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Uphill and downhill walking in unilateral lower limb amputees

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Abstract

Objective: To study adjustment strategies in unilateral amputees in uphill and downhill walking.
Design: Observational cohort study.
Subjects: Seven transfemoral, 12 transtibial unilateral amputees and 10 able-bodied subjects.
Methods: In a motion analysis laboratory the subjects walked over a level surface and an uphill and downhill slope. Gait velocity and lower limb joint angles were measured.
Results: In uphill walking hip and knee flexion at initial contact and hip flexion in swing were increased in the prosthetic limb of transtibial amputees. In downhill walking transtibial amputees showed more knee flexion on the prosthetic side in late stance and swing. Transfemoral amputees were not able to increase prosthetic knee flexion in uphill and downhill walking. An important adjustment strategy in both amputee groups was a smaller hip extension in late stance in uphill and downhill walking, probably related with a shorter step length.

In addition, amputees increased knee flexion in early stance in the non-affected limb in uphill walking to compensate for the shorter prosthetic limb length. In downhill walking fewer adjustments were necessary, since the shorter prosthetic limb already resulted in lowering of the body.
Conclusion: Uphill and downhill walking can be trained in rehabilitation, which may improve safety and confidence of amputees. Prosthetic design should focus on better control of prosthetic knee flexion abilities without reducing stability.

1. Introduction

Since our surroundings are not all at the same level, it is important to be able to walk up and down a sloping surface. Uphill and downhill walking increases the chance of falling due to slipping or losing balance [1]. However, the human locomotion pattern is highly adaptable to changes in gradient [2–5]. Studies performed on able-bodied (AB) subjects have demonstrated adjustment strategies in walking up and down a slope.

In uphill walking AB increase hip and knee flexion and ankle dorsiflexion in swing and at initial contact to provide safe foot clearance and to enable positioning of the foot at a higher level [3–8]. The increased hip flexion may also be caused by forward bending of the trunk [2,3]. The increased ankle dorsiflexion in stance is a direct result of the slope gradient [6]. Furthermore, knee flexion is reduced in midstance, which serves to lift the body and eases foot clearance of the other limb [6]. In downhill walking AB increase knee flexion from loading response to early swing. The resultant shortening of the limb lowers the body, facilitates initial contact of the other limb on the lowered surface, and reduces impact force. Moreover, knee flexion assists in rotation of the tibia in the sagittal plane and brings the body forward over the stance foot [1,2,4,6,7,9,10]. Ankle dorsiflexion is increased from late stance to midswing,

Table 1

Subjects characteristics, prosthetic components and gait velocity; Mean values and range of age, weight, height, and time since amputation, and mean values and standard deviations of Amputee Activity Scale (AAS) and gait velocity

	TF ($n = 7$)	TT ($n = 12$)	AB ($n = 10$)
Gender	Six male, one female	Ten male, two female	Nine male, one female
Age (years)	44.0 (30–71)	49.6 (27–65)	45.2 (34–63)
Weight (kg)	79.0 (67–97)	84.2 (71–98)	86.5 (72–98)
Height (cm)	182.6 (174–194)	180.9 (165–194)	184.4 (172–192)
Time since amputation (months)	210.7 (18–504)	207.8 (23–672)	
Cause of amputation	Four trauma, three oncology	Six trauma, four oncology, two vascular	
Prosthetic foot	Three Multiflex[1], two SACH[2], two C-walk[2]	Four C-walk[2], three SACH[2], two Quantum[3], one Multiflex[1], one Griessinger[2], one Safe II[4]	
Prosthetic knee	Three Graph-Lite[5], one C-leg[2], one 3R60[2], one Safe Life[6], Total knee[7]		
AAS	35.9 ± 26.9	33.8 ± 26.1	
Gait velocity (m/s)			
Level	1.03 ± 0.19	1.22 ± 0.16	1.34 ± 0.13
Slope	1.01 ± 0.23	1.21 ± 0.16	1.34 ± 0.14

Prosthetic devices by (shown in superscript numbers)—1: Endolite; 2: Ottobock; 3: Hosmer; 4: Forsee; 5: The Lin; 6: Proteval; and 7: Ossur.

whereas hip flexion is decreased from midswing to early stance, which results in a pull back of the swing limb, shortening of the step length and easier positioning of the foot on the lower surface [2,4,6,7,9].

