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Gait Asymmetry of Transfemoral Amputees Using Mechanical and Microprocessor-Controlled Prosthetic Knees

Kenton R. Kaufman, Ph.D., P.E.[a,*], **Serena Frittoli, M.S.**[a,b], and **Carlo A. Frigo, Ph.D.**[b]

[a]Mayo Clinic, Motion Analysis Laboratory, 200 First Street SW, Rochester, MN 55905, USA

[b]Polytechnic Institute of Milan, Bioengineering Department, via Golgi, 39, 20133 Milan, Italy

Abstract

Background—Amputees walk with an asymmetrical gait, which may lead to future musculoskeletal degenerative changes. The purpose of this study was to compare the gait asymmetry of active transfemoral amputees while using a passive mechanical knee joint or a microprocessor-controlled knee joint.

Methods—Objective 3D gait measurements were obtained in 15 subjects (12 men and 3 women; age 42, range 26–57). Research participants were longtime users of a mechanical prosthesis (mean 20 years, range 3–36 years). Joint symmetry was calculated using a novel method that includes the entire waveform throughout the gait cycle.

Findings—There was no significant difference in hip, knee and ankle kinematics symmetry when using the different knee prostheses. In contrast, the results demonstrated a significant improvement in lower extremity joint kinetics symmetry when using the microprocessor-controlled knee.

Interpretation—Use of the microprocessor-controlled knee joint resulted in improved gait symmetry. These improvements may lead to a reduction in the degenerative musculoskeletal changes often experienced by amputees.

Keywords

amputee; artificial limbs; gait; knee; microprocessor

1. INTRODUCTION

Mobility is an important aspect of an individual's quality of life. Walking is more difficult for transfemoral amputees to perform because they need to depend on an artificial limb for body weight support and gait mobility. Walking biomechanics is altered with the use of prosthesis. The gait of persons with a unilateral transfemoral amputation is asymmetrical (Jaegers et al., 1995). Altered load distribution may lead to back and/or intact limb pain (Burke et al., 1978, Ephraim et al., 2005) osteoarthritis in the intact limb (Burke et al., 1978, Kulkarni et al., 1998), osteopenia/osteoporosis in the residual limb (Kulkarni et al., 1998),

*Corresponding Author: Kenton R. Kaufman, Ph.D., P.E., Mayo Clinic Rochester, Motion Analysis Laboratory, 200 First Street SW, Rochester, MN 55905, Phone: 507-284-2262, Fax: 507-266-2227, kaufman.kenton@mayo.edu.

and other musculoskeletal problems (Ephraim et al., 2005). These degenerative changes can prevent the performance of everyday tasks and lead to a reduction in the quality of life.

Prosthetic knee joints for unilateral transfemoral amputees have undergone many design improvements over the past three decades. At present, transfemoral amputee prosthetic knee control is achieved through either mechanical mechanisms (non-microprocessor knee, NMPK) or microprocessor controls (MPK) (Michael, 1999). Mechanical mechanisms include single axis, constant-friction, weight activated, stance-phase control knee joints; single-hinge fluid-controlled (pneumatic or hydraulic) knee systems (with fluid swing phase control and variable methods of stance stability); and polycentric knee components that allow designers to optimize stance and swing features. Microprocessor controls regulate knee joint dynamics through analysis of several kinematic and kinetic variables, allowing more precise adjustment of knee resistance and providing the user to walk in more demanding situations such as descending stairs, step over step, or traversing a hillside.

For the above-knee amputee, the prosthetic knee joint is a critical component because it plays a complex role by providing stability in the absence of knee extensors. Several studies have compared various outcomes associated with the use of different prostheses. Most studies reported a benefit when using a MPK including lower oxygen/energy consumption (Johansson et al., 2005, Perry et al., 2004), increased walking velocity (Hafner et al., 2007, Orendurff et al., 2006, Perry et al., 2004), reduction in stumble and falls (Hafner et al., 2007, Orendurff et al., 2006, Segal et al., 2006), improved performance on stairs (Orendurff et al., 2006) and hill descent (Hafner et al., 2007), capability to adapt to any walking speed (Orendurff et al., 2006), and decreased cognitive effort (Hafner et al., 2007, Heller et al., 2000). Studies have reported kinetics and kinematics closer to the normal knee (Kaufman et al., 2007) and increased satisfaction (Hafner et al., 2007, Kaufman et al., 2008) when using an MPK. In other studies, no significant difference in the walking speed (Segal et al., 2006) or in the cognitive demand (Heller et al., 2000) was reported between the two prostheses.

