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Lower limb amputee gait characteristics on a specifically designed test ramp: Preliminary results of a biomechanical comparison of two prosthetic foot concepts

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ABSTRACT

Background: For demanding activities in daily life, such as negotiating stairs, ramps and uneven ground, the functionality of conventional prosthetic feet ("Daily Life Feet" - DLF) is often limited. With the introduction of microprocessor-controlled feet (MPF) it was expected that the functional limitations of DLF might be reduced. The purpose of the present study was to investigate biomechanical gait parameters with DLF and MPF when walking on a specifically designed ramp involving abruptly changing inclination angles as a scenario reflecting typical situations related to walking on uneven ground.

Research Question: The specific aim of the study was to answer the research question if the advanced adaptability of MPF to different ground slopes would lead to more natural motion patterns and reduced joint loading compared with DLF feet.

Methods: A specifically designed ramp was installed within a gait lab. During downward motion on this ramp biomechanical parameters – ground reaction forces, joint moments and joint angles were obtained both with DLF and MPF used by four transtibial amputees. A control group of 10 non-amputees (NA) was measured with for comparison.

Results: The NA group managed the ramp element with the abruptly changing inclination with a specific ankle joint adaptation. Compared to DLF the MPF considerably improved the ankle adaptation to the abruptly changing inclination which was reflected by a significantly increased stance phase dorsiflexion which was comparable to the NA group. The peak value of the knee extension moment on the prosthetic side was significantly increased with DLF, whereas it was almost normal with MPF (DLF: 0.71 ± 0.13 Nm/kg, MPF: 0.42 ± 0.12 Nm/kg, NA: 0.36 ± 0.07 Nm/kg, $p < 0.05$ and $p < 0.01$). The external knee adduction moment was generally reduced for the transtibial amputees and did not show differences between foot designs.

Significance: The adaptable ankle joint motion of the MPF is a crucial requirement for a more natural motion pattern and leads to a reduction of sagittal knee joint loading on the prosthetic side.

1. Introduction

Modern prosthetic feet enable transtibial (TT) amputees to live a life with a high level of mobility. Energy-Storing-and-Returning (ESR) feet are well established as state-of-the-art solutions. The specific design features of these feet allow for an almost natural general motion pattern during level walking compared to non-amputees [1,2]. For more demanding activities of daily living, such as walking stairs and slopes, the functionality of ESR feet is often limited. For these tasks, the lack of adaptability of ESR and other conventional prosthetic feet (e.g. the SACH foot) with respect to the required range of motion and the lacking lower leg muscle functions lead to specific compensatory motion patterns [3–6]. Primarily, this concerns the control of the knee joint that is characterized by an abnormal knee extension in situations in which knee flexion would be the normal motion pattern [3,7–10].

With the introduction of microprocessor-controlled feet (MPF) it was expected that the functional limitations described for ESR and other conventional feet might be reduced. In this new generation of prosthetic feet, an integrated ankle joint unit is the central functional element allowing for a sensor-controlled dorsiflexion and plantarflexion with varying ranges of motion. Furthermore, the microprocessor control enables variable motion resistance realized by hydraulic or motor

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components [8,11]. Previous studies documented the benefit of MPF for TT amputees during level walking and walking on stairs and slopes [12–16].

However, walking on uneven ground is an even more challenging situation for TT amputees. Similar to walking on stairs and ramps it requires an increased ankle joint range of motion compared to level walking. Furthermore, walking on uneven ground comprises situations with abruptly changing inclines and declines. The sensory control of MPF is expected to be particularly beneficial compared to ESR and other conventional prosthetic feet for such demanding conditions. In contrast to walking on even ground, ramps, stairs or over obstacles, biomechanical data is lacking for scenarios that reflect walking on uneven ground with abruptly changing inclinations, especially comparing prosthetic feet with different constructive features.

