Research article

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Designs and performance of microprocessor-controlled knee joints

Abstract: In this comparative study, three transfemoral amputee subjects were fitted with four different microprocessor-controlled exoprosthetic knee joints (MPK): C-Leg, Orion, Plié2.0, and Rel-K. In a motion analysis laboratory, objective gait measures were acquired during level walking at different velocities. Subsequent technical analyses, which involved X-ray computed tomography, identified the functional mechanisms of each device and enabled corroboration of the performance in the gait laboratory by the engineering design of the MPK. Gait measures showed that the mean increase of the maximum knee flexion angle at different walking velocities was closest in value to the unaffected contralateral knee (6.2°/m/s) with C-Leg (3.5°/m/s; Rel-K 17.0°/m/s, Orion 18.3°/m/s, and Plié2.0 28.1°/m/s). Technical analyses corroborated that only with Plié2.0 the flexion resistances were not regulated by microprocessor control at different walking velocities. The muscular effort for the initiation of the swing phase, measured by the minimum hip moment, was found to be lowest with C-Leg (-82.1±14.1 Nm; Rel-K -83.59±17.8 Nm, Orion -88.0±16.3 Nm, and Plié2.0 -91.6±16.5 Nm). Reaching the extension stop at the end of swing phase was reliably executed with both Plié2.0 and C-Leg. Abrupt terminal stance phase extension observed with Plié2.0 and Rel-K could be attributed to the absence of microprocessor control of extension resistance.

Keywords: amputee; dampening characteristic; exoprosthetic; gait analysis; transfemoral; velocity adaptation.

Introduction

Microprocessor-controlled exoprosthetic knee joints (MPK) are state-of-the-art medical devices that regulate stance and swing phase resistance by means of electronic sensors and complex control algorithms. Numerous MPKs are available on the market, all with unique combinations of mechanical and microprocessor-controlled mechanisms. Their complex interaction of various functional principles used for generating joint resistances are implemented by control algorithms and sensor technology that directly affect movement and safety outcomes for lower-limb amputees [2–4]. For the patient, the resultant functional quality [2, 5, 8, 12, 17, 18, 24, 30–32], and the reduction in falls some MPKs provide [3, 7, 11, 16], is of central importance. The challenge for prosthetic practitioners and patients alike is how to select the appropriate MPK to meet the functional needs of the lower-limb amputee. A comparative biomechanical analysis between C-Leg, Rheo [g], Hybrid Knee [h], and Adaptive 2 [b] revealed that C-Leg provided better swing phase flexion resistance, terminal extension damping, and stumble resistance as compared with the other MPKs [3]. Newer MPKs, including Orion, Plié2.0, and Rel-K, have not been subjected to comparative biomechanical analysis. The purpose of this study was to evaluate the quality of stance and swing phase control of Orion, Plié2.0, and Rel-K as compared to C-Leg, and determine, based on their engineering designs, which MPK device provided the optimal control profile. C-Leg is the most highly researched MPK and thus selected as the comparator device to the new MPKs in this study. Three subjects were fitted with the MPKs to investigate the relationship between design and performance during level walking. Data were recorded simultaneously by stationary and mobile motion analysis measuring systems. Additionally, a technical analysis of the MPKs was performed to identify possible explanations for deviations in their functional quality.
Methods and materials: gait lab

Materials

Four MPKs were evaluated: C-Leg (Otto Bock HealthCare [a]), Orion (Chas A Blatchford and Sons [b]), Plié2.0 (Freedom Innovations [c]), and Rel-K (Rizzoli Ortopedia [d]). For all test conditions, the same prosthetic foot, the IC60 [a], and each subject’s existing socket (ischial containment design) were used.

Subjects

Three men with unilateral transfemoral amputation participated in the evaluations. Subject demographics are described in Table 1. All subjects had a mobility grade of Medicare Functional Classification Level (MFCL) K3 and were routinely using the newly introduced Genium MPK [a]. Although the subjects entered the study on the Genium MPK, C-Leg is the most highly researched MPK and thus selected as the comparator device in this study. All subjects were experienced in performing motion analysis tests and were fitted with the knee joints of this study in previous tests.

