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Evaluation Of Impedance Control On A Powered Hip Exoskeleton

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EVALUATION OF IMPEDANCE CONTROL ON A POWERED HIP EXOSKELETON

A Thesis Presented

by

PUNITH K CONDOOR LAKSHMI

Submitted to the Graduate School of the
University of Massachusetts Amherst in partial fulfillment
of the requirements for the degree of

MASTER OF SCIENCE IN MECHANICAL ENGINEERING

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Mechanical and Industrial Engineering

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ABSTRACT

EVALUATION OF IMPEDANCE CONTROL ON A POWERED HIP EXOSKELETON

SEPTEMBER 2017

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This thesis presents an impedance control strategy for a novel powered hip exoskeleton designed to provide partial assistance and leverage the dynamics of human gait. The control strategy is based on impedance control and provides the user assistance as needed which is determined by the user's interaction with the exoskeleton. A series elastic element is used to drive the exoskeleton and measures the interaction torque between the user and the device. The device operates in two modes. Free mode is a low impedance state that attempts to provide no assistance. Assist mode increases the gains of the controller to provide assistance as needed. The device was tested on five healthy subjects, and the resulting assistive hip torque was evaluated to determine the ability of the controller to provide gait assistance. The device was evaluated at different speeds to assess the gait speed adaptation performance of the controller. Results show that hip torque assistance range was between 0.3 to 0.5 Nm/kg across the subjects, corresponding to 24% to 40% of the maximum hip torque requirements of healthy adults during walking. The peak power provided by the system is 35 W on average and a peak power of up to 45 W.

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CHAPTER 1

INTRODUCTION

The scope of the thesis project includes conceptualization and design of impedance-based control system for a novel design of a hip exoskeleton which incorporates series elastic actuation. It includes sensor integration, circuit design, necessary design alterations and functional improvements such as a shorter calibration routine and enhanced safety features. The research includes experimental setup, bench testing and control validation, human subject testing and data collection. The focus of the project was on refining and evaluating the controller performance, followed by performing a series of experimental protocols with five healthy test subjects.

1.1 Project overview

This thesis describes an impedance control strategy for powered hip exoskeleton for gait assistance. The goal is to enable the exoskeleton to provide appropriate assistance based on the user's interaction with the system. The powered hip exoskeleton used in the proposal is a single active degree of freedom that provides assistance in flexion and extension at the hip joint. The exoskeleton actuator is based on a scotch yoke mechanism [1]. Unlike most traditional rehabilitation exoskeletons that use transmissions that require reversal in the actuation direction every cycle, this mechanism aims to leverage human gait dynamics by converting the continuous rotary motion of the actuator to match the oscillatory motion of the human gait [1] [2]. Further, it incorporates a series elastic element in the design which increases the compliance of the system as well as allows measurement of interaction torque between user and exoskeleton [16]. The impedance-based control strategy has many advantages over a position based controller such as providing assistance while coping with variable gait profiles without the need to retune controller. Also, it does not require prerecorded gait data custom to the user to provide assistance. The ability of the controller to provide assistance based on user's requirement as well as its ability to cope with variations in speed will be determined by data analysis from human testing.

1.2 Research objective

The objective of this thesis is to develop and implement a control scheme on a powered hip exoskeleton that enables the user to complete their gait cycle with less effort with the assistance provided by the exoskeleton. Further, the exoskeleton must be able to provide assistance based on the user's requirements during their gait cycle. This would imply that the exoskeleton must be able to provide maximum torque during high torque requirement of user's gait cycle, and minimum torque during low torque requirement of the gait cycle.

An exoskeleton should not overpower the user and force motion against the user's intention or greatly change the user's gait kinematics. This would mean the user sets the reference gait trajectory and speed, and the exoskeleton would adjust its assistance during a gait cycle accordingly to match the user. This would enable the user to conveniently change walking speed without re-tuning the controller. Another major objective of the controller is to provide assistance to the user without the need to prerecord gait data. Prerecording gait data is time-consuming and limited in its application with regards to allowing the user to vary their gait kinematics and is less helpful with respect to the intended application of the exoskeleton. Additionally, the controller needs to detect the user's intention based on the interaction with the system and aid accordingly during the task of walking. This implies that the user should be able to slow down, speed up or completely stop walking at will.

1.3 Thesis outline

The thesis is organized as follows. The first chapter presents the motivation and objective of this thesis and introduces the exoskeleton used in this thesis. The second chapter discusses the background and prior work. Here impedance control is introduced, and its advantages and relevance for the intended application are discussed. Further, relevant commercially available exoskeletons are discussed along with a comparison of their control strategies with the control scheme being implemented in this thesis. Further, a brief description of the design of the exoskeleton in this thesis is discussed with its salient features highlighted. The third chapter discusses the control strategy being implemented on the exoskeleton, including a high impedance mode or an 'assist mode' and a low impedance 'free mode'. The fourth chapter discusses the testing and experimental protocols implemented to determine the functionality of the exoskeleton device. Further, the obtained data is plotted and presented for each test subject. The fifth chapter discusses inferences that may be made from the obtained data plots and conclusions

that may be drawn from it. Finally, possible future work that may help improve the device or the controller is discussed.

CHAPTER 2

BACKGROUND AND PRIOR WORK

Robot-assisted gait therapy has great potential to increase the rate of recovery and augment motor function for people with gait disorders [8]. However, it is assumed that a robot that assists the user when needed, as opposed to complete assist, which can lead to a faster rate of recovery through greater stimulation of muscle activity, leading to a faster recovery of motor function [15][14]. To achieve this assistance based on the user's interaction with the system an impedance control-based strategy is proposed to aid the user with their gait.

2.1 Lower limb exoskeletons

There are many different kinds of gait trainers available for a patient to undergo robot-assisted gait therapy. Each robotic gait trainer has a custom design and corresponding control scheme unique to its design. Furthermore, each control scheme may be directed at a particular subset of subjects targeting either different gait defects or different control methods to aid rehabilitation. The exoskeleton in this proposal has many similarities and differences as compared to other commercially available exoskeletons. This may be with respect to the mechanical design, actuation mechanism, power transmission, or control strategy employed. The gait trainers and exoskeletons used for comparison have been chosen based on differences in mechanical actuation, mobility, type of sensory feedback, intended application, an intended group of population and control strategies implemented.

Cyberdyne HAL (Figure 1.1) is a mobile whole-body exoskeleton with DC motors coupled to harmonic drives for actuation intended for both able-bodied and impaired individuals (such as paraplegic subjects or elderly) developed by researchers at Tsukuba and Cyberdyne Inc. It uses the users EMG signals and inverse dynamics to assist the user [3, 7, 18, 19, 20, 25].

The BLEEX (Figure 1.3) is a full body exoskeleton that uses hydraulic actuators to provide assistance to the user, intended for power augmentation function. It uses inverse dynamics to determine the torque to be provided at the joints based on joint angles and in foot pressure sensors to determine gait phase [27]. The ATLAS (Figure 1.4) is a full body exoskeleton providing actuation with DC motors and Harmonic drives. It has sensory feedback from goniometers and in foot pressure sensors and is impedance controlled with joint reference trajectories. It was intended to provide gait assistance to children with quadriplegia [30]. The Ekso by Bionics (Figure 1.5) is a lower limb exoskeleton intended for gait training of subjects with spinal cord injuries, hemiplegia or paraplegia, and other lower limb movement disorders. One uses crutches to aid in balance and is actuated by Hydraulic actuators and on-board battery pack. It uses in foot pressure sensors to determine gait phase and actuates based on changes to the user’s center of mass [29].



Figure 1.1: Cyberdyne HAL



Figure 1.2: ReWalk



Figure 1.3: BLEEX

The Lokomat (Figure 1.6) is a treadmill based gait trainer that is intended to provide rehabilitation to subjects with spinal cord injuries, stroke, neurological and orthopedic conditions that hinder them from walking normally. It uses a linear actuator to actuate the hip and knee joints, which are powered by an external power supply. High-resolution position sensors are used to measure the angle of the joints and force sensors are used to measure the interaction torques and forces between the user and the exoskeleton [13, 21, 22].

The Lopes (Figure 1.7) is also a treadmill based gait trainer similar to the Lokomat. This exoskeleton was intended for assessment of stroke subjects and for gait rehabilitation. It incorporates series elastic actuators and uses flexible Bowden cables for power transmission. The actuation is provided by off board servomotors powered by an external power supply. Force sensors were used to determine joint torques for a closed loop impedance control [6, 11, 15, 16, 24].



Figure 1.4: ATLAS



Figure 1.5: Bionics Ekso



Fig 1.6: LOKOMAT



Fig 1.7: LOPES

2.2 A reciprocating series elastic actuation for hip exoskeleton

Most gait trainers and exoskeleton systems use a direct drive mechanism for power transmission which would require actuators to reverse the direction of motion repeatedly for every gait cycle as a result of the human gait dynamics. This wastes energy as a result of changing actuator direction [2], and overcome its trailing inertial effects for every cycle. There is a need to customize and optimize designs to leverage the gait dynamics of a person. By designing a system with a variable transmission ratio that matches the torque profile makes the control of the exoskeleton more straightforward and more efficient.

