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Margins of stability of persons with transtibial or transfemoral amputations walking on sloped surfaces

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ABSTRACT

Gait is a complex motor skill. However, most falls in humans occur during gait, and people with lower limb amputation have an increased risk of falls. Thus, this study evaluated the stability of persons with unilateral amputation by quantifying the margin of stability (MoS) during gait, to contribute to understanding the strategies adopted by these people to reduce falls. The participants were divided into 3 groups: persons with transtibial amputations (n = 12, 32.27 ± 10.10 years, 76.9 ± 10.3 kg, 1.74 ± 0.06 m); persons with transfemoral amputations (n = 13, 32.21 ± 8.34 years, 72.55 ± 10.23 kg, 1.73 ± 0.05 m); and controls (n = 15, 32.2 ± 10.17 years, 75.4 ± 9.25 kg, 1.75 ± 0.05 m), who walked for 4 min on a level and sloped (8% down and up) treadmill. The pelvic and foot marker kinematic data were used to estimate the center of mass and base of support, and from these, the MoS was estimated. Although both groups of persons with amputations showed higher values for the ML MoS than did the control group (transtibial: 8.81 ± 1.79, 8.97 ± 1.74, 8.79 ± 1.76, transfemoral: 10.15 ± 2.03, 10.60 ± 1.98, 10.11 ± 1.75, control: 8.13 ± 1.30, 7.18 ± 1.85, 8.15 ± 1.57, level, down, and up, respectively), only the transfemoral group presented a significant higher value compared to the control group. Our findings suggest that the documented limitations in persons with amputations, especially with transfemoral amputation, are exacerbated in situations that require more skills, such as walking on sloped surfaces, triggering protective mechanisms.

1. Introduction

During gait, perturbations arise from internal and external sources (Bruijn et al., 2013). Thus, some individual capabilities are necessary to optimally control important factors for a safe locomotion, including balance and stability, which, in this study, refers to the dynamic ability to recover from perturbations and avoid falls (Bruijn et al., 2013; Stergiou and Decker, 2011). This is especially important for people with lower limb amputations since this population has a higher risk of falls than do healthy people (Gooday and Hunter, 2004; Sheehan et al., 2016). This situation depends on the level of amputation as people with a transfemoral amputation (TFA) present greater risk of fall than those with a transtibial amputation (TTA) (Miller et al., 2001).

Maintaining stability during walking is difficult for people with lower limb amputations. Ankle control plays an important role in medial–lateral (ML) balance, and the prosthetic side lacks distal muscles and afferent feedback from the lower limb (Gates et al., 2013; Viton et al., 2000), which can limit mobility, so that people with lower limb amputations face several walking challenges (Grumillier et al., 2008; Van Velzen et al., 2006). Thus, individuals with lower limb amputations are considered less stable than are healthy individuals (Gates et al., 2013), especially in situations with greater demands, such as slopes (Sturk et al., 2019; Vrieling et al., 2008).

Sloped surfaces, compared to level surfaces, demand different behaviors to maintain walking stability. Results have been specifically found for people with amputation such as slower walking speed, smaller step length, especially for intact limb

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during uphill walking due to reduced prosthetic limb push-off, wider step width, reduced foot clearance (Fradet et al., 2010; Heitzmann et al., 2013; Morgenroth et al., 2018; Villa et al., 2017; Wolf et al., 2012), especially in people with transfemoral amputation that were not able to increase the angle of prosthetic knee flexion in either upward and downward walking (Vrieling et al., 2008).

Several methods have been proposed in the literature for objectively estimating gait stability (Bruijn et al., 2013; Vieira et al., 2017), including the margin of stability (MoS) (Hof et al., 2007, 2005). The MoS is calculated as the distance between the extrapolated center of mass (XCoM) and the limits of the base of support (BoS), where the XCoM represents the state of the center of mass (CoM) as a function of both its position and velocity (Hof et al., 2005). To preserve stability, in the ML direction, individuals should avoid that the XCoM exceed the lateral BoS edge. In the anterior-posterior (AP) direction, the XCoM should always be ahead of the posterior BoS edge to prevent a backward (BW) fall (Hak et al., 2013b), as decreasing the risk of a backward fall appears to be prioritized over decreasing the risk of a forward fall (Hak et al., 2013b). Thus, a larger MoS can indicate that either the individual is exhibiting greater stability or a strategy was adopted to increase stability, as observed in individuals with compromised stability (Hak et al., 2013b; Hof et al., 2007).