Motion analysis technology

The measurements were performed in a motion analysis laboratory with a 12-m instrumented walkway. The kinematic parameters were recorded using an optoelectronic camera system (VICON 460 [e]). The kinetic parameters were determined by means of two force plates (Kistler 9287A [f]) embedded in the floor, one after another, in the middle of the walkway.

In parallel to these measurements, the forces and moments acting on the prostheses, as well as the femur and tibia orientation on the prosthetic side, were recorded using the mobile Oktapod measuring system [a]. Previous evaluations of data recorded by the Oktapod measuring system showed a high validity compared to the stationary gait analysis [23]. Thus, the Oktapod measuring system offered an additional method of permanent data recording of all completed steps and of the kinetic parameters during the swing phase.

Test preparation

The functional differences between the four prosthetic knee joints were investigated based on an identical prosthetic alignment. The bench alignment performed with LASAR Assembly [a] met all four manufacturers’ specifications. The optimization of the static alignment (Table 2) by means of LASAR Posture [a] was conducted according to recommendations by Otto Bock HealthCare GmbH [25]. The knee joint settings were individually adapted to the subjects by a certified prosthetist. Changes on the knee joint parameters were conducted according to the manufacturer’s recommendations to achieve a uniform and secure gait over all walking velocities. Thereby, the prosthetic alignment was not changed to keep forces and moments acting on the prosthesis comparable. Following fitting and alignment, the subjects were provided a minimum time of 30 min to accommodate to the prosthesis while walking in the laboratory. Given that the subjects are well experienced, the time required to adapt to the different MPKs could be kept short. Because none of the knee joints evaluated in this study were used in everyday life by a subject, the similar timeframe of accommodation was expected to lead to comparable results. At the discretion of the individual subjects, the accommodation time was terminated and the test series started.

Test procedure

Each subject performed four test series during level walking with each MPK. The order of the knee joints was randomized. Each test series consisted of eight to ten repetitions at three self-selected velocities: medium (comfortable), slow, and fast (see Table 3 for the measured velocities). Lastly, the subjects were evaluated while walking at a self-selected medium speed with conscious stance phase flexion.

Data processing and analysis

With the stationary gait analysis system, one gait cycle (stride) per trial was recorded. The step recognition was
fully automatic, and all steps with complete measuring data were included. By using the Oktapod measuring system, the gait data were continuously recorded. The step recognition was also fully automated. To eliminate deceleration and acceleration processes from the Oktapod measurements, a software-based step filter was used.

Kolmogorov-Smirnov tests and n-way analyses of variance (ANOVA) with selected gait parameters of the single steps were conducted. Following a positive result of the ANOVA, multiple comparison procedures with Bonferroni adjustments were performed. The analyses were executed with MATLAB. The therein implemented multiple comparison procedure is based on Hochberg and Tamhane [14]. The significance level was set to 5%.

To enable the evaluation of the conclusions from the statistic tests, figures with the estimated mean values from the fitted linear model and uncertainty intervals are shown. The maximum knee flexion angle and the minimum of the sagittal hip moment were tested for a correlation with the walking velocity using a linear regression model. Mean curves were calculated pointwise over the gait cycle, which shifts maxima and minima but did not affect the curve characteristic.

Results: gait lab

Time-distance parameters

The self-selected walking velocities varied between subjects and knee joints (see Table 3). For every subject, a tendency to walk slowest with Rel-K and fastest with C-Leg was found. Looking at the walking velocity in Figure 1, the differences between two knee joints are significant if the uncertainty intervals do not overlap. For example, significant differences can be identified between Rel-K and all the other knee joints.

Gait symmetry was evaluated by the duration of the stance phase. The mean difference between the two sides was 3.93±9.74% of the gait cycle for the duration of the stance. No significant variation of the difference could be found for any subject or MPK.