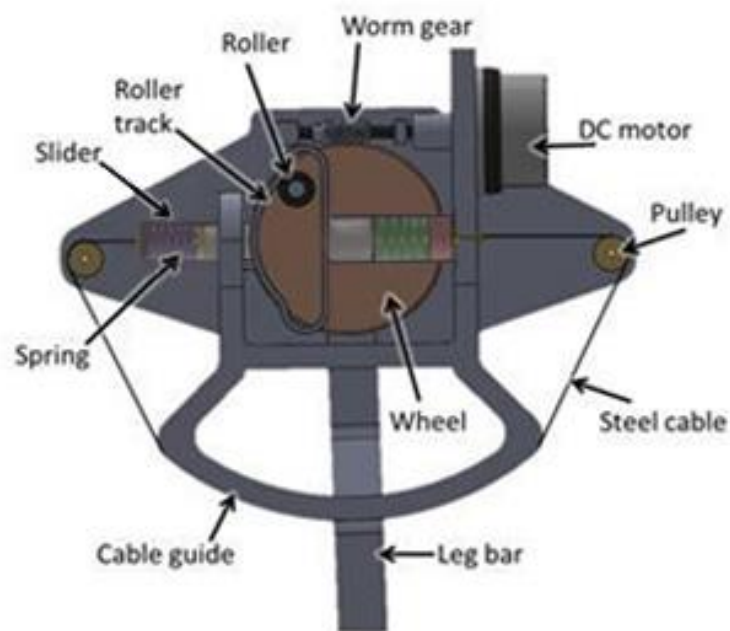


Figure 2.1 Exoskeleton to be controlled in this thesis

An exoskeleton was designed and developed with the objective of devising a mechanism that leverages gait dynamics of a human to provide a periodic oscillatory motion using continuous rotary motion [1]. The human gait cycle is periodic in nature, with the joint angles varying between a maximum and minimum angle. A scotch yoke mechanism is incorporated into the design to achieve this oscillating motion. A scotch yoke mechanism is a mechanism that converts rotational motion of a wheel into linear reciprocating motion or vice versa. It consists of a wheel of sufficiently large diameter with a fixed roller at a specified distance from the center acting like a crank about the axis of rotation. A slotted link with a slider is fixed to accommodate the roller with negligible sliding friction, within the groove and the slider and is fixed to allow only reciprocating horizontal translation.

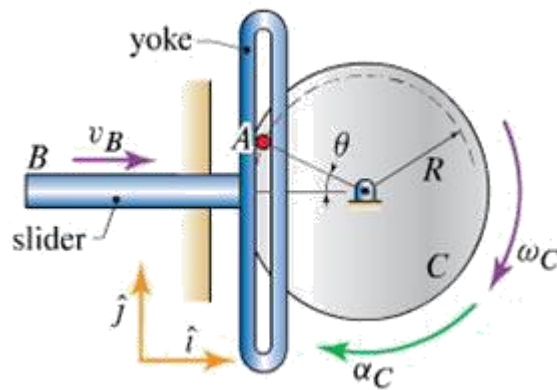


Figure 2.2 Scotch yoke mechanism

The advantage of using a scotch yoke mechanism is that actuator does not have to change the direction of rotation every cycle, provided the amplitude does not need to be changed [2]. As a result saving energy dissipation during instances where motor direction needs to be changed when a person switches between stance phase and swing phase. Further, a fixed speed of the actuator can provide variable torque output at the actuated joint.

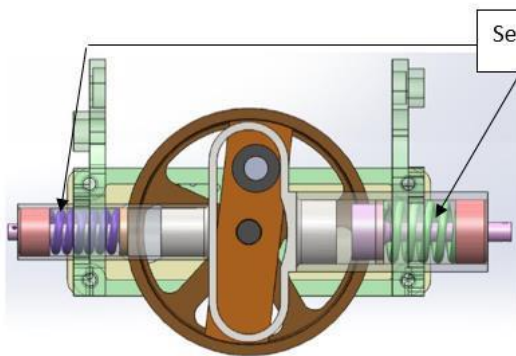


Figure 2.3 Series elastic components (springs)

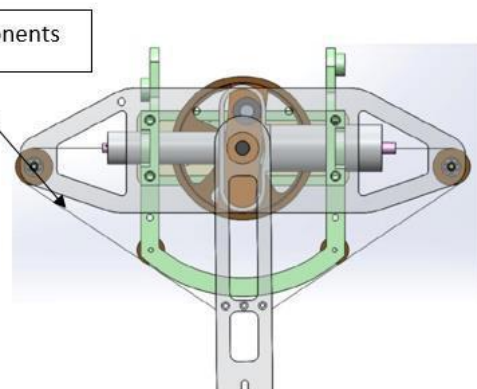


Figure 2.4 Series elastic components (Steel cable)

Another element of importance in the design is the incorporation of series elastic actuation to actuate the hip joint. This adds compliance to the entire system and makes it more comfortable for the user to interact with the exoskeleton. Series elastic actuation incorporate an elastic component such as a spring system through which the power transmission takes place between actuator and load. The series elastic components in the proposed hip exoskeleton are springs located inside the linear slider of the scotch yoke mechanism, along with the flexible steel cables used to transmit motion to the leg bar. Two springs of different spring stiffness (33000 N/m and 56000 N/m) are fixed at either end of the slider. This is in turn connected to steel cables which add to the series elastic effect before there is power transmission to the user. The spring elements in the design are used to determine the interaction torque between the leg bar and the user.

The interaction torque is inferred from the spring displacement measured with Hall effect sensors. Since the spring stiffness is constant, the spring force may be determined by considering the force-displacement relation of the spring to be linear. A combination of Hall effect sensors and magnets are used to determine spring displacement as shown in Figs. 2.5-2.8.

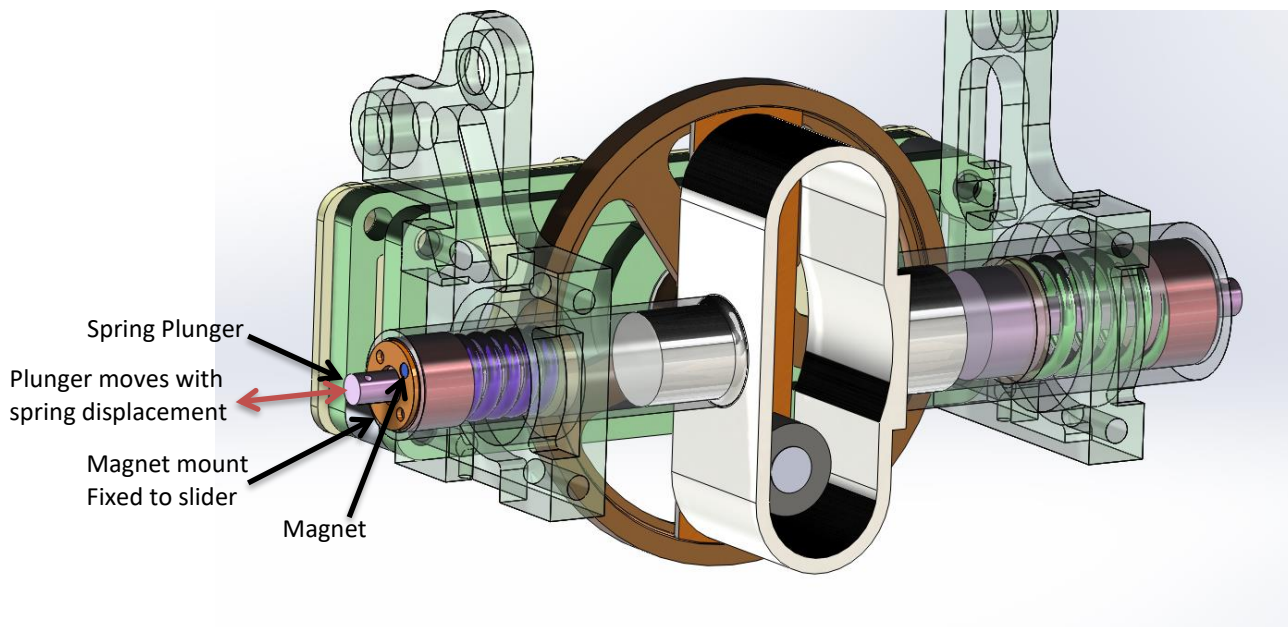


Figure 2.5 Magnet mount placement on device

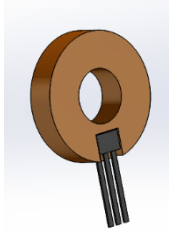


Fig 2.6 Hall effect sensor mount

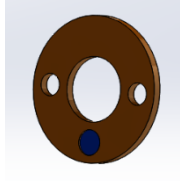


Fig 2.7 Magnet mount

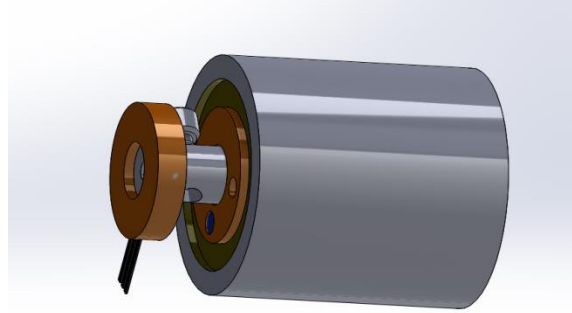


Fig 2.8 Hall effect sensor-magnet arrangement

The radial magnet and Hall effect sensor are press-fit into 3D printed mounts and fixed to the end of the slider mechanism. The magnet mount is fixed to the linear slider and hence moves along with the linear slider. The Hall effect sensor mount is press-fit on the spring plunger which moves according to the spring displacement. The relative displacement between the magnet and the Hall effect sensor determines the net spring displacement and as a result the interaction torque. The initial distance and range of the displacement are calibrated before testing, along with the change in cable tension with respect to linear slider position. The Hall effect sensor used is an Hall effect sensor (Allegro, MODEL A1324), with a sensitivity of 5 mV/G and which results in a resolution of approximately 0.68 μm corresponding to the spring displacement. This is calculated by calibrating the Hall sensor values with respect to the radial encoder values.

$$\text{Spring displacement}(mm) = \frac{\text{enc bits}}{8192} 2\pi h$$

Where enc bits corresponds to the change in number of bits of the radial encoder, h corresponds to leg bar length which is 4.5 inches or 114.3 mm. Hence if enc bits is 1, the spring displacement measured in mm is 0.0876 mm. Since the Hall effect sensor reads approximately 128 bits between 1 encoder bit recorded, the resolution of the Hall effect sensor can be approximated to 0.0876 mm divided by 128 which gives 0.00684 mm or 0.68 μm .

