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Effect of slope in an immersive virtual environment on segmental asymmetry in people with femoral amputation and a microprocessor knee

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Purpose: The continued development of microprocessor-based knee prostheses has improved the independence of people with a femoral amputation in many environments. This study aimed to describe the effect of slopes on kinematic joint variables and segmental asymmetry. *Methods*: Ten individuals with transfemoral amputation fitted with microprocessor-controlled knees performed 5 sessions of treadmill walking at their preferred speed in an immersive virtual environment in 5 incline conditions (Level, 3° and 6° Uphill, and 3° and 6° Downhill). The Human Body Model was used to quantify kinematic joint variables from motion capture system data. The perimeter-to-area method was used to determine the symmetry ratio of the trajectory of the leg segments in the sagittal plane. *Results*: There was a significant effect of the Uphill conditions on step length and width on the intact side and on all kinematic joint variables on both sides, although the changes differed according to the phase of the gait cycle. The segmental symmetry index was significantly modified in all slope conditions compared to Level. *Conclusion*: Kinematic joint variables are affected by slopes; the effect was greater for the Uphill than Downhill conditions compared to the Level condition. The perimeter-to-area symmetry ratio differed from the Level condition for all slope conditions. These results indicate that, although microprocessor knees improve the autonomy of prosthesis users, work is required to improve their capacity of adaptation to varied terrain to reduce kinematic asymmetry.

Key words: *prosthesis, virtual reality, gait, symmetry, microprocessor knee, slope*

1. Introduction

Vascular or traumatic lower limb amputation considerably reduces physical and locomotor activity in everyday life [3], [4]. Studies using wearable sensors to measure daily physical activity levels have reported that people with tibial amputation are less active than able-body participants [3]. Active prostheses have been developed to facilitate physical activity and locomotion in different conditions [2]. These prostheses consist of a mechatronic knee joint equipped with microprocessors that controls the support and swing phases of gait [29]. The aim is to improve the safety of prosthetic gait. A study of 13 people with transfemoral amputation found improvements in balance confidence and safety when using a microprocessor-enhanced knee [7].

Furthermore, studies have demonstrated functional improvements with the use of these prostheses in different locomotor conditions such as obstacle clearance and slopes [18], [38]. The use of a microprocessor prosthesis seems to improve biomechanical variables such as side-to-side asymmetry in both the prosthetic and intact limbs during ascent and descent of slopes [4].

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More recently Gholizadeh et al. [9] showed that kinetic and kinematic variables differed between uphill and downhill gait in patients with transtibial amputation using a virtual immersive environment.

To our knowledge, few studies have been performed under similar conditions in individuals with femoral amputation. The aim of this study was to describe the effect of upward and downward slopes on joint kinematics and segmental asymmetry in people with femoral amputation fitted with microprocessor-controlled knees. For this, we hypothesized that the direction of the slope has an impact on joint kinematic parameters and increases segmental asymmetry, reflecting difficulties in compensating for the loss of mobility.

2. Materials and method

2.1. Method

Ten people with a transfemoral amputation and a microprocessor controlled prosthetic knee were included (Table 1). The participants were recruited from the rehabilitation centre. The recruitment took the heterogeneity of currently commercialised prosthetic knees into account. All the knees were monocentric and microprocessor controlled, however, they differed in terms of the control systems of the swing and stance phases (hydraulic or magnetorheological systems) and the number of sensors that piloted the mechanism. Nevertheless, we considered that they functionned similarly since none generated propulsive forces, only breaking ones.

Approval was granted by our local hospital ethics committee. In accordance with the Declaration of Helsinki, all participants provided written consent for participation. All participants took part in daily gait training sessions in an immersive virtual reality system (GRAIL, Motek Force Link, Amsterdam, NL) as part of the usual rehabilitation in our centre.

2.2. Data collection

Participants performed four 40 s trials of gait at their preferred speed on the GRAIL system treadmill. The preferred speed was determined by having the patient walk a distance of 10 meters under four conditions. The patient alternated between twice their comfortable speed and twice the slow speed in a random sequence. The average of the two passages at the comfortable speed was considered the preferred speed.

The GRAIL system (Fig. 1) consists of a 4-degree--of-freedom platform equipped with a dual conveyor belt and two 6-component force platforms. Above the platforms, there is a 180° screen on which 3D scenes are projected. The projected scene corresponded to a path in the undergrowth. Motion capture was performed using an optoelectronic system (10 cameras, 100 hz, Vicon, Oxford, UK) with a set of 26 markers positioned on the anatomical points defined by the HBM model (Human Body Model, [35]) (Fig. 2). The participants' prostheses are equipped with a prosthetic foot and a mechatronic knee allowing for physiological mimicry of joint kinematics and that this is compatible with the HBM model as detailed [35 Supplementary data: https://static-content.springer.com/esm/art%3A10.1007%2Fs11517-013-1076-z/MediaObjects/11517_2013_1076_MOESM1_ESM.pdf].