Amputees may experience limitations in function in slope walking due to the loss of muscles, joint(s) and nerves in the amputated limb. Prosthetic knees and feet have different properties compared to human joints, and the length of a prosthetic limb is usually reduced. Consequently, transfemoral amputees (TF) and transtibial amputees (TT) may not be able to perform the required adjustment strategies in uphill and downhill walking, which may cause loss of balance. To date, slope walking in amputees has not been studied. The objective of this study was to determine the strategies that amputees use to adjust their gait to uphill and downhill walking.

We hypothesized that amputees would not be able to increase flexion in the prosthetic knee and dorsiflexion in the prosthetic foot in swing and early stance in uphill walking. As an adjustment strategy, flexion in swing and early stance of the intact lower limb joint(s) proximal to the amputation level would be larger in TF and TT than in AB. As an adjustment to the shorter prosthetic limb length, we expected increased flexion in the non-affected lower limb joints in TF and TT in swing and early stance, compared to AB. In downhill walking we hypothesized that prosthetic knee flexion in TF would not increase in stance and early swing, and prosthetic ankle dorsiflexion in TF and TT would not increase from late stance to midswing. To compensate for the higher impact force, we expected an increase in non-affected knee flexion in TF and TT in early stance, compared to AB.

2. Methods

2.1. Subjects

TF and TT were approached via a prosthetics workshop. Inclusion criteria were: a unilateral amputation for at least 1 year, the use of a prosthesis on a daily basis, and the ability to walk more than 50 m without walking aids. An AB control group was recruited through advertisements at the local blood bank, hospital, and television station. Subjects were excluded if they had any medical condition that could affect their mobility or balance, such as neurological, orthopaedic or rheumatic disorders, or if they had cognitive problems or severe impaired vision. In amputees, reduced sensation of the non-affected limb, wounds or pain at the stump, or fitting problems of the prosthesis were also exclusion criteria.

Seven TF, 12 TT and 10 AB agreed to participate in the study. The medical ethics committee approved the study protocol. All subjects signed informed consent before testing. The amputees used different types of prosthetic feet. TF were provided with a free moveable prosthetic knee. The Amputee Activity Score (AAS) was used to obtain information on the activity level in amputees [11,12]. A higher score on the AAS represents a higher activity level. The subject characteristics and prosthetic components are presented in Table 1.

2.2. Apparatus

The study was performed in a motion analysis laboratory. An upward and downward ramp, both 2-m long, were placed halfway an 8-m long walkway. The gradient of the slope was 5%, which is advised in building instructions as maximum

gradient [13]. The runs were videotaped with a camera[1] in the sagittal plane, which moved with the subject along the walkway. The recording frequency was 25 Hz. Subjects were provided with six electro-goniometers, for which a high accuracy and repeatability were proven.[2] The goniometers were placed bilaterally on the ankle joint (or prosthetic foot), knee joint (or prosthetic knee), and hip joint. Joint angles were measured in the sagittal plane. Prior to testing, the goniometers were zeroed while the subjects stood upright with both limbs straight and ankles in plantigrade. The beginning and end of the walkway were fitted with infrared lights, which registered the passing of a subject. Data were recorded by a portable data acquisition system (PORTI)[3] and analyzed with a custom developed Gait Analysis System (GAS).[4] The sampling frequency was 800 Hz, which was low-passed filtered and resampled to 100 Hz. The goniometer data were filtered by a low-pass second-order Butterworth filter with a cut-off of 10 Hz.

2.3. Procedure

All subjects were instructed to walk at their self-selected comfortable velocity. Three different walking conditions were performed; level, uphill and downhill walking. The study consisted of four level and four slope walking runs. In slope walking the subjects first walked up and then down the slope.

