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The influence of a user-adaptive prosthetic knee on planned gait termination

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Abstract— **: The added value of user-adaptive prosthetic knees has been predominantly evaluated in level walking or ramp/stair negotiation. Previous studies indicate that the activity pattern of individuals with an amputation mainly consists of short periods of continuous walking, indicating that a high percentage of ambulatory activity involves gait termination. The potential added value of user-adaptive prosthetic knees in gait termination has not been studied yet. Ten individuals with an amputation were measured with their own non-microprocessor controlled prosthetic knee and with the Rheo Knee II (a user-adaptive prosthetic knee). Spatiotemporal, kinematic and kinetic variables were measured. We found that the Rheo Knee II had no effect on the studied outcome parameters when compared to the non-microprocessor controlled prosthetic knee. We also found that the intact leg was responsible for producing the deceleration forces irrespective whether the last step was made by the intact or prosthetic leg. In conclusion we found that the prosthetic leg is limited in producing deceleration forces. Although user-adaptive prosthetic knees claim to increase stance stability, the added value of the Rheo Knee II on the studied outcome parameters is limited.**

I. INTRODUCTION

Developments in prosthetic knee technology have led to the introduction of user-adaptive microprocessor-controlled prosthetic knees. Examples of these prosthetic knees are the C-Leg and the Rheo Knee. User-adaptive prosthetic knees are capable of changing damping properties based on information from different sensors, such as knee angle, knee angular velocity and force. [1, 2] This variable damping is thought to lead to more optimal knee kinematics and, ultimately, be beneficial for the user.

The added value of user-adaptive prosthetic knees has been studied on a number of different tasks, but the majority studied level walking or stair or ramp negotiation. Studies looking into the activity pattern of persons with an amputation showed that majority of periods of continuous activity were only 1 or 2 minutes in length. [3] Klute et al. also found that persons with an amputation walked for 10 minutes only once or twice a day. [3] These results indicate that a high percentage of ambulatory activity of persons with an amputation involves gait initiation and termination. A recent study investigated the added value of a user-adaptive prosthetic knee in gait initiation. [4] However, to our knowledge, the added value of a user-adaptive prosthetic knee in gait termination has not been studied yet.

Previous authors who studied gait termination in persons with an amputation have found that the prosthetic leg has a reduced role in generating deceleration forces that are needed to terminate gait. [5, 6] These deceleration forces are normally generated by placing the center of pressure (CoP) in front of the center of mass (CoM). This slows the forward movement of the CoM down. [7] Vrieling et al. showed that in persons with a transfemoral amputation this mechanism is impeded. [5] First of all, the stiffness of the prosthetic ankle inhibits a smooth anterior displacement of the CoP. [5] Next to this, the absence of prosthetic knee flexion during the loading response inhibits a posterior positioning of the CoM with respect to the CoP. [5] The lack of knee flexion also inhibits absorption of energy by the knee. This is of particular importance, as the negative work of the knee dissipates the largest amount of energy of any of the lower extremity joints. [8] Finally, the duration of single limb support of the prosthetic leg is reduced when compared to the intact leg, [5, 6] which limits the time in which the braking impulse can be produced. The duration of single limb support on the prosthetic leg is thought to be decreased because stability is challenged during this phase. Because of all this, persons with an amputation heavily rely on their intact leg for the absorption of energy during gait termination. [5, 6]

The Rheo Knee II is thought to increase stance stability. This could increase the duration of single limb support on the prosthetic leg while walking with the Rheo Knee II. This potentially leads to more time to produce deceleration forces and, in turn, to a larger deceleration impulse. Next to this, the Rheo Knee II should allow knee flexion during early stance. This would enable absorption of energy at knee level and would allow a more posterior position of the CoM with respect to the CoP. The latter would mean a more efficient use of the CoP-CoM mechanism.