Table 1. Clinical, anthropometric and prosthetic characteristics of participants with M: Male F: female, PADO: obliterating arterial disease of the lower limbs, and Stump length measured in cm from anterior iliac spine

Participant	Age [years]	Sex	Amputation side	Number of years since amputation	Etiology	Stump length [cm]	Socket	Sleeves	Fixing system	Knee	Foot class	Activity level
1	58	M	right	8	traumatic	40	ischial-integrated	silicone	seal in	Rhéo	III	K4
2	51	M	left	12	traumatic	50	ischial-integrated	silicone	seal in	Genium	III	K4
3	63	M	right	4	PADO	37	ischial-integrated	silicone	seal in	C-LEG	III	K2
4	59	M	right	4	PADO	41	ischial-integrated	silicone	seal in	Rhéo	III	K3
5	71	F	right	2	PADO	37	ischial-integrated	silicone	seal in	Kénévo	II	K2
6	58	M	left	2	PADO	22	ischial-integrated	silicone	seal in	C-LEG	III	K4
7	62	M	left	2	PADO	28	ischial-integrated	silicone	seal in	Kénévo	II	K3
8	77	M	right	5	PADO	36	ischial-integrated	silicone	seal in	Kénévo	II	K4
9	42	M	left	11	traumatic	46	ischial-integrated	silicone	seal in	Rhéo	III	K3
10	60	F	left	3	PADO	35	ischial-integrated	silicone	seal in	Kénévo	II	K3

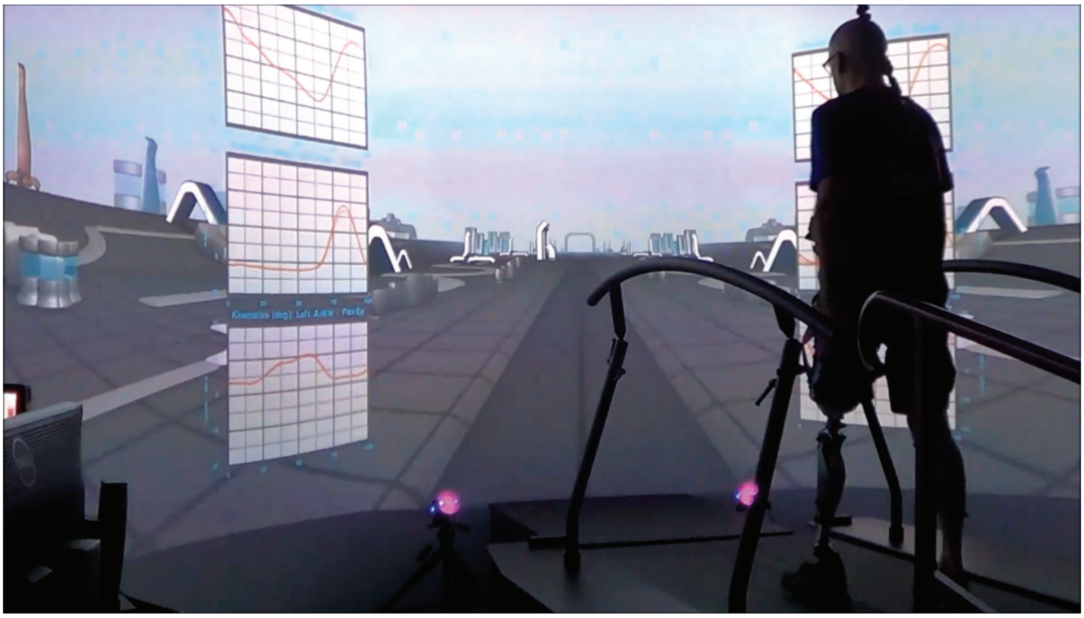

Fig. 1. Test setup on GRAIL System

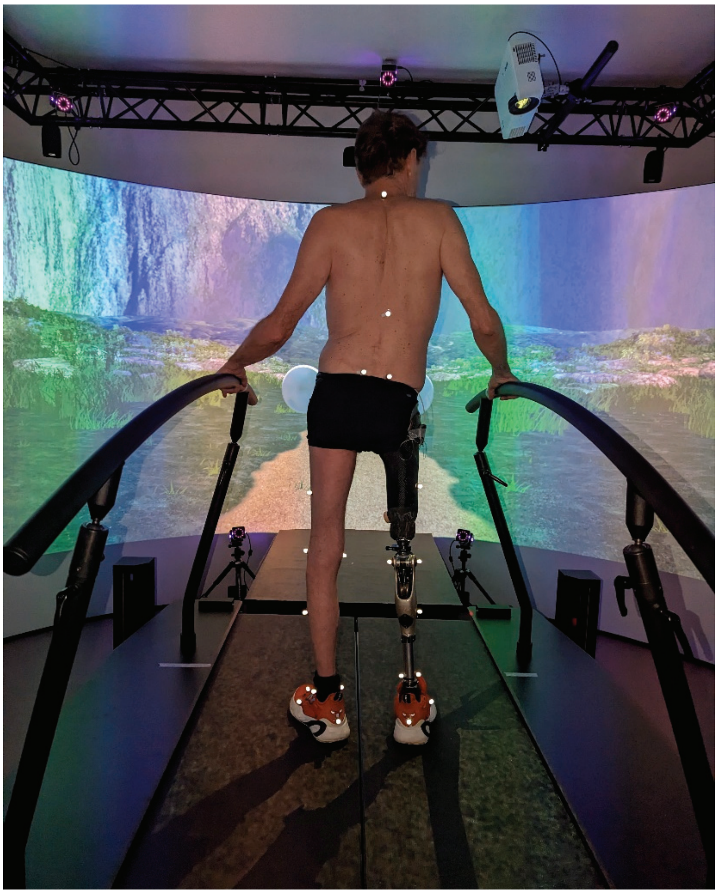

Fig. 2. Example of HBM marker placement

The ground reaction force measurement system used in the GRAIL consists of two six-component force platforms. These platforms are integrated into the treadmill, allowing for accurate measurement of the forces exerted by each foot during walking at a frequency of 1000 Hz [35]. They function by recording forces in vertical, lateral, and antero-posterior directions, as well as moments of force around these axes.