### Swing phase control

In Figure 2, the maximum knee flexion angle during the swing phase for the amputated and contralateral sides is plotted over the walking velocity for all valid trials of slow, medium, and fast walking velocity. The slope of the linear regression line was 3.5°/m/s with C-Leg, 28.1°/m/s with Plié2.0, 18.3°/m/s with Orion, and 17.0°/m/s with Rel-K. On the contralateral side, the natural knee flexion angle was similar with all tested knee joints, resulting in a mean slope of 6.2°/m/s. The low coefficients of determination (R²) on the contralateral side suggest that the linear correlation between velocity and maximum knee angle is only small. C-Leg showed the most similar correlation in respect to the contralateral side; therefore, the R² was lowest (Table 4).

In the late swing phase, the knee angle characteristics showed differences between the joints. If a knee joint reaches full extension before the initial contact of the foot, the zero crossing of the angular velocity should appear before 100% gait cycle (Figure 3). The number of steps with full extension, measured by the existence of a zero crossing of the angular velocity between 80 and 100% gait cycle, related to the number of all trials is presented in Table 5.

With C-Leg and Plié2.0, the extension stop was reached for most steps (Table 5). In contrast, with Orion and Rel-K, the extension stop was seldom reached especially at mid velocity. To evaluate the effect on the gait of the amputees, the knee angular velocity at the end of swing phase was examined (see Figure 4). At all velocities,
Plié2.0 showed the lowest absolute values, without significant differences compared to C-Leg and with significant differences to Orion and Rel-K. With Rel-K, the highest values were found.

**Stance phase control during walking with stance phase flexion**

The test series with conscious stance phase flexion was used to determine the dampening of the extension stop. A smooth dampening characteristic involves a smooth curve characteristic of the knee angle (see Figure 5, left). The curvature of the knee angle curve is given by the minimum of the angular acceleration of the knee (see Figure 5, right). C-Leg (-2565±1431°/s²) and Orion (-3149±1014°/s²) showed the smoothest characteristic and no significant differences. Compared with Plié2.0 (-4347±1181°/s²) and Rel-K (-7042±2200°/s²), significant differences can be found (Figure 6). During the trials with Rel-K, it could be noticed that it does not switch to the low swing phase flexion resistance after a stance phase flexion and continues in an...
Table 5  Number of steps with full extension in relation to all trials.

<table>
<thead>
<tr>
<th></th>
<th>C-Leg</th>
<th>Plié2.0</th>
<th>Orion</th>
<th>Rel-K</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow</td>
<td>19/20</td>
<td>22/22</td>
<td>10/22</td>
<td>17/23</td>
</tr>
<tr>
<td>Mid</td>
<td>19/19</td>
<td>21/21</td>
<td>3/17</td>
<td>6/21</td>
</tr>
<tr>
<td>Fast</td>
<td>18/21</td>
<td>22/23</td>
<td>11/24</td>
<td>14/23</td>
</tr>
</tbody>
</table>

Figure 4  Knee angular velocity (°/s) at the end of swing phase, estimated mean values, and uncertainty intervals (α=0.05) over all velocities and subjects.

extended position. A repeated gait pattern after a single step with stance flexion was therefore not possible.

Initiation of the swing phase

To evaluate the effort for swing phase initiation, the minima of the external sagittal hip moment has been examined. The results are influenced by the differing walking velocities, shown by the linear correlation between the minimal hip moment and the walking velocity (R²>0.5 for every subject and knee joint). To reduce this effect, the velocities have been analyzed separately and only the results for the mid walking velocity are displayed.

The minima of the external sagittal hip moment at swing phase initiation (50.3±2.6% gait cycle) differed significantly between the knee joints (p<0.001). Noticeably, the confidence intervals are very small in this examination (Figure 5). This is achieved by the analysis of 384 valid steps recorded by the Oktapod measuring system for the mid velocity. The highest absolute values were measured with Plié2.0 (-95.2±9.5 Nm) followed by Orion (-93.8±10.2 Nm), Rel-K (-90.7±12.5 Nm), and C-Leg (-88.7±4.7 Nm).