The aim of the current study is to investigate the added value of the Rheo Knee II in planned gait termination. Therefore we compared the Rheo Knee II condition with a non-microprocessor controlled prosthetic knee condition. We hypothesized that the prosthetic leg produced a higher deceleration impulse by increasing the duration of the single limb support phase on the prosthetic leg. In addition, we hypothesized increased energy dissipation of the prosthetic knee while using the Rheo Knee II.

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II. METHODS

A. Subjects

For this study, persons with a transfemoral amputation were recruited from the Netherlands and Belgium. The inclusion criteria were: (1) at least one year post amputation, (2) functional level from K2 (limited outdoor) to K4 (active athlete), (3) never supplied with a microprocessor-controlled knee unit before. Exclusion criteria were: (1) other musculoskeletal problems influencing walking ability, (2) problems with the residual leg/bad socket fitting, (3) body weight > 125 Kg (maximum tolerable weight of the Rheo Knee II), (4) knee centre-floor distance below 41 cm (minimal height that is needed to build in the Rheo Knee II).

B. Study protocol

The persons with an amputation were randomly assigned to start measurements with their own mechanically passive prosthesis or to start measurements with the Rheo Knee II. We used a block randomization procedure with a block size of 4 to ensure comparable groups size. We used Excel (Microsoft, Redmond, USA) to generate random numbers which were used for the randomization procedure. In both prosthetic knee conditions the Vari-Flex with EVO prosthetic foot was used. Prior to the first measurement, participants with an amputation had an appointment with the prosthesist to either install the Vari-Flex-Evo foot combined with their own prosthesis, or to install the Rheo Knee II in combination with the Vari-Flex with Evo. After 8 weeks of acclimatization the first set of measurements was performed. After these measurements, the subjects crossed-over to the other prosthetic condition. After another 8 weeks of acclimatization the second set of measurements was performed after which the prosthesis was converted to the original set up and subjects left the study. All prosthetic fittings were performed by the same prosthesist.

C. Experimental protocol

Participants were instructed to walk five steps before coming to a halt. Data were collected using a three-dimensional optic movement tracking system consisting of six Vicon cameras (Oxford Metrics Ltd., Oxford, UK) and two AMTI force plates (Advanced Mechanical Technology Incorporated, Watertown, USA). The modified Helen-Hayes markers set was used, including 35 reflective makers. Markers were placed on anatomical landmarks according to the Vicon full-body plug in gait model. The collected data were processed using Vicon Nexus 1.8 (Oxford Metrics Ltd., Oxford, UK). The marker trajectories were recorded with a frequency of 100 Hz. The force plate data were collected at a frequency of 1000 Hz. A trial was valid when both the second last step as the step of the leading leg were on a force plate each. For details of the experimental set-up see Figure 1. Data collection was stopped when eight valid trials were available. The first five trials with a walking speed within ±5% of each other were included in the data analysis.

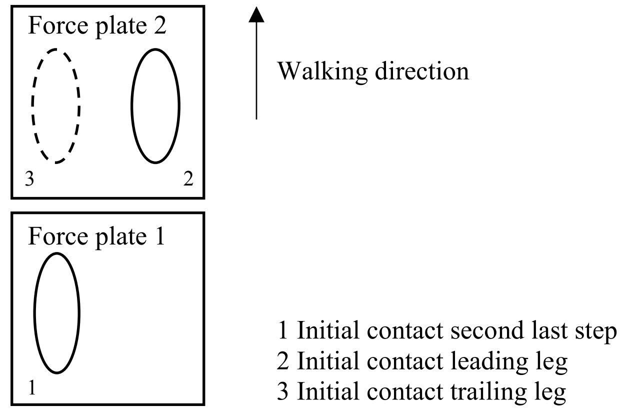

Figure 1. Details of experimental set-up

D. Data preprocessing

The first step in the data processing was identification of initial contact and initial swing of both legs. Initial contact was defined as the moment the heel marker stopped moving in a downward direction. Initial swing was defined as the moment the heel and toe marker both started moving in the upward direction. Subsequently, data were loaded into custom developed Matlab 2010b software (Mathworks, Natick, USA). Kinematic and kinetic data were filtered using a 2nd order bidirectional zero phase-shift Butterworth filter with a Low-pass filter of 10 Hz.