All participants were familiar with the GRAIL system since they used it during their rehabilitation. Before recording, they walked on the treadmill for 30 seconds with a 0° incline for the purpose of re-familiarisation.

The first trial was performed with a 0° incline (Level condition). The order of the slopes (Uphill or Downhill) was randomised by drawing lots. Only the progression of the slope was not randomised: it was always first 3° then 6°. This choice of slopes is similar to that chosen by Vrieling et al. [37] and corresponds to access slopes in buildings.

2.3. Data analysis

Motion analysis was performed post-acquisition using the GOAT software suite (Gait Off-ine Analysis Off Tool 4.0.1, Motek Force Link, Amsterdam, NL). Gait cycles (i.e., foot strike and foot off) were determined from the force platforms signals.

2.4. Spatio-temporal and kinematic variables

The following spatio-temporal variables were calculated: step and stride length, step width, gait speed, cadence, and the percentage durations of the phases of the gait cycle: initial double support, single support, pre swing and swing.

The following kinematic variables were calculated using the GOAT software associated with the HBM model: peak ankle plantar flexion during stance, peak ankle dorsiflexion during stance, ankle range of motion (ROM), knee flexion at initial double support, peak knee flexion during swing, knee ROM, peak hip flexion during stance, peak hip flexion during swing,

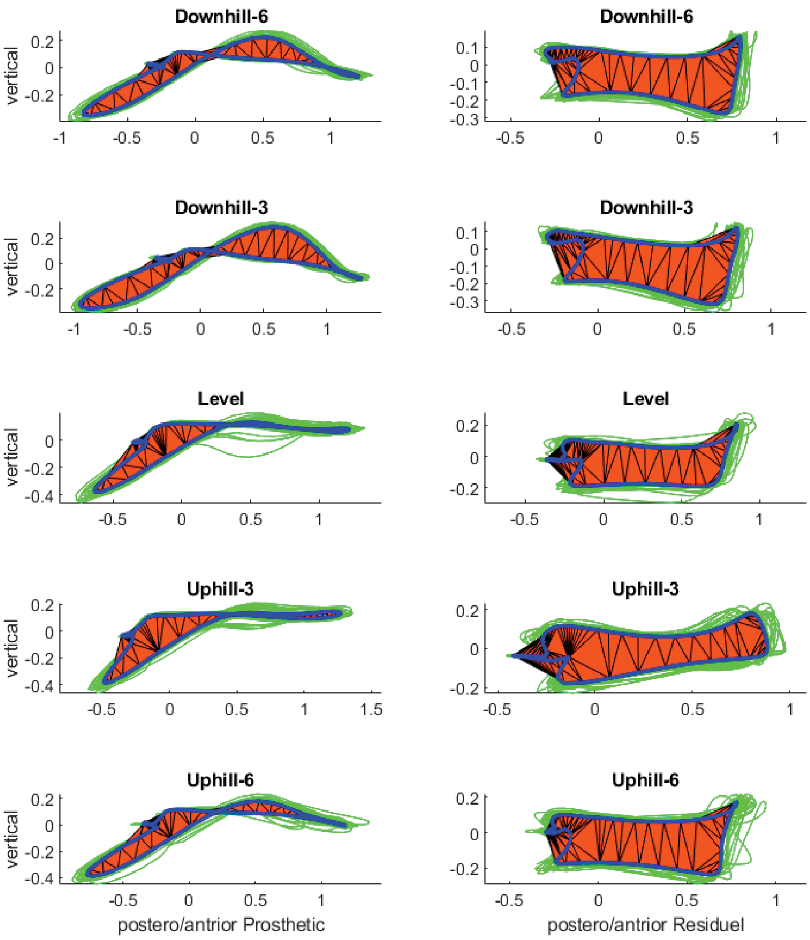

Fig. 3. Example of a graphical representation of a participant's perimeter-to-area ratio calculation. The velocity displacement on the antero-posterior axis (abscissa) and on the vertical axis (ordinate) of the shank marker for each gait cycle (green), for the average cycle (blue). In black, the triangles used to calculate the polygon area (red). In the different experimental conditions (Level, 3° and 6° Uphill, 3° and 6° Downhill) for the prothetic (left) and intact (right) sides

hip ROM and pelvic ROM (sagittal, frontal and rotation) [33].