Methods and materials: technical analysis

To identify possible technical reasons for the measured differences between the knee joints, the mechanical function was analyzed by patents, instructions for use, and other information published by the manufacturers [9, 10, 15, 26, 27]. Additional computed tomographies of the four knee joints completed the picture and provided a source of information that was independent from manufacturers’ declarations. The tomographic system uses a 450-kV X-ray source and a flat-panel X-ray detector (1024×1024 sensors) resulting in a resolution of 0.5 mm.
Results: technical analysis

Functional principle of C-Leg

With C-Leg, the generation of the joint resistances is based on the principle of a hydraulic system with two separate servo valves (1, 2) for the flexion and the extension movement (Figure 8). Each valve is controlled by the microprocessor and can continuously vary flow resistances from low to high values, including a complete closure if necessary.

When the piston immerses during flexion, the oil flows through flexion valve 1 (flow marked in red) and check valve 4a. The steel spring 3 is tensioned during flexion by the fluid displaced by the piston rod movement. The minimum of the flexion resistance is therefore given by the flexibility of the steel spring 3 and a small amount of internal resistance [2, 9]. The pressure control valve 8 avoids overloading of the hydraulics in the flexion direction.

With extension, the piston moves in the opposite direction and the oil passes extension valve 2 and check valve 4b reaching chamber A (flow marked in blue). The energy stored in the compensation reservoir B by tensioning of steel spring 3 is released now, resulting in a flow through check valve 4b to chamber A. The progressive characteristics of hydraulic 5 and a progressive spring 7 provide a smooth extension stop. Owing to a minor flow through valve 6, a slow extension movement is possible even if the battery pack is low.

The two check valves 4a and 4b switch between the flexion and extension directions so that resistances of different magnitude can be set at a time for the two movement directions. Thereby, the larger cross-section of the check valves ensures that the resistance of the respective servo valve in parallel has no influence on the resistance. In addition, the resistances may be changed under load and during motion both in the stance and swing phase. All resistances are set by means of a PC software [15, 27].

Functional principle of Plie2.0

With Plie2.0 (Figure 9), the setting of the knee joint resistances is based on a hydraulic system with a microprocessor-controlled switching valve (5) and two manually adjustable valves (1, 2). Extension assistance is provided by an air spring (C).

During flexion, the oil flows from chamber A to chamber D, passing check valve 4b (flow marked in red). The hydraulic resistance of 4b depends on the pressure difference. In Figure 9, this behavior is illustrated by the variable, passive hydraulic valve 6.

The volume of hydraulic oil displaced by the piston rod flows from chamber A to chamber B through the manually adjustable valve 1 (stance flexion) that causes a high flow resistance. In addition, the oil may flow through a bypass opened by the switching valve 5 (spool valve). Owing to its design, the microprocessor-controlled electromechanical actuator of valve 5 offers two discrete positions. If valve 5 is closed, the manually adjustable flexion resistance of
valve 1 is dominant. If valve 5 is opened, a lower flexion resistance is defined by the larger cross-section of valve 5. Now, the resistance is mainly given by valve 5, valve 6, and the compressibility of air spring C.

With every flexion movement during the swing and stance phase, energy is stored while air spring C is compressed. The pneumatic resistance and energy storage depend on the pressure in chamber C. The preset value may be changed manually, connecting a hand air pump to valve 3. Furthermore, higher flexion angles increase the air pressure, resulting in a progressive characteristic of the flexion resistance.

With the initiation of the extension movement, the energy stored in chamber C is released (Figure 9, flow marked in blue). Hence, the oil from chamber B flows back into chamber A through check valve 4c with a large cross-section. Owing to the comparatively small cross-section, the parallel electromechanical switching valve 5 is unable to influence the flow resistance during extension.

Figure 10 shows the actual design of valves 5, 4c, and 1. Although only one of the three equal inlets of valve 5 is presented, the overall cross-section dimension of valve 4c is considerably larger. Over the whole extension movement, the piston presses the oil through the manually adjustable valve 2 (swing extension) and the check valve 4a [10, 26, 33]. Therefore, the resistance for swing and stance phase extension is identical and cannot be influenced by the microprocessor.

**Functional principle of Orion**

With Orion, the generation of the knee joint resistances is based on a pneumatic and a hydraulic system. In contrast to Plié2.0, the two independent circuits are connected over one piston rod with a double piston (Figure 11). For the setting of flexion resistances by microprocessor control, a hydraulic servo valve (1) and a pneumatic needle valve (3) are provided.