Asymmetry, or lack of symmetry, appears to be a relevant aspect for differentiating a normal and pathological gait. Several methods have been used to determine asymmetry between the lower limbs. Gait asymmetry is often described as a ratio of the kinematic or kinetic parameters between the right and left sides. This has most often been assessed by calculating a symmetry index (SI) (Robinson et al., 1987), a ratio index (RI) (Ganguli et al., 1974), or a symmetry angle (SA) (Zifchock et al., 2008). All these indices have major limitations because these ratios are reported as a single point in the gait cycle. Gait asymmetry has also been reported as the difference between parameters recorded on the two limbs using a t-test, MANOVA, variance ratios (Winter and Yack, 1987), principal component analysis (Sadeghi, 2003), correlation coefficients (Arsenault et al., 1986), coefficients of variation (Hershler and Milner, 1978), cross-correlation, and root-mean-square (RMS) difference measures (Haddad et al., 2006). Unfortunately, these statistical tests do not provide a measurement of the asymmetry magnitude. Accordingly, it is not possible to quantify the asymmetry effect.

The purpose of this study was to compare the gait symmetry of active transfemoral amputees while using a passive mechanical knee joint (NMPK) or a microprocessor-controlled knee (MPK) joint. Unlike previous studies, this study used the entire gait waveform rather than a limited set of points from the gait cycle. Specifically, we looked at the effect of the prosthetic knee component on the kinematic and kinetic characteristics of walking on flat, level ground. We hypothesized that the patient would have improved gait symmetry when wearing a MPK compared to a NMPK.

2. METHODS

2.1 Subjects

The protocol was approved by the Institutional Review Board at the Mayo Clinic. These subjects were recruited on a volunteer basis. The experimental procedures were explained to the subjects and consent was obtained prior to enrollment into the study. Before inclusion in the study, an experienced ABC certified prosthetist, certified by Otto Bock Healthcare to properly fit the MPK, examined each amputee. The prosthetist verified that the socket fit was comfortable, the overall mechanical function of the prosthesis was sound and properly aligned for stability and comfort, and the attachment mechanism of the prosthetic knee to the prosthetic socket would accommodate the Otto Bock C-Leg, First Generation. Inclusion criteria to participate in this study were unilateral transfemoral amputation, age 18 years and older, amputation for any reason, at least two years' experience using a prosthesis, Medicare Functional Classification Level 3 or 4, utilization of a passive mechanical prosthetic knee, no significant fluctuation in stump volume within the last 6 months, no other neuromuscular problems or a partial amputation of the contralateral limb, no acute illness or chronic illness, assistive aids for ambulation, and no dialysis. Control subjects were recruited by word of mouth. All control subjects were screened for previous or current back, hip, knee, or ankle joint disease, pain, or injury; previous lower limb fractures; lower limb injury and/or laxity; circulatory or neurologic conditions; or any other disease or injury that may have affected their gait patterns. No restrictions were placed on gender or race for either cohort.

2.2 Study design

The study employed a repeated-measures experimental design whereby only the prosthetic knee joint was changed. The independent variable in this study was the type of prosthetic knee. The design and function of the prosthetic knee is of particular importance because it is the most proximal artificial joint that the amputee must stabilize and control to effectively ambulate (Hafner et al., 2007). The same socket, suspension, and prosthetic foot were used for both studies to eliminate any confounding effect of these variables. The inertial characteristics of the limb were unchanged for the two prosthetic knees. Subjects were tested in an array of domains, including gait biomechanics, balance, energy expenditure, activity level, and prosthetic evaluation questionnaires. Only the gait symmetry is reported in this article. Results of the balance (Kaufman et al., 2007) as well as the energy expenditure and activity level (Kaufman et al., 2008) assessments are published elsewhere.

Data collection was performed over two sessions. During the first session, subjects performed three walking trials at a comfortable, self-selected pace along a 20 m gait pathway with the NMPK. The speed averaged 1.11 m/sec (SD = 0.22 m/sec). At the end of the first session, the knee joint in the subject's prosthesis was exchanged for a MPK. Subjects were instructed to use the MPK until they felt their gait had stabilized with the new prosthesis. The acclimation time averaged 18 weeks (SD = 8 weeks). Subjects returned to the gait laboratory for a second data collection session while wearing the MPK prosthesis. Data were again collected at the self-selected pace. Speed averaged 1.19 m/sec (SD = 0.23 m/sec). This acclimation period is similar to the time reported by other studies (Hafner and Smith, 2009, Kahle et al., 2008). All subjects completed the full protocol with each type of knee prosthesis.

2.3 Fitting and alignment of prosthesis

Alignment of the prosthesis is the relative position and orientation of the prosthetic components and affects comfort, function, and cosmesis. Improper alignment can contribute to poor socket fit, and would result in undesirable pressure distribution at the residual limb/socket interface which would cause discomfort, pain, and potentially tissue damage (Yang et al., 1991). Further, poor alignment can cause difficulty with flexing or stabilizing the knee. Alignment was quantified using the Otto Bock Laser Assisted Static Alignment Reference (LASAR) system (Blumentritt, 1997).

