FACTA UNIVERSITATIS
Series: **Mechanical Engineering** Vol. 13, N° 3, 2015, pp. 193 - 203

Preliminary report

THE GAIT SIMULATOR FOR LOWER LIMB EXOPROSTHESES – OVERVIEW AND FIRST MEASUREMENTS FOR COMPARISON OF MICROPROCESSOR CONTROLLED KNEE JOINTS

UDC 621:617.3

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Abstract. *A test device for lower limb exoprostheses has been developed at the Department of Medical Engineering of the TU Berlin which is able to apply realistic loads to prostheses. Hence, the gait simulator meets the increasing demands on functional and fatigue testing of microprocessor controlled knee joints (MPK). An exemplary comparison of two MPK was performed to prove that known differences in the functional quality of the MPK can also be demonstrated in simulator tests. Significant differences between the MPK could be found. The MPK could not be tested in their full range of function though. To enable comprehensive functional and fatigue testing, the gait simulator has to be modified to achieve higher walking velocities and step lengths.*

Key Words: *Amputee, Transfemoral, Knee Joint, Motion Analysis, Gait Simulator*

1. INTRODUCTION

The gait simulator, a test device for lower limb exoprostheses, is developed at Berlin Institute of Technology to fulfill the requirements for comprehensive functional and fatigue testing of modern prosthetic knee joints, especially microprocessor controlled knees (MPK). Technological advances in exoprosthetics lead to a steady increase in the mobility of amputees. The resulting and partially complementary demands on the prostheses concerning low weight, high functional capability, and improved strength result in complex designs which are strongly optimized to their field of application. Due to that, it is important to test the prostheses under realistic conditions. The current standards DIN EN ISO 10328 and DIN EN ISO 22675 are not able to address these requirements as the amount of loading and the number of cycles does not comply with field studies [1]. In contrast, the simulator is able to apply real time series of multi-axis

Received October 12, 2015 / Accepted November 5, 2015
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loads. Furthermore, it is possible to study prostheses with a high reproducibility from an objective point of view. Especially the influence of subjects concerning inter-individual and day-to-day variability as well as accommodation time on different prostheses set ups can be eliminated. Particularly, the simulator is suited for measurements which are usually not possible due to ethical restrictions or excessive stress for the subjects, such as simulations of falls and stumbles. In this study, the simulator was used to examine differences between C-Leg 3[1] and Plié 2.0[2] to prove its abilities for functional testing of MPK.

2. Methods and Materials

The gait simulator is based on two kinematic chains (hip module A and foot module B, Fig. 1), which are linked by the leg prosthesis during stance phase and separated during swing phase [2, 3]. The simulator is designed to apply loads in a realistic manner using five servo-hydraulic actuators with volume-controlled servo valves for precise load application. The hip module consists of three serially coupled actuators for the movements of flexion/extension (A1), of adduction/abduction (A2) and of inversion/ eversion (A3). For all drives, moments are captured by integrated strain gauge based

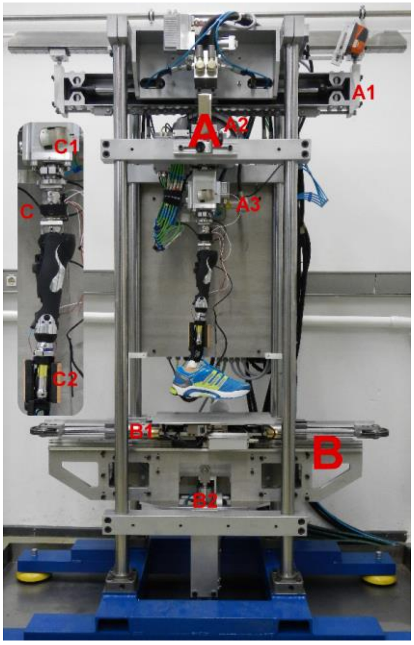

Fig. 1 Gait simulator with integrated Oktapod measuring system,
A: hip module, B: foot module, C: Oktapod