During flexion, when the pistons move downward, the oil flows from chamber A to chamber B through check valve 4a and servo valve 1 (flow marked in red). Hence, the microprocessor is able to vary the flexion resistance in the hydraulic circuit continuously from low to high values in the stance and swing phases. At the same time, the air flows from chamber C to chamber D through check valve 4c and needle valve 3. A closure of valve 3 by the microprocessor-controlled actuator enables energy storage in the pneumatic circuit during flexion. Thereby, the air pressure in chamber C increases and it decreases in chamber D. At low resistance of valve 3, the air spring is disabled and flexion resistance is given by the hydraulic circuit. Consequently, the microprocessor can control the amount of energy stored in the air circuit.

When the pistons move upward, the oil flows through check valve 4b and the manually adjustable valve 2 (flow...
marked in blue). Thus, no microprocessor control of the oil flow is provided during extension. The extension behavior of the air spring depends on the previous flexion movement. If the air spring was compressed during flexion, the energy is released during extension, balancing the air pressure between chambers C and D. If required, the microprocessor can release the stored energy immediately by opening of needle valve 3 to abort the extension support. With the decompressed air spring, further flexion movement induces an air flow from chamber D to chamber C through check valve 4d. This is also the case when the air spring was not compressed during the previous flexion. At the extension stop, the impact is caught through a hydraulic (5) and a pneumatic damping mechanism.

**Functional principle of Rel-K**

For the setting of joint resistances, Rel-K (Figure 12) provides a microprocessor-controlled servo valve (1) and a manually adjustable valve (2). Extension assistance is realized by two steel springs, one around the piston rod (5) and another in the compensation reservoir (3).

When the piston moves downward during flexion, the oil flows from chamber A to chamber C through servo valve 1 and check valve 4b (flow marked in red). At the same time, steel spring 5 is tensioned directly. The insertion of the piston rod causes that part of the flow from chamber A reach chamber B, tensioning the mechanical spring 3. The range of flexion resistance reaches from complete disabling of flexion movements as servo valve 1 is closed to a minimum given by the flexibilities of the steel springs (3, 5) and a small amount of internal resistance. The pressure control valve 7 avoids overloading of the hydraulics in the flexion direction.

During extension, the loss of volume from the piston rod movement is balanced by a flow from chamber B to chamber A. Thereby, the energy stored in springs 3 and 5 is released. Effectively, the microcontroller cannot vary the extension resistance as the oil passes check valve 4c, which is in parallel to servo valve 1.

Despite that, the extension resistance changes mechanically. The movement of the piston during extension induces a flow from chamber C to chamber A. At first, most of the oil passes check valve 4a, resulting in a low extension resistance. When the piston moves forward, the inflow 6 to valve 4a is covered at a specific knee angle. After that, the flow can only pass the manually adjustable valve 2 and the check valve 4c, resulting in a higher extension resistance. The transition between the low and the high extension resistance could be very abrupt, as there are no elements supporting a progressive transition characteristic similar to the extension stop of C-Leg (Figure 8, element 5) and Orion (Figure 11, element 5).

**Discussion**

Compared with conventional mechanical knee joints, most microprocessor-controlled prosthetic knee joints intend to offer the user a reliable recognition of the transition between the stance and swing phase, and safe switching of the required joint resistances. In addition, some knees allow an adaptation of the resistances via microprocessor control to different movements and situational requirements (Table 6).

An asymmetry in stance time between the amputated and the contralateral leg has been shown in several studies [21, 28]. The shortened stance time on the amputated leg (3.93±9.74%) could be found for all MPKs in this study. Although the self-selected walking velocity is also not a predictor for functional quality, it was analyzed too. For all subjects, the mean walking speed with Rel-K was significantly slower than with C-Leg (p<0.0001), which showed the highest mean value. Because the velocity influences other gait parameters, its effect was always considered in the analysis. At higher walking speeds, as measured with C-Leg, Plie2.0, and Orion, the higher dynamics of the gait requires an adequate dampening of the load response, for example.