2.4 Gait analysis

Kinematic parameters were acquired with a computerized video motion analysis system utilizing ten infrared cameras (EvaRT 4.0, Motion Analysis Corporation, Santa Rosa, CA, USA). The spatial distribution of the cameras was optimized to yield reliable motion data at the hip, knee, and ankle, bilaterally. The motion capture system recorded and processed the locations of passive reflective markers placed at bony prominences for establishing anatomic coordinate systems for the pelvis, thigh, shank, and foot. A modified Helen Hayes marker configuration was used. One set of data corresponding to the standing position (static data) were recorded in order to calculate the location of the joint centers. Ground reaction forces were measured using four force plates (two AMTI and two Kistler) embedded in a 10m walkway synchronized to the video system. Kinematic and ground reaction force data were collected at 120 and 360 Hz, respectively. The 3D marker coordinates and force plate data were used as input to a commercial software program (OrthoTrak 5.0, Motion Analysis Corp., Santa Rosa, CA, USA) to calculate the 3D joint kinematics and kinetics. Gait cycle periods were selected by heel strike to heel strike events. Timing of all intra-cycle gait events was expressed as a percentage of the gait cycle, irrespective of the actual time for a stride, to yield a normalized gait cycle (Kaufman et al., 2007).

2.5 Symmetry index

The symmetry index compared the kinematics and kinetics of the non-prosthetic leg (NPL) to the prosthetic leg (PL) for each type of prosthesis used. The symmetry index was calculated during the stance and swing phase of the gait cycle for each subject (Shorter et al., 2008). The method utilized expanded the method proposed by Crenshaw and Richards (Crenshaw and Richards, 2006), which uses the Singular Value Decomposition. Each gait variable was translated by subtracting its mean value from every value in the waveform. Translated data points from the NPL and PL waveforms were entered into a 2xn matrix (M=[NPL; PL]). The eigenvalues and the eigenvectors were then calculated from the matrix M. The symmetry index was calculated as the ratio between the variance about the eigenvector (second eigenvalue squared and divided by n-1) and the variance along the eigenvector (first eigenvalue squared and divided by n-1). The Crenshaw and Richards method (Crenshaw and Richards, 2006) was expanded by subtracting the obtained value from 1.0 and the sign associated with the slope of the eigenvector was assigned. A final value of +1 indicated perfect symmetry between the two waveforms, while a value of −1 indicated perfect asymmetry. A value of 0 indicated that the waveform shapes were unrelated. For each trial, the symmetry index of selected variables between the PL and NPL was calculated for the kinematics and kinetics at the three joints (ankle, knee and hip) in the sagittal plane.

2.6 Statistical analysis

The analysis focused on the sagittal plane only because the prosthetic knees are one degree-of-freedom devices which only allow motion in the sagittal plane. For each joint, the symmetry index was calculated for the two phases of gait (stance and swing) and for the two different knee components (NMPK and MPK). Statistical analysis was performed using a commercial statistical analysis package (SAS 9.1, Cary, NC, USA). A two-way, repeated measures Analysis of Variance (ANOVA) (2 gait phases × 2 knees) was used for determining whether the subject's gait symmetry changed when wearing the different prosthetic knees. Statistical significance was set at $p \leq 0.05$. The gait biomechanical variables that demonstrated statistically significant differences between the two knee prosthetic knee conditions were then compared to able-bodied subjects to determine how closely the prostheses approximated normal walking. A two sample t-test was used to compare the demographics of the amputee cohort to the control group.

3. RESULTS

3.1 Participants

The study cohort consisted of 15 subjects [12 men and 3 women; mean age of 42 (SD = 9 years, range 26–57); and mean BMI of 24 kg/m^2 (SD = 4 kg/m^2)] who had a unilateral above-knee amputation due to trauma (7), cancer (6), peripheral vascular disease (1), or congenital factors (1). All subjects were long-term prosthesis users with an average age of 20 years (SD = 10 years). They were tested with a mechanical fluid-controlled knee prosthesis (11 Mauch SNS, 2 CaTech, 1 Black Max, 1 Century 2000) and retested with a microprocessor-controlled knee joint (Otto Bock C-Leg) after an acclimation period. The average acclimation period was 18 weeks (SD = 8 weeks) (Table 1). We chose to first test the subjects with the mechanical prosthesis since the amputees were already acclimated to it and it reproduced the clinical experience of most transfemoral amputees.

As a basis for comparison, 20 able-bodied healthy subjects were also studied. This group consisted of 9 males and 11 females. Subjects ranged in age from 20 to 42 with a mean age of 28 (SD = 9 years) and mean BMI of 23 kg/m^2 (SD = 3 kg/m^2). These subjects had no history of osteoarthritis, joint instability, or major lower extremity joint surgery. These individuals had normal strength, full range of motion of the lower extremities, and no neurologic deficits. While the controls were significantly younger than the amputees ($p < 0.01$), they were similar in body mass index ($p = 0.19$).