Few studies have assessed gait stability of people with amputation in challenging situations using the MoS (Gates et al., 2013; Hak et al., 2013b; Sinitski et al., 2021; Sturk et al., 2019). Although there are common findings for amputee groups, such as greater ML MoS, smaller BW MoS, asymmetric changes in ML MoS on the prosthetic and intact side, lower walking speed, increased step width for amputee groups that further increased in challenging situations, these studies are not completely comparable. Only Sturk et al. (2019) tested the individuals on sloped surfaces, however only TFA individuals were tested and the MoS was calculated only in ML direction. Some results are contrasting: in the challenging situations, Gates et al. (2013) reported that TTA individuals decreased their ML MoS on their prosthetic limbs, but Sturk et al. (2019) reported that TFA individuals increased their ML MoS on their prothetic limbs; however the level of amputation and the protocols were different. Gates et al. (2013), Hak et al. (2013), and Sinitski et al. (2021) tested only TTA individuals and they did not include sloped surfaces in the protocols. Only Hak et al. (2013) computed the MoS in both ML and BW directions. Hak et al. (2013), Sturk et al. (2019), and Sinitski et al. (2021) conducted the experiments in a virtual reality environment, and Sturk et al. (2019) recommended that future studies should verify the results in a nonvirtual environment.

Therefore, to fill some gaps in literature concerning the level of amputation, the inclination of the surface, and the direction in which the stability descriptor is calculated, this study aimed to assess gait stability in people with transtibial amputations (TTA), people with transfemoral amputations (TFA), and a control group (CT) walking on level and sloped surfaces using the MoS, calculated in both backward (BW) and medial–lateral (ML) directions. Based on previous studies and on the second MoS interpretation, we hypothesized that people with amputation present a greater MoS and that sloped surfaces will further increase it, especially in the downward condition, which presents higher biomechanical demands. Furthermore, we hypothesized that a higher level of amputation is directly related to a larger increase in the MoS.

2. Methods

2.1. Participants

A total of 40 age-matched subjects (12 with TTA, 13 with TFA, and 15 CT) participated in this study. Table 1 shows an overview of each group's characteristics. The lower limb amputations were due to injuries in all cases (25). Eight people with amputations had undergone amputations of the right limb, and seventeen people with amputations had undergone amputations of the left limb. All the people with amputations wore their prostheses daily and walked with good experience, without musculoskeletal impairment on the intact side. They reported making adjustments/alignments of the prosthesis by a prosthetist regularly. Eleven of them reported having experienced falls in the past twelve months. The inclusion criteria for the TTA and TFA groups were as follows: to have undergone an amputation of the lower limb unilaterally at the transfemoral or transtibial level (traumatic amputation), to be in the age group from 18 to 55 years old, to walk without assistance, to have no skin lesions, and to be free from phantom sensation or pain. The exclusion criteria were vestibular problems, known neurological dysfunction, not making prosthesis alignment regularly, severe visual impairment, impaired cognitive function, a recent history of trauma, a peripheral arterial disease affecting the lower limbs, fractures and surgeries in the lower limb (self-reported). The inclusion criteria for the CT group were as follows: to be in good health, to have the ability to walk independently without an assistive device, to be without neurological impairments, to be without a history of musculoskeletal surgery, to be without any injury or pain at the time of data collection, and to have age and anthropometric characteristics matched with those of the participants with amputation (Table 1). The participants voluntarily signed an informed consent form. Next, they participated in testing protocols previously approved by the Local Research Ethics Committee (approval number: 1.003.935–2018).

The TFA, TTA, and CT groups did not present significant differences in age, body mass, height, BMI, amputation time, time using prosthesis, or residual limb length (Table 1).

2.2. Equipment

For all the participants, reflective markers were attached to the lateral malleoli, the heels, and the top of the head of the second and fifth metatarsals. For the margin of stability calculation, four reflective markers in a squared cluster were attached in the lumbar region (between the left and right posterior superior iliac spines) for estimating the CoM. The markers were used for gait assessment, and a kinematic analysis was performed using a 3D motion capture system consisting of 10 infrared cameras and a sampling rate of 100 samples/s (Vicon Nexus, Oxford Metrics, Oxford, UK).

2.3. Protocol

The participants walked on a level and sloped treadmill at their preferred walking speed (PWS). First, their PWS was determined following a previously reported protocol (Kang and Dingwell, 2008) at each inclination level. Next, each participant performed a total of three 4-min trials (one in each condition), walking in treadmill inclinations of −8% (DOWN – downward walking), 0% (HOR-horizontal), and 8% (UP- upward walking) in a randomized order. The participants rested for 2 min between trials. An inclination of 8% is the standard for civil engineering and architecture

Table 1
Participant characteristics (mean ± standard deviation).

