

Design of An Expert System to Automatically Calibrate Impedance Control for Powered Knee Prostheses

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Abstract—Many currently available powered knee prostheses (PKP) use finite state impedance control to operate a prosthetic knee joint. The desired impedance values were usually manually calibrated with trial-and-error in order to enable near-normal walking pattern. However, such a manual approach is inaccurate, time consuming, and impractical. This paper aimed to design an expert system that can tune the control impedance for powered knee prostheses automatically and quickly. The expert system was designed based on fuzzy logic inference (FLI) to match the desired knee motion and gait timing while walking. The developed system was validated on an able-bodied subject wearing a powered prosthesis. Preliminary experimental results demonstrated that the developed expert system can converge the user's knee profile and gait timing to the desired values within 2 minutes. Additionally, after the auto-tuning procedure, the user produced more symmetrical gait. These preliminary results indicate the promise of the designed expert system for quick and accuracy impedance calibration, which can significantly improve the practical value of powered lower limb prosthesis. Continuous engineering efforts are still needed to determine the calibration objectives and validate the expert system.

Keywords—powered knee prosthesis; impedance control; auto-tuning; fuzzy controller

I. INTRODUCTION

Transfemoral amputees see hope to regain natural locomotion through recent developments of powered knee prosthesis [1-6]. Majority of these experimental system relied on finite state impedance control, which adjusts the impedance of knee joints based on gait phases. To ensure high performance of the controller, parameters of the controller, especially the desired impedance in each gait phase, had to be fine-tuned for each subject due to inter-subject variation. Currently the related tuning procedures (usually through trial and error) were conducted based on gait analysis and responses of patients [5]. Our previous experience showed that the parameter tuning procedures were time consuming even under the direction of experienced experimenters. Any effort to automate and shorten the tuning procedures will have positive clinical impacts.

Three approaches have been explored to shorten the impedance calibration procedure for lower limb prostheses: (1) improving experimenters' skill through training, (2) decreasing the number of parameters needed for tuning, and (3) developing automatic calibration approaches. Most of leading manufactures use the first approach and develop their own training programs and graphic user interfaces (GUIs) to help prosthetists to calibrate prostheses control parameters for individual patients. Although the training programs for prosthetists and patients work well when the number of adjustable parameters is limited, they are not expected to be efficient solutions if the number of adjustable parameters increases. For the second approach, the commonly used method to reduce the number of tuning parameters is modeling. The basic idea is to use measurable or known parameters to estimate some of unknown impedance values through biomechanical models. However, the success of using biomechanical modeling to simplify the calibration procedure has been limited. For example, Eilenberg and Herr [7] proposed control logic based on neuromuscular model and successfully reduced the number of tuning parameters for an ankle-foot prosthesis. Currently, this patented technology is only applied on transtibial prostheses. Actually, modeling human lower limb joint impedance during dynamic movements, such as walking, is inevitably difficult. This is because human joint impedance varies with angular position [8-9], neural activation [10], and externally applied torque [11]. In addition, an amputee-prosthesis integrated multibody system is very complex. Furthermore, measuring human joint impedance in dynamic movements, such as walking on uneven terrain, is experimentally difficult and expensive, which makes the validation of joint dynamic model challenging. The third reported approach is to calibrate the prosthesis control parameters using online learning or rules. Blaya and Herr realized automated parameter tuning for an active ankle-foot orthosis [12]. Impedance parameters in two gait phases were adjusted to maintain a constant frequency of slaps and avoid orthosis oscillation. Although foot drops were avoided, the event (slap and oscillation)-driven calibration rules had to rely on predefined events which can only be triggered or avoided if the impedance parameters were at their targeted values. An auto-tuning system helped a semi-active knee prosthesis to gain the capacity to adapt to different walking speed [13]. Success of this system on tuning the joint damping ratio relied