3.2 Kinematic symmetry

The kinematic symmetry differed by joint level and gait phase (Figure 1). In comparison, the symmetry indices of the control subjects were >0.99 for the sagittal plane kinematics of all lower extremity joints. At the hip joint in the amputee population, the sagittal motion exhibited good symmetry both in stance and swing phases were the symmetry index was >0.98, and approximated the control subject’s symmetry. There was no significant difference in hip kinematic symmetry when the subjects wore either of the two knee prostheses (p=0.15). At the knee joint, there was a significant difference in symmetry between stance and swing (p<0.008) with the greatest asymmetry during stance. The stance phase kinematic asymmetry, in most cases, depended on differences in the knee position during loading response. In the swing phase, most of the subjects had symmetrical kinematics. There was no significant difference in knee kinematic symmetry when the subjects wore either of the two knee prostheses (p=0.38). At the ankle joint, there was a large range of ankle symmetry during stance, while in the swing phase the symmetry index was close to −1, thus indicating an almost perfect asymmetry. This was primarily due to the lack of plantarflexion of the prosthetic foot, which resulted in significant differences between these two phases of gait (p<0.001). No significant differences were found in the symmetry indexes for ankle kinematics between NMPK and MPK (p=0.07). The data from a representative subject will more fully demonstrate the gait differences between a NMPK and a MPK (Figure 2). The amputee lacked a knee flexion loading response when wearing the NMPK and achieved a knee flexion loading response when using the MPK. Accordingly, the stance-phase knee asymmetry index changed from −0.012 when using the NMPK to 0.441 when using the MPK. In contrast, the symmetry index remained essentially unchanged at the knee during swing and at the hip and ankle during stance and swing (Table 2).

3.3 Kinetic symmetry

The joint kinetics exhibited a symmetrical behavior for all joints (symmetry index close to 1) except for the knee in stance phase (Figure 3). The hip moment symmetry was significantly higher in stance than in swing ($p=0.008$). In contrast, the amputees' knee and ankle moments were significantly more symmetrical during swing that during stance ($p<0.01$). All subjects demonstrated significant improvement in gait symmetry at the hip, knee, and ankle after receiving the MPK ($p<0.002$). Data from a representative subject (Figure 4) shows the reason for the significant change in kinetic symmetry when using the MPK. When using the NMPK, the knee moment remained an internal flexion moment throughout all of stance as the subject maintained the force vector in front of the knee to assure stance phase stability. In contrast, the subject adopted a more symmetrical gait when using the MPK. A knee internal extension moment was generated during stance, thereby indicating more reliance on the prosthetic knee for stance phase stability and resulting in greater symmetry between the prosthetic and non-prosthetic limb during stance. The greater difference in the stance phase knee moment between the prosthetic and non-prosthetic limb when using the NMPK is reflected with a symmetry index of 0.459 as compared to a symmetry index of 0.640 when using the MPK. To a lesser extent, there were also increases (improvements) in the swing phase knee moment and the hip moment symmetry (Table 3). For comparison, the symmetry indices of the control subjects were >0.99 for the sagittal plane kinetics of all lower extremity joints.

4. DISCUSSION

This study analyzed the gait symmetry of transfemoral amputees wearing two different kinds of prostheses (NMPK and MPK). We hypothesized that differences in the control of the two prostheses could have an effect on gait symmetry. For simplicity, normal gait can be considered symmetrical. There is contrasting evidence in the literature regarding the advantages provided by MPK mechanisms over a NMPK, specifically in terms of gait symmetry. In this study, we used a new method to calculate gait symmetry that does not simply compare data points along the stride cycles, but uses the whole waveforms for comparison. This study demonstrated that amputees have improved gait kinetic symmetry when using a C-Leg. Previous research on non-amputee subjects reported global symmetry when the general behavior of the limbs was considered (Sadeghi, 2003).

Our results indicate that there was no significant difference in joint kinematics when using the different kinds of prostheses. However the prosthetic limb peak knee-flexion angle during swing decreased when using the MPK compared to the NMPK. Similar findings have been reported by Segal et al. (Segal et al., 2006) who indicated this behavior is caused by the higher damping of the MPK. During stance phase, the knee flexion during loading response was greater than zero for the MPK in some subjects (Kaufman et al., 2007). These results differ from other studies (Johansson et al., 2005, Segal et al., 2006) that show an incapacity to achieve knee flexion during loading response when using either kind of prosthesis.

The results demonstrated a statistically significant improvement in gait symmetry of the sagittal plane moments when using a MPK. We have previously reported an improvement in the knee extensor moment when converting from a NMPK to a MPK (Kaufman et al., 2007). Segal (Segal et al., 2006) similarly reported that the stance knee-flexion moment increased for the MPK compared to the NMPK. This improvement in the stance phase knee moment is very important because the knee joint is the most important joint for the stability during stance. The heel strike and the loading response phases of the prosthetic limb are recognized as the most critical phases of an amputee’s gait (Schmid et al., 2005). The results of this study suggest that the MPK improves amputee gait through more natural movements and could explain the improved balance and stability found in a number of previous studies (Highsmith et al., 2010, Kaufman et al., 2007).