The purpose of the present study was to investigate biomechanical gait parameters when walking on a specifically designed ramp involving abruptly changing inclination angles. The ramp was designed to mimic a typical situation encountered while walking on uneven terrain and was used to compare the functionality of MPF and conventional feet used by TT amputees. The specific aim was to answer the research question if the advanced adaptability of MPF to different ground conditions would lead to more natural motion patterns (i.e. kinematic and kinetic parameters closer to normal values of non-amputees) and reduced joint loading compared with ESR feet. Furthermore, due to of well-known effects of different sagittal prosthetic foot designs on sound side knee loading in level walking [17], these biomechanical parameters are also of interest. A control group of non-amputees (NA) was studied for comparative evaluation of the results and for describing the natural motion pattern.

2. Methods

2.1. Patients and control group

The study included four male unilateral TT amputees with K-Level 3 and 4 (Table 1). Amputation etiologies were peripheral arterial disease in one patient and trauma in the other three patients. All participants were experienced users of TT prostheses and were currently fitted with modern TT prostheses including a total surface bearing socket and a conventional prosthetic foot (“Daily Life Foot” - DLF). Further specific inclusion criteria were that the socket fit enabled pain free walking, no comorbidities that limited activity in the lab to less than 4 h, residual limb length > 12 cm and body weight < 100 kg. A control group of ten NA reference subjects (6 male, 4 female) was used for comparison (Table 1). The inclusion criterion for these participants was absence of any orthopedic or neurological impairment.

The study was conducted in accordance with the Declaration of Helsinki and was approved by the ethics committee of the University Medical Center Göttingen (UMG, study number 17/7/16).

2.2. Measurements devices

For the simulation of walking on uneven ground with abruptly changing inclinations, a specific ramp was designed (“Göttingen Parcours”, Fig. 1). The parcours was installed within the calibrated measurement volume of the gait lab. It consists of a 3 m downhill walkway (10° inclination) followed by specific uphill and downhill elements with opposite inclination angles of 10°. The uphill element was fixed on a force plate (9287A, KISTLER, Winterthur, CH, 1 KHz sampling frequency) for measuring ground reaction forces (GRF). An optoelectronic motion capture system was used to measure the 3D coordinates of retroflective markers placed on anatomical landmarks (12 BONITA cams, VICON, Oxford, GB, measurement frequency 200 Hz).

2.3. Prosthetic feet

All DLF used were current ESR feet primarily designed for lower limb amputees of K-Level 3 and 4. The MPF studied for comparison was the hydraulic Meridium foot (Otto Bock, Germany). The Meridium provides 36° range of motion in the sagittal plane (14° dorsiflexion and 22° plantarflexion). This allows for a better adaptability on ramps by providing more dorsiflexion on upward ramps and more plantarflexion on downward ramps as compared to ESR feet. Its specific constructive feature, a four-bar polycentric linkage, enables additional plantarflexion in terminal stance. The foot adapts in real-time (ROM and resistances adapt to the current gait situation within the same step). A detailed description of the design and functionalities of Meridium has previously been published [11,16].

2.4. Test procedure

The TT amputees attended two gait lab visits. During the first session, the participants were informed about study details and provided their informed consent. Then they practiced walking on the parcours with their DLF to familiarize with the test situation before the actual biomechanical testing. Afterwards, a set of 17 passive markers described in a previous paper [19] was attached to anatomical landmarks. The ankle marker placement is crucial in amputee gait analysis. For the Meridium the ankle marker was aligned with the mechanical axis of the device. For the DLF, the marker was attached to the foot shell at the approximate location of the ankle. The TT amputees walked the 3 m downhill section with their self-selected comfortable speed. Eight trials were performed with the prosthetic side contacting the uphill element first followed by eight trials with the sound side contacting the uphill element first.

Thereafter, an experienced and product-certified CPO fitted the participants with the MPF. The static prosthetic alignment was adjusted using the LASAR posture system in accordance with the biomechanically based recommendations by Blumentritt [18]. This was followed by a dynamic alignment optimization and a training session with the new component. After the first visit, the TT amputees used the MPF for two weeks in their home environment.

Table 1
Demographic data of patients and NA group.