	Transfemoral	**Transtibial**	**Control**	***p***
N/sex	13 (1♀,12♂)	12 (2♀,10♂)	15 (2♀,13♂)	NA
Age (years)	32.2 ± 8.3	32.3 ± 10.1	32.2 ± 10.2	0.746
Mass (kg) *	72.6 ± 10.2	76.9 ± 10.3	75.4 ± 9.3	0.576
Height (m)	1.73 ± 0.05	1.74 ± 0.06	1.75 ± 0.05	0.825
BMI (kg/cm^2)	23.8 ± 2.6	25.6 ± 2.9	24.7 ± 3.1	0.473
Time since amputation (years)	11.1 ± 7.8	10.8 ± 6.7	NA	0.783
Time using prosthesis (years)	9.4 ± 5.1	9.8 ± 7.9	NA	0.739
Residual limb length (cm)	36.1 ± 7.8	26.6 ± 13.9	NA	0.218
Hydraulic/Mechanical Knee	3R15/3R20 Ottobock	NA	NA	NA
Dynamic Foot	1D10 Ottobock	1D10 Ottobock	NA	NA

BMI: Body mass index, * including prosthesis, NA- not applicable. *p*: one-way Anova.

structures (Americans with Disabilities Act homepage, 2019, Secretaria Nacional de Promoção dos Direitos da Pessoa com Deficiência, 2019).

2.4. Data analysis

Before data analysis was performed, the kinematic data were low-pass filtered with a fourth-order, zero-lag, Butterworth filter with a cutoff frequency of 8 Hz. First, the initial and final 15 s of each trial were discarded (Hak et al., 2013a), after which all steps were detected according to the zero-cross of the heel-marker velocity in the AP direction (Souza et al., 2017). The intermediate 150 strides were then selected, and the initial and final strides exceeding 150 were excluded. The data analysis was performed using a custom MATLAB (version 2018a, MathWorks, Natick, MA) code.

2.4.1. Spatiotemporal parameters

To interpret the MoS results, as MoS depends on the step width (Hof et al., 2005), step length, and step frequency (Hak et al., 2013a), related spatiotemporal parameters were computed. First, step frequency (SF) was determined as the inverse of the average duration of the steps. Next, the average step length (SL) was computed from the average step frequency and the average treadmill speed (SL = PWS/SF) (Souza et al., 2017). Then, the walk ratio (WR) was calculated as the SL normalized by the SF (WR = SL/SF) and represents the estimated SL when SF is equal to 1 step/s, taking into account the invariant relationship between SL and SF, regardless of the walking speed (Sekiya et al., 1996). A decrease in the WR indicates that the SL decreased more than the SF did, a sign of increased cautiousness (Terrier and Reynard, 2015). We calculated the WR because the groups presented different PWSs at different inclinations. Step width (SW) was determined as the ML distance between heel markers within two subsequent heel strikes.

2.4.2. Gait stability

Gait stability was assessed using the MoS. The MoS was estimated using the method proposed in (Hak et al., 2013a) and it is defined as the distance of the XCoM to the edge of the BoS (check supplementary material for MoS calculation– Appendix A).

Additionally, the MoS for prosthetic and intact limb was calculated separately taking the average MoS obtained from steps with each corresponding limb (check supplementary material for results– Appendix B).

2.5. Statistical analysis

As the variables presented normal distributions (Shapiro-Wilk test, $p > 0.05$), mixed repeated-measures analysis of variance (mixed ANOVA) was used to assess the main effects of inclination and groups (and limb for MoS) and the interaction effect between group and inclination. A *post hoc* test with Bonferroni correction was performed when a main or interaction effect was significant. Paired T-test was conducted to compare the MoS on prosthetic vs intact limb. The statistical analysis was performed using SPSS software (SPSS Inc., Chicago, IL), and the significance level was set at $p < 0.05$.

3. Results

3.1. Spatiotemporal parameters

For walking speed, significant differences were observed between groups for all slopes, as well as between slopes within all the groups (Table 2).

Only the SW presented significant interaction effect between inclination and group (Fig. 1-A). A significant simple main effect of inclination was only observed for the TFA group (Table 2), whereas a significant simple main effect of group was observed for all inclinations (Table 2). The post hoc tests revealed a wider SW at all inclinations for the TFA group than for either the TTA or CT group (Table 2).

For the WR, significant main effects of inclination and group, but not the interaction effect between inclination and group, were observed (Fig. 1-B). A significant main effect of inclination was observed for all groups (Table 2), whereas a significant main effect of group was observed for the DOWN and UP conditions (Table 2). The post hoc tests revealed that the CT group presented a significantly larger WR in the DOWN and UP conditions than did transtibial and transfemoral groups, and the DOWN condition induced the smallest walk ratio for all groups (Table 2). Considering the walking speeds presented in Table 2, the results for WR indicate that the SL decreased more than any change in the SF in all groups, especially the transfemoral group, and in the DOWN condition rather than in the UP and HOR conditions (Table 2).

3.2. Margin of stability

The MoS (average minimum value of the MoS within each step over 150 strides) presented a significant interaction effect in both the ML and BW directions ($p = 0.001$ and $p < 0.001$, respectively) (Fig. 2). The simple main effect of group was significant for MoS ML at all inclinations but was significant only in the DOWN condition for MoS BW (Table 3). Additionally, the simple main effect of inclination was significant for the TFA and CT groups for MoS ML, and it was significant for all groups for MoS BW (Table 3).

