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Stabilizing moments of force on a prosthetic knee during stance in the first steps after gait initiation

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A B S T R A C T
In this study, the occurrences of stabilizing and destabilizing external moments of force on a prosthetic knee during stance, in the first steps after gait initiation, in inexperienced users were investigated. Primary aim was to identify the differences in the external moments during gait initiation with the sound leg leading and the prosthetic leg leading. A prosthetic leg simulator device, with a flexible knee, was used to test able-bodied subject, with no walking aid experience. Inverse dynamics calculations were performed to calculate the external moments. The subjects learned to control the prosthetic leg within 100 steps, without walking aids, evoking similar patterns of external moments of force during the steps after the gait initiation, either with their sound leg loading or prosthetic leg leading. Critical phases in which a sudden flexion of the knee can occur were found just after heelstrike and just before toe-off, in which the external moment of force was close to the internal moment produced by a knee extension aiding spring in the opposite direction.

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1. Introduction

Amputee subjects who learn to walk with a prosthetic leg with an artificial knee, perform poorly during the initial gait training, hence the use of parallel bars, supported by therapists and other safety measures. In the weeks that follow, subjects develop adjustment strategies to improve obstacle crossing, gait initiation and gait termination [17]. It is suggested that the most significant gait adaptations occurred following receipt of a functional prosthesis. Research does not show a clear benefit in gait patterns at discharge following use of generic prosthetic devices (early walking aids with limited functionality) during the initial rehabilitation process [1]. Therefore, it is of value to study the gait initiation strategies, with fully functional prosthetic legs, in inexperienced prosthetic leg users in the early phase of motor learning. In the current study we investigated the stabilizing external extension moments on the prosthetic knee in the first steps after gait initiation with the sound leg or prosthetic leg leading in inexperienced prosthetic leg users.

The preference of experienced transfemoral (TF) amputee subjects to initiate gait with their prosthetic leg leading, indicates that they have implicit knowledge of the active control possibilities in their sound ankle, which they use to gain forward velocity [15,18]. Because of these active control possibilities it seems advisable to initiate gait with the prosthetic leg leading. When considering the first step after gait initiation, in which the leading prosthetic leg becomes the stance leg, the leading leg has to be placed in such a manner that sufficient knee stability is reached when loading the leg. The ground reaction force (GRF) under the prosthetic foot results from the angle at which the leg is placed, the internal moment of force around the hip joint and gravitational forces on the body segments. When this GRF generates an external moment of force around the knee joint that remains within the limits of the knees stability, the knee will not buckle and stable stance will be achieved.

In contrast to experienced prosthetic leg users, inexperienced patients are taught to initiate gait with their sound leg leading in the initial stage of therapy in our rehabilitation facility. This strategy ensures a stabilizing external extension moment on the prosthetic stance leg during gait initiation and minimizes the risk of falling during the first step, as the sound leg, with more control possibilities, becomes the stance leg. Consequently, in the second step the prosthetic leg becomes the stance leg again, with the same need to stabilize the knee.

Based on differences in step length and velocity of the prosthetic leg in the steps after the gait initiation with either the prosthetic leg or the sound leg leading, we expect different ground reaction forces under the prosthetic foot. These forces generate external flexion or extension knee moments which may stabilize or destabilize the prosthetic knee. During the swing phase knee flexion is necessary for ground clearance. At the end of the swing phase, an internal hip extension moment can be applied to extend the prosthetic knee, using the inertial properties of the lower part of the prosthetic...
2.2. Apparatus

