

# Impedance Control of the Knee Mechanism of a Transfemoral Prosthetic for Level Walking

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## Introduction

Powered prosthetics have been shown to increase a user's walking efficiency and enable higher walking speeds in comparison to the non-powered counterparts. Yet, powered prosthetics suffer from lack of use owing to poor kinematic compatibility, bulkiness, and low adaptability to changes in the terrain. Additionally, current powered prosthetics mandate tedious calibration and tuning procedures, which prohibit their general usage. A possible approach to some of these problems is to focus on improving user-comfort by replicating able-bodied walking with such prosthetics. Many researchers have done this by adopting impedance control strategies. Under this control scheme, the gait is generally divided into 4-6 phases with a unique set of constant impedance control parameters—stiffness, damping, and equilibrium angles—assigned to each phase [1]. But studies such as [2] have revealed that a healthy human's joint impedance varies continuously throughout the stance phase of the gait cycle. Since the objective of a prosthetic's controller is to mimic able-bodied walking, it stands to reason that control parameters must also be human-inspired. Studies such as [3] have implemented algebraic curves within some phases of the control scheme. While the parameters vary continuously and smoothly within the phases, there are likely non-smooth variations during transitions between phases. Further, the controller proposed in [3] relies on a load cell that measures vertical ground reaction force. Such load cells can be expensive and increase the prosthetic's weight. A prior study by the authors proposes an impedance control scheme that varies the impedance parameters continuously and smoothly throughout the stance phase of the gait cycle [4]. Unlike [3], this scheme does not rely on a load cell. The impedance parameters are estimated using a least squares approach like that used in [1]. In the study, [4], the proposed controller was only implemented on a transfemoral prosthetic's ankle joint. This paper documents preliminary attempts at extending the control scheme to the prosthetic's knee joint.

## Methods

The proposed control scheme sections the gait cycle into 4 phases: heel strike (0%) to flat foot (13%), flat foot to heel off (42%), heel off to toe off (62%), and toe off to the end of the gait cycle (100%). The torque generated by the impedance controller is represented as follows.

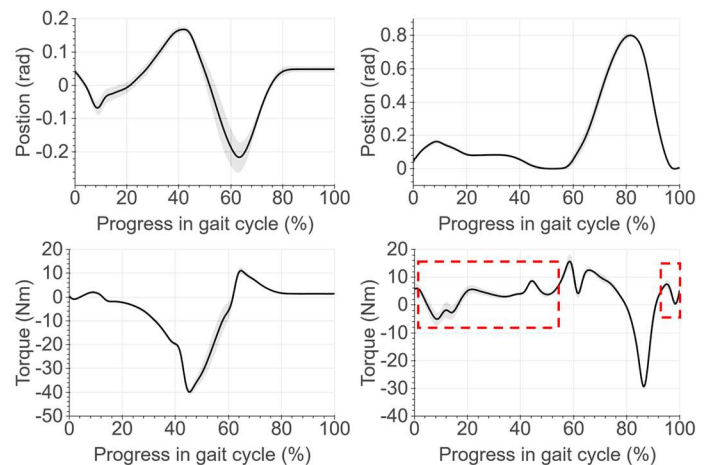
$$\tau(t) = K(t) (\theta(t) - \theta_{eq}(t)) + D(t)\dot{\theta}(t)$$

where  $K(t)$  and  $D(t)$  are the stiffness and damping parameter at the instant  $t$  ( $0\% \leq t \leq 100\%$ ). The term  $\theta_{eq}$  is the equilibrium angle, while  $\theta(t)$  and  $\dot{\theta}(t)$  represent the joint's position and velocity. Both  $K(t)$  and  $D(t)$  are represented by 4<sup>th</sup> order polynomials during the stance phase and a constant value during the swing phase. The polynomials were estimated using a least squares method detailed in [4] and were tuned by scaling and adding an offset. This controller was tested on a transfemoral prosthesis (AMPRO II [4]) with a healthy participant (male, 31 yrs., 1.70 m, 70 kg) using a L-shaped simulator. While the ankle was controlled using impedance control, the knee was controlled

using impedance control during the stance phase, followed by trajectory tracking during until 90% of the gait cycle, and low gain PD control during 90%-100% of the gait.

## Results and Discussion

The generated kinematics and kinetics have been presented in Figure 1. Both ankle and knee kinematics resemble able-bodied walking. The slightly lessened ankle dorsiflexion during mid-stance can be countered with further tuning. While the ankle torque also followed a humanlike trend, the peak torque is significantly lesser than that of healthy human's. This is due to the torque limitations of the prosthetic's actuators.



**Figure 1:** Top left: Ankle position, Top right: Knee position, Bottom left: Ankle torque, Bottom right: Knee torque.

Unlike the ankle, the knee torque deviated from that of a healthy human's. These deviations have been highlighted with red boxes in Figure 1. The first discrepancy is attributed to the controller's high damping parameters. It is believed that the results will improve by constraining the magnitude of damping. The second discrepancy is due to the low gain PD controller.

## Significance

In addition to generating near human-like gait, this controller requires lesser tuning. Additionally, the involved tuning process is easy to carry out [4]. With improvements, the proposed control scheme can undoubtedly make prosthetics easier to use for both amputees and therapists/practitioners. Future efforts are directed at eliminating the entire tuning process via auto-tuning methods such as fuzzy-logic.

## References

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