The post hoc tests revealed that the MoS ML was significantly larger for the TFA group than for the CT group at all inclinations

Table 2
Mean and standard deviation of the spatiotemporal parameters between groups (transfemoral, transtibial, and control).

		Groups			
		Transfemoral (*n* = 13)	**Transtibial (*n* = 12)**	**Control (*n* = 15)**	p^*
Step Width (cm)	**DOWN**	$16.40 \pm 5.89^{a,b,1}$	11.71 ± 2.45^{a}	9.67 ± 2.89^{b}	**<0.001**
	HOR	$16.26 \pm 6.09^{c,d,2}$	11.06 ± 3.17^{c}	9.79 ± 3.98^{d}	**0.002**
	UP	$17.92 \pm 5.95^{e,f,1,2}$	11.24 ± 3.27^{e}	9.53 ± 2.29^{f}	**<0.001**
	p^+	**<0.001**	0.308	0.855	
Walk Ratio (cm/Hz)	**DOWN**	$29.58 \pm 4.53^{g,3,4}$	$30.08 \pm 4.04^{h,5,6}$	$34.43 \pm 3.78^{g,h,7,8}$	**0.006**
	HOR	34.96 ± 5.18^{3}	34.85 ± 4.54^{5}	38.11 ± 3.60^{7}	0.102
	UP	$34.26 \pm 5.93^{j,4}$	$35.95 \pm 4.63^{i,6}$	$39.28 \pm 3.48^{j,i,8}$	**0.025**
	p^+	**<0.001**	**<0.001**	**<0.001**	
Preferred Walking Speed (km/h)	**DOWN**	$2.78 \pm 0.60^{k,l,9,10}$	$3.99 \pm 0.72^{k,m,12}$	$4.68 \pm 0.68^{l,m,14}$	**<0.001**
	HOR	$3.27 \pm 0.69^{n,o,9,11}$	$4.45 \pm 0.76^{n,12,13}$	$4.94 \pm 0.74^{o,14,15}$	**<0.001**
	UP	$3.03 \pm 0.69^{p,q,10,11}$	$4.10 \pm 0.81^{p,13}$	$4.66 \pm 0.65^{q,15}$	**<0.001**
	p^+	**< 0.001**	**< 0.001**	**< 0.001**	

Values expressed as mean ± standard deviation. p^* Significant differences between groups (mixed ANOVA), Bonferroni test *post hoc*: a = 0.018, b < 0.001, c = 0.022, d = 0.002, e < 0.001, f < 0.001, g = 0.011, h < 0.001, i < 0.001, j < 0.001, k < 0.001, l < 0.001, m = 0.035, n < 0.001, o < 0.001, p = 0.002, q < 0.001. p^+ Significant differences between inclinations (mixed ANOVA), Bonferroni test *post hoc:* 1 = 0.003, 2 = 0.001, 3 < 0.001, 4 < 0.001, 5 < 0.001, 6 < 0.001, 7 < 0.001, 8 < 0.001, 1 < 0.001, 2 = 0.025, 3 = 0.029, 4 < 0.001, 5 < 0.001, 6 < 0.001, 7 < 0.001. Pairs of lowercase letters (between groups) and numbers (between inclinations) indicate significant pairwise comparisons.

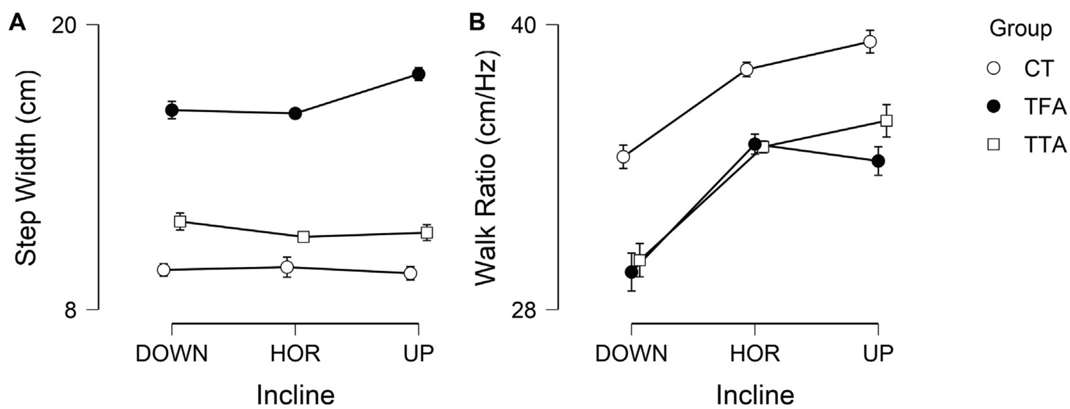

Fig. 1. (A) Average step width (cm) and (B) Walk ratio (cm/Hz) variables for each condition (DOWN, HOR, and UP inclinations). TFA - Transfemoral amputees, TTA - Transtibial amputees, and CT - Control group. A - ML: medial–lateral direction, B - BW: backward direction. Error bars indicate the data standard deviation.