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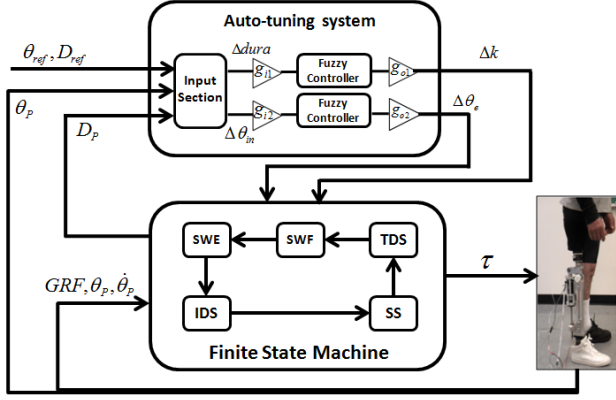


Fig. 1. The architecture of powered prosthesis control with auto-tuning system. The finite-state machine consisted of five states: IDS, SS, TDS, SWF, and SWE. θ_{ref} and D_{ref} represented the reference knee angle and reference phase duration, respectively; while θ_p and D_p represented PKP knee angle and phase duration, respectively. $\dot{\theta}_p$ represented the PKP knee angular velocity.

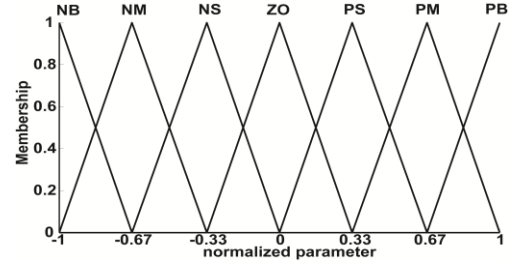
on a linear relationship between damping and walking speed. This type of relationship is hard to establish and validate when more than one adjustable impedance values exist. Despite the success of auto-tuning approaches in semi-active knee prosthesis and ankle-foot orthoses and prosthesis, research effort for developing automated impedance calibration system has been very limited for PKP, partly due to the lack of the precise relationship between the impedance parameters and certain measurements from PKP.

In this paper, we presented a new expert system based on fuzzy logics to automatically calibrate the knee impedance for finite state impedance control of a PKP. Fuzzy logic inference was selected because it has the advantages of emulating the behavior of the experienced expert (e.g. experimenters or prosthetists) and of dealing with the uncertainty in the system. In addition, the fuzzy logic system avoids the requirements of previous clinical knowledge to formulate the precise relationship between the inputs and outputs. The results of this study may pave a new way for efficient impedance calibration for PKPs and advance the practical value of PKP for daily use.

II. METHODS

A. Design and Control of a Powered Knee Prosthesis

In our group, we designed a prototype of PKP [14]. A moment arm and a pylon were used to construct the knee joint. The joint angle was driven by a DC motor (Maxon, Switzerland) through a ball screw. A potentiometer was instrumented on the knee joint to measure the knee joint angle, and an encoder was connected with the motor to obtain the knee joint angular velocity. A 6 degrees of freedom load cell (ATI, NC) was mounted on the pylon to measure the ground reaction force. The designed powered prosthesis was tethered and controlled by a desktop PC. A multi-functional DAQ card (National Instruments, TX) collected all the sensor measurements at 100Hz. It also provided a D/A for control output to drive the DC motor through a motor controller (Maxon, Switzerland).



If Δd_{ura} is NB	Then Δk is PB	If $\Delta \theta_{in}$ is NB	Then $\Delta \theta_e$ is NB
If Δd_{ura} is NM	Then Δk is PM	If $\Delta \theta_{in}$ is NM	Then $\Delta \theta_e$ is NM
If Δd_{ura} is NS	Then Δk is PS	If $\Delta \theta_{in}$ is NS	Then $\Delta \theta_e$ is NS
If Δd_{ura} is ZO	Then Δk is ZO	If $\Delta \theta_{in}$ is ZO	Then $\Delta \theta_e$ is ZO
If Δd_{ura} is PS	Then Δk is NS	If $\Delta \theta_{in}$ is PS	Then $\Delta \theta_e$ is PS
If Δd_{ura} is PM	Then Δk is NM	If $\Delta \theta_{in}$ is PM	Then $\Delta \theta_e$ is PM
If Δd_{ura} is PB	Then Δk is NB	If $\Delta \theta_{in}$ is PB	Then $\Delta \theta_e$ is PB

Fig. 2. Membership function for the normalized parameter and the linguistic rules. Note: NB, NM, NS, ZO, PS, PM, and PB are negative big, negative middle, negative small, zero, positive small, positive middle, and positive big, respectively.