2.4. Outcome parameters

Average gait velocity in level and slope runs was calculated from the length of the 8-m walkway divided by the necessary time to walk over this walkway. In slope walking the mean gait velocity of uphill and downhill walking was assessed. The joint angles of the prosthetic and non-affected limbs in amputees were analyzed separately. In AB the mean joint angles of the right and left limbs were used in the analysis to minimize the influence of asymmetry. In slope walking, the middle stride on the slope trajectory was selected. In level walking three successive strides in the middle of the walkway were analyzed. The joint angles of the hip, knee and ankle were assessed at several events in the gait cycle and these are shown in Fig. 1.

2.5. Statistical analysis

Data were tested with the Kolmogorov–Smirnov test and were normally distributed. Significant differences in
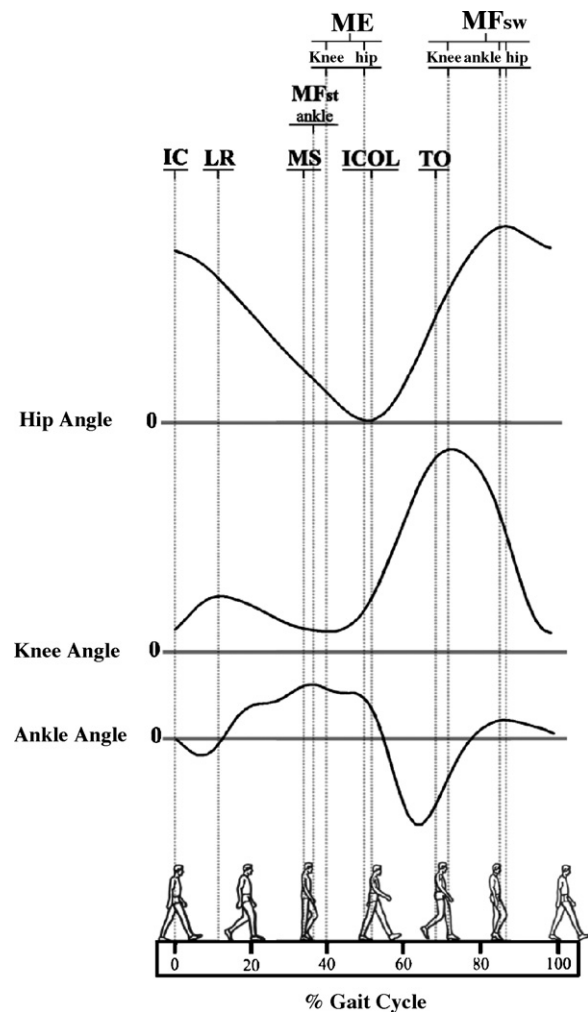

Fig. 1. The analyzed angles of the hip, knee and ankle at initial contact (IC), loading response (LR), midstance (MS), maximum dorsiflexion in stance (MFst), maximum extension in stance (ME), initial contact of the opposite limb (ICOL), toe-off (TO) and maximum (dorsi)flexion in swing (MFsw). LR was assessed at maximum knee flexion in stance, and MS when the ankle of the swing limb passed the stance limb.

subject characteristics were tested with ANOVA. A mixed design ANOVA with repeated measures was performed on gait velocity and joint angles, followed by post hoc analysis according to the least-significant difference (LSD) method. This analysis allowed adjustment for differences in outcome parameters among study groups that already existed in level walking and detection of differences in outcome parameters that were specific to uphill and downhill walking. The joint angles in both the prosthetic and non-affected limbs of TF and TT were compared with the non-affected limb of AB. Within-subject variables included the outcome parameters in the different walking conditions (level-uphill and level-downhill), and the between-subject factors were the study groups. The level of significance was set in all analyses at $p \leq 0.05$.

3. Results

No statistically significant differences were found in the characteristics of the study groups. Gait velocity was similar in level and slope walking within the study groups. Gait velocity was increased in AB compared to TF and TT, and TT walked faster than TF (Table 1). The lower limb joint angles are shown in Figs. 2–4. Only significant interaction effects in joint angles between the study groups in uphill and downhill walking compared to level walking are reported in this section.