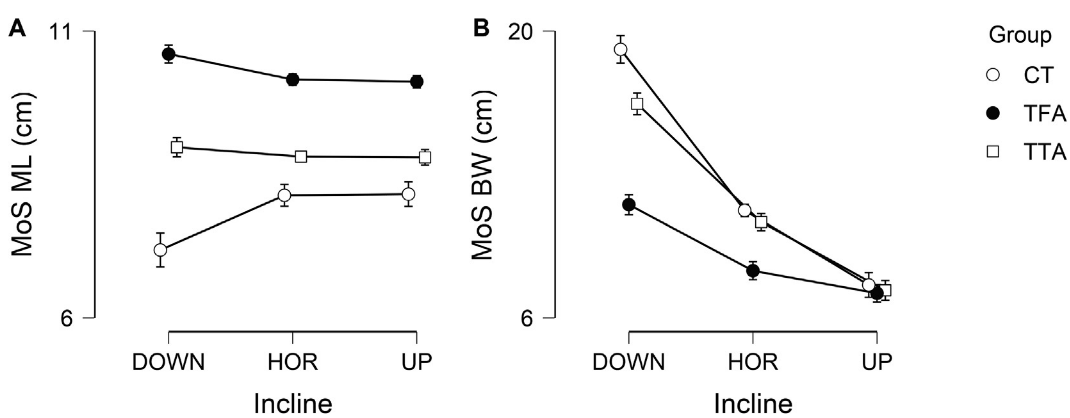

Fig. 2. The Margin of stability in the medial–lateral (A) and backward (B) directions for each condition (DOWN, HOR, and UP inclination). TFA - Transfemoral amputees, TTA - Transtibial amputees, and CT - Control group. A - ML: medial–lateral direction, B - BW: backward direction. Error bars indicate the data standard deviation.

Table 3
Mean and standard deviation of MoS between groups (transfemoral, transtibial, and control).

		Groups			
		Transfemoral (n = 13)	**Transtibial (n = 12)**	**Control (n = 15)**	p^*
MoS - ML (cm)	**DOWN**	10.60 ± 1.98^{a,1}	8.97 ± 1.74	7.18 ± 1.85^{a,2,3}	**<0.001**
	HOR	10.15 ± 2.03^{b}	8.81 ± 1.79	8.13 ± 1.30^{b,2}	**0.013**
	UP	10.11 ± 1.75^{c,1}	8.79 ± 1.76	8.15 ± 1.57^{c,3}	**0.014**
	p^+	**0.020**	0.575	**0.010**	–
MoS - BW (cm)	**DOWN**	11.52 ± 3.37 ^{d,e,4,5}	16.44 ± 4.32^{d,6,7}	19.11 ± 4.56 ^{e,9,10}	**<0.001**
	HOR	8.30 ± 4.10^{4}	10.67 ± 4.40^{6,8}	11.24 ± 5.18^{9,11}	0.229
	UP	7.20 ± 4.79^{5}	7.34 ± 5.58^{7,8}	7.61 ± 5.94^{10,11}	0.980
	p^+	**<0.001**	**<0.001**	**<0.001**	–

Values expressed as mean ± standard deviation. p^* Significant differences between groups (mixed ANOVA), Bonferroni test *post hoc*: a < 0.001, b = 0.011, c = 0.012, d = 0.016, e < 0.001. p^+ Significant differences between inclinations (mixed ANOVA), Bonferroni test *post hoc*: 1 = 0.035, 2 = 0.025, 3 = 0.021, 4 < 0.001, 5 < 0.001, 6 < 0.001, 7 < 0.001, 8 < 0.001, 9 < 0.001, 10 < 0.001, 11 < 0.001. MoS: margin of stability; ML: medial–lateral; BW: backward. Pairs of lowercase letters (between groups) and numbers (between inclinations) indicate significant pairwise comparisons.

($p < 0.001$, $p = 0.011$, $p = 0.012$, for DOWN, HOR, and UP conditions, respectively) and significant differences for UP-DOWN ($p = 0.035$) inclinations in the TFA group and for UP-DOWN ($p = 0.021$) and DOWN-HOR ($p = 0.025$) inclinations for the CT group (Table 3). MoS BW was significantly smaller for the TFA group than for the CT group only in the DOWN condition ($p < 0.001$). In addition, MoS BW was significantly larger in the DOWN condition than in both UP and HOR conditions for the TTA group ($p < 0.001$, $p < 0.001$, respectively) and CT group ($p < 0.001$, $p < 0.001$, respectively).