	Age [years]	Mass [kg]	Height [cm]	Follow-up after amputation [years]	Prosthetic foot
NA group [n = 10] **mean ± SD**	23 ± 3	71 ± 13	173 ± 8		
Patients					
P1	48	68	183	3	1C30 Trias
P2	74	84	174	15	1C40 C-Walk
P3	48	80	177	13	1C60 Triton
P4	56	87	178	35	1C40 C-Walk
mean ± SD	56 ± 12	79 ± 8	178 ± 4		

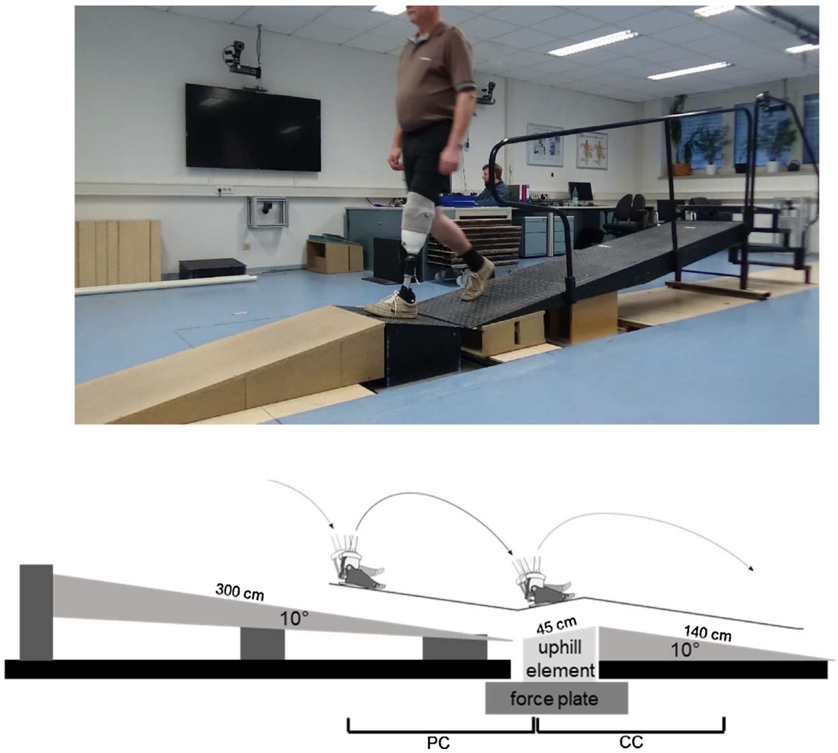

Fig. 1. The "Göttingen Parcour" (GP) used to test ramp walking with an abruptly changing inclination. (top: whole arrangement, center: scheme of the MPF adaptation to the different conditions, bottom: sagittal contour scheme with geometric sizes; PC: "preparing cycle"; CC: "changing cycle").

During the subsequent visit, the TT amputees practiced walking on the parcours in order to get accustomed to the test scenario using the MPF. The biomechanical testing was repeated under identical conditions as in the first session.

In a similar way, the biomechanical parameters were determined for the control group. After arriving in the lab, the participants received instructions and provided their informed consent. In a 15-minute period, they adapted to walking on the parcours and the biomechanical measurements were conducted with the same marker set. Eight repeated trials with self selected comfortable speed were recorded.

2.5. Evaluation parameters and data processing

All biomechanical parameters were calculated with customized algorithms using VICON Body Builder software. This includes the calculation of external joint moments as described previously [20]. For kinematic analysis, the flexion angles of the lower limb joints were calculated. From the set of available kinetic parameters, the external ankle (sagittal) and knee (sagittal and frontal) moments were selected for evaluation. These parameters are considered as essential parameters for the characterization of functionality of prosthetic feet as well as their impact on biomechanical parameters of the knee joint [1,21,22]. The external knee adduction moment was used as a well-accepted proxy for characterizing knee loading and overloading effects in TT amputees during level walking [1,21,23]. The kinematic parameters were time-normalized to two consecutive gait cycles: The last gait cycle before the uphill contact ("preparing cycle" - PC) and the gait cycle starting with uphill element contact ("changing cycle" – CC, Fig. 1). The PC was used to investigate preparatory adjustments prior to the abruptly changing ramp inclination. The kinetic parameters were normalized to CC. Group means were calculated from the mean values across all single trials for each patient and each parameter. For local maxima and minima of the biomechanical parameters (Table 2) comparisons were performed between TT amputees and the control group using the Mann-Whitney *U* test (Table 2). The Student's *t*-test applicable for very small sample sizes was used to identify significant differences between prosthetic feet [24]. The significance level was set at $p < 0.05$ for two tailed tests.