The leg bar on the exoskeleton connects to the user's leg and has a range of motion of 29 degrees which usually corresponds to 24 degrees flexion and 5 degrees extension. A salient feature of the leg bar is that it has a backlash region or a region of free play within which the exoskeleton cannot provide any force on the user's leg. This region is used to add compliance to the system and provides the user with a small region of free play. The

exoskeletons previously introduced will be compared and contrasted with the exoskeleton in this thesis on its mechanical design as well as its control approach.

2.3 Exoskeleton mechanical design comparison

The HAL by Cyberdyne, ReWalk by ARGO Medical Technologies, Berkeley lower extremity exoskeleton (BLEEX), and Bionics Ekso are all mobile exoskeletons. These exoskeletons provide the advantage of traveling from one place to another without being constrained to a single location. Treadmill gait trainers with body weight support such as Lopes, Lokomat and ReoAmbulator are intended for subjects with injuries such as spinal cord injuries which render them unable to support body weight. Regarding actuation, exoskeletons such as HAL, ReWalk and Lopes use DC motors to transmit power to the user. DC motors are light yet powerful and do not rely on bulky equipment such as compressors, or storage tanks. This is ideal for a mobile exoskeleton with no external support which needs to be as light as possible for the user. Most commercially available exoskeletons such as the HAL uses a direct transmission from the motors via harmonic drives [1]. The lower extremity HAL-3 powers the hip and the knee joint using separate motors with harmonic drive gear heads at the corresponding joints. Although this enables greater control of the joints, it requires the motors change direction with every stride to accommodate the person's gait. The ReWalk is similar in its force transmission with the HAL, which also depends on DC motors at each joint fitted with harmonic drive heads for force transmission to each of the joints [2].

The ATLAS is another exoskeleton that depends on harmonic drive transmission for limb actuation [30]. The exoskeleton in this thesis uses a scotch yoke mechanism, which converts the rotary motion of the actuator to linear reciprocating motion. This linear motion is then converted to a periodic oscillatory rotation, which is directly transmitted to the user's leg. This has the advantage of producing oscillating motion of the human hip with continuous rotation of the DC motor [3][4]. Exoskeletons like HAL, ATLAS, and ReWalk have direct force transmission to user's joints through a medium such as a harmonic drive. However, exoskeletons such as LOPES which incorporate series elastic elements have a greater advantage on user comfort. They make the actuation system compliant and more comfortable to the user, as well as adding an element of safety to the design. The LOPES uses springs and flexible Bowden cables connected to the user's legs, controlled by off board servomotors [11] [16]. Another advantage of using series elastic elements are a measurement of forces during operation. The LOPES uses springs to measure the forces in the joints [16]. In contrast, the HAL and Lokomat use force sensors at the joints to determine torques [1] [20], which add more expenses to the design.

2.4 Exoskeleton control comparison

The control of an exoskeleton is dependent on many factors such as the specific design unique to itself such as the type of actuator, sensors used, the user's needs and so on. Exoskeletons such as HAL and BLEEX are model-based controls where a predefined model controls input-output parameters. The input parameters such as angular position, or myoelectric signals, are fed into the model which estimates output parameters such as torque or force. The HAL is controlled by EMG sensors strategically placed on the user extracting information on the user's intent and estimating joint torques using inverse dynamics. It estimates the intended joint torque and amplifies it at the user's joint, and receives actual torque feedback from strain gauges [1]. The HAL implements impedance control [7], where reference torque is determined by myoelectric signals, and impedance control-based adjustments are incorporated based on joint encoder parameters. The BLEEX is also model based and determines joint torques using dynamic modeling with the help of force and motion sensors, and pressure sensors under the foot to determine the user's gait phase. The model of the system that determines joint torques is based on the corresponding gait phase of the user. Each joint is fitted with encoders and linear accelerometers that determine its angle, velocity, and acceleration and determines the required torque at each joint using inverse dynamics to minimize the error between actual and ideal torques [27].

The Lokomat has multiple control strategies implemented on it, the first mode being position control wherein the subject's legs are guided along a fixed reference trajectory. This trajectory is predetermined according to the user's physical characteristics, and the subject's interacting force and muscle activity are ignored [21]. This strategy may be implemented on users undergoing their initial stages of rehabilitation where complete assistance is required. This control has limitations with regards to varying speed and trajectory and hence was not selected for implementation on the exoskeleton in this thesis.

The second mode is the cooperative assist strategy which incorporates active force contribution to the user's gait by determining the user's torque along the gait and muscle activity and provides assistance accordingly. This is intended to force the subject's active participation in stimulating muscles, hence increasing the rate of rehabilitation. Under cooperative mode, there are two separate modes that stimulate the subject. The first is an adaptive control algorithm that adapts the reference gait pattern depending on the interaction forces between the user and the exoskeleton. The second is an impedance control-based strategy which determines the user's deviation

from reference gait trajectory and corrects their gait with a variable virtual spring damper element. Inverse dynamics is used to determine desired torques at joints and determine the effect of spring damper system [13].

This is very similar to the control scheme presented in this thesis, which uses impedance control. However, the difference is that the proposed control scheme in this thesis does not use a reference gait trajectory but relies on the user to set the gait trajectory. The LOPES has many aspects similar to the exoskeleton in this thesis. It incorporates impedance control which is dependent on the position sensing with force actuation to control the user's leg [11]. The LOPES can function with a zero impedance mode [11] where the subject walks with no assistance or resistance (normal walking). The LOPES also implements an assist mode [11]. It implements impedance control through a model known as "Virtual model control" [24] which uses the interaction forces between the user and the robot to determine the amount of assistive torque to be provided and provides virtual stiffness and damping based on gait phase and user's need. This concept is similar to that being proposed in this thesis, where the interaction of user and exoskeleton determines the level of assistive torque. However, VMC incorporates different levels of control based on gait phase and is not a pure impedance control, unlike the control method in the exoskeleton in this thesis.

CHAPTER 3

IMPEDANCE BASED CONTROL FOR HIP EXOSKELETON

The previous controller developed for the scotch-yoke hip exoskeleton was a position control strategy [3]. The exoskeleton was operated in a low impedance mode, and the user's hip joint angle data was recorded during gait and then played back to ensure user follows the reference trajectory in the assist mode. The position control scheme posed a limitation of gait profile variability requiring the user to follow a set reference trajectory at the recorded walking speed. The control scheme developed for this thesis is impedance control based which eliminates the need to record gait data prior to assist mode. An impedance-based control scheme enables the user to change walking pace at will without re-tuning or rerecording a sample gait dataset. This is because there is no set gait trajectory that the user must comply with, and controller simply reacts to the user's interaction with the device. This means that for a well-tuned controller, the device would provide corresponding torque required for different walking speed which in turn would correspond to different gait trajectory with respect to time. Another feature of the new control scheme is it enables the user to slow down or stop walking without an external switch. The position control scheme relied on an external switch to ensure exoskeleton ceases exerting a torque on user's leg. This is convenient and has an added safety element, avoiding unintended injuries to the user.