4. Discussion

In the present study, we investigated the gait stability of both individuals with transfemoral and transtibial amputations when walking on an sloped treadmill at their preferred walking speed. We hypothesized that sloped surfaces increase MoS and that this change would be more pronounced in groups of people with amputations and in the DOWN condition. For this purpose, we used surfaces sloped to −8, 0, and 8% to investigate gait stability. Based on the second assumption that a larger MoS can indicate that a strategy was adopted to increase stability, our hypotheses were partially supported. In the ML direction, the transfemoral group was more stable in downward walking, whereas the control group was more stable in upward walking. In the BW direction, however, the lowest values were exhibited by the participants with amputations, especially those with transfemoral amputations. Overall, these findings reveal the strategies adopted by amputees to improve stability, in line with the second interpretation of margin of stability.

Previous studies have reported kinematic adjustment strategies consistent with our results in both transtibial and transfemoral amputees in upward and downward walking (Vrieling et al., 2008), as well as margin of stability adjustments in transtibial (Hak et al., 2013b) and transfemoral amputees in unstable virtual environments (Sturk et al., 2019); although in the latter study, only MoS ML was assessed. In both amputee groups, there was a smaller hip extension in late stance in both upward and downward walking, probably due to a shorter step length (Vrieling et al., 2008). In an unstable virtual environment, the transtibial group walked slower than the control group, with a lower step frequency and wider step width, resulting in a larger MoS ML and smaller MoS BW (Hak et al., 2013b). Similarly, in a virtual environment, the transfemoral group walked slower than the control group, with a shorter step length and wider step width, resulting in an increased MoS ML on the prosthetic side (Sturk et al., 2019). Thus, the present study is the first to investigate both transtibial and transfemoral amputees walking on slopes and compute MoS in both the ML and BW directions, revealing the relationship between the level of amputation and stability on slopes.

For all inclinations, the MoS ML was larger for the transfemoral group than for the transtibial and control groups, indicating that at first glance, this group would have greater stability from a biomechanical point of view (Table 3), although there were significant differences only between the transfemoral and control groups. A wider step width and slower gait speed accompany this result for the transfemoral group (Table 2); this behavior remained on all inclinations. This result suggests that to ensure stability during gait, some adaptations were required: a wider base of support and a slower gait speed, and they denote a more cautious gait compared to that adopted by the control group (Table 2).

The highest MoS ML value (Table 3) for the transfemoral group in comparison to the other groups indicated that the XCoM moved farther from the base of support edge at all levels of inclinations, especially for prosthetic limb in the DOWN condition (Table Appendix B). The study by Hof et al. (Hof et al., 2007) revealed that transfemoral amputees have less accurate foot positioning, which results in a wider step width (Table 3), as evidenced by the significant differences (*post hoc*) between the transfemoral group and the transtibial and control groups at all inclinations. This result can be explained by compensatory movements of the contralateral limb due to reduced functionality in the prosthetic limb. Additionally, the difference between the transfemoral and the other groups, for both the MoS ML and step width, was significant due to the high level of amputation, which compromises the positioning of the foot in a greater extension; the transtibial group has better adaptability and control of the prosthetic limb than does the transfemoral group. However, the increase in MoS ML observed here may not necessarily indicate an increase in gait stability in the transfemoral group compared to the control group. This finding was mainly due to an increase in the base of support and a decrease in gait speed, which resulted in a reduction in the oscillation of the COM, leading to a more conservative gait pattern under the controlled protocol. However, these results do not indicate that the transfemoral group is able to resist greater perturbations than is the control group; instead, the transfemoral group adopts a gait pattern that improves stability.

In addition, the DOWN condition, compared to the other conditions, induced larger MoS in the ML and BW directions for both transfemoral and transtibial groups. However, this increase in the MoS can be attributed to the prosthetic limb only for TFA group (check supplementary material for results– Appendix B). Furthermore, this condition was the condition in which the participants reported greater difficulty in performing the tests. In contrast, the control group had lower MoS ML in the DOWN condition, indicating less stability. This result may be related to the decrease in the walk ratio (Table 2), in agreement with the results of previous studies (Hak et al., 2013b; Sivakumaran et al., 2018). However, as discussed above, it is not possible to infer that the control group is less stable than the transfemoral and transtibial groups. Indeed, it seems that the groups of people with amputations adopted a strategy that increases stability, which is critical for these groups.

The fact that no significant differences between groups were found for MoS BW in either the UP or HOR condition suggests that a similar level of stability was maintained in both conditions. However, the MoS BW in the DOWN condition was significantly smaller in the transfemoral group than in the transtibial and control groups, showing a 39% reduction compared to the control group. This suggests that the transfemoral group adopted a more cautious gait in the DOWN condition, reducing the walking speed and the advance of the XCOM relative to the foot contact during the progression in favor of the gravity (Hak et al., 2013a).