We used a PRIMAS 3D motion capture camera system, a Bertec force plate and a kneewalker prosthetic leg for AB subjects. The 3D motion capture camera system consists of six infrared cameras recording at 100 Hz. Seven retroreflective markers, positioned on the socket (3: upper leg, knee joint, lower leg), the shaft (2: proximal, distal) and the foot (2: heel, toe) (Fig. 1) were used to record the motion of the prosthetic leg. The GRF and center of pressure (COP) under the prosthetic foot were recorded at 100 Hz with the Bertec force plate. The marker data and force plate data were rotated around the vertical (y) axis and projected in the sagittal plane through the artificial knee joint, enabling us to calculate the external flexion and extension moment on the knee. The motion and force data were filtered with a second order 5 Hz low pass zero time-lag Butterworth filter and processed in MATLAB with custom made software for the 2D inverse dynamics calculations. The outcome parameters were analyzed with SPSS. The kneewalker prosthetic leg is a prosthesis for AB subjects [8] which consists of an Otto Bock Habermann modular four bar linkage knee joint (3R36), an Otto Bock dynamic foot with toes (1D10, size 26) and a shoe (size 43/9, toe–heel length 0.30 m) (Fig. 1). The artificial knee is equipped with an extension aiding spring. This spring has two main functions. Firstly, the spring supports the forward motion of the foot and shaft at the end of the swing phase, reducing the swing time. Secondly, the spring enables a prosthetic leg user to raise the prosthetic leg forward against gravity without flexion of the knee, assumed that the motion is not performed with high accelerations. This second feature provides a prosthetic leg user control over the passive knee when positioning the prosthetic foot for the stance phase at low speed. By making use of the extension spring the prosthetic knee remains locked in full extension. The spring generates an internal moment between 45 and 0° flexion. The magnitude of the moment is inversely related to the amount of flexion, decreasing down to 0 Nm at 45° flexion. Hyperextension of the prosthetic knee is prevented by a mechanical stop, i.e. a very high stiffness. The spring produces a maximal internal extension moment of 12.4 Nm in full extension. The length of the shaft can be adjusted to match the contralateral leg length. The mass of the knee–shaft–socket system is 2.08 kg. The prosthetic ankle-foot system of the kneewalker prosthetic leg is relatively stiff. The leg socket of the kneewalker is constructed in such a way that the prosthetic leg is connected to the upper and lower leg, which is fixed in 90° flexion at the knee joint. Because of this construction, the AB subjects are able to put weight on the kneewalker via their knee and the socket/leg connection. In this way the prosthesis can be used in a comparable manner as a prosthesis for knee-exarticulation amputees. All subjects used the same shoe under the prosthetic foot. The heel–toe length was 0.3 m.

2.3. Procedure

Before the measurements the subjects were allowed to walk with the kneewalker prosthetic leg without walking aids. The subjects were not allowed to make more than 100 steps. No other instructions were given. All subjects were informed and they experienced that the kneewalker was equipped with a flexible knee that not only can flex during the swing phase, but also flexes in loaded condition when used inadequately.

Before gait initiation measurements started, the subjects had to balance on the kneewalker with little support from the contralateral foot for at least 3 s to determine a midstance position per subject, based on the angles of the joints and segments. After this measurement all subjects were tested in two conditions. In the first condition subjects initiated gait with their sound leg leading, placing their prosthetic leg on the force plate in the second step. In the second condition subjects initiated gait with their prosthetic

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**Fig. 1.** Kneewalker prosthetic leg for able-bodied subjects. The black dots indicate the optical marker positions.
2.4. Outcome parameters

The forces and markers on the prosthetic leg in both leading leg conditions were used to determine (1) the ground reaction force in horizontal \( F_x \) and vertical \( F_y \) direction, (2) the position of the CoP over time under the prosthetic foot \( \text{CoP}_x \), (3) the segment angle of the shank, (4) the joint angle of the knee, (5) the velocity of the heel just before heel strike, (6) the external moment \( M \) on the center of rotation (CoR) of the knee (Fig. 2), and (7) the horizontal \( u_x \) and vertical \( u_y \) linear acceleration and angular \( \omega \) acceleration of the knee–shank–foot system. For the analysis of the data we made a distinction between subjects that used knee flexion during the swing phase with the prosthetic leg and subjects that kept their prosthetic knee in extension during the swing phase.

The external moment on the center of rotation of the knee \( M_{\text{CoR}} \) was calculated based on static and dynamic components in 2D inverse dynamics. The position of the CoR of the modelled four bar linkage knee joint was calculated by the intersection of the lines of the two grounded links (Fig. 3). The static components are formed by the moment arm of the knee–shank–foot system \( r_{\text{mCoR}} \) to the center of rotation, the static moment of the ground reaction force \( F_{\text{GRF}} \) and the moment arm \( r_{\text{CoPm}} \) on the mass of the system \( m \), and the gravitational force \( F_{\text{gm}} \) on the system. The dynamic components are the moments needed to produce the angular \( \alpha \) and linear \( \omega \) accelerations of the system with its inertial properties \( I \). For the inverse dynamics calculations we modeled the knee and shank, and the foot as two slender rods. The inertia of the knee–shank–foot system of the kneewalker was calculated based on the properties of the components and was on average 0.05 kg m² in relation to the CoM of the system depending on the length of the shank.