2.5. Segmental displacement asymmetry: perimeter-to-area ratio approach

To quantify segmental asymmetry and cycle-to-cycle variability in asymmetry during gait, we used the calculation methods associated with the characterization of geometric shapes. This method makes it possible to characterize surfaces that can be compared. These methods are used in medical imaging [31] and geography [36]. Segmental displacement over time of the prosthetic leg, compared to the intact leg, was calculated to quantify the use of the mechatronic capacities of the prosthesis. This analysis was performed on sagittal plane motion for two reasons: 1) treadmill gait is essentially rectilinear and 2) the prosthetic joints were all medial-pivot. Many calculation methods exist to characterize a shape in 2D: we chose to calculate the perimeter to area ratio (p-to-a ratio). In this study, we used the displacement velocity of the marker positioned on the shank of the HBM model. The speed of movement on each axis was calculated from the marker coordinates recorded during the gait trials. Displacement velocity on the antero-posterior axis and on the vertical axis was time-normalized for each gait cycle and centred on the values of the coordinates of the first cycle. We calculated the area and perimeter of the polygon described by the movement of the marker in the sagittal plane. The perimeter of this shape corresponds to the sum of the norm of the vectors of the velocity displacement of the shank marker. The area corresponds to the sum of the triangles making up this polygon. The triangles were determined using Delauney's triangulation method. A script was specifically developed in Matlab for all the calculations (Matlab 2018a, Mathworks, MA, US).

We defined a symmetry index from the ratio between the p-to-a ratio for the prosthetic side and the p-to-a ratio for the intact side (Eq. 1).

$$\text{Symmetry Index} = \frac{\text{Prothetic side } p \text{ to a ratio}}{\text{Intact side } p \text{ to a ratio}}. \quad (1)$$

An index value of 1 indicates perfect symmetry, values below 1 indicate a higher ratio on the intact side than on the prosthetic side and conversely. This index was calculated every two consecutive gait cycles from a total of 60 to 70 cycles for each condition.

2.6. Statistical analysis

Given the small sample size, a non-parametric statistical analysis was performed. The Friedman test was used to compare variables that do not follow a normal distribution. Post-hoc analysis of significant differences was performed using a Wilcoxon test for two-by-two comparisons of the effect of slope (Uphill, Downhill) compared to Level for each variable. A p-value <0.05 was considered significant for all analyses.

3. Results

3.1. Spatial and temporal variables

Small, but significant between-conditions differences were found for step length on the prosthetic side in the Uphill 3° and 6° conditions (Table 2): compared to the Level condition, step length was by 1 cm longer in the Uphill 3° and 6° conditions. There were no changes on the intact side. There was also a small but significant difference in the step width, which was by 1 cm wider in the Uphill 3° and 6° conditions than the Level condition.

3.2. Kinematic variables

Different kinematic variables were altered by the Uphill and Downhill conditions on the prosthetic and intact sides (Table 3).

On the prosthetic side, ankle ROM, pelvic rotation ROM in the frontal plane, and peak hip and knee flexion in swing phase were significantly greater in both the Uphill 3° and 6° conditions than Level. Hip ROM was significantly smaller in the same conditions. For the Downhill conditions, the only change was a small but significant decrease in peak knee flexion during swing phase in the 6° condition.

On the intact side, ankle, hip and pelvic frontal rotation ROM, peak hip flexion in swing and peak hip and ankle flexion during initial double support increased significantly in the 3° and 6° Uphill conditions. For the 6° Downhill condition, knee flexion decreased, and hip flexion increased significantly at initial contact. In the 3° Downhill condition, only hip flexion at initial contact increased significantly compared to the Level condition.

Table 2. Median values [1st and 3rd Quartiles] of the spatio-temporal parameters of the walking cycle under the conditions: Level, Uphill (UP) 3°, Uphill 6°, Downhill (DS) 3°, Downhill 6° for the prosthetic side (Pro) and the intact side (Int), * $p < 0.05$

PRO	Velocity [$m.s^{1}$]	Stride Length [m]	Step Length [m]	Step Width [m]	Stance Phase [%]	Swing Phase [%]	Initial Double Support Phase [%]	Single Support Shase [%]
Level	0.61 [.51 .81]	0.93 [.85 1.02]	0.49 [.47 .51]	0.21 [.18 .21]	68.4 [65.7 71.4]	31.6 [28.6 34.3]	20.7 [19.1 23.9]	27.8 [26.9 29.8]
DS 3	0.60 [.49 .76]	0.94 [.88 1.01]	0.48 [.46 .52]	0.20 [.19 .22]	68 [64.3 71.7]	32 [28.3 35.7]	21.5 [19.3 27.1]	28.1 [26.1 30.2]
DS 6	0.62 [.49 .77]	0.95 [.84 .99]	0.47 [.44 .53]	0.20 [.20 .22]	67.9 [64.2 71.9]	32.1 [28.1 35.8]	19.5 [18.1 26.9]	28.5 [26.3 31]
UP 3	0.61 [.51-.83]	0.94 [.85 1.05]	0.50 [.44 .52]	0.21 [.19 .22]*	68.4 [65.5 71.4]	31.6 [28.6 34.5]	21.1 [18.9 24.6]	28.2 [26.9 30.7]
UP 6	0.62 [.51 .83]	0.96 [.85 1.05]	0.50 [.44 .54]	0.22 [.19 .23]*	68.5 [65.5 71.4]	31.5 [28.6 34.5]	21.9 [18.9 24.6]	28.3 [26.9 30.9]
INT	Velocity [$m.s^{1}$]	Stride Length [m]	Step Length [m]	Step Width [m]	Stance Phase [%]	Swing Phase [%]	Initial Double Support Phase [%]	Single Support Shase [%]
Level	0.70 [.62 1.03]	0.93 [.87 1.02]	0.44 [.35 .51]	0.21 [.18 .21]	72.1 [70.2 73.1]	27.9 [26.9 29.8]	18.2 [15.3 19.8]	31.6 [28.5 34.2]
DS 3	0.71 [.66 1.1]	0.94 [.89 1.01]	0.43 [.39 .51]	0.20 [.19 .22]	72 [69.8 73.9]	28 [26.1 30.2]	16.5 [14.3 19.4]	31.9 [28.2 35.6]
DS 6	0.73 [.66 1.11]	0.95 [.84 .99]	0.41 [.37 .48]	0.20 [.20 .22]	71.5 [69 73.7]	28.5 [26.3 31]	15.7 [14.4 20.3]	32.1 [28.1 35.7]
UP 3	0.69 [.63 1.03]	0.94 [.88 1.05]	0.45 [.39 .52]*	0.21 [.18 .22]*	71.8 [69.2 73.1]	28.2 [26.9 30.8]	17.4 [15.7 18.7]	31.5 [28.6 34.4]
UP 6	0.69 [.63 1.04]	0.96 [.88 1.09]	0.45 [.39 .54]*	0.22 [.19 .22]*	71.7 [69.2 73.1]	28.3 [26.9 30.8]	17.4 [15.6 18.7]	31.4 [28.6 34.4]