Uphill walking—prosthetic limb in amputees: In the hip an interaction effect was shown at IC ($F_{9.8}, p < 0.01$), ICOL ($F_{5.5}, p = 0.01$) and MFsw ($F_{11.4}, p < 0.01$). In uphill walking hip flexion at IC and MFsw increased in AB and TT, which was not seen in TF. Hip extension at ICOL decreased in TT in uphill walking, but not in TF and AB. In the knee an interaction effect was demonstrated at IC ($F_{6.8}, p < 0.01$)

and MFsw ($F_{4.1}, p = 0.03$). Knee flexion at IC in uphill walking was enlarged in AB and TT, but not in TF. In TF knee flexion at MFsw was reduced in uphill walking, which was not demonstrated in AB and TT. The ankle showed no interaction effects.

Uphill walking—non-affected limb in amputees: The hip showed an interaction effect at MEst ($F_{3.8}, p = 0.04$). In TF hip extension was decreased at MEst in uphill walking, but not in AB and TT. In the knee an interaction effect was seen at LR ($F_{3.7}, p = 0.04$) and TO ($F_{6.4}, p < 0.01$). TF and TT increased knee flexion more than AB did at LR in uphill walking. TF increased knee flexion at TO in uphill walking, which was not shown in AB and TT. In the ankle an interaction effect was demonstrated at TO ($F_{5.5}, p = 0.01$). TF increased ankle plantar flexion at TO in uphill walking, AB and TT did not.

Downhill walking—prosthetic limb in amputees: In the hip an interaction effect was shown at ICOL ($F_{6.1}, p < 0.01$)
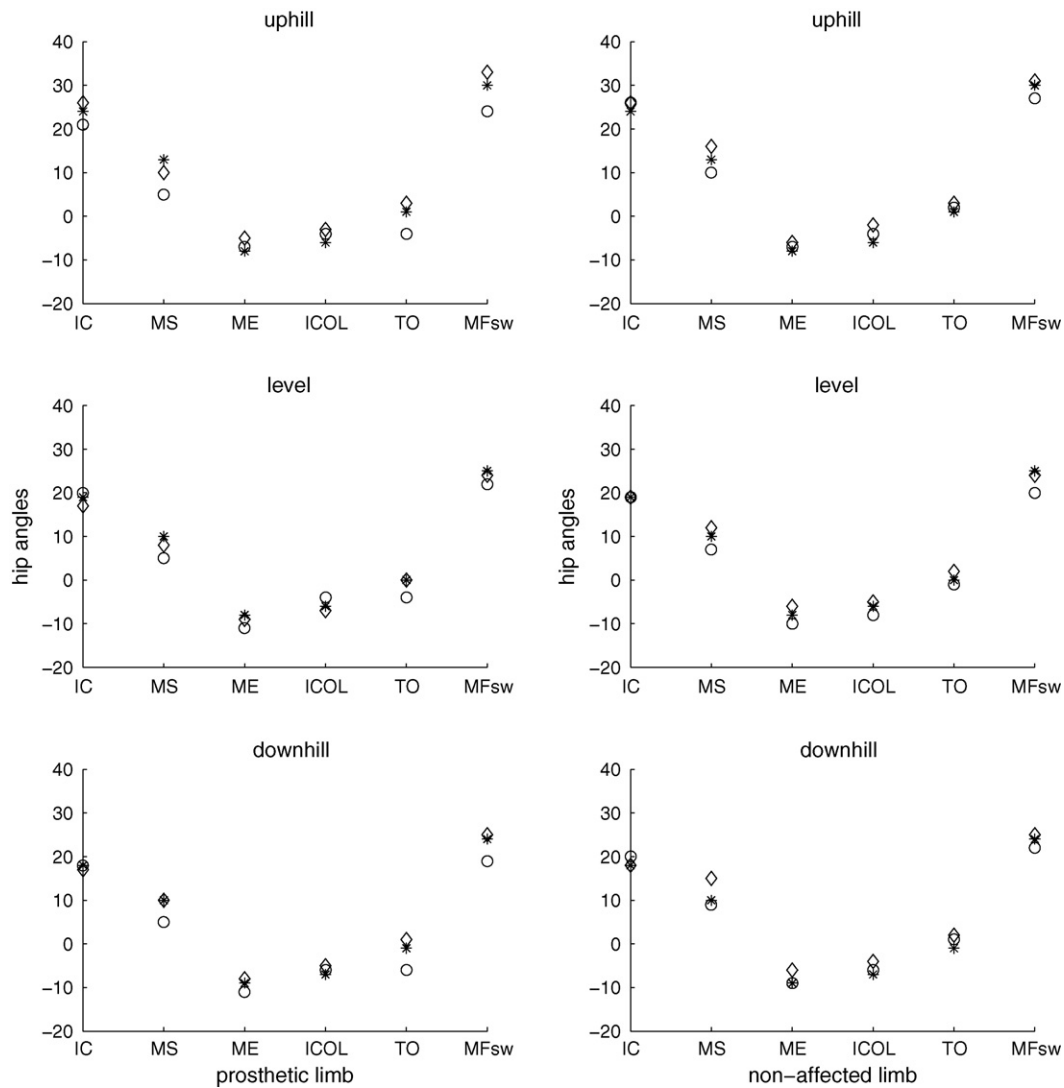