The control of PKP was based on finite-state impedance control. For level ground walking mode, the impedance controller consisted of five states (gait phases): initial double support (IDS), single support (SS), terminal double limb support (TDS), swing flexion (SWF), and swing extension (SWE). The definitions of the phases were based on the gait phases in [15] (IDS included initial contact and loading response; SS included mid stance and terminal stance; TDS was the same as pre swing; SWF was the same as initial swing; SWE included mid swing and terminal swing). The transitions between the states in finite state machine (FSM) were triggered by the ground reaction force (GRF), knee joint angle, and knee joint angular velocity measured from the prosthesis [5-6].

In each state, desired prosthesis joint impedances were defined to mimic the knee impedance characteristic in healthy subjects. The appropriate joint torques τ was calculated based on (1) and generated by the motor.

$$\tau = k(\theta_p - \theta_e) + d\dot{\theta}_p \quad (1)$$

, where k , θ_e , and d denoted the linear stiffness, equilibrium position, and damping coefficient; θ_p and $\dot{\theta}_p$ represented measured knee joint angle and angular velocity, respectively.

B. Architecture of the Auto-tuning System

Fig.1 showed the architecture of the powered prosthesis control with auto-tuning system. Walking knee profile of normal people[15] was used as the reference of the system. The inputs θ_p and D_p of the system were the measured PKP knee angle and phase duration obtained from the FSM, respectively. They were sent into the input section to extract two features. The first feature, Δd_{ura} , was the difference of phase duration (percentage of gait cycle) between the measured PKP phase duration and reference. The second feature, $\Delta \theta_{in}$, was the difference of specific targeted knee angular parameter between PKP knee profile and reference. The targeted knee

angular parameter was different for each phase: maximum stance flexion angle for IDS; maximum stance extension angle for SS; maximum swing flexion angle for SWF; maximum swing extension angle for SWE [13]. The reference values used to calculate the two features were obtained from the normal gait knee profile [15]. Currently the system only tuned stiffness and equilibrium position and a small constant value was set to the damping coefficient using previous manually tuning result. The ignorance of damping tuning was based on the facts that the damping coefficient was very low at knee joints for healthy subjects[5]. Since some damping already existed at the PKP joint, we didn't tune the damping coefficient in the system.

In the fuzzy controller, the scaling gains g_{i1} , g_{i2} , g_{o1} , and g_{o2} (Fig.1) were employed to normalize the inputs (duration error $\Delta dura$ and angle error $\Delta\theta_{in}$) and outputs (stiffness adjustment Δk and equilibrium position adjustment $\Delta\theta_e$), and map them to [-1, 1] [16]. In this study, the scaling gains were determined by monotonically increasing and decreasing the impedance parameters until PKP reached its physical constraints (e.g. maximum flexion angle and maximum extension angle). The normalized inputs were then sent to the fuzzy controller. In this study, the fuzzy controller based on the Mamdani implication function and the mean of maxima defuzzification method was used, because of the capability to produce a faster convergence to the setpoint and a smoother output [17-18]. For the normalized inputs and outputs, seven triangle membership functions were used, distributed evenly with 50% overlap. Membership functions were given in Fig.2.

Based on the previous study on our PKP, we simplified the tuning rules as: phase duration decreases with the increase of stiffness, while target knee angle increases with the increase of equilibrium position. So the error of phase duration between reference knee profile and PKP knee profile was used for the tuning of stiffness. The error of the knee angle parameter was used for the tuning of equilibrium position. The linguistic rules were given in Fig.2. The tuning procedure terminated when the difference between PKP knee profile and reference knee profile was within one standard deviation of the reference profile[19].