In addition, the increase in MoS BW in the DOWN condition that occurred for all groups was probably due to a corresponding decrease in the walk ratio (Table 2), which led to a larger displacement of the XCoM relative to the foot contact during the progression in favor of gravity (Hak et al., 2013a). This finding may indicate an increased risk of a forward loss of balance in the DOWN condition. Our findings support previous results that have shown that a lower walk ratio leads to increased MoS BW and hence increased stability (Hak et al., 2013a); forward loss of balance requires a relatively small adaptation of the next steps to recover in level condition (Hak et al., 2013a). This may not be the case in the DOWN condition because the progression occurs in favor of gravity. In line with this interpretation, the groups presented a descending order according to their MoS BW: the control, transtibial, and transfemoral groups (Table 3). This finding suggests that the amputees adopted a more cautious gait according to the level of amputation.

The results for transtibial group indicated that this level of amputation has less influence on MoS since significant differences were not observed between the transtibial and control groups in either the BW or ML direction (Table 3). Additionally, no differences were found when comparing prosthetic vs intact limb in the transtibial group, confirming this conclusion. These results are in agreement with those reported by Curtze et al. (Curtze et al., 2011), who studied gait changes in 18 people with transtibial amputations while walking on an uneven flat surface. These authors observed a slight reduction in the gait speed of the people with transtibial amputations and no significant differences in MoS compared to a control group (Curtze et al., 2011). Although the challenges imposed on the people with transtibial amputations were different in the present study, the results suggest the same concepts.

The transtibial group was more likely to adapt locomotor strategies to meet the demands of different walking conditions than was the transfemoral group, suggesting that the level of amputation was causally related to the results found. A possible explanation for this finding is related to the integrity of the knee joint in the transtibial group; the foot placement mechanism (Bruijn and Van Dieën, 2018; Hof et al., 2007), important for stability, is less compromised in this group than in the transfemoral group. In addition, foot clearance, important when walking on slopes and even on level ground, is less compromised in the transtibial group than in the transfemoral group. Due to reduced knee flexion on the prosthetic limb, transfemoral group uses compensatory movements of the trunk and pelvis (Michaud et al., 2000) to achieve proper foot clearance during the swing phase.

This study has some limitations. As the groups were homogeneous in terms of age, mass, height, activity level, cause of amputation, time since amputation, and time using prosthesis, the results should be interpreted with caution when generalizing them to other individuals with lower limb amputations who are older, less active, or underwent an amputation not related to trauma. In addition, the alignment of the prosthesis can influence the results. Although all participants with lower limb amputations reported to align the prosthesis regularly, we cannot guarantee that this alignment was the best possible at the moment of data collection. Another limitation is the estimation of the CoM using a markers cluster. This is not a precise representation of the CoM, but errors introduced using this approximation were likely similar for all groups across conditions, and therefore would not affect differences in MoS for all groups and conditions (Hak et al. 2013).

5. Conclusion

Our findings revealed that people with and without amputations show significant differences in spatiotemporal parameters and MoS. Overall, the strategies adopted by the people with lower limb amputations compensate the functional changes to maintain dynamic stability during gait, and the level of amputation was directly related to the strategies.

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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Appendix A

Fig. A1.