2.5. Statistical analysis

All outcome parameters were distributed normally. Paired \( T \) tests for differences between conditions were used for the analysis of outcome parameters. To determine differences between two subject groups that used or did not use knee flexion during swing subjects we used independent samples \( T \) tests between groups. The level of significance was set to \( p < 0.05 \).

3. Results

One out of the eleven subjects needed one extra trial in the sound leg leading condition to produced the five successful trials per leading leg condition, because of a sudden flexion of the knee.

Similar external moment production patterns were found between subjects. The patterns within the subjects were very consistent in the two conditions (Fig. 4). The data showed no significant intra-individual differences in the temporal values of the gait cycle stance phase between the two conditions. On average the total duration of the foot contact was 1.1 s (±0.2). Midstance was reached at 56% (±4) of the stance phase after heel strike. Critical knee flexion moment phases, which were close to the 12.4 Nm produced by the extension aiding spring, were found at 9% (±4) of the stance phase in the sound leg leading condition and at 13% (±8) of the
stance phase in the prosthetic leg leading condition, and at 93% (±6) of the stance phase for both leading leg conditions.

Table 1 shows the average outcome parameters. Significant differences between the two conditions were found in the external knee moments at heel strike and at midstance. At heel strike a larger flexion moment on the prosthetic knee was found in the sound leg leading condition. In midstance a larger extension moment on the knee was found in the sound leg leading condition. Significant differences were also found in the horizontal velocity of the prosthetic foot just before heel strike and the shank angle at heel strike. The velocity of the foot was higher in the sound leg leading condition, while the angle of the shank with the vertical was smaller, compared to the prosthetic leg leading condition. No significant differences were found in the vertical velocity and the knee moment before and after midstance.

In the swing phase, three groups could be distinguished. Three subjects used knee flexion strategy when swinging their foot forward (Fig. 5). Four subjects used a knee extension and circumduction strategy when moving their foot forward for the next step.

The external knee moments used in these two groups differed. The remaining four subjects in the last group used both strategies, with a preference for the knee flexion strategy. One of the four subjects used the knee flexion strategy three times out of the five trials. The other three subjects used the knee flexion strategy four times out of five times. The data from the last group were excluded from the analysis.

Table 2 shows the averaged linear and angular acceleration outcome parameters of the knee–shank–foot system of the knee-walker prosthetic leg during stance phase. A significant difference was found in the linear horizontal acceleration. In the sound leg leading condition the minimal horizontal acceleration was higher compared to the prosthetic leg leading condition.

Fig. 5 shows the graphs of a subject during the stance phase after gait initiation. A flexion moment (−13 Nm) on the knee is seen after midstance at 1.00 s, resulting in flexion of the knee. This moment results from the horizontal and vertical forces ($F_x$ and $F_y$) and their point of application (CoP) under the prosthetic foot. Fig. 6 shows that the external moment on the center of rotation of the
Table 1
Outcome parameters of prosthetic leg motion and external moments of force on the prosthetic leg during gait initiation in sound leg leading (second step) and prosthetic leg leading (first step) conditions (n = 11). Negative moments (Nm) indicate that a flexion moment on the knee is applied. Negative angles indicate that the shank is rotated with the distal end in front of the proximal end. An angle of 0 radians indicates that the shank is in vertical position. Negative velocities indicate that the motions are downwards (vertical z-axis) and backwards (horizontal y-axis).

<table>
<thead>
<tr>
<th>Outcome parameters of prosthetic leg side</th>
<th>Sound leg leading condition (2nd step)</th>
<th>Prosthetic leg leading condition (1st step)</th>
</tr>
</thead>
<tbody>
<tr>
<td>External knee moment at heel strike</td>
<td>−2.5 Nm (±2.8)</td>
<td>0.2 Nm (±2.0)</td>
</tr>
<tr>
<td>Horizontal velocity of the heel just before heel strike</td>
<td>0.13 m/s (±0.11)</td>
<td>0.08 m/s (±0.07)</td>
</tr>
<tr>
<td>Vertical velocity of the heel just before heel strike</td>
<td>−0.02 m/s (±0.05)</td>
<td>−0.04 m/s (±0.02)</td>
</tr>
<tr>
<td>Shank angle at heel strike</td>
<td>−14.90 (±2.86)</td>
<td>−16.62 (±4.58)</td>
</tr>
<tr>
<td>Minimal external moment on the knee in the first phase</td>
<td>−10.4 Nm (±3.6)</td>
<td>−8.9 Nm (±2.8)</td>
</tr>
<tr>
<td>External knee moment at midstance</td>
<td>46.1 Nm (±1.00)</td>
<td>32.8 Nm (±9.7)</td>
</tr>
<tr>
<td>Minimal external moment on the knee in the second phase</td>
<td>KF: −14.9 Nm (±0.49)</td>
<td>KF: −13.7 Nm (±0.97)</td>
</tr>
<tr>
<td></td>
<td>KE: −11.9 Nm (±1.9)</td>
<td>KE: −10.6 Nm (±2.7)</td>
</tr>
</tbody>
</table>