Unilateral leg amputation leads to gait problems. Previous studies have shown that the gait of amputees has asymmetry in the temporal parameters (Jaegers et al., 1995, Nolan et al., 2003), ground reaction forces (Engsberg et al., 1993), and center of pressure trajectories (Schmid et al., 2005). Asymmetrical gait results in increased loading of the intact leg (Suzuki, 1972). In transfemoral amputees, the knee extensor moment which contributes to shock absorption during weight acceptance is increased in the intact limb when compared to a control group (Beyaert et al., 2008, Nolan and Lees, 2000). Amputees also have increased musculoskeletal disorders (Burke et al., 1978, Kulkarni et al., 1998, Norvell et al., 2005) when compared to a control group. The results of this study and a companion study (Kaufman et al., 2007) demonstrate that use of a MPK improved gait and balance. These improvements may lead to reduced degenerative changes. The long term effects of advances in prosthetic care, such as a MPK, will need to be confirmed in a future study.

This study had several limitations. First, the order of testing the prosthetic knees was not randomized. The MPK knee was fit after the NMPK. This reflects the typical clinical scenario for the patient on an NMPK, who may subsequently be fit with an MPK. However, the lack of randomization may have resulted in an effect bias. Second, the amputee subjects chose to walk at self-selected speeds during the trial and there were different speeds when they were wearing the MPK and the NMPK. Lelas et al. (Lelas et al., 2003) reported that alterations in walking speed resulted in systematic changes in peak kinematic and kinetic variables, especially related to knee flexion during stance. Specifically, 60 to 72 % of the variance for knee kinematics and kinetics are associated with walking speed. Similarly, Nolan et al (Nolan et al., 2003) reported changes in loading asymmetry of amputees when walking speed increased. This change in walking speed could have affected our results. However, the change in walking speed in these other studies was greater than 1 m/s. In contrast, the change in walking speed between the two test conditions tested in this study was very small (0.08 m/s). This difference in walking speed, most likely, had little effect on the conclusions of this study.

5. CONCLUSION

The results of this study indicate that a MPK has significantly improved kinetic symmetry over a NMPK for unilateral transfemoral amputees walking at self-selected speeds on level ground. The results of this investigation not only highlight measured differences between the MPK and NMPK, but also offer a new method for assessing gait symmetry. This type of analysis will be useful for clinicians to better detect gait impairments and to quantitatively monitor change in gait as a function of prosthetic component utilization.

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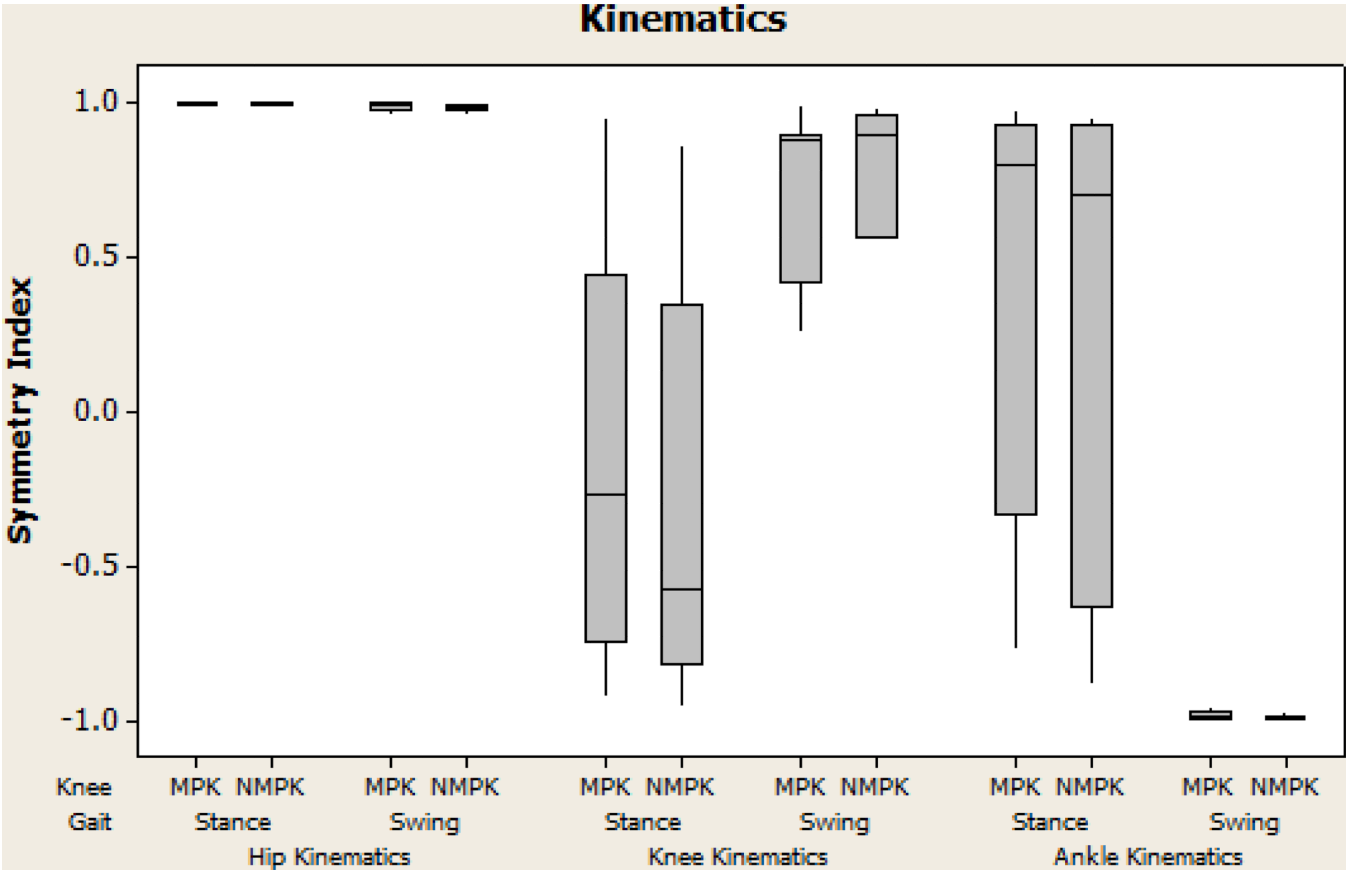