E. Data processing

We calculated spatiotemporal, kinematic and kinetic outcome measures, and the margins of stability. Firstly, spatiotemporal variables included duration of the first double limb support of the leading leg, duration of single limb support on the leading leg, and duration of the moment from initial contact of the trailing leg to end of gait termination. The moment of gait termination was retrospectively defined as the moment the anteroposterior component of the ground reaction force was below 1% of the body weight in Newton for 100 frames (0.1 sec). Secondly, we compared kinematics of the knee of the leading leg by determining knee angle at initial contact and knee yielding. Knee yielding was defined as the difference between maximum knee flexion during the loading response and knee flexion at initial contact. Thirdly, we calculated joint work of the ankle, knee, and hip of the leading leg as indication of energy dissipation on individual joint level. These were calculated by numerical integration of joint power. In addition, we calculated the deceleration impulse of the second last step and of the leading and trailing leg by numerical integration of the anteroposterior component of the ground reaction force. Finally, we calculated the margin of stability from the moment of initial contact of leading leg to the end of gait termination. The margin of stability is defined as distance between the CoP and the extrapolated centre of mass (XcoM). [9] The XcoM is calculated by adding the ratio of the CoM velocity and the eigen frequency of the pendulum to the CoM position. [9]

F. Data analysis

Data were analyzed using IBM SPSS statistics 22 software (IBM, Chicago, USA). Because of the small sample size, non-parametric statistics were chosen. For the within-subject comparisons of the persons with an amputation, the Wilcoxon Signed Rank Sum Test was used. For the comparisons of the persons with an amputation with the control subjects, the Mann-Whitney U test was used. Statistical significance was set at $P \leq 0.05$.

III. RESULTS

A. Subject recruitment

A group of 61 potential candidates were contacted of which 52 met the inclusion criteria. A total of twelve persons were willing to participate in the study. Two participants dropped out before the first measurement: one was not able to satisfactorily adjust to the Rheo Knee II and one developed problems with the residual leg. The remaining ten participants completed both measurements. Characteristics of participants are displayed in Table I.

TABLE I. CHARACTERISTICS OF PARTICIPANTS

Variables	Participants
Age[a]	56 (23 – 67)
Sex (male/female)[b]	6/4
Time since amputation (years)[a]	33.5 (1 – 41)
Level of amputation[b]	Transfemoral (7) Knee disarticulation (3)
Reason for amputation[b]	Trauma (7), Infection (2), Osteosarcoma (1)
Functional level[b]	K2 (2), K3 (5), K4 (3)
Own prosthetic knee[b]	3R60 (4), 3R80 (1), 3R95 (1), Mauch SNS (1), Graph Lite (1), CaTech (1), Total Knee 2000 (1)

a. Data presented as median (range)

b. Data presented as counts

B. Comparison of the leading prosthetic leg and leading intact leg condition within a prosthetic knee condition

When the leading intact leg condition is compared with the leading prosthetic leg condition within a prosthetic knee condition, similar results are visible for the non-adaptive prosthetic knee and the Rheo Knee II. For results see Table II and Figure 2, and 3.

On spatiotemporal variables the following statistically significant differences were found for both the non-adaptive prosthetic knee and Rheo Knee II condition. The duration of the single limb support of the leading leg was significantly larger in the leading intact leg condition, when compared to the leading prosthetic leg condition. In addition, the duration from the moment of initial contact of the trailing leg to the end of gait termination was significantly longer in the leading prosthetic leg condition when compared to the leading intact leg condition. CoM velocity at initial contact of the second-last step, the leading leg, and the trailing leg were also statistically significantly different between the leading intact and leading prosthetic leg condition (See Figure 2). For the leading prosthetic leg condition CoM velocity at initial contact of the second-last step (intact leg) and at initial contact of the trailing leg (intact leg) were significantly higher when compared to the leading intact leg condition. CoM velocity at initial contact of the leading leg was significantly higher for the leading intact leg condition when compared to the leading prosthetic leg condition. text heads.