C. Participants and Experiments

This study was approved by Institutional Review Board (IRB) of the University of Rhode Island and with informed consent of the subject. One male able-bodied (AB) subject, free from orthopedic or neurological pathologies, participated in this study. The AB subject was 24 year-old, 181 cm, and 83 kg. A special adaptor was designed for him so that the subject could walk with the powered prosthesis. The subject had received 10 hours of treadmill walking training with the PKP prior to the experiment. The appropriate impedance parameters for this AB subject were manually calibrated by an experienced experimenter.

During the experiments, the subject walked on the treadmill at a speed of 0.6 m/s wearing the PKP. The initial impedance parameters were randomly chosen at the beginning of each trial. The auto-tuning system adjusted the impedance

parameters every five steps. Each trial lasted for 2 minutes and totally five trials were conducted. An insole pressure measurement system (Novel Electronics, Germany) was used to measure the GRF under both feet to evaluate the gait symmetry. The GRF signal was sampled at 100 Hz and synchronized with the measurements from powered knee prosthesis.

D. Evaluation

In each trial, the normalized PKP knee profiles before and after tuning were compared with the reference knee profile. The PKP knee profile before tuning was average from five continuous gait cycles when the subject walked with the initial impedance parameters, and the PKP knee profile after tuning was averaged from five continuous gait cycles at the end of the trial. For the phase duration parameters, we calculated the error ratio(ER) based on

$$ER = \frac{D_r - D_p}{D_r} \quad (2)$$

, where D_r represented the reference phase duration value, D_p represented the measured PKP phase duration value. Zero error ratio means the PKP phase duration was identical to the reference value.

We also studied the effect of parameter tuning on the gait symmetry. Magnitude of the vertical GRF(vGRF), duration of stance and swing were obtained based on the GRF data. Symmetry index (SI) for these three parameters was calculated for each step[20].

$$SI = \frac{(I - P)}{0.5(I + P)} \quad (3)$$

, where I and P indicated the measured parameter from the intact limb and prosthesis limb, respectively. Zero SI value means the gait is symmetrical.

III. RESULTS

The auto tuning procedure helped the PKP to generate a knee profile closer to the reference one. Fig. 3 showed the PKP knee profiles before tuning and after tuning comparing with the reference knee profile. The dash black line was the PKP knee profile before tuning, the solid black line was the PKP knee profile after tuning, and the dash grey line was the reference

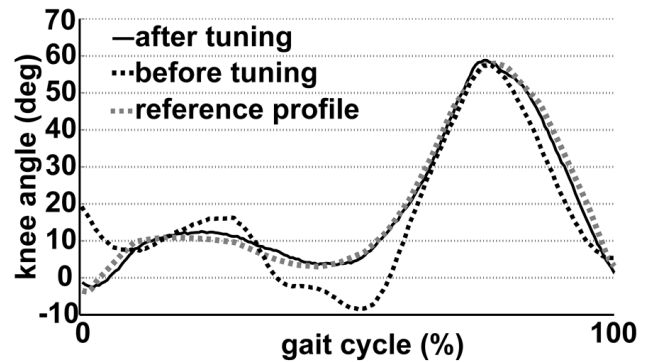


Fig. 3. The powered prosthesis knee profiles before tuning and after tuning from one trial. The dash black line was the PKP knee profile before tuning, the solid black line was the PKP knee profile after tuning, and the dash grey line was the reference knee profile.