Fig. 2. Mean values of hip angles of the prosthetic limb and non-affected limb in TF (○), TT (◇), and AB (✳) during uphill, level and downhill walking. Hip flexion is positive, extension negative.
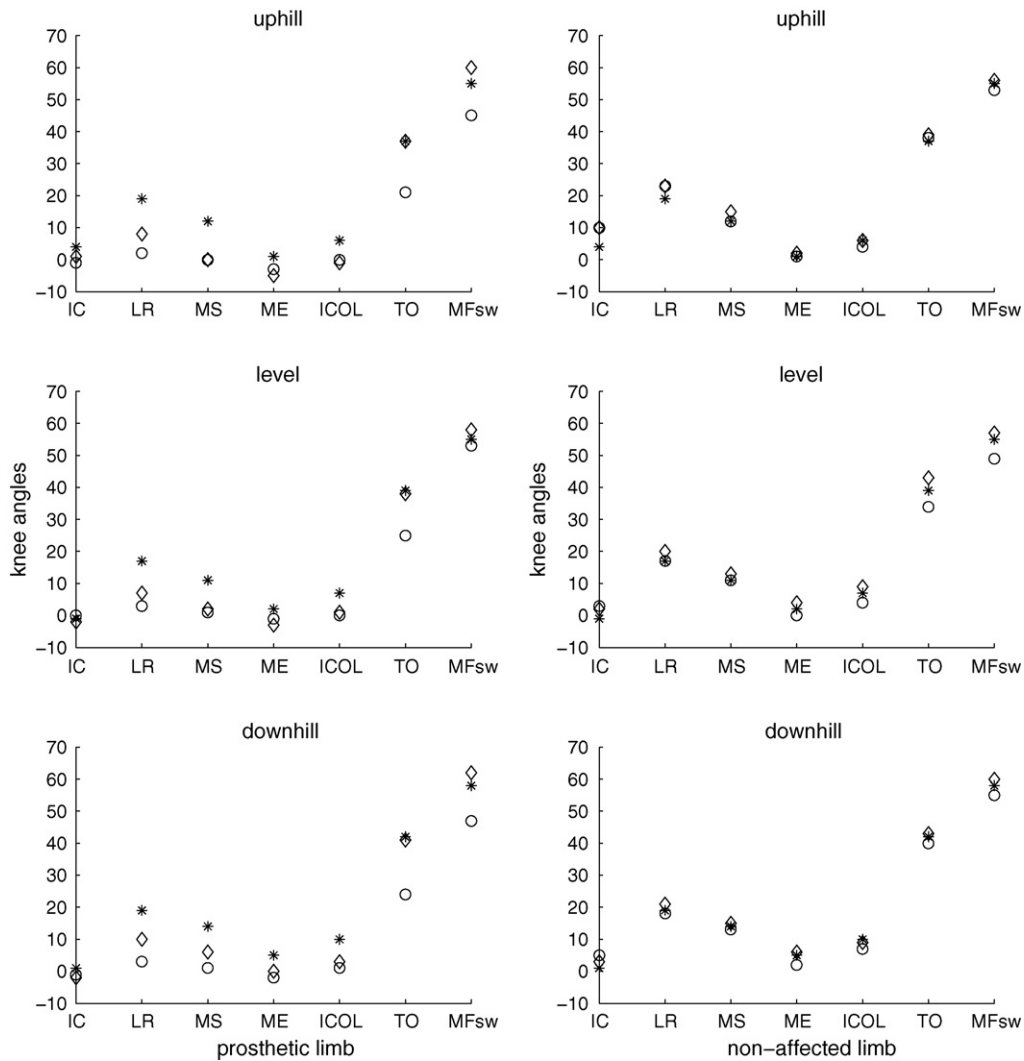
Fig. 3. Mean values of knee angles of the prosthetic and non-affected limb in TF (○), TT (◇), and AB (✳) during uphill, level and downhill walking. Knee flexion is positive, extension negative.