Table 3. Median values [1st and 3rd Quartiles] of joint kinematic parameters: Peak Ankle plantar flexion during initial double contact (Peak Ankle IDC), Peak Ankle dorsiflexion during stance phase (Peak Ankle ST), Ankle range of motion (Ankle Range), Knee flexion at initial contact (Knee IC), Peak knee flexion during swing phase (Peak Knee SW), Knee range of motion (Knee Range), Peak hip flexion during initial double contact (Peak Hip IDC), Peak hip flexion during swing phase (Peak Hip SW), Hip range of motion (Hip Range), Pelvis range of sagittal motion (Pelvis Sagittal), Pelvis range of frontal motion (Pelvis Frontal), Pelvis range of rotation motion (Pelvis Rotation), dans les conditions: Level, Uphill (UP) 3°, Uphill 6°, Downhill (DS) 3°, Downhill 6° for the prosthetic side (Pro) and the intact side (Int), * $p < 0.05$

PRO	Knee Range	Ankle Range	Hip Range	Pelvis Sagittal	Pelvis Frontal	Pelvis Rotation	Knee IC	Peak Ankle IDC	Peak Hip IDC	Peak Ankle ST	Peak Knee SW	Peak Hip SW
1	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**	**11**	**12**	**13**
Level	49.1 [45.5 52.7]	13.1 [11.9 14.8]	37.4 [36.6 39.5]	7.5 [6.6 10]	7.5 [6 11.4]	10.5 [9.2 12.1]	4.5 [3.1 5.6]	2.7 [1.6 3.4]	32.6 [23.3 36.7]	15.8 [14.3 17.9]	51.4 [47.4 57.2]	33.3 [23.5 36.8]
DS 3	47.1 [43.5 52.8]	13.3 [11.5 15.4]	37.9 [34.7 39.4]	7.6 [6.8 12.4]	8 [6.2 9]	11.4 [9.8 13.3]	4.6 [3 5.8]	2.2 [1.5 3.5]	34.2 [20.8 37.5]	15.6 [13.8 18.0]	51.2 [46.3 57.2]	34.4 [22.6 37.7]
DS 6	49.2 [41.4 51]	13.2 [12.5 15.2]	34.7 [31.1 38.7]	8.1 [7.2 10.3]	7.7 [7 8.5]	11.2 [9.4 12.7]	4.6 [3 6.4]	2.2 [1.1 4.1]	32.0 [21.6 40.1]	15.3 [13.8 17.9]	51 [44.4 54]	33.2 [22.7 37.7]
UP 3	44.5* [44 50.5]	13.6* [12.2 15.6]	38 [36.7 39.2]	8 [6.2 11.3]	9.2* [6.7 12.5]	9.7 [8.6 12.5]	4.5 [3 5.7]	2.8 [1.7 3.7]	35.0* [25.2 37.5]	16.3 [14.4 18.7]	46.4* [45.5 57]	35.2* [26 37.7]
UP 6	42.7* [39 51.9]	14.2* [12.4 16.5]	38.7 [37.1 40]	8.7 [7.4 11.5]	12.4* [7.7 13.7]	10.7 [9.2 12.7]	4.5 [3 5.7]	3 [1.9 3.8]	36.5* [29.0 40.2]	16.7 [14.9 19.1]	44.6* [40.1 54.6]	36.8* [30.2 41.2]
INT	Knee Range	Ankle Range	Hip Range	Pelvis Sagittal	Pelvis Frontal	Pelvis Rotation	Knee IC	Peak Ankle IDC	Peak Hip IDC	Peak Ankle ST	Peak Knee SW	Peak Hip SW
Level	53.2 [48.6 59.7]	22.5 [19.3 25.4]	37.3 [36.8 40.6]	7.6 [6.5 9.7]	7.6 [6 .1 11.5]	10.7 [9.5 12.7]	12.4 [7.6 16.6]	2.5 [1.7 3.6]	35.0 [26.4 39.0]	17.7 [15.9 20.3]	63.2 [57.4 66.2]	36.9 [31.6 41.9]