On kinematic variables the following statistically significant differences were found in both prosthetic knee conditions. Both the knee angle at initial contact and knee yielding of the leading leg were smaller in the leading prosthetic leg condition when compared to the leading intact leg condition.

On knee work, we found the following statistically significant differences. For the non-adaptive prosthetic knee condition, ankle and hip work of the leading intact leg were higher when compared to the leading prosthetic leg. For the Rheo Knee II condition, knee work of the leading leg was significantly higher in the leading intact leg when compared to the leading prosthetic leg. When looking at the deceleration impulse the following statistically significant differences were visible for both prosthetic knee condition. For the leading prosthetic leg condition, the deceleration impulse of the second-last step (intact leg) and the deceleration impulse from initial contact of the trailing leg (intact leg) to the end of gait termination were higher when compared to the deceleration impulse produced by the prosthetic leg in the leading intact leg condition. The deceleration impulse of the leading leg was significantly higher in the leading intact leg condition when compared to the leading prosthetic leg condition.

The margin of stability is graphically represented in Figure 3. For both prosthetic knee conditions, the margins of stability were statistically significantly higher in the leading prosthetic leg condition when compared to the leading intact leg condition.

C. Comparisons of prosthetic knees

When the leading intact leg condition while walking with the non-adaptive prosthetic knee is compared to the leading leg condition while walking with the Rheo Knee II, no differences were found on all studied outcome parameters. Comparison of the leading prosthetic leg conditions of the non-adaptive prosthetic knee and Rheo Knee II showed one statistically significant difference. Knee flexion of the leading leg at initial contact was significantly higher for the Rheo Knee II condition when compared to the non-microprocessor controlled prosthetic knee.

IV. DISCUSSION

The aim of the current study was to study the effect of the Rheo Knee II on biomechanical variables of gait termination. We hypothesized that the prosthetic leg produced a higher deceleration impulse by increasing the duration of the single