Figure 1.
Symmetry index for kinematics in the sagittal plane for three joints (hip, knee, ankle) and two different prostheses. There were no significant differences between the MPK and NMPK. There was a significant difference between stance and swing phase gait symmetry.

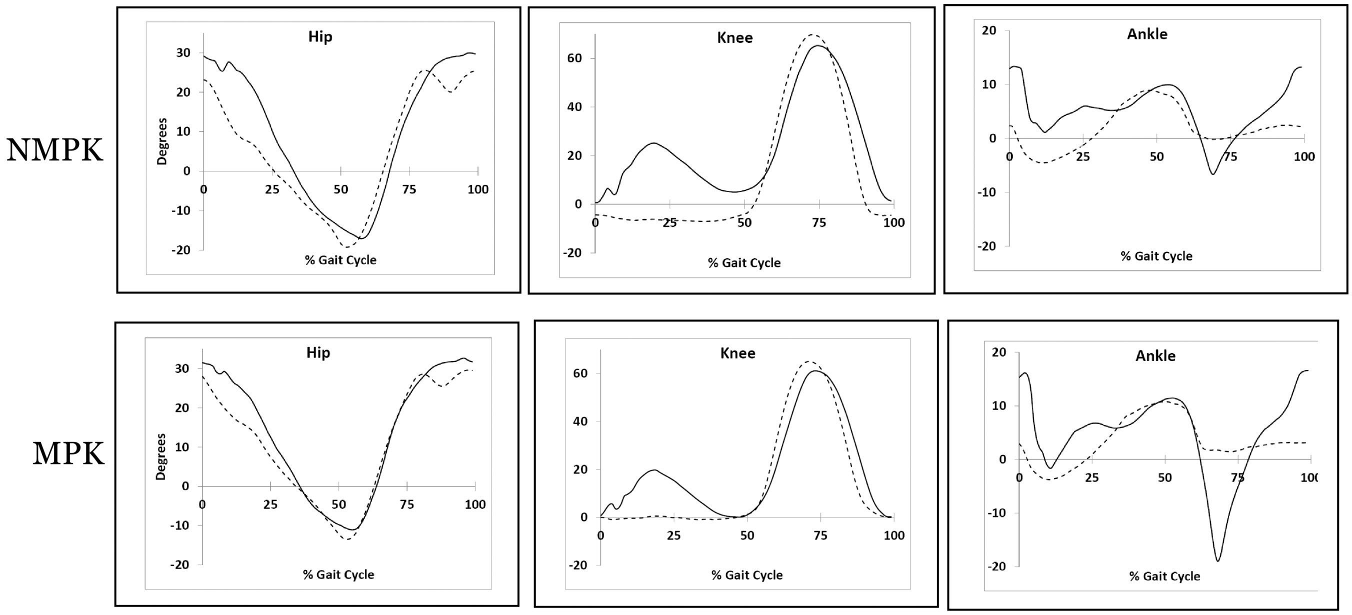

Figure 2.
Sagittal plane kinematics of the intact leg (solid line) versus the prosthetic leg (dashed line) of a representative subject for the three joints (hip, knee, ankle). The subject is wearing a NMPK knee in the first row and a MPK in the second row. Graphs are plotted as percentage of gait cycle (% Gait Cycle), where 0% is heel strike and 100% is subsequent heel strike. A positive value indicates hip flexion, knee flexion, and ankle dorsiflexion.