3.1 Impedance control

The basis of the control strategy employed on the exoskeleton being referred to in this thesis is impedance control. It helps achieve variable assist method and enables the user to control stride speed and reference gait profile while walking. Impedance control helps control the exoskeleton based on the user interaction with the system as well as aids compliant actuation as opposed to position control which relies on high precision and comparatively more rigid systems for effective control [15]. Impedance control treats the environment as the admittance and the robot system as impedance [5, 9, 26]. Impedance is the transfer function that takes in flow parameters such as position and its derivatives as input to the system and relates it to effort parameters such as force or torque [26].

$$\Sigma T = M(\theta)\dot{\theta} + H(\theta, \dot{\theta}) + V(\dot{\theta}) + G(\theta) \quad (1)$$

where θ denotes the angular position vectors for the actuator, T is the torque vectors acting on the human-robot system, $M(\theta)$ is a symmetrical and positive definite matrix that represents the inertial effects at joints, $H(\theta, \dot{\theta})$ is a square matrix that represents the Coriolis and centrifugal effects induced on joints, $V(\dot{\theta})$ denotes the diagonal matrix of viscous friction coefficients, $G(\theta)$ is a vector that represents the gravitational effect at joints.

$$\sum T = T_r + T_f \quad (2)$$

where T_r is the torque exerted by the robot, and T_f the torque exerted on the robot by the environment. The desired dynamic behavior of the robot is expressed as

$$T_r = M\ddot{\theta} + B\dot{\theta}_e + K\theta_e \quad (3)$$

where θ_e represents the angular error with respect to the reference angle, M is the inertial coefficient, B the damping coefficient and K the stiffness coefficient of the system. Neglecting the inertial effects of the system, results in

$$T_r = B\dot{\theta}_e + K\theta_e \quad (4)$$

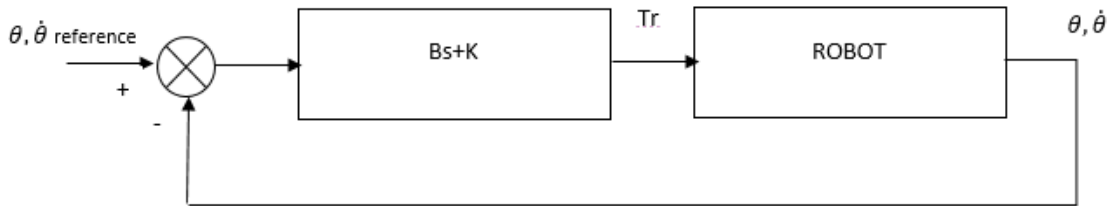


Figure 3.1: The basic impedance control framework

The output torques of the robot are based on the positional and velocity component of the system and respond based on the external interaction with the system. Impedance control has the advantage of controlling the system via compliant actuation. This implies that unlike positional or trajectory control of a robot which depends on the high output impedance of the system. An impedance controlled system can adjust its apparent stiffness and damping. Further, this control method can be used to estimate the user's intention and can help the user according to

their need. This is achieved through determining magnitude and direction of the displacement of the user's leg with respect to equilibrium position corresponding to zero interaction force between environment and system. Hence, this can help determine parameters for the impedance controller to provide the corresponding level of assistance based on the user's interaction with the exoskeleton.

The exoskeleton used in this thesis contains aspects that are unique in nature compared to other lower limb exoskeletons. The scotch yoke mechanism imposes a level of mechanical control on the system and has unique requirements of the electrical control system. The introduction of the backlash region or free play region is another factor that adds to the complication of the exoskeleton control. The series elastic components in the system is used to measure the interaction torque pose a limitation with respect to their measurement accuracy. A major limitation in the exoskeleton design is the fixed range of actuation of the leg bar. This angle of rotation cannot be changed while walking and forces the user's hip joint motion within this range. Any attempt to change the direction of rotation without completing a full rotation would force the motor to change directions, which would be counterproductive to the design consideration of the scotch yoke mechanism. Controlling this design to provide walking assistance to the user could result in reduced control bandwidth due to the optimized design. The effectiveness of this design in aiding the users was determined by data collected from human testing, i.e., the ability of the exoskeleton to provide assistance to the user without greatly altering gait pattern.

3.2 Controller Modes

The modes of control schemes implemented on the hip exoskeleton are:

- Low impedance- free mode
- High impedance- assist mode

The low impedance free mode does not allow the exoskeleton to assist or resist the user, but simply tries to track the user's movement. This mode is mainly used to determine the user's gait characteristics. The user is asked to walk with the exoskeleton in this mode, and the data is recorded. This data is then used as the subjects reference gait at a particular walking speed for further analysis and comparison with the subsequent data recorded for different modes.

The assist mode enables the user to walk while receiving torque assistance from the exoskeleton. The user's inability to give out sufficient torque to complete gait cycle is compensated with the help of the exoskeleton. The amount of assistance provided by the exoskeleton is determined by the user's interaction with the system. This mode ensures that the exoskeleton provides variable assistance to the user based on the user's gait phase. It incorporates a high stiffness impedance control where torque is provided based on user's interaction with the system indicated by displacement and speed.

3.3 Controller Design

Impedance control takes flow parameters corresponding to kinematics into account, such as velocity or acceleration and produces an effort corresponding to kinetics in terms of force or torque as the output [5]. This output may either be assistive where the direction of applied output torque is same as the direction of motion or resistive where the direction of applied output torque is in the opposing direction. This depends on the application and takes into account various factors such as stiffness, inertia or damping of the system. This simplified form of impedance control varies stiffness and damping and ignores inertia as follows

$$T_r = B\dot{\theta}_e + K\theta_e \quad (4)$$

where T_r is the torque exerted by the exoskeleton, θ_e represents the angular error with respect to the reference angle, K is the stiffness coefficient, and B the damping coefficient [5].

A PD (Proportional-Derivative) controller is implemented to adjust the stiffness component of the control. The gains of the PD controller contribute to the magnitude or level of assistance to be provided by the controller by adjusting the virtual stiffness of the system. The damping parameter is adjusted by the damping coefficient B which provides added stability to the system ensuring safety to the user. It takes in the angular velocity of the leg bar ($\dot{\theta}$) and reduces the output torque based on this value hence ensuring system does not continuously increase output torque, making system go out of bounds.

A moving average filter is used to remove high-frequency output and ensures a smoother transition between low and high torque output. This filter further adds to the stability of the system at the cost of decreasing the bandwidth of the actuator. Further, to ensure that the motor provides torque only in one direction all negative components of the reference are negated to zero. Hence only positive values of the of reference torque are provided

by the controller. This is done to optimize controller to the particular design of the exoskeleton system where a continuous unidirectional rotation of the motor

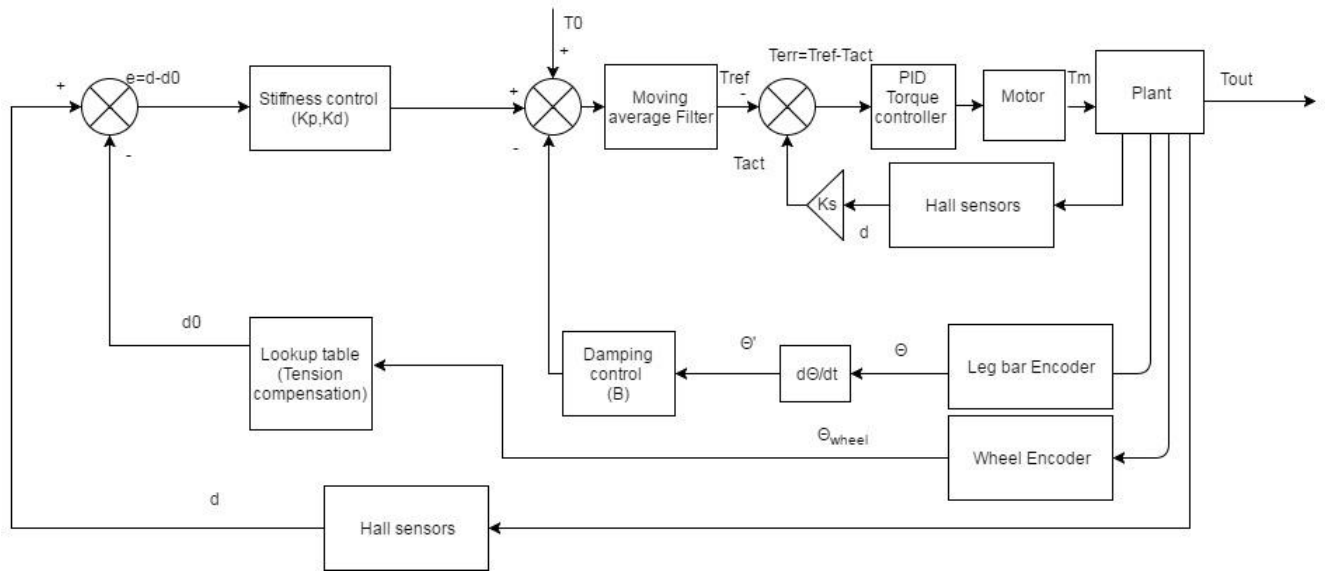


Figure 3.2 Control diagram of implemented scheme

enables oscillating motion of the leg bar over a fixed range of motion [2]. Further, an inner closed loop torque control is implemented to ensure actual output torque exerted on the user's leg is close to the reference torque provided by the controller. This is achieved by implementing a PID controller with pre-tuned gains set experimentally through bench testing.

The controller is based on the simplified impedance controller equation Eq (4), where the stiffness is determined by a PD controller and the damping by the rate of change of the leg bar. The negative damping factor indicates opposing the direction of damping factor resulting in a reduction of reference torque at higher velocities. This is to prevent the system from going unstable at higher walking speeds. The minimum torque T_0 is set to prevent back drivability of the motor and to ensure there is a minimum torque exerted by the exoskeleton on the user when the user is stationary. Further, it serves as a method to provide a smooth output torque on the user's leg as the interaction between the user, and the exoskeleton is measured as an offset from zero interaction torque during assist mode, hence preventing leg bar oscillating around zero torque position. The torque offset value depends on the operating mode where it is set to zero in low impedance mode and set to required value for a high impedance mode. The required value corresponds to the preset minimum torque exerted by the motor, which can correspond to a value up to 5 Nm by the leg bar of the exoskeleton on the user.

$$T_{ref} = T_0 + K_p(e) + K_d(\dot{e}) - B(\dot{\theta}) \quad (5)$$

where T_{ref} is the reference value of torque sent to the motor, T_0 is the minimum torque offset to be provided by the motor, e is the displacement error given by $d-d_0$, \dot{e} is the rate of change of displacement error, θ is the angular position of the leg bar, $\dot{\theta}$ is the rate of change of the leg bar angle, K_p and K_d are the gains for the stiffness component of the system, B is the damping coefficient of the controller, T_{act} is the actual torque or the torque detected by the spring compression, T_{err} is the torque error between the actual and the reference torque, K_s is the spring stiffness of the corresponding spring, K_t is the torque constant of the motor, I_{ref} is the reference current value sent to the motor, T_m is the motor output torque and T_{out} is the output torque of the exoskeleton leg bar.