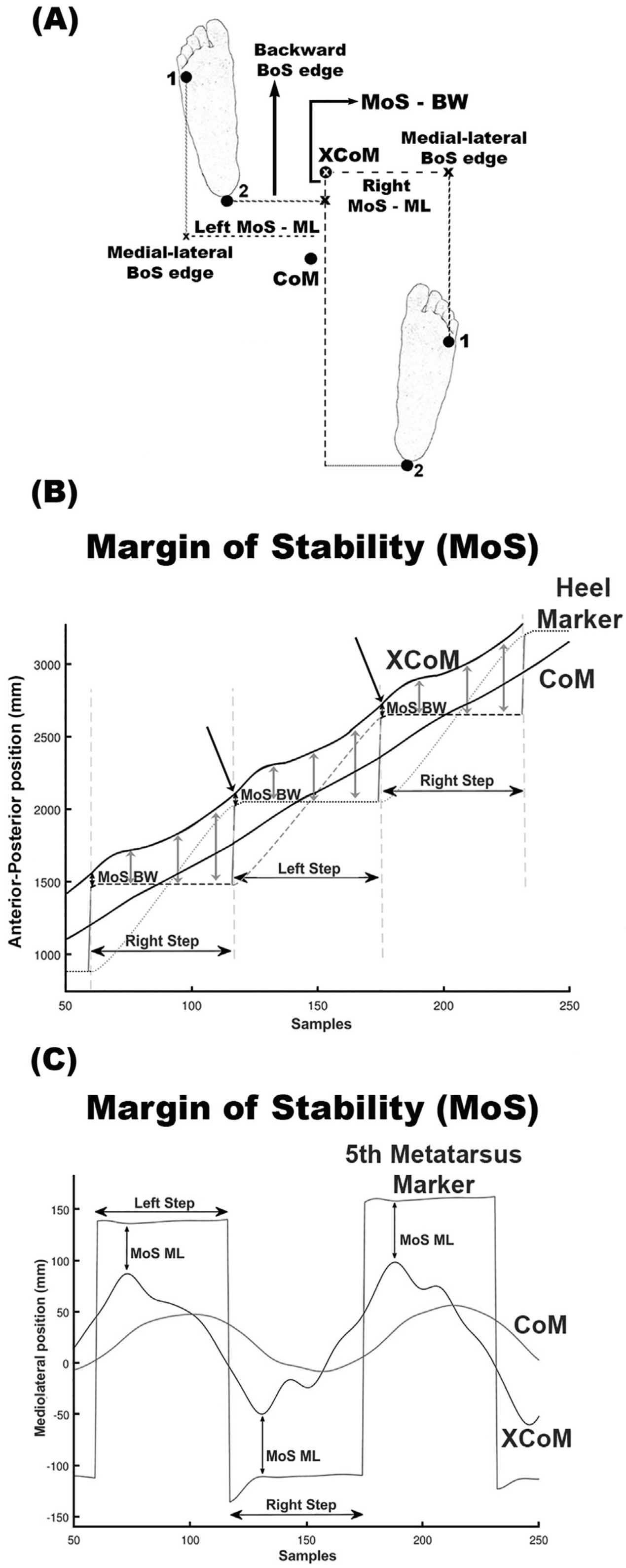

Fig. A1. Representation of the: (A) margin of stability projection (ML e BW), (B) backward margin of stability calculation (BW MoS), and (C) medial–lateral margin of stability (ML MoS) calculation. The border of the base of support (BoS) is the heel marker in (B), and the metatarsal marker in (C). In (B), black dashed and dotted horizontal lines indicate right and left stance phase, respectively, gray dashed and dotted lines indicate right and left swing phase, respectively. Double horizontal arrows indicate right and left stance phase. Black diagonal arrows indicate the point where anterior-posterior base of support edge change from one foot to another. Double vertical arrows indicate BW MoS: black double vertical arrow is the minimum BW MoS.

Appendix B

Table 4.

Table 4
Mean and standard deviation of MoS between prosthetic and intact leg in persons with amputation (transfemoral and transtibial), and between non-dominant and dominant leg in control group.

Groups		Transfemoral (*n* = 13)			Transtibial (*n* = 12)			Control (*n* = 15)				
		Prosthetic leg	**Intact leg**	**p***	**Prosthetic leg**	**Intact leg**	**p***	**Non-dominant leg**	**Dominant leg**	**p***	**p# Prost**	**p# Intact**
MoS - ML (cm)	**DOWN**	12.11 ± 2.35^{1}	9.09 ± 2.53	**<0.001**	9.28 ± 2.43	8.67 ± 1.53	1.000	6.78 ± 2.78^{1}	7.58 ± 1.48	1.000	**<0.001**	0.107
	HOR	10.93 ± 2.45^{2}	9.37 ± 1.81	0.969	9,00 ± 2.27	8.62 ± 1.67	1.000	8.11 ± 1.75^{2}	8.15 ± 1.13	1.000	**0.005**	0.128
	UP	11.00 ± 2.35^{3}	9.23 ± 1.53	0.334	8.96 ± 2.26	8.63 ± 1.67	1.000	7.96 ± 2.30^{3}	8.34 ± 1.27	1.000	**0.005**	0.291
	***p*⁺**	0.324	0.941		0.935	0.996		0.164	0.301	–		
MoS - BW (cm)	**DOWN**	15.18 ± 3.59^{4,a}	7.87 ± 4.53^{5,6}	**<0.001**	17.71 ± 3.22^{b,c}	15.18 ± 5.66^{6,d}	0.396	19.20 ± 4.79^{4,e,f}	19.00 ± 4.70^{5,g,h}	1.000	**0.039**	**<0.001**
	HOR	9.75 ± 4.30	6.85 ± 4.61	0.055	10.64 ± 3.99^{c}	10.71 ± 4.95	1.000	11.22 ± 5.21^{e}	11.26 ± 5.31 ^{g}	1.000	0.700	0.057
	UP	8.94 ± 3.75^{a}	5.47 ± 6.18	**0.004**	7.27 ± 5.28^{b}	7.42 ± 5.94^{d}	1.000	7.58 ± 6.14^{f}	7.63 ± 5.80 ^{h}	1.000	0.690	0.591
	p⁺	**<0.001**	0.473		**<0.001**	**0.011**		**<0.001**	**<0.001**			

Values expressed as mean ± standard deviation. p* Significant differences between limbs. p# Significant differences between groups. p⁺ Significant differences between inclinations (mixed ANOVA). *post hoc* Bonferroni test: 1 < 0.001, 2 = 0.048, 3 = 0.027, 4 = 0.035, 5 < 0.001, 6 = 0.021, a = 0.028, b < 0.001, c = 0.010, d = 0.024, e < 0.001, f < 0.001, g = 0.007, h < 0.001. MoS: margin of stability; ML: medial–lateral; BW: backward. Pairs of lowercase letters (between inclinations) indicate significant pairwise comparisons.

Appendix C. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2021.110453.

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