* Knee Flexion group (KF; 3 subjects) and Knee Extension group (KE; 4 subjects).

Four subjects were excluded for this second phase analysis because they made use of both knee strategies in the two conditions. No statistical analysis was performed in this second phase analysis because of the small group sizes.

*1 Significant difference (p < 0.01) between two leading leg conditions.

*2 Significant difference (p < 0.05) between two leading leg conditions.

The knee (M<sub>CoR</sub>) is mainly influenced by the ground reaction force in the static moments. When the knee starts flexing at the end of the stance phase (1.00 s), the influence of the dynamic moments can be seen when the ground reaction force is reduced substantially.

4. Discussion

In this study, we investigated whether an inexperienced prosthetic leg user with only limited training (100 steps) is able to initiate gait and walk two steps without walking aids on a prosthetic leg with a flexible knee, without the occurrence of a sudden flexion of that knee during stance, either with their sound leg leading or their prosthetic leg leading. The flexible knee is equipped with an extension aiding spring, which can deliver an internal extension moment. We wondered if the evoked GRF delivers an external flexion moment that is close to the internal extension moment.

We used inexperienced young AB subjects instead of recently amputated TF subjects, to ensure that our subjects had no experience with the use of a prosthetic leg and because of the absence of experience and comorbidity in this group, which otherwise could influence the outcome of the experiments. Indeed, physical differences and age differences between the AB subjects and TF amputee subjects can be expected, but a study by Lemaire et al. [8] has shown that the AB subjects have to learn to deal with mechanical properties of a body extension in a similar way as inexperienced prosthetic

users. Therefore we believed it was justified to use AB subjects instead of TF amputee subjects to get first insights in gait initiation strategies in inexperienced prosthetic leg users.

In order to be able to walk with the kneewalker prosthetic leg, with conventional mechanical components, the subjects have to learn to control external moments on the passive prosthetic leg. These moments are influenced by the angle in which the leg was placed on the ground, the applied internal moments by the hip muscles and gravity, and the mechanical properties of the artificial knee.

The results of our experiment showed no clinical differences between the two leading leg conditions. All the subjects were able

Table 2
Mean and standard deviation of the minimum and maximum values of the angular (ω) and the horizontal (τ<sub>r</sub>) and vertical (τ<sub>z</sub>) linear accelerations of the CoM of the knee–shank–foot system of the kneewalker prosthetic leg (m = 2.1 kg; I = 0.05 kg m<sup>2</sup>) during the stance phase.

<table>
<thead>
<tr>
<th>Accelerations of the prosthetic knee–shank–foot system</th>
<th>Sound leg leading condition (2nd step)</th>
<th>Prosthetic leg leading condition (1st step)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Min</td>
<td>Max</td>
</tr>
<tr>
<td></td>
<td>Max</td>
<td>Min</td>
</tr>
<tr>
<td>ω&lt;sub&gt;r&lt;/sub&gt; (m/s&lt;sup&gt;2&lt;/sup&gt;)</td>
<td>−1.9 (±1.4)</td>
<td>5.5 (±0.74)</td>
</tr>
<tr>
<td>ω&lt;sub&gt;z&lt;/sub&gt; (m/s&lt;sup&gt;2&lt;/sup&gt;)</td>
<td>−0.67 (±0.43)</td>
<td>1.37 (±0.50)</td>
</tr>
<tr>
<td>τ&lt;sub&gt;r&lt;/sub&gt; (rad/s&lt;sup&gt;2&lt;/sup&gt;)</td>
<td>−12.2 (±3.6)</td>
<td>6.0 (±1.7)</td>
</tr>
</tbody>
</table>