The outcome of stance phase flexion for amputees is somewhat controversial. In spite of that, the gait pattern appears more natural with stance phase flexion as it is
Table 6  Designs of microprocessor-controlled knee joints, overview.

<table>
<thead>
<tr>
<th></th>
<th>C-Leg</th>
<th>Plié2.0</th>
<th>Orion</th>
<th>Rel-K</th>
</tr>
</thead>
<tbody>
<tr>
<td>Microprocessor controlled elements</td>
<td>Two hydraulic servo valves</td>
<td>One hydraulic switching valve (open/close)</td>
<td>One hydraulic servo valve, one pneumatic needle valve</td>
<td>One hydraulic servo valve</td>
</tr>
<tr>
<td>Stance phase flexion resistance</td>
<td>Microprocessor-controlled hydraulic servo valve</td>
<td>Manually adjustable hydraulic valve</td>
<td>Microprocessor-controlled hydraulic servo valve and microprocessor-controlled pneumatic needle valve</td>
<td>Microprocessor-controlled hydraulic servo valve</td>
</tr>
<tr>
<td>Stance phase extension resistance</td>
<td>Microprocessor-controlled hydraulic servo valve</td>
<td>Same setting as the swing phase extension</td>
<td>Same setting as the swing phase extension</td>
<td>Same setting as the swing phase extension</td>
</tr>
<tr>
<td>Swing phase flexion resistance</td>
<td>Microprocessor-controlled hydraulic servo valve</td>
<td>Manually adjustable hydraulic valve and manually adjustable air spring</td>
<td>Microprocessor-controlled hydraulic servo valve and microprocessor-controlled pneumatic needle valve</td>
<td>Microprocessor-controlled hydraulic servo valve</td>
</tr>
<tr>
<td>Swing phase extension resistance</td>
<td>Microprocessor-controlled hydraulic servo valve</td>
<td>Manually adjustable hydraulic valve</td>
<td>Manually adjustable hydraulic valve</td>
<td>Check valve or manually adjustable hydraulic valve (depending on flexion angle)</td>
</tr>
<tr>
<td>Extension assistance</td>
<td>Steel spring</td>
<td>Air spring</td>
<td>Microprocessor-controlled air spring</td>
<td>Two steel springs</td>
</tr>
</tbody>
</table>

an inherent phenomenon of the human gait reducing the modulation of the center of gravity. A biomechanical advantage of knee flexion during the stance phase is that the time interval between heel contact and foot flat is decreased [6]. Hence, the foot is longer in full contact with the ground, enabling a more stable gait.

If a knee joint allows for stance phase knee flexion, not only the flexion but also the extension movement should be adequately damped to avoid abrupt movement transitions, in particular at the extension stop. The measured minima of the angular acceleration state that the Rel-K and the Plié2.0 joints do not meet the demand for adequate stance phase extension damping. In addition, Rel-K does not switch to the low swing phase resistance after a step with stance phase flexion. During the following swing phase, the high stance phase resistance acts and a normal walking pattern is not possible. As the technical analyses gave no evidence for this behavior, it is probably caused by the control algorithms. In contrast, C-Leg and Orion show improved extension stop damping characteristics so that walking with a stance phase flexion is supported better. The technical analysis confirmed that the extension resistance cannot be influenced by the microprocessor with Plié2.0 and Rel-K, and therefore meets only the requirements for swing phase extension. Orion also has no means for microprocessor control of the hydraulic extension resistance. However, the progressive hydraulic extension stop yields a good stance extension behavior.

At the end of the stance phase, the swing phase is initiated. Transfemoral amputees have to use the hip flexors in this phase to flex the residual hip joint and the artificial knee joint. Thus, the muscular effort required from the patient to initiate the swing phase should be as low as possible. The measured external hip moments can be used as an indicator for the activity of the residual limb musculature. The hip moment at swing phase initiation has to compensate the external knee extension moment and the flexion resistance of the knee joint to induce a knee flexion. Schmalz et al. [29] named an increased hip flexion moment at swing phase initiation as the likely cause of an increase in metabolic energy consumption.