TABLE II. REULTS OF STUDIED OUTCOME PARAMETERS

Variables	***Non-microprocessor controlled prosthetic knee***			***Rheo Knee***			***Comparison prosthetic knees***	
	Leading IL	Leading PL	*P*	Leading IL	Leading PL	*P*	LIL	LPL
Spatiotemporal							*P*	*P*
Double limb support leading leg (s)	0.19 [0.16-0.22]	0.18 [0.17-0.23]	0.721	0.20 [0.18-0.22]	0.19 [0.18-0.24]	0.959	0.475	0.646
Single limb support leading leg (s)	0.51 [0.43-0.56]	0.33 [0.28-0.37]	**0.005**	0.50 [0.43-0.58]	0.34 [0.27-0.36]]	**0.005**	0.953	0.139
Duration IC trailing leg to end gait termination (s)	0.52 [0.43-0.60]	0.80 [0.69-0.87]	**0.005**	0.53 [0.42-0.67]	0.78 [0.74-0.99]	**0.009**	0.799	0.678
CoM velocity IC second last step (m/s)	0.78 [0.66,0.85]	0.84 [0.78,0.94]	**0.011**	0.75 [0.61,0.82]	0.84 [0.78,0.92]	**0.008**	0.678	0.767
CoM velocity IC leading leg (m/s)	0.77 [0.72,0.80]	0.63 [0.57,0.69]	**0.005**	0.77 [0.71,0.82]	0.62 [0.53,0.68]	**0.005**	0.721	0.721
CoM velocity IC trailing leg (m/s)	0.10 [0.07,0.16]	0.32 [0.25,0.36]	**0.005**	0.10 [0.05,0.16]	0.32 [0.27,0.37]	**0.005**	0.799	0.139
Kinematics								
Knee angle at IC (degrees)	5.13 [1.81,10.08]	-3.84 [-5.54,.23]	**0.007**	4.24 [2.09,7.30]	-.67 [-2.19,2.89]	**0.047**	0.285	**0.028**
Knee yielding (degrees)	9.60 [7.80,16.00]	.42 [0,4.27]	**0.005**	9.03 [6.86,11.06]	.38 [0,2.96]	**0.007**	0.722	0.646
Kinetics								
Ankle work leading leg ($JKg^{-1}s^{-1}$)	-.056 [-.065,-.030]	-.020 [-.030,-.009]	**0.015**	-.040 [-.062,-.020]	-.020 [-.054,-.006]	0.314	0.173	0.139
Knee work leading leg ($JKg^{-1}s^{-1}$)	-.070 [-.092,-.050]	-.038 [-.076,-.022]	0.139	-.090 [-.109-.033]	-.042 [-.081,-.026]	**0.015**	0.374	0.678
Hip work leading leg ($JKg^{-1}s^{-1}$)	-.056 [-.065,-.030]	-.020 [-.030,-.009]	**0.015**	-.040 [-.062,-.020]	-.020 [-.054,-.006]	0.314	0.346	0.139
Brake impulse second last step (Ns^{-1})	-16.43 [-23.17,-12.90]	-32.79 [-37.69,-27.24]	**0.008**	-16.99 [-18.40,-1.06]	-31.60 [-38.11,-26.03]	**0.011**	0.214	1.000
Brake impulse leading leg (Ns^{-1}) – one foot	-63.21 [-72,05,-53.65]	-34.07 [-50.72,-28.86]	**0.008**	-62.23 [-65.09,-56.53]	-33.11 [-39.04,-23.56]	**0.008**	0.953	0.173
Brake impulse trailing leg (Ns^{-1}) – two feet	-9.22 [-14.38,-5.92]	-27.92 [-35.19,-21.46]	**0.008**	-9.79 [-14.80,-6.22]	-31.41 [-37.32,-26.63]	**0.008**	0.767	0.086
Margin of stability								
Max MoS first double support LIL (cm)	5.23 [4.01-7.31]	8.06 [6.32-9.18]	**0.021**	6.06 [4.75-6.79]	9.41 [7.90-10.29]	**0.021**	0.153	0.110
Max MoS single limb support LIL (cm) – one foot	4.80 [4.35-5.58]	11.81 [10.04-15.93]	**0.008**	4.54 [3.92-5.61]	14.19 [12.13-17.37]	**0.008**	0.678	0.110
Max MoS IC trailing leg to end gait termination (cm) – two feet	5.25 [4.15-5.67]	9.82 [6.90-11.89]	**0.015**	4.79 [4.58-5.34]	10.75 [7.96-13.15]	**0.008**	0.515	0.173

All data are presented as median (inter quartile range) with the exception of the *P* values.

Abbreviations: IL: intact leg; PL: prosthetic leg; LIL: leading intact leg; LPL: leading prosthetic leg; CoM: center of mass; IC: initial contact; Max: maximal; MoS: margin of stability

limb support phase on the prosthetic leg. In addition, the results disproved our hypotheses and showed that the transition towards the Rheo Knee II had a very limited effect on the studied outcome parameters

When looking at the first hypothesis, we found that the duration of single limb support of the leading leg was significantly higher in the leading intact leg condition when compared to the leading prosthetic leg condition. This was also found by previous studies. [5, 6] The Rheo Knee II had no influence on the duration of single limb support of the prosthetic leg in both the leading intact and leading prosthetic leg condition when compared to the non-microprocessor controlled prosthetic knee.