[1] Provider: Otto Bock HealthCare GmbH, Max-Näder-Str. 15, 37115 Duderstadt, Germany
[2] Provider: Freedom Innovations, 30 Fairbanks, Ste. 114, Irvine, CA 92618, USA

measuring devices. The foot module consists of two serially coupled actuators for vertical (B1) and horizontal movement (B2) of the instrumented foot plate which records ground reaction forces. The kinematic chains are coupled within the stance phase by the leg-prostheses. An additional six DOF force and moment sensor [1] (Oktapod measuring system C with data logger C1 and Li-Ion battery C2) was integrated in the prosthesis to obtain reference data for comparison of actual and nominal values of the simulator control.

2.2. Control of the simulator

The control of the gait simulator has a robust design and enables a quick setup of new testing scenarios. The required test data are either directly collected during measurements with subjects or calculated from test records.

Nine independent closed loop controls with PD controllers are used to control the five actuators of the simulator. During the swing phase all the actuators are position controlled. During the stance, a force/moment control is used. Only the actuator for extension and flexion movement is always position-controlled because of the need for correct knee angles for the reproduction of physiological loads during gait. The sequence control ensures correct switching between the force/moment-controlled stance phase and the position-controlled swing phase. For the correct transition between the two phases the sequence control is designed as a finite state machine with three states, which are processed in a defined order during simulation of a gait cycle. An overview of the states is given in Fig 2.

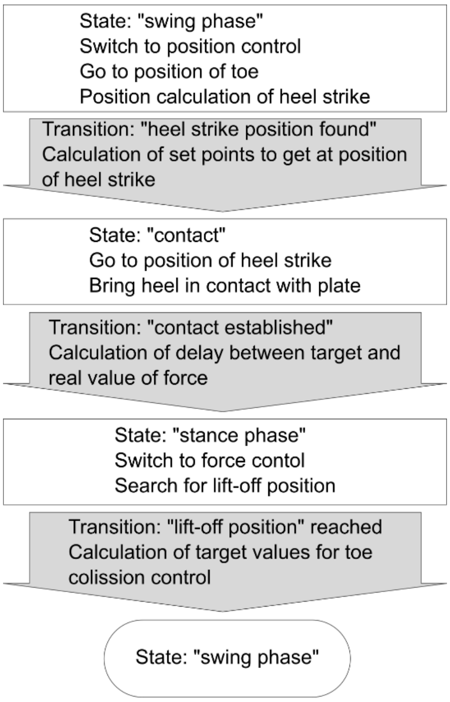

Fig. 2 Overview of the states of the finite state machine

The first state "swing phase" starts when the foot detaches from the base plate. In this state, a real-time analysis is performed to estimate the position of heel strike. If the calculation of the estimated position finishes, the next state "contact" is entered, where the heel and the base plate are brought into contact. After the initial contact, the ground reaction force rises. When a defined force is reached, the control switches to the state "stance phase" and back to the force/moment-controlled movement. This phase ends when the plate gets to the position "toe off" and the swing phase starts again.

Due to the coupling of the actuators during the stance phase, a movement of one actor also changes the measured forces and moments of the other control loops. That effect depends on the individual characteristics of the prostheses like geometry, flexibility and damping. Therefore, a decoupling is implemented in the two most affected control loops.

2.2.1. Decoupling of the force control of the vertical foot actuator from the flexion/extension angle

For decoupling of the vertical force from the flexion/extension angle, the compliance of the prostheses is needed. It is acquired by experimental examination and recording of so called foot-compliance-curves (Fig. 3). To that end, the position of the foot during a constant vertical force and variable shank angle is determined. The resulting field of curves is used to calculate the variable compliance of the system during a single step (Fig. 3, purple trace).