1	2	3	4	5	6	7	8	9	10	11	12	13
DS 3	52.5 [47.6 61.2]	23.1 [19.1 24.9]	39.4 [34.2 42.3]	7.6 [6.7 12]	8.1 [6.1 9.3]	11.7 [10 13.5]	10.3 [7 15.2]	–0.1 [–0.6 2.7]	35.8* [24.3 41.8]	18.4 [15.6 20.2]	64.2 [55.6 67.2]	37.9 [30 44.5]
DS 6	52.7 [48.2 62.8]	22.9 [18.8 24.3]	37.6 [33.8 40.7]	7.9 [7.2 10.3]	7.7 [7 8.5]	11.7 [9.8 12.8]	11.3 [8 14.9]	–0.5 [–2.1 2.3]	34.9* [23.7 41.5]	18.3 [16.5 20.3]	64.2 [57.6 69.8]	37.3 [28.8 44.7]
UP 3	53.7 [47.3 56.9]	25.2* [20.9 28.4]	41* [40 42.7]	8.1 [6.6 10.7]	9.4* [6.6 12.7]	9.9 [9 13.1]	17.6* [11.4 19.8]	4.2* [3 5.9]	38.6* [32.9 44.4]	18.9 [16.2 20.9]	62.2 [56.4 65.3]	40.2* [33.9 47.2]
UP 6	52 [47.3 56.9]	28.3* [21.5 29.3]	43.7* [40.1 45.5]	8.7 [7.2 10.7]	12.6* [6.6 13.6]	11.3 [9 13.2]	21.4* [17.6 25.2]	7.1* [3.9 8.9]	42.0* [40.3 48.2]	20.0* [18.6 22.3]	61.1 [57.5 65.7]	44.3* [41.4 51.3]

3.2. Perimeter-to-area ratios and Segment Symmetry Index

The perimeter-to-area ratio on the prosthetic side was significantly smaller in the Uphill 6° and the Downhill 3° conditions than the Level condition. On the intact side, the p-to-a ratio was significantly smaller in the Downhill 3° and Uphill 3° and 6° conditions (Table 4). The segmental symmetry index was significantly smaller than Level in all conditions, with –17% for Downhill 3° and +16% for Downhill 6°; and with +27% Uphill 3° and +1% for Uphill 6° (Table 4).

Table 4. Median values [1st and 3rd Quartiles] of the p-to-a ratios on the prosthetic side (p-a Pro) and intact side (p-a Int) and of the symmetry index (Ind SY) under the conditions: Level, Uphill (UP) 3°, Uphill 6°, Downhill (DS) 3°, Downhill 6° for the prosthetic side (Pro) and the intact side (Int). * $p < 0.05$

Condition	p-a Pro	p-a Int	Ind SY	
Level	10.8 [10.1–11.4]	10.5 [10.2–11.8]	1.06 [1.00–1.11]	
DS 3	7.0 [6.8–7.1]*	7.6 [7.6–8.2]*	0.88 [0.86–0.91]*	–17%
DS 6	13.3 [10.7–16.1]	11.6 [10.6–12.4]	1.23 [0.95–1.59]*	+16%
UP 3	10.4 [8.2–15.7]	8.1 [6.8–9.2]*	1.35 [1.20–1.68]*	+27%
UP 6	9.3 [8.5–9.8]*	9.1 [7.9–10.0]*	1.08 [1.04–1.11]*	+1%

4. Discussion

This study sought to identify the effect of positive and negative slopes on biomechanical gait variables in people with femoral amputation fitted with a microprocessor knee prosthesis.

The three-dimensional assessment of gait in individuals with amputations, coupled with force platforms, is the preferred method for providing both kinematic and spatiotemporal data (motion capture) as well as kinetic and even electromyographic data [1]. There are simpler tools available, such as quantifying the trajectory of the center of pressure during stance, which have already been studied in individuals with lower limb amputations [14], [25]. The literature has identified trajectory asymmetries between lower limbs in a population of transfemoral amputees [12], [27], however, quantifying these adaptations is challenging. In this study, we proposed the calculation of a symmetry index to provide a global and straightforward measure of segmental kinematic adaptations. We believe that this index indirectly quantifies the adaptations associated with the loss of joint mobility on the amputated side.

Spatiotemporal variables

Only the Uphill slope modified the spatiotemporal variables compared to Level. Furthermore, the change was relatively small with an increase of only 1 cm in intact step length and step width (Table 2). This statistically significant difference must be interpreted in terms of its clinical significance [16], [34]. It is unlikely that a change of 1 cm in the step length and width would give rise to a major clinical change. We also consider that the effect of slope did not lead to any major changes in the spatio-temporal parameters. This lack of change can be explained by the fact that the treadmill speed, defined as comfortable by the participant in the Level condition, was maintained for the other levels. Irrespective of the minor changes we observed, the participants performed less well than people in other studies. In fact, we did not find the values of the recent study by Sturk. These authors observed a significant effect of ascent and descent on walking speed, and specifically of ascent on step width. Although our participants were in similar categories (categories K3 and K4), they differed in age and aetiology of amputation. These authors have a mainly traumatic population and are younger (43 ± 8.6 years) than our participants (60 ± 10 years, vascular amputation). The characteristics of their population may explain why their preferred walking speed is almost double that of our participants (1.16 m/s).