and MFsw ($F_{3.6}$, $p = 0.04$). Hip extension at ICOL in AB and TF increased slightly in downhill walking, whereas in TT a decrease was shown. In downhill walking hip flexion at MFSW was reduced in TF, but not in AB and TT. The knee showed a significant interaction effect at MEst ($F_{3.7}$, $p = 0.04$) and MFsw ($F_{4.1}$, $p = 0.03$). In downhill walking AB and TT decreased knee extension at MEst, but this was not seen in TF. In AB and TT knee flexion at MFsw in downhill walking was increased, in TF reduced. In the ankle no interaction effects were found.

Downhill walking—non-affected limb in amputees: No significant interaction effects were demonstrated in this limb.

4. Discussion

The main goal of this study was to establish how amputees adjust their gait pattern to uphill and downhill

walking. Adjustment strategies can involve the prosthetic or the non-affected limb. The adjustment strategies on the prosthetic side of TT were very similar to those in AB. In uphill walking hip and knee flexion at initial contact and hip flexion in swing were increased in the prosthetic limb of TT. In downhill walking TT showed more knee flexion on the prosthetic side in late stance and swing.

As hypothesized, TF were not able to increase prosthetic knee flexion in uphill and downhill walking. For example, maximum swing knee flexion in the prosthetic limb in uphill walking was 8° lower than in level walking. Prosthetic knee flexion in swing relies on activity of the hip flexors and the properties of the prosthetic knee. In stance, prosthetic knee flexion is only possible to a limited