The displacement of the leg bar is determined by Hall effect sensors at either end of the linear slider measuring corresponding spring's displacement. The Hall effect sensor is pre-calibrated with the help of capacitive radial encoders of a lower resolution (0.09 mm resolution). The points in between are assumed linear and extrapolated using MATLAB curve fitting tool box. Further, the Hall sensors are calibrated to compensate for cable tension throughout the rotation of the worm wheel, where d_0 denotes the Hall sensor displacement considering cable tension and d denotes the actual Hall sensor's displacement. The value of d_0 is determined by a lookup table which stores the relation between the worm wheel angle θ_{wheel} and the Hall Effect sensor displacement corresponding to no external force and only considering the tension in the cable which changes with angle of worm wheel, in terms of a polynomial equation. The error, e is given by $d-d_0$ or the difference in displacement of spring and the displacement due to cable tension, where d_0 is a function of θ_{wheel} or the angle of worm wheel. To determine the relation between d_0 and θ_{wheel} , the value of d_0 is recorded while moving θ_{wheel} from 0 deg to 360 deg when there is no interaction with the device, and these values are stored in the lookup table where d_0 may be determined based on θ_{wheel} value.

The controller senses the user's intent by determining the velocity and displacement of the leg bar with respect to the offset torque position T_0 . This is given by Eq (5), where e is the displacement of the leg bar and \dot{e} is the velocity or the rate of change of the leg bar with respect to torque offset (T_0) position. Further, to ensure the motor rotates in a single direction, any negative reference torque corresponding to counter clockwise direction of the motor is set to zero. Tuning the gains of the controller can vary the impedance of the system resulting in corresponding levels of assistance provided by the system. This makes it possible for the user to operate the

exoskeleton in a low impedance free mode or a high impedance assist mode. To operate the exoskeleton in the low impedance free mode, the gains of the system are lowered accordingly. By setting the minimum torque T_0 to zero and reducing the value of the stiffness parameters (K_p and K_d) of the system, the exoskeleton can track the user's movements with minimal interaction torque. Ideally, a zero interaction torque between the user and the system indicates no assistance or resistance by the exoskeleton on the user. Data collected during free mode corresponds to the user's natural unassisted gait and is used as the user's reference gait profile where it is used to compare gait parameters plot for same walking speed with assisted and unassisted modes.

CHAPTER 4

METHODS

The impedance-based control system designed is implemented on the exoskeleton device through a microcontroller which interfaces the mechanical design of the device with the electronic components that are sensing and actuating the device. The exoskeleton device has electronic components embedded in it that include sensors, motor driver, microcontroller, and data communication system. The device is run by a 16-bit PIC microcontroller (MicroChip, dsPIC33FJ64MC202). This microcontroller runs that sample time of 2 ms. The motor is controlled by a motor driver (ESCON 50/5) running a 200 W brushless DC motor (Maxon EC 4 pole 305014) which is supplied 32 V and can be supplied up to 12 A. The worm wheel angle and leg bar angle are determined by incremental encoders with 10-bit resolution. The torque is measured with Hall effect sensors (Allegro, A1324), with a sensitivity of 5 mV/G. These are read through a 12-bit ADC on the microcontroller. Data is collected from the microcontroller with serial communication (UART) and transferred to a PC using a serial terminal (Putty/Real term). The data was logged at a rate of 250 Hz and processed using Matlab.

4.1 Exoskeleton calibration

The exoskeleton requires calibration to ensure that every time the device is turned on and off the key parameters are stored. There is a fixed relationship between worm wheel angle (θ_{wheel}) and leg bar angle (θ) when it is being operated. The incremental encoders determine the θ_{wheel} and θ values in terms of number of counts. This relation is stored in the controller in terms of an n^{th} order polynomial equation of the form.

$$P(x)=a_0 +a_1x^1+a_2x^2+\dots+a_nx^n \quad (6)$$

where $P(x)$ is the leg bar position, n is the highest order, a_n is the coefficient, and x is the variable corresponding to the worm wheel position.

The data to parameterize Eq (6) were obtained by recording and plotting the leg bar position with respect to the worm wheel at either end of the backlash region. These plots were segmented, and equations were curve fitted (MATLAB curve fitting toolbox) to obtain the corresponding set of equations in the form of Eq (6). The plots were segmented and curve fitted based on the fitting of a lower order polynomial. The roller diameter is smaller than the

guideway of the slider and results in the backlash region that are dependent on the position of the roller with respect to the guideway. Further, each subject's upright standing position was preset mechanically and corresponding equation updated at calibration. The spring displacement magnitude was determined by Hall effect sensors at either end of the linear slider. These sensors were pre-calibrated and determine the spring displacement and rate of spring displacement, taking into consideration the variation in cable tension through the complete range of motion of the roller.

4.2 Experimental protocols

The experimental protocols are divided into bench testing and human testing protocols where bench testing is done to determine the functionality of the controller to a certain extent before human subject testing. This was an initial validation stage as well as a safety stage that helps determine if components of the controller are working as expected as well as to determine the gains and range of operational gains that controller may be configured to during human testing. The second stage of testing was the human subject testing stage which helps evaluate the controller's functionality during actual operation conditions. Further, the human testing stage involves multiple test subjects to determine functionality and hence validate the controller.

4.3 Bench testing

Before testing the device on human test subjects, it was evaluated with a series of bench tests. The device was fixed to a stationary bench and made to interact with the user's hand in an attempt to simulate device's movement during gait. From Figure 3.1 and Eq (5), it can be seen that the major control parameters affecting torque output are stiffness, damping and torque offset as well as variable factors such displacement error due to the interaction between system and environment as well as speed.

The bench testing steps involved tuning the gains of the controller according to the bench test being performed, ensuring device was firmly fixed to a stationary and sturdy table and proper safety mechanisms are in place such as an emergency stop switch, a timed shutoff of motor control in case device is left unattended and padding was provided on the leg bar to ease strain on hands during testing. The device was calibrated first followed by the bench test to be performed where the user interacts with the device by grasping leg bar and attempting to

simulate interaction with the device. The data was recorded using serial communication and processed to evaluate performance.

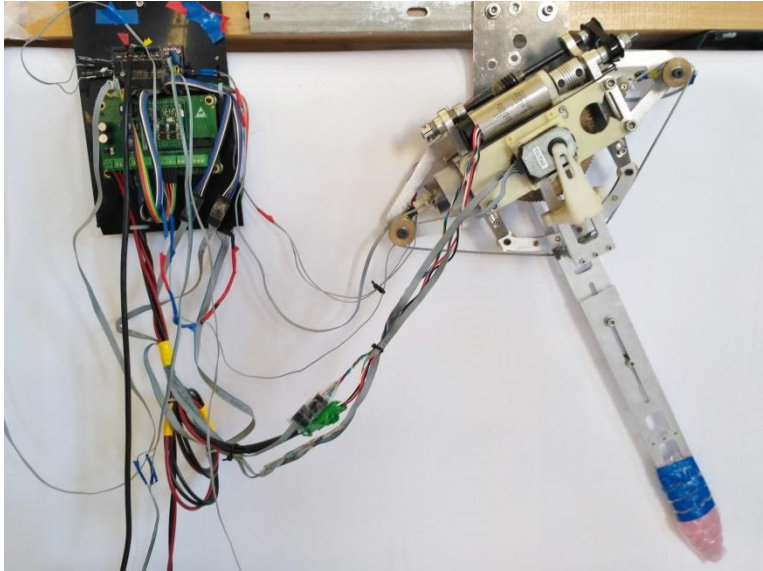


Figure 4.1 Bench testing setup

Bench test 1: Stiffness testing test was used to evaluate the ability to control the compliance or stiffness of the controller by adjusting the stiffness gains of the impedance controller. Two sets of gains were preselected with low stiffness values and high stiffness values. The device was tested by interaction with the hand for 10 seconds for a set of gains.

Expected outcome: The torque for high stiffness gains was expected to be greater than the torque for the low stiffness gains according to the difference in the selected gains. Further, power for high stiffness gains was also expected to be greater.

Bench test 2: Safety of the device on compliance was tested where the gains are preselected as per Table 5.1 and interacted with for 10 seconds after which the user resists it to stop motion, followed by a resuming motion of the leg bar.

Expected outcome: The torque output was not expected to spike when the user stops moving the leg bar and only slightly resists motion. The device is not expected to force movement of leg bar by providing a high torque spike to overpower the user.

Bench test 3: The gains for assist mode human testing are determined by attempting to simulate leg motion with the hand to determine approximately the range of torque and power provided for selected impedance gains.