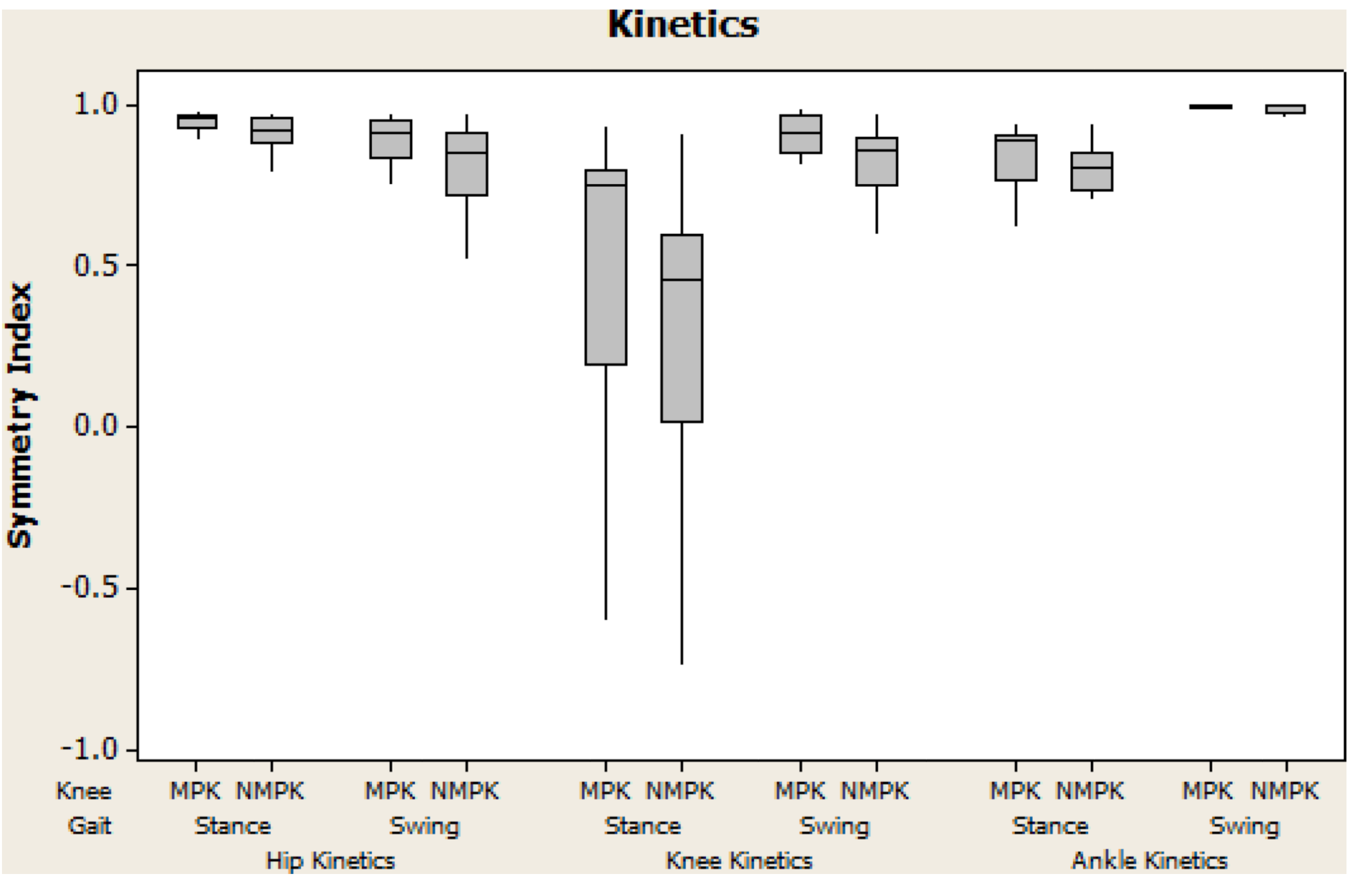

Figure 3.
Symmetry index for kinetics in the sagittal plane for three joints (hip, knee, ankle) and two different prostheses. There was a significant improvement in the symmetry index for all joints when the MPK was used. There was also a significant difference in the symmetry index between stance and swing.