It is thought that individuals with an amputation shorten the single limb support duration on the prosthetic leg because stability is challenged during this phase due to the lack of active ankle control. [10] When looking at the margin of stability, we found that the maximal margin of stability was significantly higher in all phases of the leading prosthetic leg condition when compared to the leading intact leg condition. This was seen for both the non-microprocessor controlled prosthetic knee and Rheo Knee II condition. The main reason for the increased margin of stability is the reduced anterior shift of the CoP under the prosthetic foot in the leading prosthetic leg condition (see Figure 3). This has two consequences: (1) the leading prosthetic leg condition seems to be much more unstable when compared to the leading intact leg condition; (2) in the leading prosthetic leg condition, the mechanism of placing the CoP in front of the CoM by which braking forces are produced is impaired. The increased margin of stability in the leading prosthetic leg condition could have attributed to the reduced duration of single limb support of the prosthetic leg. However, whether the increased margin of stability and the reduced duration of single limb support are causally related, needs to be addressed in future studies.

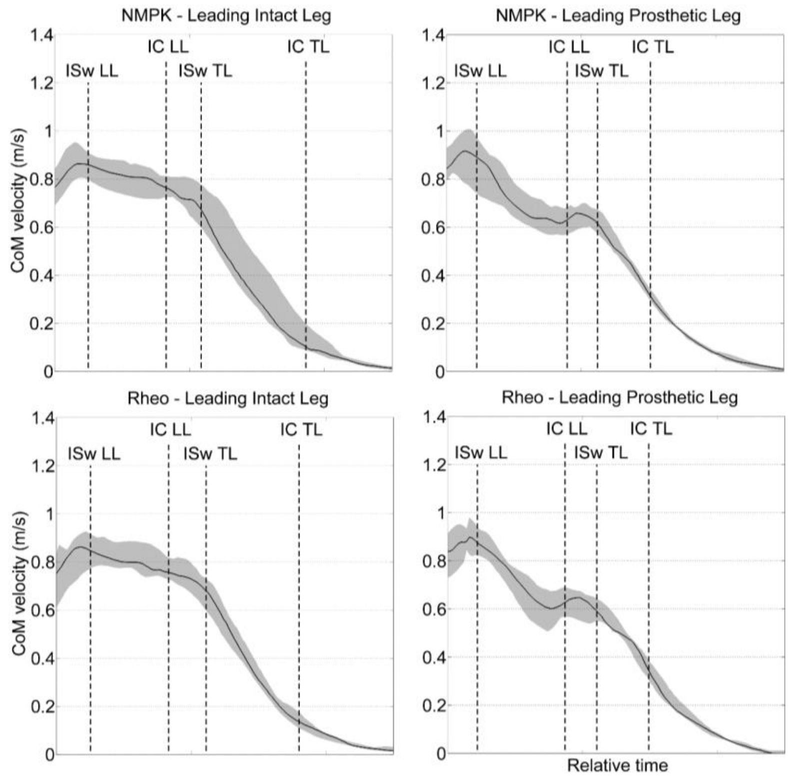

Figure 2. CoM velocity during the different conditions. The data are normalized on a relative time frame in which 0% is the moment of initial contact of the second last step and 100% is the end of gait termination. Abbreviations: IC: initial contact; LL: leading leg; ISw: initial swing; TL: trailing leg.

The fact that the CoP-CoM mechanism is impaired in the prosthetic leg was also clearly visible in the deceleration impulses generated by the intact and prosthetic leg. The deceleration impulse of the intact leg related to total deceleration impulse (combined deceleration impulse of the leading and trailing leg) was 87% for both prosthetic knee conditions. A previous study found that the leading intact leg of individuals with an amputation accounted for 85% of the total impulse. [6] Individuals with an amputation seem to increase the deceleration impulse of the leading intact leg by increasing the duration of the single limb support of the leading leg. This duration was significantly longer for the leading intact leg when compared to the leading prosthetic leg, which was also found by Vrieling et al. [5]