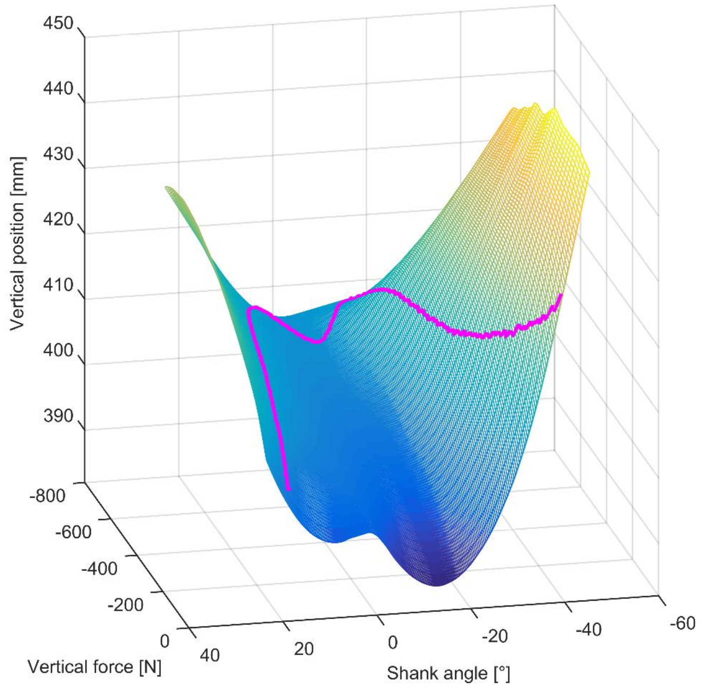

Fig. 3 Resulting field of curves after examination of the foot compliance

2.2.2. Decoupling of the moment control of horizontal foot actuator from flexion/extension angle

The compliance of the prosthesis in horizontal direction is smaller than in vertical direction. Hence, only the geometry of the leg prostheses is needed for decoupling of the sagittal moment from the flexion/extension angle. Additionally, the control loop is supported by an iterative learning pre-control (Fig. 4), which adjusts manipulated variable Y after a few runs of target values w_t to minimize the control deviation.

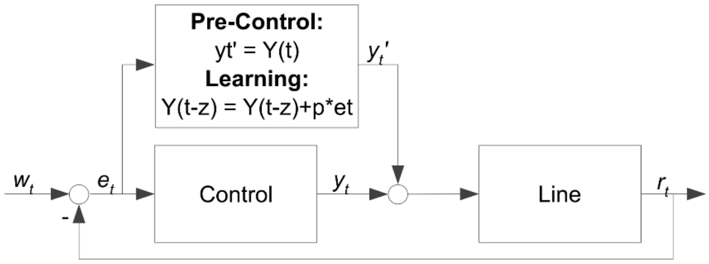

Fig. 4 Scheme of the closed-loop control with the self-learning pre-control

2.3 Prosthesis

To determine whether the current development level allows the use of the gait simulator to identify differences in the functional quality of MPK joints, measurements were first performed with the C-Leg 3 (Ottobock Healthcare) and the Plié 2.0 (Freedom Innovations) only. These joints were selected on the basis of the results of a gait lab analysis, which demonstrated that the quality of the swing phase damping between these two joints differed significantly [4].

2.4. Test procedure

For controlling of the simulator, kinematic and kinetic reference data were used, which were recorded by the Oktapod six DOF force and moment sensor system [4]. Via the servo-hydraulic drives of the simulator, movements were induced and forces and moments were applied simultaneously. Thereby, the measured data were also recorded by means of the Oktapod measuring system, which was integrated in the prosthesis (Fig. 1).

A significant limitation of the set up was the maximum step length of the simulator. Due to constructive restrictions, it is limited to 0.8 meter. Acceleration and deceleration of the foot plate before the heel strike and after toe off additionally consume useable length of the foot module. Hence, the subjects were asked to walk at medium velocity with short steps, and at slow velocity with normal steps to obtain reference data for the simulator (Table 1). Thereby, the C-Leg 3 and the Plié2.0 knee joints were used in combination with the prosthetic foot VariFlex with Evo[3]. Prosthetic alignments and setup of the prostheses were conducted by a certified prosthetist. The same prostheses with identical knee joint settings were used in the gait lab and during simulator measurements.