3. Results

3.1. Time-distance parameters

No significant differences between prosthetic feet were found for the mean walking speed. The NA group walked slightly faster (MPF: 1.00 ± 0.03 m/s, DLF: 1.03 ± 0.05 m/s, NA: 1.19 ± 0.04 m/s). The mean stance phase duration during the force plate element contact showed no significant group differences (MPF 61.8 ± 1.4%GC, DLF 61.0 ± 1.2%GC, NA 61.4 ± 0.03%GC).

3.2. Kinematics

The kinematic characteristics of NA revealed different motion patterns for PC and CC. For PC, the flexion angles of the ankle, knee and hip joint were similar to the motion pattern previously reported for descending ramps [3,12]. The CC kinematic pattern for knee and hip was similar to that in level walking. A distinct difference was seen for the ankle joint with a fast and prolonged dorsiflexion (Fig. 2).

The plantarflexion angles for the TT amputees showed typical foot-related differences for the prosthetic limb. The initial MPF plantarflexion during PC is increased by approximately 5° compared to DLF. The final PC stance phase plantarflexion for push-off was considerably reduced for both prosthetic feet in comparison to NA. In the following CC stance phase the MPF shows a significantly increased mean dorsiflexion of 16.9° ($p < 0.05$), comparable to the NA parameter (Fig. 2, Table 2). Moreover, the MPF remains in a dorsiflexed position of approximately 7° during the swing phase (Fig. 2). The terminal plantarflexion both for PC and CC was reduced by approximately 10° for MPF and DLF compared with NA (Fig. 2). The sound side plantarflexion angles showed no significant differences between both prosthetic situations.

Table 2
Selected mean peak values of the biomechanical data measured (in bold print: patient values significantly different compared with NA group; Δ: significant difference between prosthetic feet (p ≤ 0.05), moment values represent external joint moments; max_CC: maximum during CC; ne: non-existent).

	Prosthetic side		Sound side		NA group
	MPF	DLF	MPF	DLF	
Kinematics					
Ankle dorsiflexion angle (max_CC) [°]	16.4 ± 2.9 **Δ**	**6.9 ± 0.7**	18.4 ± 5.2	16.7 ± 2.0	15.6 ± 2.3
Knee flexion angle (stance_CC) [°]	**9.8 ± 6.1**	**11.5 ± 6.6**	22.7 ± 1.6	20.7 ± 4.9	21.3 ± 4.2
Hip extension angle (max_CC) [°]	**17.4 ± 3.1 Δ**	10.5 ± 3.4	11.3 ± 5.3	9.3 ± 3.6	10.9 ± 3.4
Kinetics					
GRF_vert_(1st peak) [%BW]	**102 ± 10**	**106 ± 18**	128 ± 5	127 ± 12	124 ± 10
GRF_ap (min) [%BW]	13 ± 6	16 ± 9	19 ± 3	21 ± 3	18 ± 4
GRF ap (max) [%BW]	**9 ± 4**	**14 ± 4**	18 ± 6	17 ± 3	22 ± 4
Ankle dorsiflexion moment (max) [Nm/kg]	1.49 ± 0.13	1.36 ± 0.10	1.56 ± 0.31	1.46 ± 0.08	1.38 ± 0.09
Knee flexion moment (max) [Nm/kg]	ne	ne	0.51 ± 0.28	0.67 ± 0.42	0.52 ± 0.19
Knee extension moment (max) [Nm/kg]	0.42 ± 0.12 **Δ**	**0.71 ± 0.13**	0.46 ± 0.12	0.38 ± 0.08	0.36 ± 0.07
Knee adduction moment (1st peak) [Nm/kg]	0.28 ± 0.19	0.29 ± 0.19	**0.59 ± 0.14**	**0.59 ± 0.18**	0.43 ± 0.10