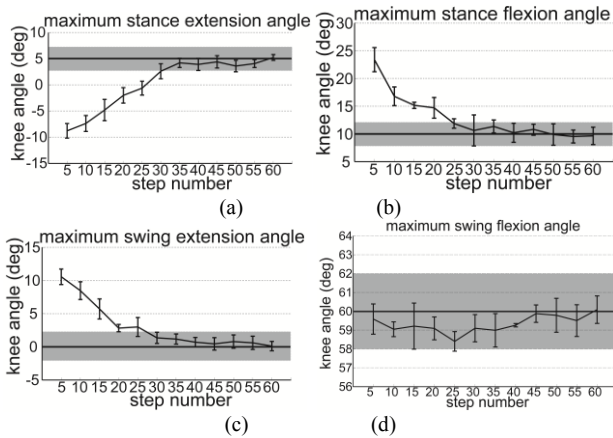


Fig. 4. The change of four target knee angles during the auto-tuning procedure from one trial. (a) maximum stance extension angle; (b) maximum stance flexion angle; (c) maximum swing extension angle; (d) maximum swing flexion angle. The black horizontal line in each figure indicated the reference value for each angle. The grey area indicates ± 2 deg (one standard deviation of the normal gait knee profile)

knee profile. It could be observed that after the tuning procedure, the PKP knee profile became smoother and closer to the reference one. The sum of square errors (SSE) between a normalized knee profile from PKP and the reference knee profile decreased by 95% during this tuning procedure. At the same time, both unnatural stance hyperextension and swing extension limitation under the initial impedance parameters were removed. The subject also reported a more comfort gait at the end of each trial.

While the subject was walking, the targeted knee angles converged to their reference values monotonically. The targeted knee angles from one trial were shown in Fig. 4. The targeted knee angles were averaged every five steps and its reference values were indicated by the horizontal solid black lines. The grey area indicated the ± 2 deg (one standard deviation of the kinematic patterns at knee joints estimated based on normal subjects [19]). The maximum swing extension angle (Fig.4a), maximum stance extension angle (Fig.4b), and maximum stance flexion angle (Fig.4c) all converged to their reference values in 35 steps. Due to the selection of initial parameters, the maximum swing flexion was very close to its reference value at the beginning of the trial. After the tuning procedure was finished, the impedance parameters were kept constantly. The subject walked with these impedance parameters for another 25 steps and the targeted knee angles were kept in ± 2 deg around their reference values. For the other trials, the tuning procedures were stopped at 45, 40, 45, and 40 steps.

Fig.5 demonstrated that the error ratio of phase duration decreased significantly after tuning in all phases. Here, these error ratios were averaged from the first five steps and last five steps across five trials. The lower absolute value of error ratio indicated that phase duration was closer to the reference value after auto tuning and the changes were statistically significant ($P < 0.05$).

Fig.6 showed the significant improvements of the gait symmetry indicated by smaller absolute value of averaged SI for vGRF, stance duration, and swing duration after auto tuning. The results were averaged from the first five steps and

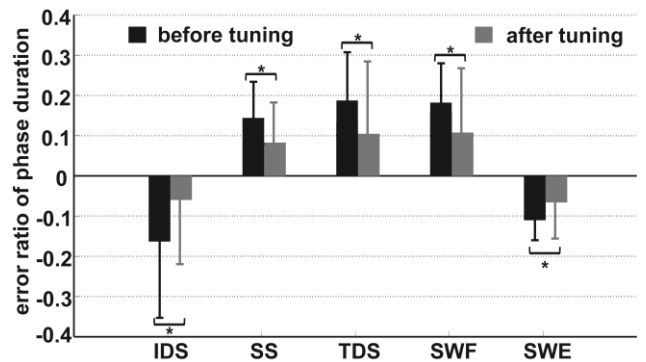


Fig. 5. The error ratio of phase duration before tuning and after tuning for each phase. * indicates that there are significant difference between the result ($P < 0.05$).

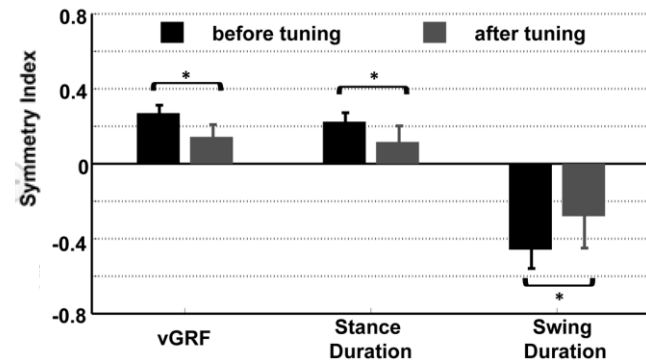


Fig. 6. The change of three SI before and after tuning. * indicates that there are significant difference between the result ($P < 0.05$).

last five steps across five trials. All the absolute values of these three SI decreased significantly ($P < 0.05$).