extent, depending on the type of prosthetic knee, to prevent unlocking during weight bearing. In this study one subject was fitted with a microprocessor-controlled prosthetic knee joint. Advantages of such a knee joint are the ability to flex in the beginning of stance and to decrease damping in late stance,
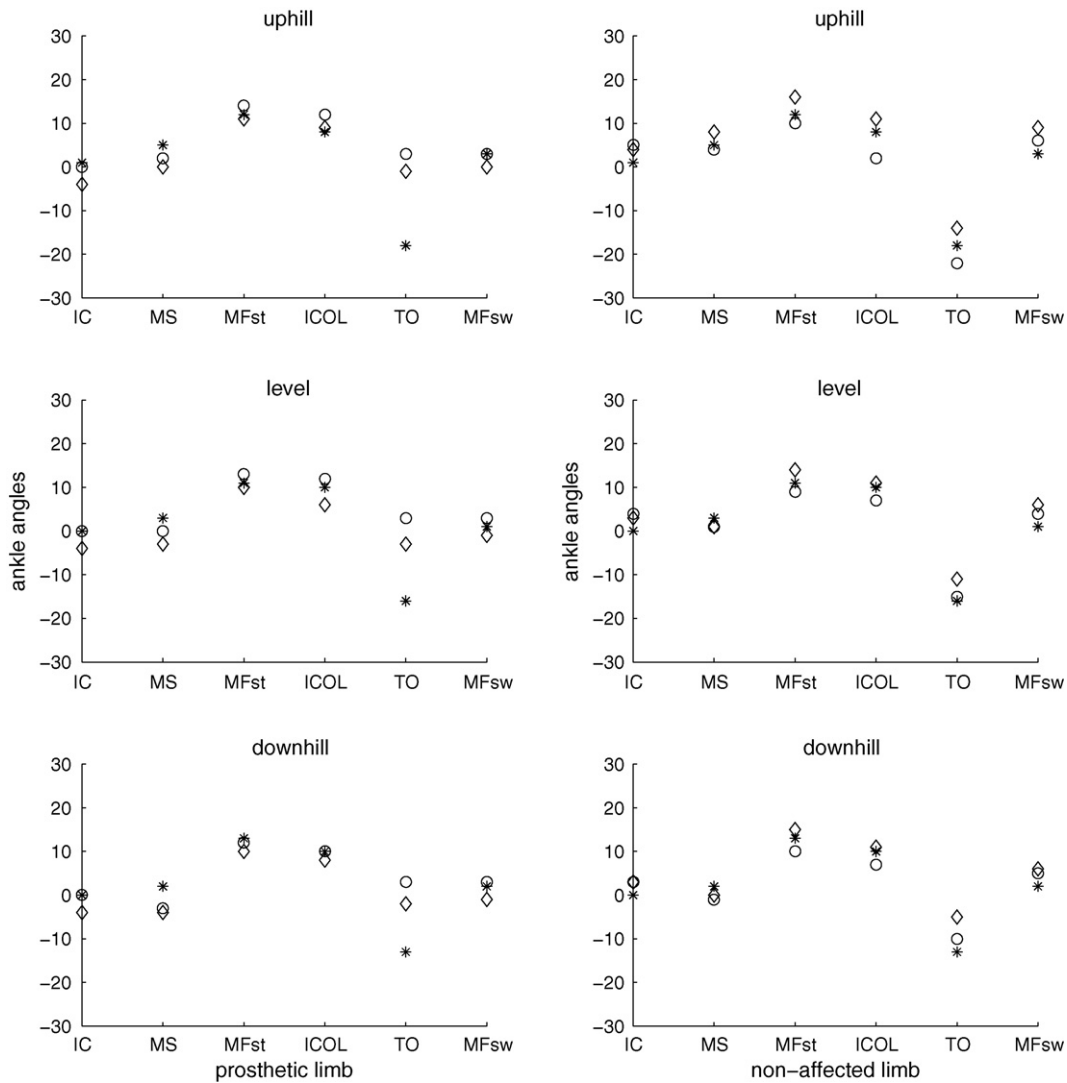
Fig. 4. Mean values of ankle angles of the prosthetic and non-affected limb in TF (○), TT (◇), and AB (✳) during uphill, level and downhill walking. Ankle dorsal flexion is positive, plantar flexion negative.

which respectively contribute to shock absorption in loading response and to knee flexion in swing. The subject who was provided with a C-leg showed an increased knee flexion in late stance and swing in level, uphill and downhill walking compared to TF with a conventional prosthetic knee, but at loading response no differences were observed.

Since the prosthetic knee offers no possibility to adjust gait to slope walking, TF are required to make use of other strategies. One area where adjustment strategies can be generated is the hip on the prosthetic side. We hypothesized that TF would increase hip flexion in the prosthetic limb in swing and initial contact to provide safe foot clearance and foot positioning in uphill walking, but this adjustment strategy was not found. In downhill walking TF showed a smaller hip flexion in the prosthetic limb in swing, which could indicate a shorter step length. By reducing the step length, positioning of the prosthetic foot on the lowered surface is eased, because the difference in height is smaller.