[3] Provider: Össur hf., Grjothals 1-5, 110 Reykjavik, Iceland

Table 1 Walking velocity and double step length of the reference values

	Velocity [m/s]	Double step length [m]
Walking at slow velocity	0.69	1.11
Walking at medium velocity	0.97	1.23

3. RESULTS OF GAIT SIMULATOR MEASUREMENTS

3.1. Optimizations after the first measurements

The first measurement data acquired by the simulator show two main deviations from the target values. First, the movement of the shank generates a moment when the knee reaches its extended position at the end of the swing phase. The impact results in a high stop moment which leads to oscillations of the sagittal moment (Fig. 5, red line). Also, the shank is reflected at the extension stop resulting in an oscillation of the knee flexion angle. Second, the knee angle maximum produced by the simulator deviates from the reference values as seen in Fig. 6. Those two problems were improved by an optimization of the thigh angle. Based on the original angle curve, a fifth degree polynomial was fitted through four defined points, while the transitions between fitted and original curves were continuously differentiable (Fig. 7). Within those borders, the angular speed and acceleration could easily be manipulated. The results are shown in Fig. 5 and Fig. 6. The stop moment was decreased by half, the knee-angle maximum was raised by 12° and the reflection at the extension stop was nearly eliminated.

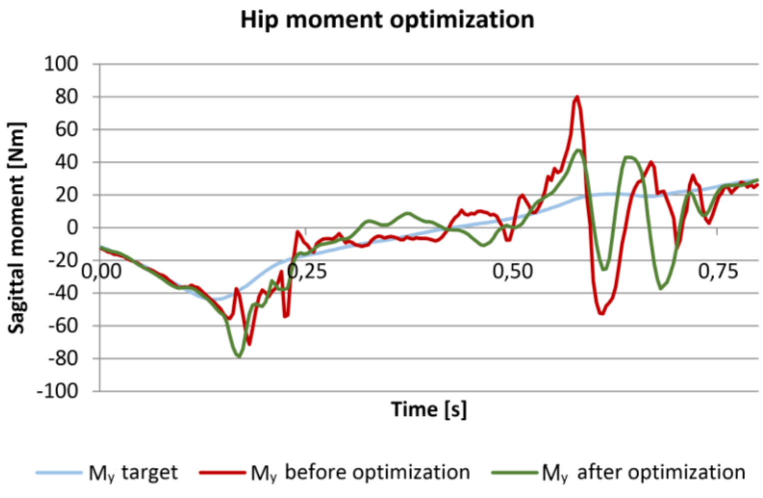

Fig. 5 Sagittal hip moment (M_y) before and after optimization of thigh angle

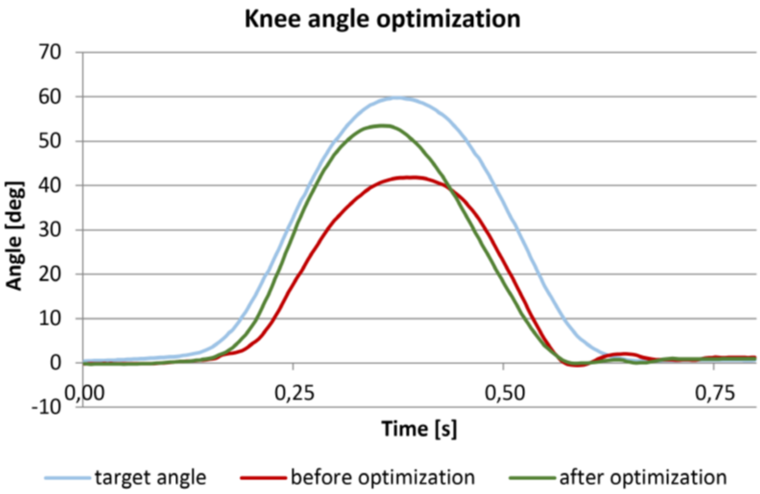

Fig. 6 Knee flexion angle before and after optimization of thigh angle

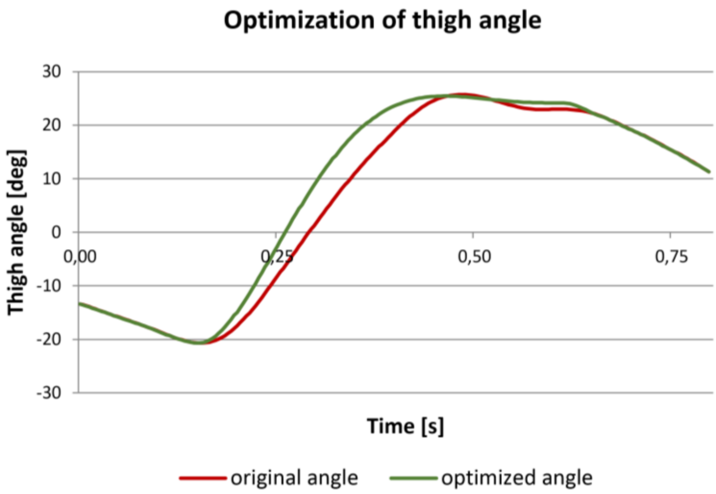

Fig. 7 Thigh angle before (red) and after optimization (green)

3.2. Reproducibility of measurement

Successive steps of the gait simulator were recorded with the Oktapod measuring system and compared to each other. The mean correlation factor over all signals was calculated from the pairwise comparison of each step with a reference step. The results are presented in Table 2. The variation is slightly increased at higher velocities. However, the minimum of the correlation factor of 0.982 is still very high, showing very high reproducibility.