Kinematics

The effect of the slopes was much larger on joint kinematics than on spatiotemporal variables.

The Uphill condition requires raising the foot higher during swing phase to clear the rising ground whereas the Downhill condition requires placing the foot lower than the previous foot. We found that the kinematic parameters were more modified by the Uphill than the Downhill condition (Table 3) with greater modifications on the intact side than the prosthetic side. This effect appeared to be caused by the direction of the slope (up or down), with little effect of slope magnitude (3° or 6°).

More changes occurred on the intact than the prosthetic side (Table 3), indicating that the intact limb adapts more to the change in terrain than the prosthetic limb. Compared to people with transtibial amputation, people with transfemoral amputation are less able to generate propulsive forces [17], [26], [37]. This functional limitation remains present even with a microprocessor knee because the knee is not motorised. As a result, the intact side is forced to compensate for the lack of adaptive capacity of the prosthetic limb. Our results revealed compensation challenges, especially during the swing phase, in the uphill conditions compared to the literature on able-body participants. Indeed, transfemoral amputees exhibited greater variability in trunk and pelvic movements during walking on uneven and sloping surfaces, indicating an affected gait pattern compared to able-body participants [24]. Hip flexion increased on both sides along with pelvic rotation in the frontal plane. This kinematic pattern shortens the functional limb length to maintain sufficient clearance and prevent foot catching. On the prosthetic side, the increase in hip flexion was associated with a decrease in knee flexion, which may be linked to the increase in pelvic tilt, requiring the knee to be bent less to ensure the stride is taken. These results differ from those of Vrieling et al. [37], who found no change in flexion in the trans-femoral amputee group did not modify their knee's flexion and that they were unable to increase the flexion of the prosthetic knee compared with the tibial amputee group. Finally, the significant increase in frontal pelvic rotation observed seems to confirm the more proximal adaptation to the slope than at knee or ankle level. Indeed, walking with an increase in pelvic obliquity during the oscillation phase is a known locomotor pattern reflecting a compensatory strategy for more distal motor deficits [10]. These findings are also consistent in the transfemoral amputee population compared to able-body participants. Trunk and pelvic movement variability in transfemoral amputees is significantly higher during walking on uneven and sloping surfaces than in able-body participants [24]. These observations suggest a proximal compensation in the amputee population due to less optimal use of the prosthetic knee and foot compared to the knee and ankle in able-body participants. Our results also differ from those of by Lura et al. [18] in terms of the magnitude of knee flexion of the Genium knee versus the C-LEG knee in the Uphill condition. The two participants in our sample with a C-LEG knee had greater flexion in the Level and Uphill conditions than the with a Génium knee, but the change in amplitude was smaller, suggesting better use likely more stable with less variability. However, we could not perform a formal comparison because of the small number of participants with these types of knee.

Fewer changes occurred in the Downhill than the Uphill condition compared to Level. This contrasts with other studies that found more changes in knee joint kinematics during downhill gait [9]. An analysis of the characteristics of the population in these studies reveals a younger age, a higher walking speed and, for some studies, a traumatic amputation. However, there are similarities with the study carried out by Lura et al. [18]. These authors imposed different walking speeds in each slope condition. In the slow gait condition, prosthetic knee flexion during swing phase did not change in the 5° downhill condition but it increased/decreased in the 10° downhill condition. Furthermore, knee flexion increased/decreased for all uphill conditions. The spontaneous walking speed of the participants in our study was close to the slow speed of the participants in the study by Lura et al. [18]. Therefore, the changes in joint kinematics we found may be strongly related to the gait speed. For rehabilitation or clinical assessment purposes, consideration of gait speed is essential to analyse the patient's locomotor compensations [30].

Symmetry

More changes occurred in the joint kinematics of the intact than the prosthetic limb with the slope. Dorsiflexion was greater on the intact than the prosthetic side in Level and increased by a further 4.2° in the Uphill 6° condition. This difference is expected because of the stiffness of the carbon “ankle ” joint that limits deformation to allow propulsion of the body. Similarly, knee flexion in swing increased on the intact but not the prosthetic side. Jaegers et al. [15] also found lower peak prosthetic knee flexion in swing than in non-amputated people. They suggested this difference was caused by the slower gait speed. Finally, the greater increase in hip flexion in swing on the intact than the prosthetic limb could relate to the architecture of the integrated ischium socket of the prosthesis and the stump length. The reduction or absence of prosthetic knee flexion during the stance phase alters the kinematics in the sagittal plane of the hip compared to asymptomatic individuals, as the hip does not remain in constant flexion at the beginning of stance. Therefore, to ensure knee extension locking and safety during support, the residual hip tends to move earlier from a flexion to extension motion. Indeed, Jaegers et al. [15] reported that most amputees had greater hip extension at the end of the stance phase than non-amputees, except for those with very short stumps and limited flexion. Our results, combined with the existing literature, indicate that the change in kinematic parameters during uphill gait seems to be linked to the need for greater propulsion to climb the slope because of the passive, non-motorised prosthetic lower limb. Overall, people with trans-femoral amputation have an asymmetric gait pattern. This asymmetry seems to be accentuated when walking uphill or downhill and corresponds to a risky situation for the patient. This raises the question of the appropriateness of prosthetic adjustment to promote symmetry and/or rehabilitation to work on destabilising situations in order to reduce asymmetry in situations of physical stress. The benefits of symmetrizing the gait of an amputee are not conclusively established in the literature. However, in studies of normal gait, there is a tendency toward symmetry between the hemi-bodies with the goal of reducing energy consumption [27]. It is also known that amputees exhibit an increased energy consumption during walking [20]. It appears that the strategy may involve moving towards a more symmetrical gait to enable the amputee to walk longer by reducing energy expenditure. Possible impacts of this symmetrization include prosthesis alignment adjustment or prosthetic knee improvement, as well as a reduction-based approach to train the patient in load transfer to improve symmetry [5]. When an amputee faces challenging situations (uneven terrain, uphill or downhill slopes, etc.), they tend to decrease their speed and increase their asymmetry [15]. In such cases, it is important that they can replicate the effects of rehabilitation to navigate obstacles with greater safety and reduce the risk of falling.