The external hip moments measured at swing phase initiation suggest that with C-Leg, less muscular activity is required by the user to initiate flexion than with the other joints. Statements according to which the initiation of swing phase with Plié2.0 is easier in direct comparison to C-Leg cannot be confirmed [1]. As the prosthetic setups were identical so that the external knee moments are kept constant, it can be assumed that the basic flexion resistances of the joints vary. The technical analyses showed that with opened flexion valves, the minima of the flexion resistances are mainly given by the resistances of the extension assistance mechanism. Well-tuned extension assistance should therefore fulfill at least two requirements: a reliable reaching of the extension stop and a low flexion resistance.
The controlled damping of the swing phase flexion is one of the most important advantages of microprocessor-controlled knee joint systems as it enables amputees to walk with different velocities without the need to change their gait pattern. Inadequate knee flexion angles require compensation by the user with hip flexion moments. Otherwise, toe-stubbing in mid-swing or incomplete extension at heel contact can occur, especially when changing walking speeds.

Several studies examined the influence of walking speed on the maximum knee flexion angle on healthy subjects. Some studies have reported a significant correlation [19, 20, 22]; others claim that the knee angle does not vary significantly [13, 34] across walking speeds. Even in those studies that state significance, the predictability of the knee angle is found to be low, with R² values of about 0.43 [19, 20]. Many of these studies report high intersubject variability, which explains the low R² values [13, 19, 22]. The change in the predicted mean knee flexion values based on the regression equations vary greatly for the different studies: Hanlon and Anderson 1.4°/m/s, Oberg et al. 6.8°/m/s, Kirtley et al. 8.6°/m/s, and Lelas et al. 24.5°/m/s. As the literature cannot be used to predict the knee angle at different velocities, the angles of the contralateral side were used to evaluate the swing phase flexion behavior of the knee joints.

As shown in Figure 2, the knee angle of the contralateral leg changes only 6.2°/m/s on average. Compared with the other knee joints, this behavior is met best by C-Leg. With an increase of 3.5°/m/s, its maximum knee flexion angle changes only slightly over walking speed. The increase of 17.0°/m/s and 18.3°/m/s with the Orion and Rel-K joints, respectively, is similar to other microprocessor-controlled knee joints [3]. With 28.1°/m/s, Plié2.0 clearly exceeds the values of the other knee joints within this group and does not comply with the requirement of natural swing phase behavior. The technical analyses showed that from the knees under investigation, only Plié2.0 does not possess means for microprocessor control of swing flexion angle to adapt to different walking speeds.

The control of the following swing phase extension is a further decisive criterion for a functional prosthetic knee joint. In this context, the knee joint should reach the extension stop at the end of the swing phase before a new initial contact. Thus, reproducible positioning of the prosthetic foot on the ground is ensured. During natural walking, the knee joint reaches the maximum extension at ~97% of the gait cycle. From the tested knee joints, this is achieved best by Plié2.0 and C-Leg at all tested walking velocities. The highest motion just before the initial contact, measured by the highest angular velocity, is demonstrated by Rel-K followed by Orion. However, no stumbling situation with Orion and Rel-K could be observed, as walking on even ground does not pose high demands on foot positioning. Besides, the reliable reaching of the extension stop with Plié2.0 is linked to the highest hip moments in swing phase initiation resulting from the non-controlled air spring of the extension assistance.

Limitations of the study

A significant limitation of the study design is the low number of subjects that participated in the gait laboratory measurements. In addition, the recruited subjects represented only functional level K3, which has different demands compared with lower functional levels. As the different designs underline the results in the gait laboratory, this limitation was acceptable. A comprehensive examination of functions of the knee joints should also examine stability during standing and especially stumbling and falling situations. From ethical aspects, falling tests require extensive security measures, which could, therefore, not be accomplished. The test situation in the laboratory can influence the gait of the subject. Preferable field measurements could not be realized as only selected kinematic parameters are recorded with the mobile Oktapod measuring system. Functions of the tested prosthetic components deviating from that of the previously existing prosthesis could also have restrictive effects. To reduce the appearance of those effects, the gait was evaluated by the prosthetist and the subject after accommodation time. To identify the differences in the performance of the knee joints, the prosthetic alignments and the knee joint parameter settings should be comparable. With the use of the alignment tools [a], the first requirement could be fulfilled. In consequence of different parameters and parameter scales for every knee joint type, the fulfillment of the second requirement could only be assumed indirectly. The mostly reliable reaching of the extension stop in the terminal swing phase at slow and fast velocities indicates that the individual setting of the joint parameters makes a good compromise for the full speed range. However, this compromise works best for a single subject. To cover the whole range of parameter settings, a huge number of subjects would be needed.