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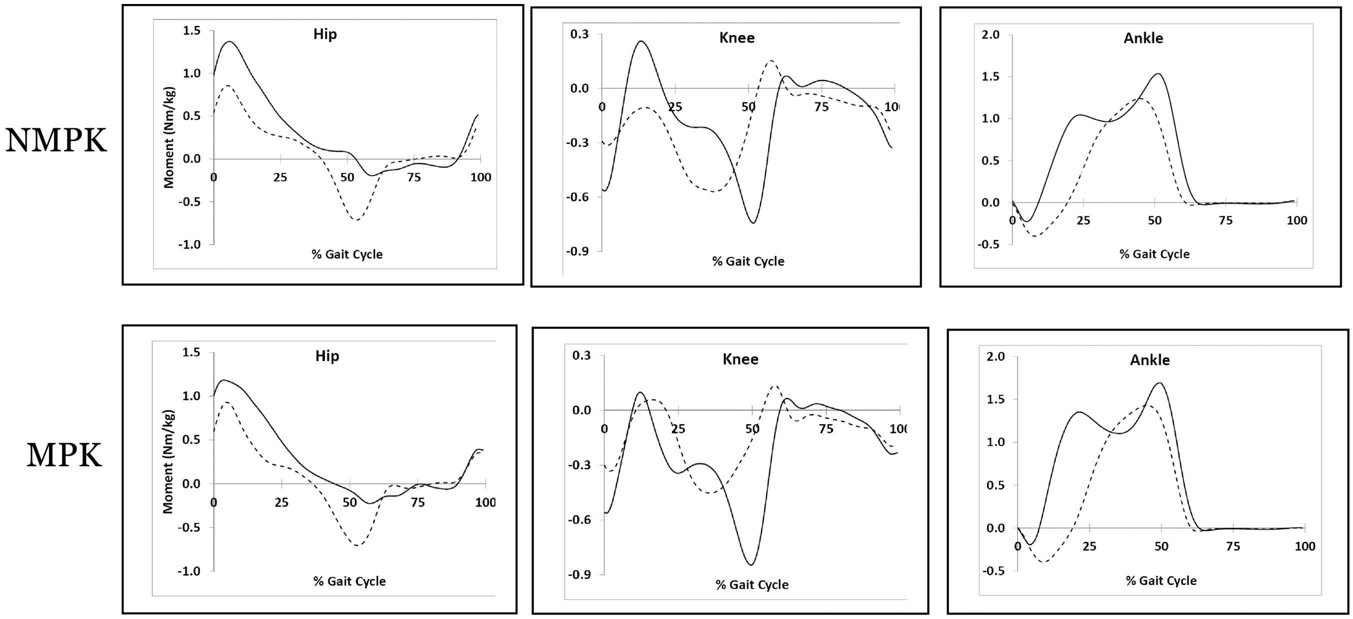

Figure 4.

Sagittal plane Kinetics of the intact leg (solid line) versus the prosthetic leg (dashed line) for a representative subject at the three joints (hip, knee, ankle). The subject is wearing a NMPK knee in the first row and a MPK in the second row. Graphs are plotted as percentage of gait cycle (% Gait Cycle), where 0% is heel strike and 100% is subsequent heel strike. A positive value represents an internal hip extensor, knee extensor, and ankle plantarflexor moment.

Table 1

Characteristics of participating transfemoral amputees

Subject	Age (y)	Sex	Etiology	Years as amputee	NMPK	Prosthetic foot	Acclimation (weeks)*	Employed
A	37	F	CA	25	CaTech	Luxon Max	14.4	Y
B	30	M	CA	11	Mauch SNS	Dynamic Plus	16.9	Y
C	57	M	PVD	6	Mauch SNS	Luxon Max	10.4	Y
D	46	F	Trauma	33	Mauch SNS	Luxon Max	18.1	N
E	46	M	CA	24	Mauch SNS	Luxon Max	10.7	Y
F	55	M	Trauma	35	Mauch SNS	Luxon Max	17.0	N
G	44	M	Trauma	25	Black Max	College Park	10.1	Y
H	37	M	Trauma	17	Mauch SNS	Axtion	20.6	Y
I	39	M	Trauma	12	Mauch SNS	Axtion	39.3	Y
J	54	M	CA	9	Centuary2000	Axtion	18.0	Y
K	31	M	Congenital	31	Mauch SNS	Axtion	33.0	Y
L	26	F	CA	2	Mauch SNS	Axtion	16.4	Y
M	46	M	Trauma	14	Mauch SNS	Axtion	18.7	Y
N	44	M	CA	28	CaTech	Axtion	14.0	Y
O	42	M	Trauma	20	Mauch SNS	Axtion	14.7	Y

*Weeks of acclimation period with MPK

Abbreviation: CA, cancer; PVD, peripheral vascular disease; R, right; L, left; Y, yes; N, no.

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Table 2

Symmetry Index of the hip, knee and ankle sagittal plane motion for a representative subject (Reported in Figure 2).

HIP				KNEE				ANKLE			
Stance		Swing		Stance		Swing		Stance		Swing	
NMPK	MPK	NMPK	MPK	NMPK	MPK	NMPK	MPK	NMPK	MPK	NMPK	MPK
0.987	0.995	0.980	0.992	−0.012	0.441	0.957	0.953	0.713	0.773	−0.995	−0.989

A value of +1 indicates perfect symmetry and a value of −1 indicates perfect asymmetry

Table 3

Symmetry Index of the hip, knee and ankle sagittal plane moments for a representative subject (Reported in Figure 4).

HIP				KNEE				ANKLE			
Stance		Swing		Stance		Swing		Stance		Swing	
NMPK	MPK	NMPK	MPK	NMPK	MPK	NMPK	MPK	NMPK	MPK	NMPK	MPK
0.942	0.976	0.846	0.943	0.459	0.640	0.768	0.915	0.835	0.900	0.974	0.987

A value of +1 indicates perfect symmetry and a value of −1 indicates perfect asymmetry