In the leading prosthetic leg condition the trailing intact leg also had a substantial role in the generation of the deceleration impulse. The deceleration impulse after initial contact of the trailing intact leg was significantly higher in the leading prosthetic leg condition when compared to the leading intact leg condition. Contrastingly, the deceleration impulse of the leading prosthetic leg was significantly lower when compared to the leading intact leg. When looking at deceleration impulse of the leading prosthetic leg related to the total deceleration impulse, we found that this was 55% for the non-microprocessor controlled prosthetic knee condition, and 51% for the Rheo Knee II condition. These values are in line with those of van Keeken et al., also reporting 55% in their study. In the leading prosthetic leg condition, individuals with an amputation seem to place their trailing intact leg as soon as possible, after which the majority of deceleration forces are produced.

The reliance on the intact leg for the generation of braking forces led to distinct differences in CoM velocity patterns between the leading intact and the leading prosthetic leg condition (see Figure 2). Individuals with an amputation seem to lower their CoM velocity before initial contact of the prosthetic leg during the last two steps. This was found irrespective of which leg was leading.

Our second hypothesis was that energy dissipation of the prosthetic knee would increase while walking with the Rheo Knee II compared to walking with the non-microprocessor controlled prosthetic knee. Our results disproved our hypothesis, as no differences in knee work were seen in the leading prosthetic leg between the non-microprocessor controlled prosthetic knee and Rheo Knee II condition.

We did find that intact leg reliance was reflected on all joint work outcome parameters. On all joint levels, the joint work of the leading intact leg was higher than the leading prosthetic leg. Although differences were marked, they were not significant for all comparisons. For the non-microprocessor controlled prosthetic knee condition, ankle and hip joint work of the leading leg were significantly higher in the leading intact leg condition when compared to the leading prosthetic leg condition. In the Rheo Knee II condition, only the knee work of the leading leg was significantly higher in the leading intact leg condition when compared to the leading prosthetic leg condition. The joint work of the leading prosthetic leg were

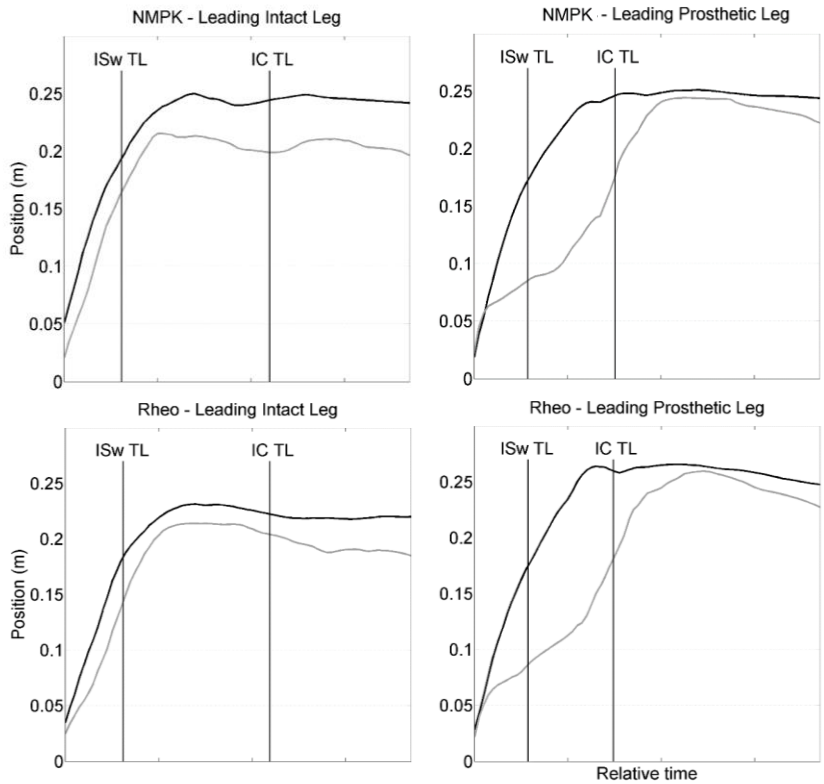

Figure 3. Margin of stability during the different conditions. The gray line represents the center of pressure and the black line represents the extrapolated center of mass. Data are normalized to a relative time frame in which 0% is initial contact of the leading leg and 100% is the end of gait termination. Abbreviations: ISw: initial swing; TL: trailing leg; IC: initial contact.