Outlook

In the gait laboratory, this study only examines aspect of level walking on even ground. The technical analyses,
however, showed differences in the generation of joint resistances that might be safety relevant during gait situations. Thus, the performance of the different designs concerning safety should be further examined.

To overcome the restrictions of gait laboratory tests concerning safety and reproducibility, a testing machine for exoprostheses was developed at the TU-Berlin. This gait simulator reproduces the gait of transfemoral amputees, including flexion/extension, adduction/abduction, and inversion/eversion of the hip. A further intention is to configure the simulator for tests that provide information on the potential of prosthetic knee joints to prevent falls. In the future, the combination of measurements from a stationary gait analysis and simulator tests could provide the basis for comprehensive and reproducible functional testing of prosthetic components.

**Summary**

With the microprocessor-controlled knee joints compared in this study (C-Leg, Plié2.0, Orion, and Rel-K), differences in the quality of functions required for level walking could be identified. The technical analyses of the functional principles showed that the differences in joint design and control corroborate those differences.

The stance phase control is one of the central functions of prosthetic knee joints. While walking with stance flexion, the gait pattern appears more natural and the full ground contact of the foot is reached faster. The analysis of the knee angular acceleration showed that damping of the extension movement after stance phase flexion is significantly smoother with the C-Leg and the Orion joints than with the Plié2.0 and the Rel-K joints. The technical analysis showed that Plié2.0 and Rel-K do not possess technical means to support proper hydraulic stance extension damping. With Rel-K, the following swing phase cannot be initiated, making stance phase flexion impossible.

Owing to the low hip flexion moment that had to be provided by the hip musculature, the swing phase could be initiated most easily at all walking velocities with C-Leg in comparison to all other tested joints. During swing phase flexion, significant differences in the maximum knee flexion angle were identified at the different walking velocities. C-Leg shows the least variation of the maximum knee flexion angle with increasing walking velocity, thus meeting the demand for optimal swing phase flexion control. With the Orion and the Rel-K joints, the peak knee flexion angles increase significantly with increased walking speed, whereas the knee flexion angle on the contralateral side was nearly constant. Plié2.0 deviates most from an optimal swing phase flexion control. The technical analysis showed that Plié2.0 does not possess technical means to support microprocessor control during the swing phase. Having accurate and dependable swing angles during variable cadence ensures that toe clearance occurs and extension stop is reached timely in the swing phase. That may reduce the tendency to stumble.

Reaching of the extension stop during the end of the swing phase allows reproducible positioning of the prosthetic foot. The extension stop was reached reliably with Plié2.0 and C-Leg. The Orion and Rel-K joints more seldom reached the extension stop, especially at mid velocity.

The technical analysis showed that C-Leg, Orion, and Rel-K are using a servo-hydraulic valve that allows microprocessor-controlled variation of the flexion resistance. The design of Plié2.0 allows switching between a lower manually adjustable and a higher manually adjustable flexion resistance. With Plié2.0, the flexion resistances are not adapted to different walking velocities by microprocessor control, which is likely the reason for the significant increase of the maximum flexion angle at higher walking velocity. Abrupt terminal stance phase extension could be attributed to the fact that the microprocessor cannot control extension resistances with Rel-K and Plié2.0.

The significant differences in function found in this study suggest that patients’ benefits may also vary remarkably between the devices tested. If there are significant differences between the MPKs in level walking, even more pronounced differences may be expected on more challenging terrains and in safety-relevant situations. In light of these findings, it does not appear adequate to generalize clinical trial results with one MPK to the whole group of these devices. Further research is required to elucidate the differential function and safety of microprocessor-controlled prosthetic knees.

**Providers**

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