In TT the hip on the prosthetic side was used for another adjustment strategy. TT decreased hip extension on the prosthetic side at initial contact of the non-affected limb in uphill and downhill walking. In uphill walking the reduced hip extension in late stance facilitates positioning of the opposite non-affected foot on the higher surface, because the body remains more lifted. The reduced hip extension can be caused by a shorter step length, which decreases the height difference that the prosthetic limb has to adjust to. Furthermore, a forward bending of the trunk may add to the decrease in hip extension. In downhill walking the reduced hip extension is most likely explained by a shorter step length. However, reduced hip extension in stance is not beneficial for lowering of the body, which is required in downhill walking.

Uphill and downhill walking did not result in major adjustments in the prosthetic ankle of amputees. Dependent on the stiffness of the prosthetic foot, a passive adaptation to

the slope gradient is possible in stance. Thirteen amputees in our study used relatively flexible prosthetic feet, whereas six amputees were provided with more rigid prosthetic feet. Due to the majority of flexible prosthetic feet, amputees could achieve an increase in dorsiflexion that was similar to AB in early stance in uphill walking and in late stance in downhill walking.

Amputees can also use the non-affected limb to carry out adjustment strategies. In uphill walking TF decreased maximum hip extension in the non-affected limb, which led to an easier positioning of the prosthetic foot. The decrease in hip extension in this limb can be interpreted similarly to the prosthetic limb, which we mentioned earlier. A second adjustment strategy that TF applied in the non-affected limb in uphill walking was increasing knee flexion and ankle plantar flexion at toe-off, which may assist in loading of the prosthetic limb. Since more height has to be overcome, it is harder in uphill walking to shift the body weight above the prosthetic limb.

In both amputee groups, knee flexion in the non-affected limb at loading response was increased in uphill walking. Since the shorter length of the prosthetic limb restricted lifting of the body, amputees needed to place the non-affected limb in a more flexed position on the elevated surface. In downhill walking no specific adjustment strategies were seen in the non-affected limb of amputees. We hypothesized that an increase in non-affected knee flexion would compensate for the higher impact force. Shortening of the stance limb to lower the body is an important feature in downhill walking and the more extended position of the prosthetic limb would force the non-affected limb to arrive at the ground from a higher position. However, in amputees the shorter prosthetic length already ensured lowering of the body.

Knee extension in stance results in positive mechanical work in uphill walking, which is essential for moving the body up the slope against gravity. The already straightened position of the prosthetic knee at loading response made a large increase in positive work in TF not possible. A conventional prosthetic knee can only provide a small amount of positive work due to energy storage in the extension spring. The larger knee flexion of the non-affected limb at loading response in amputees increased the amount of positive work in this limb and enabled an adequate lift of the body. In this way, one adjustment strategy, namely increased non-affected knee flexion at loading response, compensated for both the shorter length and the limited flexion of the prosthetic limb. In downhill walking knee flexion in late stance created negative mechanical work, which is necessary for lowering of the body. In TF the increase in flexion in the prosthetic knee was lower than in AB and TT. However, TF were the only study group that showed an increase in the production of negative work on the prosthetic side when walking downhill compared to level walking, because in TF maximum prosthetic knee extension was not decreased in downhill walking.

Since the differences in lower limb joint angles between the conditions and study groups were mostly limited to several degrees, the clinical relevance can be questioned. However, the slope of the gradient was only 5% and walking up and down steeper slopes may give rise to more clinically important changes. Increasing the steepness of the slope may result in the usage of other adjustment strategies in amputees and AB. Finally, the study was limited by the different types of prosthetic knees and feet, which may have led to the application of different adjustment strategies. Due to the small sample size we were not able to study the influence of the diverse prosthetic devices on the outcome measures.

5. Conclusion

An intact knee joint is important for walking up and down a hill safely. TT increased knee flexion in the prosthetic limb to adjust gait to the slope gradient, whereas in the prosthetic knee of TF this strategy was absent. To improve safety and confidence of amputees during walking up and down a hill, training of these motor tasks in rehabilitation is recommended, in order to practise adjustment strategies on slopes of different gradients. Prosthetic knee design should focus on better flexion properties in stance and swing without compromising stability requirements.

Acknowledgements

Conflict of interest

Authors state that no conflicts of interest are present in the research.

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