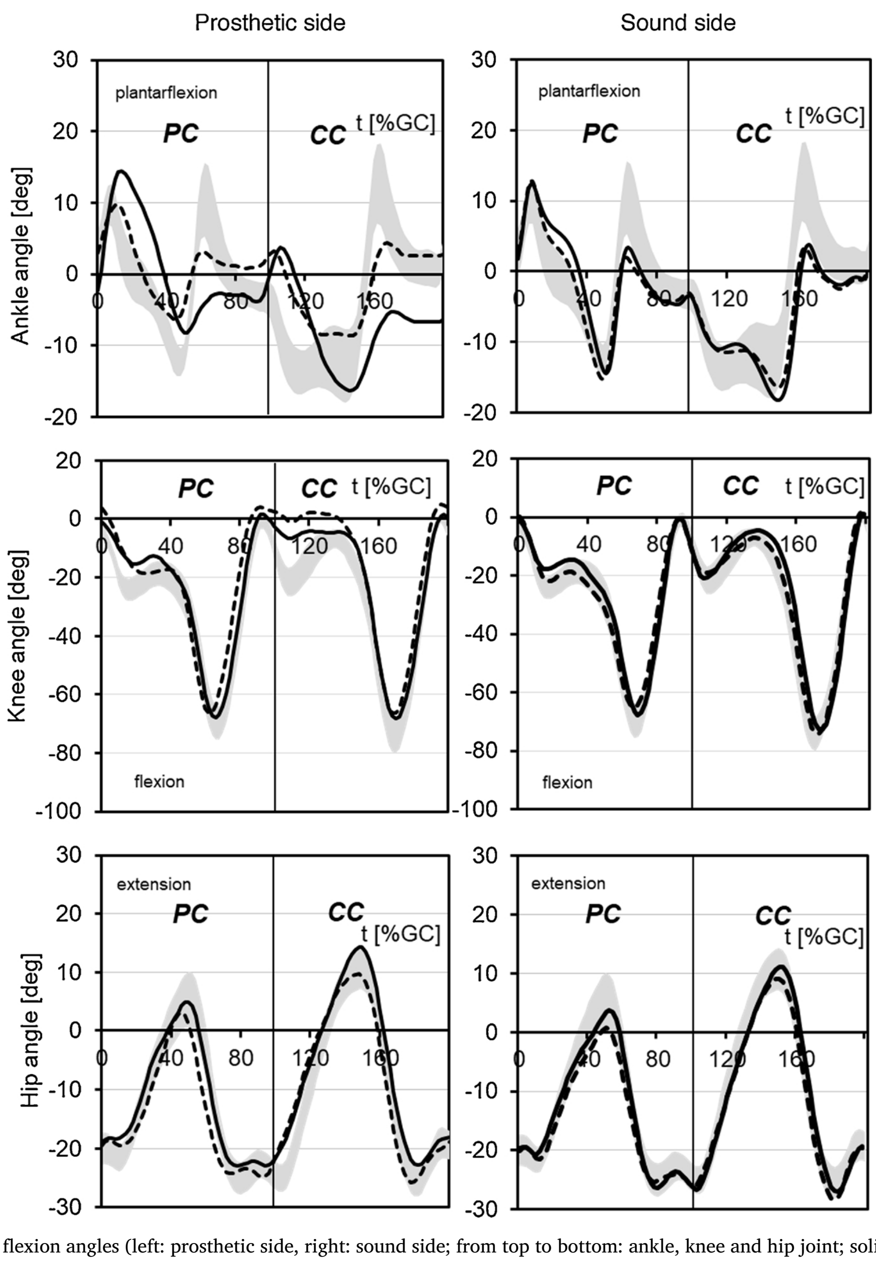

Fig. 2. Mean lower limb joint flexion angles (left: prosthetic side, right: sound side; from top to bottom: ankle, knee and hip joint; solid line: MPF, dashed line: DLF; shaded curve: NA range (mean ± SD); PC: preparing gait cycle, CC: changing gait cycle).

The CC stance phase knee flexion was significantly reduced for both prosthetic conditions compared to NA (MPF: 9.8°, DLF: 11.5°, NA: 21.3°, $p < 0.05$; Fig. 2). A significantly increased hip extension was seen for MPF compared to DLF and NA (MPF: 17.4°, DLF: 10.5°, NA: 10.9°, $p < 0.05$). The sound side knee and hip angles were comparable to NA for both prosthetic conditions (Fig. 2).