Expected outcome: The torque output was expected to be in the range of ± 20 Nm and max power greater than 30 W, so as to determine approximate gains for human testing with the required range of torque and power.

Bench test 4: The inner loop torque controller was tested and its bandwidth determined by fixing the leg bar and sending reference torque signals and comparing reference to actual values. A series of sine waves of 1 Hz, 2 Hz, 4 Hz and 8 Hz were sent as a reference and output data collected. Further, a chirp signal of frequencies varying from 0 Hz to 10 Hz was sent as a reference to determine the frequency response of the controller.

Expected outcome: The magnitude of the actual signal is expected to drop as well as lag or phase difference was expected to increase from the cut-off frequency which will be determined by the respective bode plot.

4.4 Human testing

The exoskeleton was tested on five healthy individuals where the exoskeleton parameters such as length of leg bar and stride length match with that of the individual's build as well as zero angles of leg bar aligns with the upright standing position of the subject. The hip orthosis was worn as tightly as possible to fit the person's build to ensure maximum force transmission from device to the user. The controller gains were preset accordingly based on bench testing values for reference with assist mode ($K_p=5.3$, $K_d=35$, $B=3$, $T_0=3$) and low impedance mode ($K_p=1.1$, $K_d=18$, $B=1.5$, $T_0=0$). The individuals used a treadmill with necessary safety measures taken such as emergency shut off switch. Walking speed of the user was set by setting the treadmill speed to match the desired walking speed. Institutional Review Board (IRB) approval for these experiments was obtained. Gait profile comprising of hip angle, interaction torque and power were collected through serial communication and a data logging terminal on the computer.

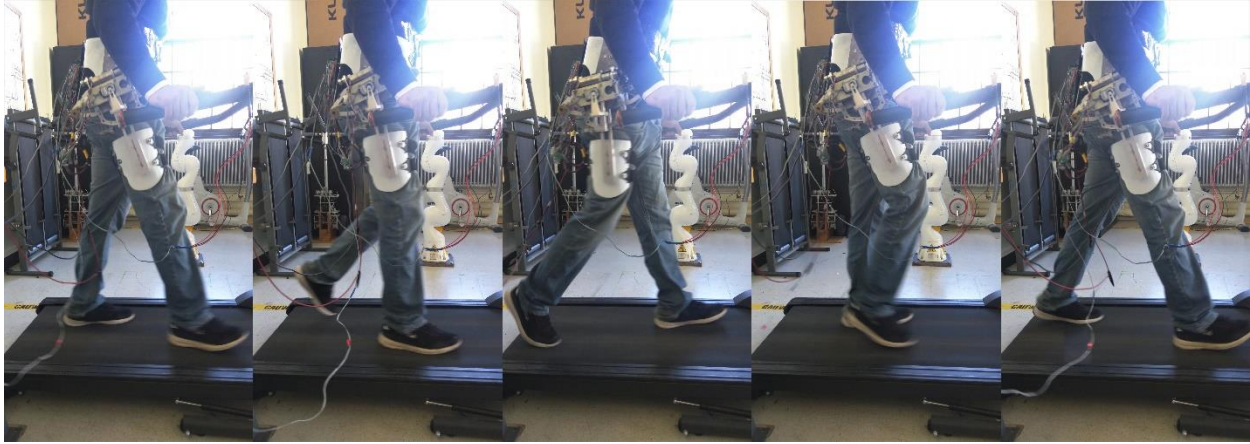


Figure 4.2 Time lapse shot of test subject- heel strike to heel strike

Experiment 1: The user was made to walk with exoskeleton operating in the low impedance mode for three sets of constant speeds (0.9, 1.1, and 1.3 m/s) to indicate controller functionality over a wide range. The low impedance control of the device was to make sure that the exoskeleton follows the user's movement and represent an individual's unassisted gait profile. The data includes leg bar position, the power of the system, and the interacting torque with respect to time were recorded. The effectiveness of the low impedance control scheme was determined by the interaction torque profile between the user and the exoskeleton denoted by the extension of the springs. The obtained gait data was plotted to obtain the user's reference gait profile.

Expected outcome: The obtained gait profile at a particular walking speed was used as a reference profile for comparison with the assist mode at the same walking speed. The obtained torque profiles were expected to be near zero indicating that the system was not assisting or resisting the user. This is plotted with the assist mode torque profile as well to contrast the difference in interaction torque profile.

Experiment 2: The assist mode was tested on the user after obtaining their reference gait profile at specified constant walking speeds from the low impedance mode test (Walking speed of 0.9, 1.1, and 1.3 m/s). The level of assistance was set by adjusting the gains of the impedance controller. The user was made to walk at the same speed in assist mode with the corresponding preset level of assistance which corresponds to the maximum assistance gains. The gait profile which includes angular position, the power, and the interaction torque was recorded with respect to time. The obtained gait trajectory in assistance mode was compared with the individual's gait trajectory in low impedance mode to determine the net change of gait pattern. The effectiveness of the assist mode was evaluated based on the obtained torque profile and power plot.

Expected Outcome: The expected outcome was a gait profile with very little variation with respect to the recorded gait data in the free mode. The hip angle trajectory should have little variation with respect to reference hip angle trajectory in free mode. Further, the torque profile should have a smooth transition from maximum torque to minimum torque through the gait cycle and should be primarily assistive torque during motion with a region of resistive torque at the end regions of linear slider corresponding to energy storage in the springs. The assistive or resistive nature of torque was determined by the power profile where a positive power corresponds to assistive and negative power corresponds to resistive torque.

Experiment 3: The user was to test the assist mode at varying speed (between 0.9 m/s to 1.3m/s), and the gait profile which includes angular position, the interaction torque, and power was recorded with respect to time. The gait profile was plotted per stride with respect to time to indicate the net change in profile with varying walking speed. The resulting plot should show a transition of the gait profile from a speed of 0.9 m/s to 1.3 m/s.

Expected Outcome: The expected outcome was varying torque profile that would be in phase with the corresponding hip angle as the walking speed increases. This would indicate the controller’s ability to provide required assistance with varying speed without re-tuning the controller or greatly varying the torque provided. Further, the gait profile should be similar with each stride and should be scaled with respect to time as the time per stride varies with the walking speed.

Table 4.1 Test subject data:

Subject Number	Height (cm)	Weight (Kg)	Mid-thigh diameter (cm)	Length of femur (hip joint to knee) (cm)
Subject No.1	178	75	58.9	54.86
Subject No.2	188	94.5	60.96	61.72
Subject No.3	182	82	58.4	58.93
Subject No.4	180.3	84	59.1	57.4
Subject No.5	172.7	67	51.8	48.76

4.5 Data Processing

The data collected for constant walking speed tests consist of the subject's gait profile for each specified speed of 0.9, 1.1, and 1.3 m/s, which were set by the treadmill. Each subject's gait data for assist mode and low impedance mode was collected and averaged over ten cycles or ten strides from heel strike to heel strike. This obtained data was further averaged over the five test subjects obtaining a gait profile for each walking speed and each mode. The mean and standard deviation for this data were calculated and plotted for both the operating modes. The plots were depicted with respect to gait percentage where 0% and 100% corresponds to heel strike. This gait phase assumption was made since the exoskeleton device functions over a fixed range of motion, hence approximately mapping the device's leg bar angle to the gait phase. However, this is an estimation and is not extremely accurate as the device's leg bar may not exactly correspond to the user's hip angle. The sign convention was taken as flexion is positive and extension is negative. Hence, positive torque corresponds to flexion torque, and negative torque corresponds to extensive torque. A hip angle greater than zero was considered flexion, and a negative hip angle is considered extension. Positive power is considered injection into the system, or assistive and negative power was considered resistive or power absorption from the system. The plot for variable speed testing shows the gait profile of the user from a walking speed of 0.9 to 1.3 m/s. The gait profile was plotted with respect to time to indicate stride by stride change in duration. Further, the gait profile for 0.9 m/s and 1.3 m/s were highlighted to indicate maximum and minimum speeds.

CHAPTER 5

RESULTS

5.1 Bench testing results

Figure 5.1 shows device angle, torque, and power for two different stiffness parameter values indicating the most significant effect of varying the stiffness parameters. Figure 5.1 shows the devices angle, torque and power profile for 10 seconds for two sets of impedance gains. The torque output range is ± 10 Nm, and max power is 15 W for low stiffness parameters ($K_p=2.2$, $K_d=23$, $B=3$, $T_0=2$) and torque output range of ± 18 Nm and max power of 20 W for high stiffness parameters ($K_p=3.4$, $K_d=33$, $B=3$, $T_0=2$).

