, no. 12, pp. 2689–2693, 2014.
[21] M. Riveras, E. Ravera, A. F. Shaheen, D. Ewins, P. C. Formento, and O. Verde, “Spatio Temporal Parameters and Symmetry Index in Transtibial Amputees Wearing Prosthetic Feet with and without Adaptive Ankles,” 2019 Glob. Med. Eng. Phys. Exch. Pan Am. Heal. Care Exch., pp. 1–6, 2019.
[22] U. G. P. Office, “HCFA Common Procedure Coding System HCPCS 2001,” 2001.
[23] A. R. De Asha, R. Munjal, J. Kulkarni, and J. G. Buckley, “Walking speed related joint kinetic alterations in trans-tibial amputees: impact of hydraulic ‘ankle’ damping.,” J. Neuroeng. Rehabil., vol. 10, no. 1, p. 107, 2013.
[24] J. A. Zeni, J. G. Richards, and J. S. Higginson, “Two simple methods for determining gait events during treadmill and overground walking using kinematic data,” Gait Posture, vol. 27, no. 4, pp. 710–714, May 2008.
[25] J. Cámara Tobalina, “Gait analysis: phases and spatio-temporal variables,” Entramado, vol. 7, no. 1, pp. 160–173, 2011.
[26] O. J. Dunn, “Multiple comparisons using rank sums,” Find this Artic. online, no. April 2013, pp. 37–41, 1964.
[27] M. Tomczak and E. Tomczak, “The need to report effect size estimates revisited. An overview of some recommended measures of effect size,” Trends Sport Sci., vol. 1, no. 21, pp. 19–25, 2014.
[28] C. Roy, “Managing Delphi surveys using nonparametric statistical techniques,” Decis. Sci., vol. 28, no. 3, pp. 763–774, 1997.
[29] C. P. Orsbon, R. S. Kaiser, and C. F. Ross, “Physician opinions about an anatomy core curriculum: A case for medical imaging and vertical integration,” Anat. Sci. Educ., vol. 7, no. 4, pp. 251–261, 2014.
[30] I. Kovac, V. Medved, and L. Ostoji, “Spatial, temporal and kinematic characteristics of traumatic transtibial amputees’ gait.,” Coll. Antropol., vol. 34, no. Supplement 1, pp. 205–213, 2010.
[31] S. J. Olney, “Kinematic and Kinetic Variations of Below-Knee Amputee Gait Take Online Quiz for PCEs,” vol. 14, no. 1, pp. 1–13, 2014.
[32] M. J. Highsmith, B. W. Schulz, S. Hart-Hughes, G. A. Latlief, and S. L. Phillips, “Differences in the spatiotemporal parameters of transtibial and transfemoral amputee gait,” J. Prosthetics Orthot., vol. 22, no. 1, pp. 26–30, 2010.
[33] L. Nolan, A. Wit, K. Dudziñski, A. Lees, M. Lake, and M. Wychowañski, “Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees,” Gait Posture, vol. 17, no. 2, pp. 142–151, Apr. 2003.
[34] M. Blazkiewicz, I. Wiszomirska, and A. Wit, “Comparison of four methods of calculating the symmetry of spatial-temporal parameters of gait,” Acta Bioeng. Biomech., vol. 16, no. 1, pp. 29–35, 2014.
[35] A. Dinno, “Nonparametric pairwise multiple comparisons in independent groups using Dunn ’ s test LP the Stata Journal at,” Stata J., vol. 15, no. 1, pp. 292–300, 2015.
[36] S. a. Gard, “Use of Quantitative Gait Analysis for the Evaluation of Prosthetic Walking Performance,” JPO J. Prosthetics Orthot., vol. 18, pp. P93–P104, 2006.
[37] T. Varrecchia et al., “Common and specific gait patterns in people with varying anatomical levels of lower limb amputation and different prosthetic components,” Hum. Mov. Sci., vol. 66, no. October 2018, pp. 9–21, 2019.

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