The symmetry index calculated from the p-to-a ratios showed significant changes in all the conditions compared with the Level condition. These changes can be explained by the difference between focal and global quantification. Gait analysis performed in the clinical setting aims to describe gait problems by quantifying joint kinematic variables at specific points in the gait cycle. In contrast, a more global approach aims to characterise joint or segmental kinematics over the entire gait cycle [13]. The two approaches are not mutually exclusive. On the contrary, they offer clinicians a better understanding of gait disorders. In our study, joint angles changed by only a few degrees and asymmetrically for each limb (Table 3) however, the symmetry index based on the p-to-a ratio was significantly different for each Uphill and Downhill condition compared to the Level condition. The shape of the leg movement velocity trajectory in the sagittal plane is likely affected by the mechatronic characteristics of the prosthetic knee. Indeed, prosthetic knees are not designed to be propulsive. Although, in their mechatronic design, these knees do not have any motorisation that can assist the production of propulsive force by the various residual muscle groups. These knees have been developed to produce a braking force to control knee flexion. This mechanical capacity enables the foot to adapt to the length of the stride and also to variations in ground height. Rehabilitation involves performing exercises and learning to master the mechatronic capabilities of the prosthesis. However, daily life usually involves more level walking than slope ascent or descent. This means that prosthesis-users may be less able to adapt to slopes. Although manufacturers are developing solutions involving onboard sensors so that mechatronic triggering is more appropriate, the attentional load of slope ascent and descent remains greater than that of level walking [22], [23]. This is, at least partly, because of having to ensure that the load on the limb during stance is sufficient to trigger the braking mechanism [8].

Perspectives

Many studies are currently focusing on monitoring physical activity in daily life. However, there is no consensus on the variables to evaluate [19]. Recently, Griffiths et al. [11] proposed a classification algorithm based on the measurement of thigh and leg acceleration. This pilot study carried out on fourteen able-body participants and one participant with a femoral amputation highlights the importance of the method for classifying 4 different types of activity: sitting, lying down, standing and walking. The symmetry index proposed in our study can be used to monitor the use of the prosthesis. One of the prospects of this work would be to enable clinicians to determine the number of locomotor movements requiring asymmetry other than walking in a straight line on flat ground. In this way, asymmetry monitoring may enable clinicians to suggest different prosthesis settings when the patient expresses difficulty in using the prosthesis in his daily environment.

Limitations

This study has two main limitations. The first one is the small sample size, which is typical of studies of people with amputation. It is difficult to include large numbers of individuals in rehabilitation centers because they are discharged as soon as they become sufficiently autonomous. Retrospective study designs could be a solution for this issue, particularly since the protocol used in this study is the same as that used in clinical practice in our centre. The second limitation concerns the tool used, namely the treadmill. There is still some debate about the value of using a treadmill to analyse spontaneous gait. Nevertheless, we believe that this tool currently represents the most suitable solution for studying walking conditions such as slope, as it enables the same level of stress to be reproduced between participants over a large number of cycles. Furthermore, although different from spontaneous walking, these tools are also used daily in rehabilitation.

5. Conclusions

The direction of the slope has an impact on joint kinematic parameters and increases segmental asymmetry reflecting difficulty compensating for the limitations of the prosthetic limb. The changes in joint kinematics when walking uphill seem to be related to the need to propel oneself to adapt to the slope because the prosthetic lower limb is passive and not motorized. In contrast, during the descent, the changes are related to the need to slow down the movement of the body to adapt to the slope and take advantage of the mechatronic characteristics of the prosthesis. Although the prosthesis breaks knee flexion, the symmetry ratio showed a significant loss of symmetry in all slope conditions, suggesting difficulty of locomotor compensation. The quantification of these biomechanical parameters seems appropriate to identify locomotor adaptations to slopes to provide objective elements for the achievement of a personalized gait rehabilitation program. In addition, this quantification of segmental symmetry may be a method for future work on the evaluation of compensation strategies with the aim of optimizing these strategies, depending on the locomotor situations encountered.

Conflict of interest disclosure

The authors declare that there is no conflict of interest.

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