comparable in both prosthetic knee conditions. We believe that a number of factors played a role in the limited differences that we found between prosthetic knee conditions. First of all, we did not provide a training program when walking with the Rheo Knee II. As the majority of non-microprocessor controlled prosthetic knee do not allow early stance flexion, the majority of the participants learned to walk with an extended knee throughout the stance phase during their rehabilitation. It is doubtable whether participants were able to unlearn this walking pattern without training. As it is not common in the Netherlands to provide a training program after provision of a microprocessor-controlled prosthetic knee we chose to stick to usual care as closely as possible. Next to this, we had a small study population which affected statistical power which makes it harder to find statistically significant differences. However, if marked differences between prosthetic knee conditions were present, these could have been identified. Secondly, users had eight weeks to get accustomed to the Rheo Knee II which might have been too short for full customization.

Future research could focus on the added value of active prosthetic feet that might be able to shift the CoP to a more anterior position under the prosthetic foot. If so, individuals might be able to use the CoP-CoM mechanism by which braking forces are generated to a greater extent.

In conclusion, the Rheo Knee II does not reduce intact leg reliance and has no added value during gait termination when compared to non-microprocessor controlled prosthetic knees.

REFERENCES

[1] H. Stinus, "Biomechanik und beurteilung des mikroprozessor-gesteuerten exoprothesengelenkes C-Leg," *Z Orthop*, vol. 138, pp. 278-82, 2000.

[2] H. Herr, and A. Wilkenfeld, "User-adaptive control of a magnetorheological prosthetic knee," *Ind Robot*, vol. 30, pp. 42-55, 2003.

[3] G. K. Klute, J. S. Berge, M. S. Orendurff, R. M. Williams, and J. M. Czerniecki, "Prosthetic intervention effects on activity of lower-extremity amputees," *Arch Phys Med Rehabil*, vol. 87, pp.717-722, 2006.

[4] E. C. Prinsen, M .J. Nederhand, H. F. J. M. Koopman, and J. S. Rietman, "Added value of a user-adaptive prosthetic knee in gait initiation: off to a good start?" *IEEE Int Conf Rehabil Robot 2015* pp. 31-36.

[5] A. H. Vrieling, H. G. van Keeken, T. Schoppen, B. Otten, J. P. K. Halbertsma, A. L. Hof, and K. Postema, "Gait termination in lower limb amputees," *Gait Posture*, vol. 27, pp. 82-90, 2008.

[6] H. G. van Keeken, A. H. Vrieling, A. L. Hof, K. Postema, and B. Otten, "Controlling horizontal deceleration during gait termination in transfemoral amputees: measurements and simulations," *Med Eng Phys*, vol. 35, pp. 583-590. 2012.

[7] Y. Jian, D. A. Winter, M. G. Ishac, and L. Gilchrist, "Trajectory of the body COG and COP during initiation and termination of gait," *Gait Posture*, vol. 1, pp. 9-22, 1993.

[8] J. Lynch, and D. G. E. Robertson (eds), "Biomechanics of planned gait termiation", *XXI Congress of the International Society of Biomechancis*, 2007, Taipei, Taiwain.

[9] A. L. Hof, M. G. J. Gazendam, and W. E. Sinke, "The condition of dynamic stability," *J Biomech*, vol. 38, pp. 1-8, 2005.

[10] Y. Hermodsson, C. Ekdahl, B. M. Persson, and G. Roxendal, "Standing balance in trans-tibial amputees following vascular disease or trauma: a comparative study with healthy subjects," *Prosthet Orthot Int*, vol. 18, pp. 150-158, 1994.