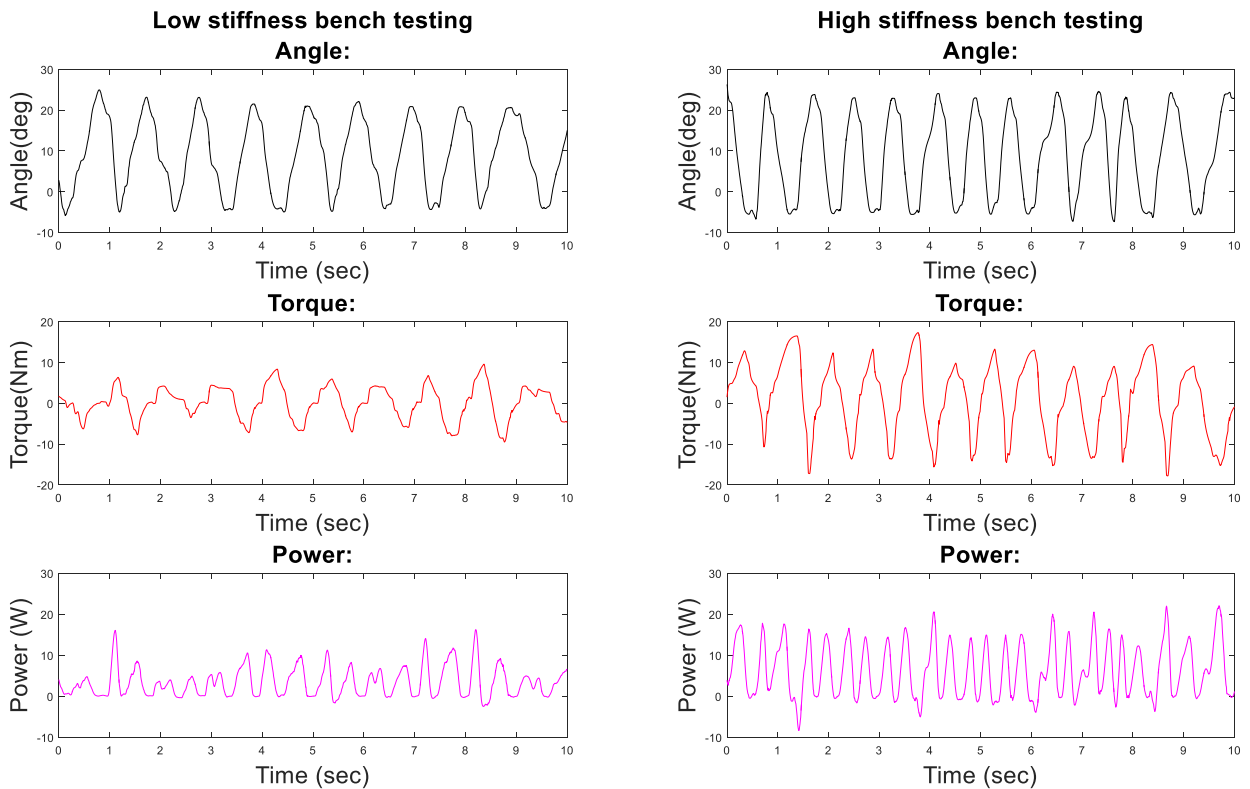


Figure 5.1 Bench testing- stiffness parameter evaluation

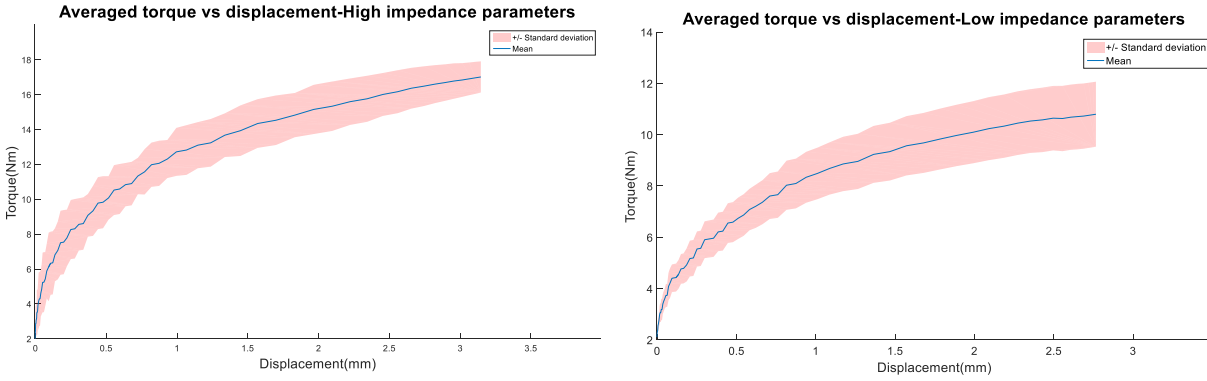


Figure 5.2 Averaged torque vs displacement curves

Table 5.1 Experiment testing- controller gains:

Experiment name	K_p	K_d	B	T_0
Low stiffness testing	2.2	23	3	2
High stiffness testing	3.4	33	3	2
Damping effect testing	2.5	23	5	2
Safety testing	3.4	33	3	2
Assist mode for human testing	5.3	35	3	3

Figure 5.3 demonstrates the safety aspect of the controller which enables the user to slow down or stop the device from running without significant resistance. The controller parameters are preset ($K_p=3.4$, $K_d=33$, $B=3$, $T_0=2$) and the device was interacted with for 10 seconds, following which it was stopped for 4 seconds. The torque in this region shows it is 2 Nm as a result of minimum torque offset T_0 .

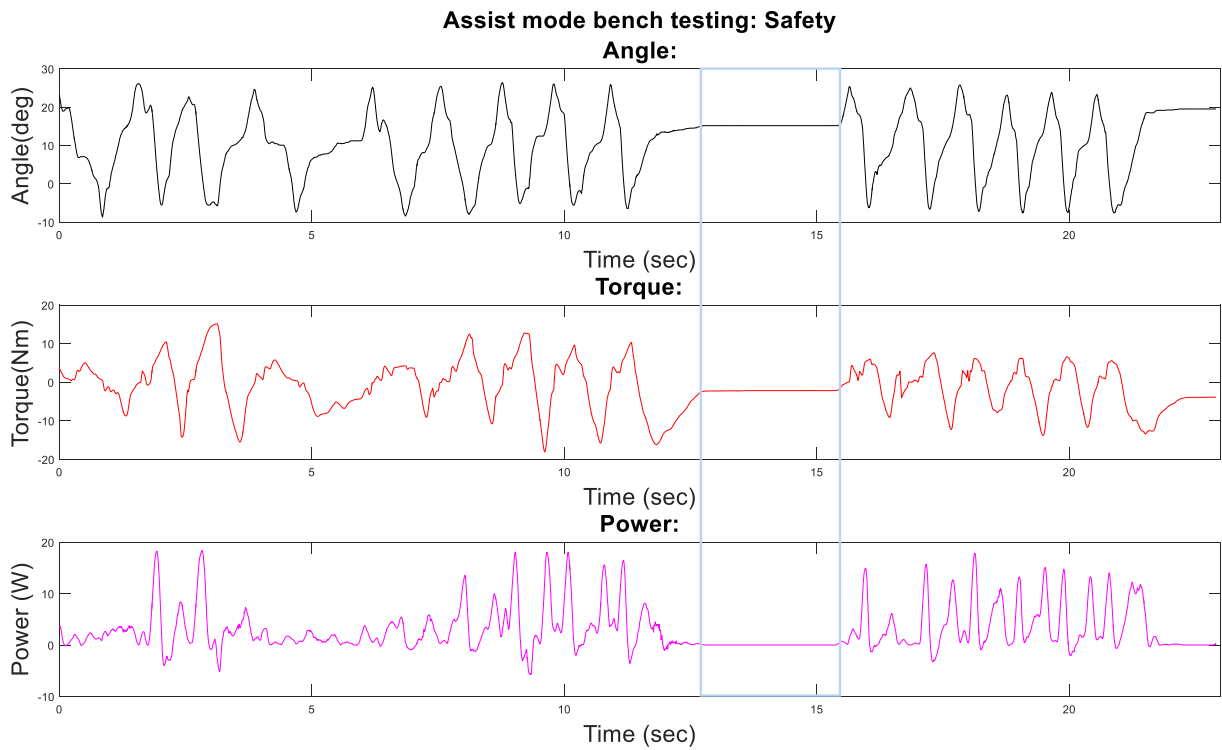


Figure 5.3 Bench testing - Safety evaluation.

To determine the bandwidth of the controller, a chirp signal of frequencies 0 Hz to 10 Hz is passed as reference and tracked. The magnitude and phase difference of actual to reference torque was determined. Figure 5.4 shows the torque controller bode plot determined by tracking a chirp signal as reference torque from 0 Hz to 10 Hz. The system begins to show phase lag at 3.95 Hz and goes unstable above 5.8 Hz which is the bandwidth of the system.