3.3. Kinetics

The vertical GRF component for NA during CC shows an increased first peak (124%BW) compared to the second one (100%BW), whereas the horizontal GRF component is generally comparable to level walking. The TT amputee GRFs during CC were affected for the prosthetic side compared to NA without any significant differences between both prosthetic conditions. The first peak of the vertical force component was significantly lower for TT amputees than for NA (MPF 102 ± 10%BW, DLF 106 ± 18%BW, NA 124 ± 10% BW, $p < 0.05$). The horizontal decelerating forces in the first half of stance were only slightly reduced whereas the horizontal accelerating forces in the second half of stance were significantly reduced (MPF: 9 ± 4%BW, DLF: 14 ± 4%BW, NA 22 ± 4%BW; $p < 0.01$ and $p < 0.05$). The sound side GRFs of the TT group showed no significant changes in comparison to NA (Table 2).

The NA external ankle moments during CC were similar to the ascending ramp pattern [25]. Compared to level walking the dorsiflexion moment shows an abrupt rise to an intermediate plateau between 20 and 30% of the CC. The sagittal and frontal external knee moments are similar to level walking [26,27] (Fig. 3, Table 2).

The prosthetic side ankle moments of the TT group revealed foot-specific differences. The dorsiflexion moment for DLF showed a discontinuous rise with a slightly reduced peak compared to NA. The dorsiflexion moment with MPF showed a gradual increase reaching a peak value similar to NA. For the sagittal knee moment, considerable differences were found between the TT group and NA as well as between the prosthetic feet. The flexion moment in the first half of stance, clearly identified for NA, was not seen for the TT group in general. The peak extension moment is significantly increased for the TT with DLF, whereas it was almost normal for TT with MPF (DLF: 0.71 ± 0.13 Nm/kg, MPF: 0.42 ± 0.12 Nm/kg, NA: 0.36 ± 0.07 Nm/kg, $p < 0.05$ and $p < 0.01$). The external knee adduction moment was generally reduced for the TT group during the CC stance phase without any foot-specific characteristics (Fig. 3, Table 2).

The peak external sagittal moment in the ankle and knee for the TT sound side were similar to NA. Increased values for the TT group compared to NA were identified for the external knee adduction moment during the whole stance phase. The first peak was significantly increased with both feet (DLF: 0.59 ± 0.18 Nm/kg, MPF: 0.59 ± 0.14 Nm/kg, NA: 0.43 ± 0.10 Nm/kg; $p < 0.05$; Fig. 3, Table 2).

4. Discussion

The ramp configuration designed for this study simulated a challenging real-life situation for TT amputees walking on uneven terrain. Thus, we were able to biomechanically evaluate a specific activity of daily living that is not easily managed by lower limb amputees in order to compare the functionality of different functional principles of prosthetic feet.

The kinematic data of NA during PC indicated that the motion pattern is not altered prior to an abruptly changing inclination in comparison to descending ramps [3,12]. Only a slightly increased dorsiflexion (about 3°) and knee flexion (about 2°) between 90 and 100% of PC can be attributed to an anticipatory motion pattern. The analysis of the CC reveales a specific adaptation of the ankle joint as the main mechanism for managing the abruptly changing inclination. The similarities of knee and hip joint kinematics with level walking are comparable with adaptations measured for NA during standing on inclined grounds. For inclinations up to 10°, adaptations occur mainly in the ankle joint, the knee and hip flexion angles remain in an almost constant, extended position as in level standing [28].