* Significant difference (p < 0.01) between two leading leg conditions.
to prevent a sudden flexion of the flexible passive prosthetic knee, with the exception of one subject who needed one extra trial in the sound leg leading condition to have five successful trials. The duration of the stance phase and the patterns of the external moments on the CoR of the knee that were evoked were similar in both conditions. The only clear differences between the two conditions could be found just before the beginning of the stance phase. The horizontal velocity and the angle of the shank differed. These differences had to be dealt with by the subjects. Remarkably, these differences had no consequences for the patterns of the external moments during the stance phase after the heel strike. The results from our study show that although different velocities of the prosthetic leg were found, the patterns of the external moments applied on the prosthetic knee in the steps after gait initiation did not differ between the two leading leg conditions.

We found two moments during stance phase in which a sudden flexion of the knee could occur. The first flexion moment occurred at the beginning of the stance phase, just after heel strike in both conditions. This flexion moment was remarkably close to the critical moment produced by the extension aiding spring. It is remarkable since only a shift of 1 mm of the CoP position backwards would create an unstable situation. We identified this as a critical phase of the stance phase after gait initiation. This phase was accomplished by a diminished loading rate, compared to normal loading patterns, which has also been described as an awareness effect to prior slip experiences [6], showing a cautious loading pattern. In this phase the CoP is positioned relatively more backward under the heel of the prosthetic foot. When the CoP is moved forward, less flexion moment is produced. Notice the relation in Fig. 5 between the shank angle and CoPx position during gait initiation. This relation is the result of the stiff prosthetic ankle properties. The CoP moves forward as a result of an increased external moment on the ankle caused by the rotation of the shaft relative to the foot. The range of this CoP motion increases with increasing stiffness of the prosthetic ankle. It seems that using a prosthetic foot with a relatively flexible heel or rocker will limit the hazard of a sudden flexion of the knee, as the CoP is moved forward more quickly. The outcome from this study applies on the design of this specific prosthetic leg with a stiff ankle–foot. Therefore some considerations have to be taken into account when applying these finding. The second flexion moment in which the external moment on the prosthetic knee was close to the moment of the extension aiding spring was found just before toe off. In this part of the stance phase some subjects evoked a moment that was used to flex the knee for the upcoming swing phase. The other subjects evoked less external moment on the knee, resulting in maintaining the prosthetic knee in extension during the upcoming swing phase.

The external moments on the knee are mainly the result of the static moments produced by the GRF. Fig. 6 shows that the total moment on the center of rotation of the knee \( M_{\text{CoR}} \) consists of static \( M_{\text{CoRst}} \) and dynamic \( M_{\text{CoRdyn}} \) moments during the stance phase after gait initiation with the prosthetic leg leading (\( M_{\text{CoR}} = M_{\text{CoRst}} + M_{\text{CoRdyn}} \)). The subject uses a knee flexion strategy during the swing phase. This flexion strategy starts at the end of the stance phase, at 1.00 s after the heel strike at 0 s.

![Fig. 6. An example of the external moment on the center of rotation of the knee \( M_{\text{CoR}} \) consisting of static \( M_{\text{CoRst}} \) and dynamic \( M_{\text{CoRdyn}} \) moments during the stance phase after gait initiation with the prosthetic leg leading (\( M_{\text{CoR}} = M_{\text{CoRst}} + M_{\text{CoRdyn}} \)). The subject uses a knee flexion strategy during the swing phase. This flexion strategy starts at the end of the stance phase, at 1.00 s after the heel strike at 0 s.](image-url)

5. Conclusions

The experiment showed that the AB subjects learned to control external moments on the knee leading or the prosthetic leg leading. Therefore we think that preference of leading leg should not be focus of attention during therapy.
efficient manner within 100 steps of gait. In both the prosthetic leg and sound leg leading condition the subjects evoked patterns of external moments of force that were similar. The external moments that were evoked were mainly the result of GRF components evoked during the stance phase. The inertial properties of the prosthetic leg during stance had only limited influence on the external moment on the knee. The critical phases in which a sudden flexion of the knee can occur were found just after heel strike and just before toe off. Based on the control strategy that the subjects used during the stance phase on the prosthetic leg after gait initiation, it makes no difference which leg is used as the leading leg during gait initiation to avoid sudden knee flexion.

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Conflict of interest statement
No conflict of interest.

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