IV. DISCUSSION

The appropriate impedance parameters were important for the PKP users. Usually they are customized for individual users by experimenters or prosthetists using manual trial-and-error approaches. In this study, a new expert system based on fuzzy logic inference was designed to replace the experts and calibrate the control parameters automatically. Our preliminary results showed that the designed expert system can simplify and shorten the entire tuning procedure; therefore, our design can potentially reduce the healthcare cost of users for visiting prosthetists for prosthesis control calibration and increase the practical value of powered artificial legs.

One interesting finding of this study was that when the calibration objective for prosthetic knee was to match the desired knee profile as well as gait timing, obtained from healthy subjects, the tested subject presented improved gait symmetry (i.e. enhanced symmetry between the unimpaired leg and prosthetic leg) after the desired impedance was obtained and applied to PKP. The preliminary result derived from one subject indicated that our intuitive selection of calibration objective was sound. In addition, it implied that the user adapted to the dynamic change of lower limb prosthesis via human-prosthesis interactions and modified walking pattern on the unimpaired side of lower limb. This observation brought another interesting question: if the calibration objective involves the measurement of unimpaired legs or other body parts of the user, how is the calibration convergence affected?

This can be an interesting topic for our future study. We also observed that the stance/swing duration of both sound leg side and PKP side changed during the tuning of the impedance parameters. These observations can lead to further study about that during the calibration, whether the system is being adjusted to the user, or the user is being adapted to the device.

In this preliminary study, the expert system showed its capacity of generating a near-normal knee profile in powered prosthesis and improved gait symmetry of a PKP user by tuning the impedance parameters automatically. However, this study was preliminary and had many limitations. First, the modification of the knee impedance was conducted every 5 steps because we considered the adaptation of the user to the new impedance parameters. However, whether or not five steps were enough for adaptation was not clear. In the future, we will design a better protocol to determine the interval for impedance adjustment during calibration procedure. Secondly, our selection of calibration objective was based on the idea that prosthetic knee can replace the function of the missing knee joint. In the future, additional research efforts will be spent to investigate and understand how calibration goals affect the mobility and stability of human-prosthesis integrated multibody system. Thirdly, now we only studied the performance of PKP side during the calibration. The kinematic of sound leg would also be studied in the future. Additionally, this study only tested one AB subject wearing the powered prosthesis. It was important to validate the auto-tuning system on lower limb amputees and in different walking conditions.

V. CONCLUSION

This study introduced an expert system based on fuzzy logic inference that automatically tuned the desired impedance values for the impedance control of PKP. The system mimicked the tuning procedure of experimenters and automatically calibrated the impedance parameters to generate a near-normal knee profile. The system was tested on one AB subject wearing our designed PKP. The preliminary results showed this system could quickly and accurately find the appropriate impedance parameters to duplicate desired knee profile in several minutes. Compared with our previous manually tuning procedure, this method can significantly simplify and shorten the calibration procedure for PKP, which was a significant step forward to make the PKP more clinically friendly.

REFERENCES

[1] R.R. Torrealba, G. Fernandez-Lopez, and J.C. Grieco, "Towards the development of knee prostheses: review of current researches," *Kybernetes*, vol. 37, (no. 9-10), pp. 1561-1576, 2008.

[2] M.J. Highsmith, J.T. Kahle, S.L. Carey, D.J. Lura, R.V. Dubey, and W.S. Quillen, "Kinetic differences using a power knee and C-Leg while sitting down and standing up: a case report," *JPO: Journal of Prosthetics and Orthotics*, vol. 22, (no. 4), pp. 237-243, 2010.