The differences of the prosthetic "ankle" angles measured during PC and CC are due to the different constructive principles of the prosthetic feet and reveal basic limitations of the ESR feet in the test situation. The flexibility of the ESR feet used in this study is a result of their carbon structure properties. This especially affects the dorsiflexion during the CC stance phase that is drastically reduced compared to the natural adaptation of NA. The concept of the MPF used in this study, enabling a sensory-controlled range of motion, is beneficial during the whole CC. The weight-bearing dorsiflexion is in good accordance to NA. In the subsequent swing phase, the MPF remains slighty dorsiflexed. This is design-related and represents a benefit compared with the ESR feet because of a reduced distance between toe and knee joint axis indicating an effective "shortening" of the shank-foot segment. It might lead to an increased toe clearance during crossing the highest vertical position of the uphill element [29]. However, that was not investigated in the present study. The adaptation is only possible because the MPF adapts quickly to specific ground conditions within the same step ("real time"). MPFs for lower limb amputees may differ with respect to this feature [28]. It can be hypothesized that for MPF that require several gait cycles in order to adapt, no benefits can be expected compared to ESR on short inclinations. Moreover, when the user is confronted with the abruptly changing inclination (CC), such MPF might be fully adapted to the downhill motion and could therefore be in an even less favorable state.

The different roll-over behavior of the MPF and ESR feet investigated in the present study affect prosthetic knee joint loading in the sagittal plane. The limited dorsiflexion of ESR feet causes an excessive external knee extension moment during the first half of the CC stance phase, similar to results measured for descending stairs of TT amputees [7,11]. Electromyographic studies, conducted during prosthetic alignment optimizations, reported that TT amputees did not reveal increased compensatory hamstring activity when large external extension moments are acting on the joint [18]. Therefore, a considerable overloading of the posterior joint capsule and ligaments around the knee joint may be assumed [18]. In this context, the quick ankle angle adaptation of the MPF enabling dorsiflexion during weight-bearing conditions appears to be responsible for the substantial reduction of the knee extension moment. A specific conclusion is that this biomechanical effect also potentially reduces loading of the posterior knee capsule and ligament structures. Despite these clear differences resulting from the different constructive principles the prosthetic side knee joint remains in an abnormally extended position compared to NA during the CC stance phase for both prosthetic situations. This specific result is similar to effects measured during descending stairs [7,9] and descending ramps [12,13] showing that TT amputees avoid the normal knee flexion during weight bearing usually measured in these motions. A possible explanation for this phenomenon could be that an extended knee represents a compensatory mechanism which is necessary due to the lack of neuromuscular ankle joint control and weakened knee extensors of long-term TT amputees [30,31].

Previous studies reported that different constructive prosthetic foot principles may affect ground reaction forces and joint moments at the sound leg of unilateral TT amputees during descending ramps and stairs [8,9]. For the specific test situation of this study, no significant differences were found for sound side loading between prosthetic situations. However, the limited dorsiflexion of ESR feet during the PC stance phase appears to be related to slightly increased deceleration forces and knee flexion moments in the first half of CC stance phase (Table 2, Fig. 3). The significantly increased values of the sound side knee adduction moment compared to NA, measured for both foot types, are noteworthy. It is generally accepted that this parameter is correlated with knee contact forces acting on the medial compartment [32]. Therefore, the parameter is seen as a biomechanical risk factor for the