Table 2 Mean deviation between six successive steps of the gait simulator; mean correlation factor over all signals

	Mean correlation factor, slow velocity	Mean correlation factor, medium velocity
C-Leg	0.997	0.982
Plié2.0	0.994	0.986

3.3. Deviation between reference values and measured values

At first, a comparison of the nominal values for simulator control and the measured actual values was conducted to assess the performance of the simulator control. Exemplary results for Plié2.0 are presented in Fig. 8. In general, the curve characteristics of the forces and moments applied by the gait simulator show a high correspondence to the gait analysis data. However, in swing phase, the highest deviations could be found for the frontal and the sagittal hip moment. Also the movements recorded for the thigh and the shank differ significantly from the target values in swing phase, which causes the low maximum knee flexion angle. Remaining optimization potentials are described in the discussion section.

3.4. Comparison of the swing phase control of the two knee joints

Previous gait lab tests have shown that significant differences could be found for the swing phase control of different MPK [4]. For the investigated slow, comfortable and fast walking velocity, C-Leg 3 and Plié 2.0 showed the largest differences regarding to the maximum knee flexion angle. With Plié 2.0 a strong correlation between maximum and walking velocity with a high slope of the regression line of 28.1°/m/s was identified for level walking. In contrast, with C-Leg 3 the slope of 3.5°/m/s was very small. With able bodied subjects as well as with amputees on the contralateral side constant maximum knee flexion angles with different speeds of level walking ensure low energy consumption and sufficient toe clearance to avoid falls [4, 6-10]. During functional testing of the two MPK with the gait simulator, similar tendencies to the results of the gait lab tests could be found. The increase of the maximum knee flexion angle with rising walking velocity measured with the simulator was 37.8°/m/s with the Plié2.0 and 21.4°/m/s with the C-Leg 3 (Fig. 9). The higher slopes resulting from simulator tests could be related to the smaller velocities simulated on the testing machine compared to the gait lab tests.