[3] R. Borjian, J. Lim, M.B. Khamesee, and W. Melek, "The design of an intelligent mechanical active prosthetic knee," in Book *The design of an intelligent mechanical active prosthetic knee*, Series *The design of an intelligent mechanical active prosthetic knee*, Editor ed.^eds., City: IEEE, 2008, pp. 3016-3021.

[4] B.G.A. Lambrecht and H. Kazerooni, "Design of a semi-active knee prosthesis," in Book *Design of a semi-active knee prosthesis*, Series

Design of a semi-active knee prosthesis, Editor ed.^eds., City: IEEE, 2009, pp. 639-645.

[5] F. Sup, A. Bohara, and M. Goldfarb, "Design and Control of a Powered Transfemoral Prosthesis," *Int J Rob Res*, vol. 27, (no. 2), pp. 263-273, Feb 1 2008.

[6] E.C. Martinez-Vilalpando and H. Herr, "Agonist-antagonist active knee prosthesis: A preliminary study in level-ground walking," *J Rehabil Res Dev*, vol. 46, (no. 3), pp. 361-373, 2009.

[7] M.F. Eilenberg, H. Geyer, and H. Herr, "Control of a powered ankle-foot prosthesis based on a neuromuscular model," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on*, vol. 18, (no. 2), pp. 164-173, 2010.

[8] P.L. Weiss, R.E. Kearney, and I.W. Hunter, "Position dependence of ankle joint dynamics--II. Active mechanics," *J Biomech*, vol. 19, (no. 9), pp. 737-51, 1986.

[9] P.L. Weiss, R.E. Kearney, and I.W. Hunter, "Position dependence of ankle joint dynamics--I. Passive mechanics," *J Biomech*, vol. 19, (no. 9), pp. 727-35, 1986.

[10] M.M. Mirbagheri, H. Barbeau, and R.E. Kearney, "Intrinsic and reflex contributions to human ankle stiffness: variation with activation level and position," *Exp Brain Res*, vol. 135, (no. 4), pp. 423-36, Dec 2000.

[11] R.E. Kearney and I.W. Hunter, "System identification of human joint dynamics," *Crit Rev Biomed Eng*, vol. 18, (no. 1), pp. 55-87, 1990.

[12] J.A. Blaya and H. Herr, "Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait," *Ieee T Neur Sys Reh*, vol. 12, (no. 1), pp. 24-31, Mar 2004.

[13] H. Herr and A. Wilkenfeld, "User-adaptive control of a magnetorheological prosthetic knee," *Ind Robot*, vol. 30, (no. 1), pp. 42-55, 2003.

[14] M. Liu, P. Datsis, and H. Huang, "A Prototype for Smart Prosthetic Legs-Analysis and Mechanical Design," *Advanced Materials Research*, vol. 403, pp. 1999-2006, 2012.

[15] J. Edelstein, "Gait Analysis - Normal and Pathological Function - Perry,J," *J Rehabil Res Dev*, vol. 29, (no. 4), pp. 137-138, Fal 1992.

[16] J.R. Layne and K.M. Passino, "Fuzzy model reference learning control," in Book *Fuzzy model reference learning control, Series Fuzzy model reference learning control*, Editor ed.^eds., City: IEEE, 1992, pp. 686-691.

[17] E. Lembessis, "Dynamic learning behaviour of a rule-based self-organising controller," in Book *Dynamic learning behaviour of a rule-based self-organising controller, Series Dynamic learning behaviour of a rule-based self-organising controller*, Editor ed.^eds., City: Queen Mary, University of London, 1984.

[18] H.B. Kazemian, "The SOF-PID controller for the control of a MIMO robot arm," *Fuzzy Systems, IEEE Transactions on*, vol. 10, (no. 4), pp. 523-532, 2002.

[19] D.A. Winter, "Kinematic and kinetic patterns in human gait: variability and compensating effects," *Human Movement Science*, vol. 3, (no. 1), pp. 51-76, 1984.

[20] L. Nolan, A. Wit, K. Dudzinski, A. Lees, M. Lake, and M. Wychowski, "Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees," *Gait Posture*, vol. 17, (no. 2), pp. 142-151, Apr 2003.