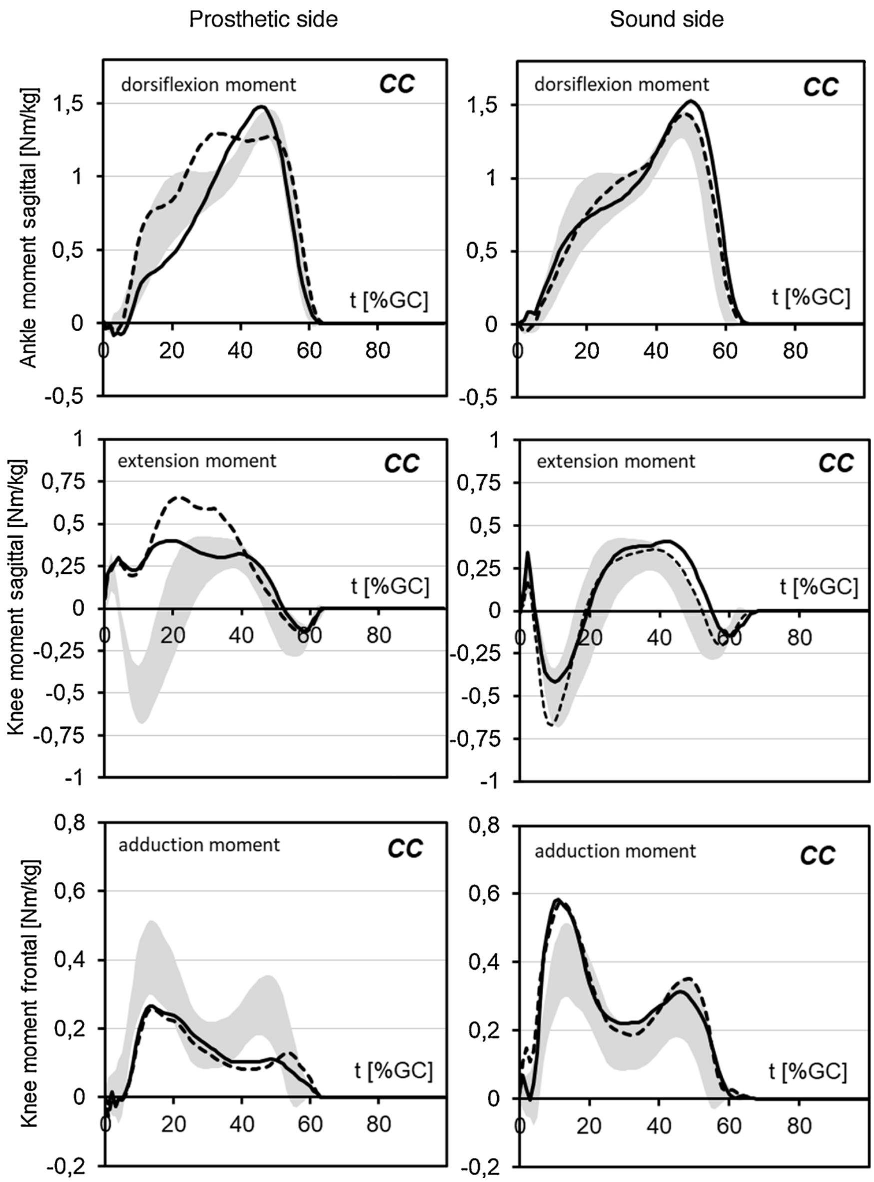

Fig. 3. Selected mean kinetic parameters (left: prosthetic side, right: sound side; from top to bottom: external sagittal ankle moment, external sagittal knee moment and external frontal knee moment; solid line: MPF, dahed line: DLF; shaded curve: NA range (mean ± SD), CC: changing gait cycle).

development of knee osteoarthritis [32,33]. The TT amputee's knee adduction moment is usually determined in level walking. Previous investigations on level walking revealed no significantly increased values for neither the prosthetic nor the sound side knee joint compared to NA [1,21,23]. The results of the present study indicate that for walking on abruptly changing inclination the knee adduction moment on the sound side is markedly increased. Therefore, an increased loading of the medial knee compartment is likely. This also holds true for the MPF used in the present study which shows a quick adaptation of the sagittal ankle joint ROM but cannot influence the sound side knee adduction moment.

As a potential limitation, it should be noted that the sample size N = 4 is small. However, all subjects showed the same trend in the main parameters between MPF and DLF, so that the results of all patients point in the same direction. In such small samples, the use of Student's *t*-test may be questioned. However, according to deWinter [24], the effect sizes (Cohen's d) of 4.4 for ankle angle adaptation with a statistical power of 0.8 and 2.3 for the sagittal knee moment with a lower statistical power of 0.6 justify the use of that test. For substantiating the effects, the investigation of a larger patient group with a similar study design should be targeted in prospective studies.

5. Conclusion

The results of the present study show that the MPF can facilitate partly normalized walking biomechanics in TT amputees on terrain with changing inclination conditions compared to regular ESR feet. The real-time adaptable ankle joint motion of the MPF seems to be the crucial functionality for a more natural motion pattern and a reduction in sagittal knee joint loading on the prosthetic side. However, the typical increase in knee loading in the medial compartment of the sound side of TT amputees does not appear to be affected by the MPF.

Conflict of interest

Thomas Schmalz, Bjoern Altenburg, Michael Ernst, Malte Bellmann and Dieter Rosenbaum work for the "Clinical Research & Services" department of OttoBock Health Care GmbH. The authors alone are responsible for the content and writing of the paper.

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