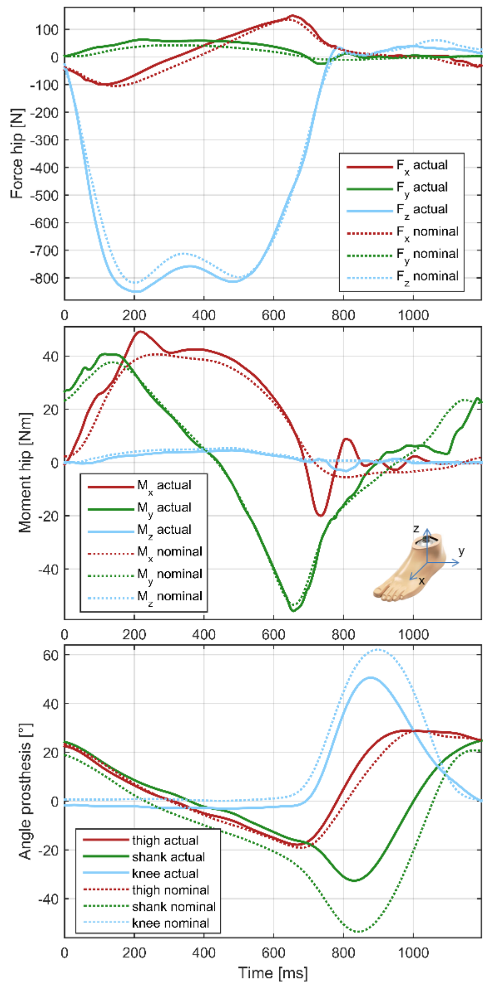

Fig. 8 Comparison of target and actual values measured with the simulator for Plié2.0, coordinate system located in the hip center with shown orientation

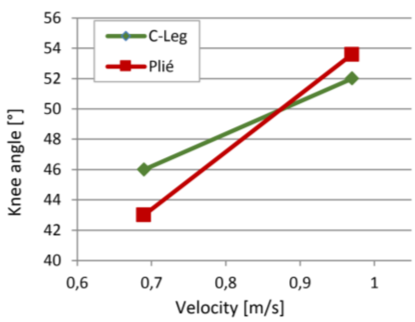

Fig. 9 Knee flexion angle [°] over walking velocity [m/s] measured with the simulator

4. DISCUSSION

The first measurements with the simulator showed a high reproducibility comparing successive gait cycles. In some parameters, there was a good concordance between the reference data used for control and the measured data. This could be observed for the values of the ground reaction forces and the hip moment in the sagittal plane, which represents the main functional plane of prosthetic knee joints. Thigh and shank angles and the resulting knee flexion angle showed higher deviations. The reason for the differences could be found in the missing translatory movement of the hip module during swing phase which also accelerates the shank.

Due to these restrictions, the prosthetic knee joints could not be tested within the optimum operation range of their swing phase control. This is also ascribed to the tested walking velocities and step lengths that were below the comfortable values measured in the gait lab tests. Both the tested knee joints are subject to this system-dependent influence as they are controlled by the gait simulator with the same reference values. For that reason, the differences between the knee joints in the shank angle and the resulting knee flexion angle can be attributed to different joint characteristics. Comparing the maximum of the knee flexion angles in the swing phase, the Plié2.0 has shown a higher dependence on the walking velocity than the C-Leg. Comparable results could also be found in the previous gait lab tests [4, 5, 11]. The results of this study suggest that the functional differences between knee joints can be shown with the gait simulator too. However, further research is needed to enable realistic test conditions.

5. OUTLOOK

The high reproducibility of the simulator tests may enable the examination of the influence of changes in the setup of the prosthesis. Thus, the interaction of different prosthetic components may be investigated, like the influence of different prosthetic feet used with the same MPK. The measured foot-compliance-curves already enable characterizations of prosthetic feet. To examine the interaction with the MPK, further measurements with different prosthetic components are necessary.

Furthermore, the optimization of the gait simulator should focus on increasing the achievable walking velocities and step lengths to allow testing of the prosthetic knee joints under conditions that correspond to everyday usage. To extend the scope of the investigations, the simulation of other gait situations in addition to level walking is planned. Especially the tests providing information on the potential of a prosthetic knee joint to prevent falls could be safely conducted with the simulator. In the future an optimized gait simulator could provide the basis for comprehensive and reproducible functional testing of prosthetic components.

Acknowledgements: *The paper is a part of the research done within the project BeMobil (FKZ 16SV7069K), founded by the Federal Ministry of Education and Research Germany. The author want to thank Wulf Wulff for designing and developing the gait simulator during his PhD project at the TU Berlin. The knee joints were provided by courtesy by Otto Bock Healthcare GmbH.*

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