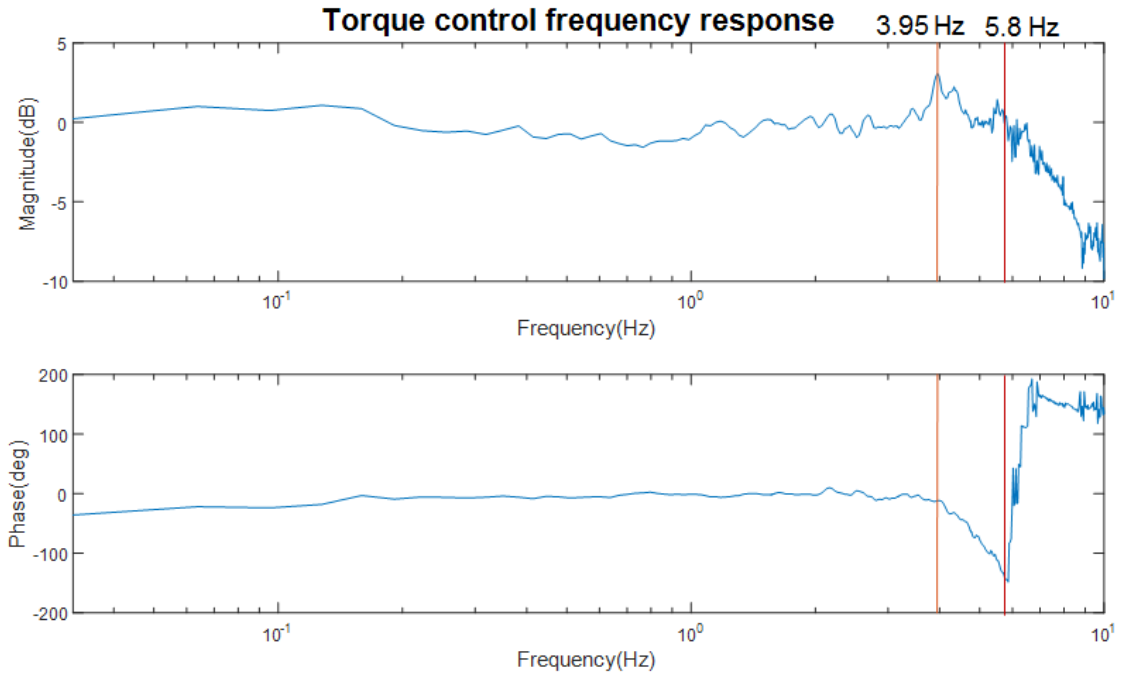


Figure 5.4 Torque control frequency response

Figure 5.5 shows a bench test with the controller gains set to gains for human testing ($K_p=5.3$, $K_d=35$, $B=3$, $T_0=3$) and the device interacts with an intention to simulate walking gait as closely as possible. The device was interacted with at a frequency of 0.9 Hz which approximately corresponds to 0.9 m/s which was the lower end of the preselected speed range for human testing. The peak torque values are in the range of 18 Nm in both flexion and extension and a peak power of 30 W.

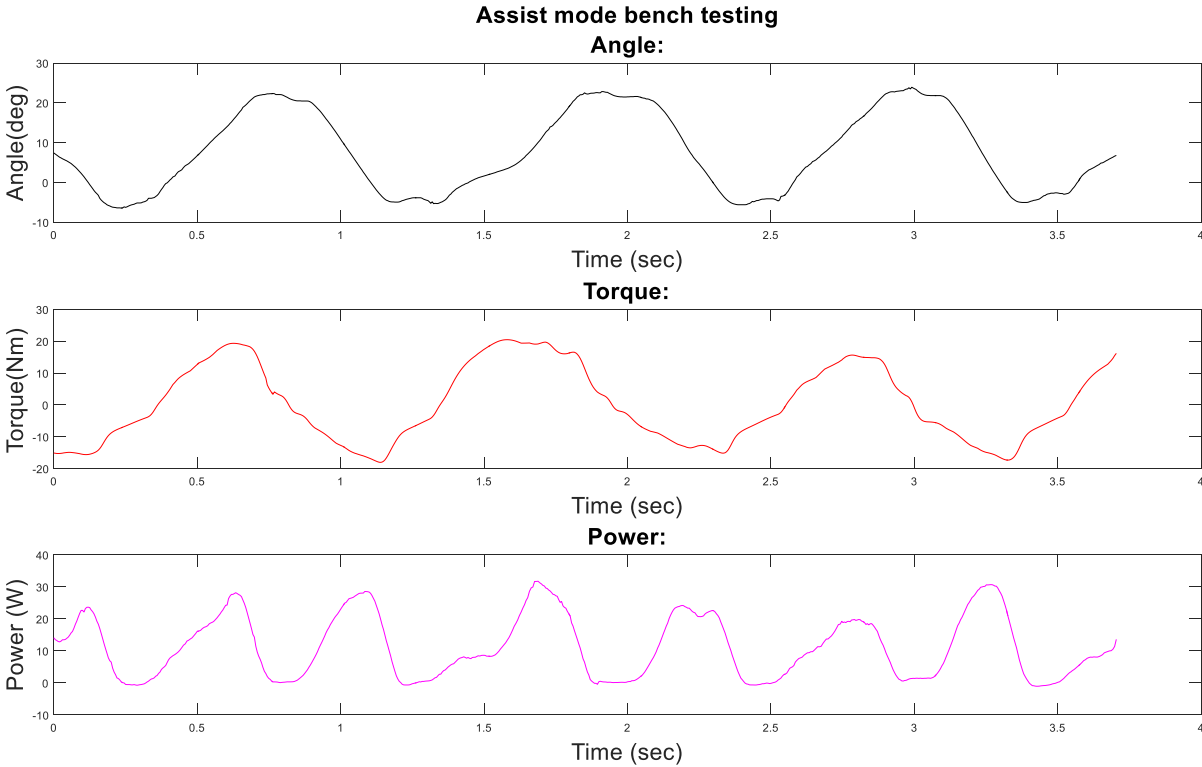


Figure 5.5 Bench testing- controller gains evaluation

5.2 Human testing results

The Figure 5.6, Figure 5.7 and Figure 5.8 show the data for fixed walking speed tests, for both assist mode and low impedance mode for five test subjects. The averaged gait profile for assist mode and low impedance mode are compared for all three walking speeds (0.9, 1.1, and 1.3 m/s) as seen in Figure 9.6. The mean and standard deviation for the data was depicted. The averaged gait data for assist mode shows torque in the range of ± 30 Nm and ± 8 Nm for low impedance mode. The averaged up to 45 W of power generated to 12 W of power dissipated or stored for assist mode and was bounded between ± 6 W for the low impedance mode. The end region corresponding to minimum device angle shows power dissipated or stored in the range of 1.5 W to -12 W, which can be related to compressive torque or energy stored in the spring.

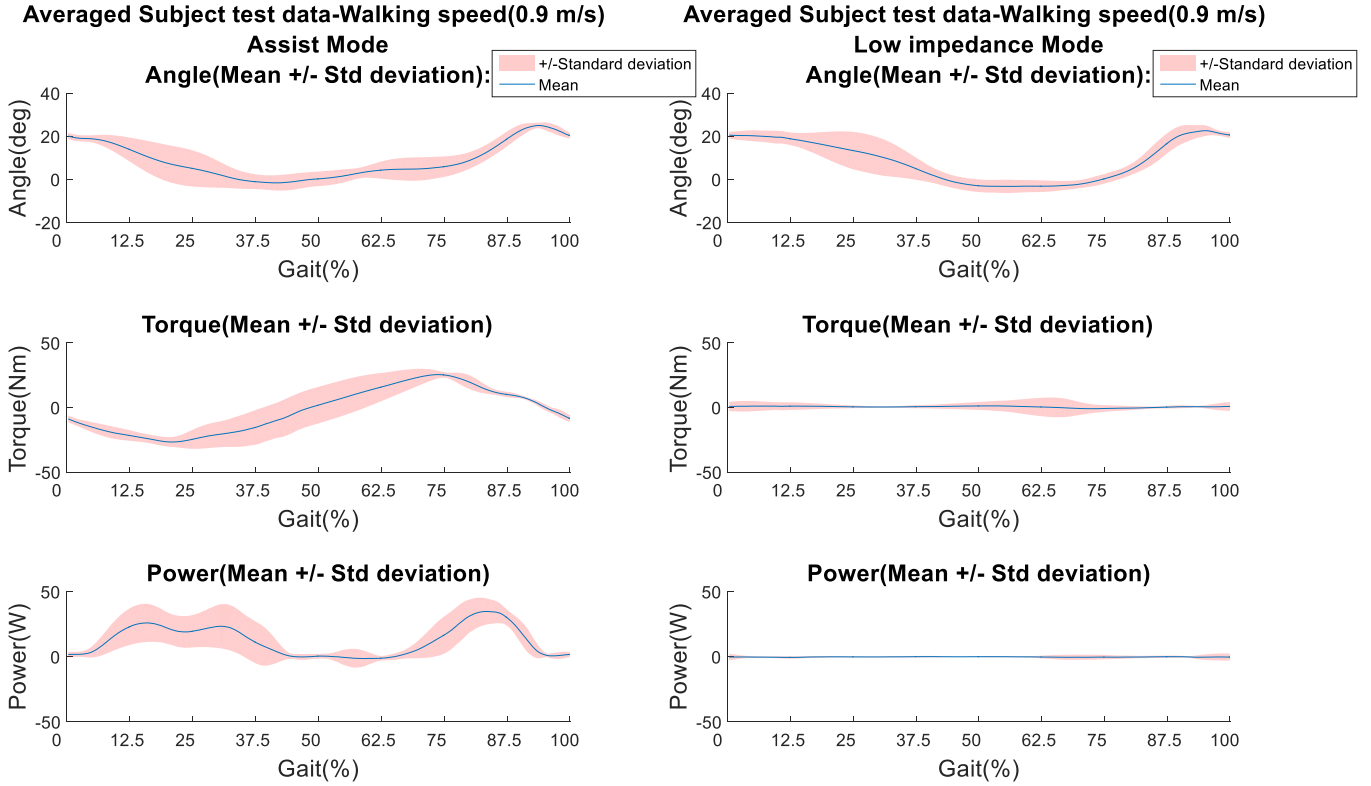


Figure 5.6 Averaged gait data for speed 0.9 m/s - Assist mode and low impedance mode

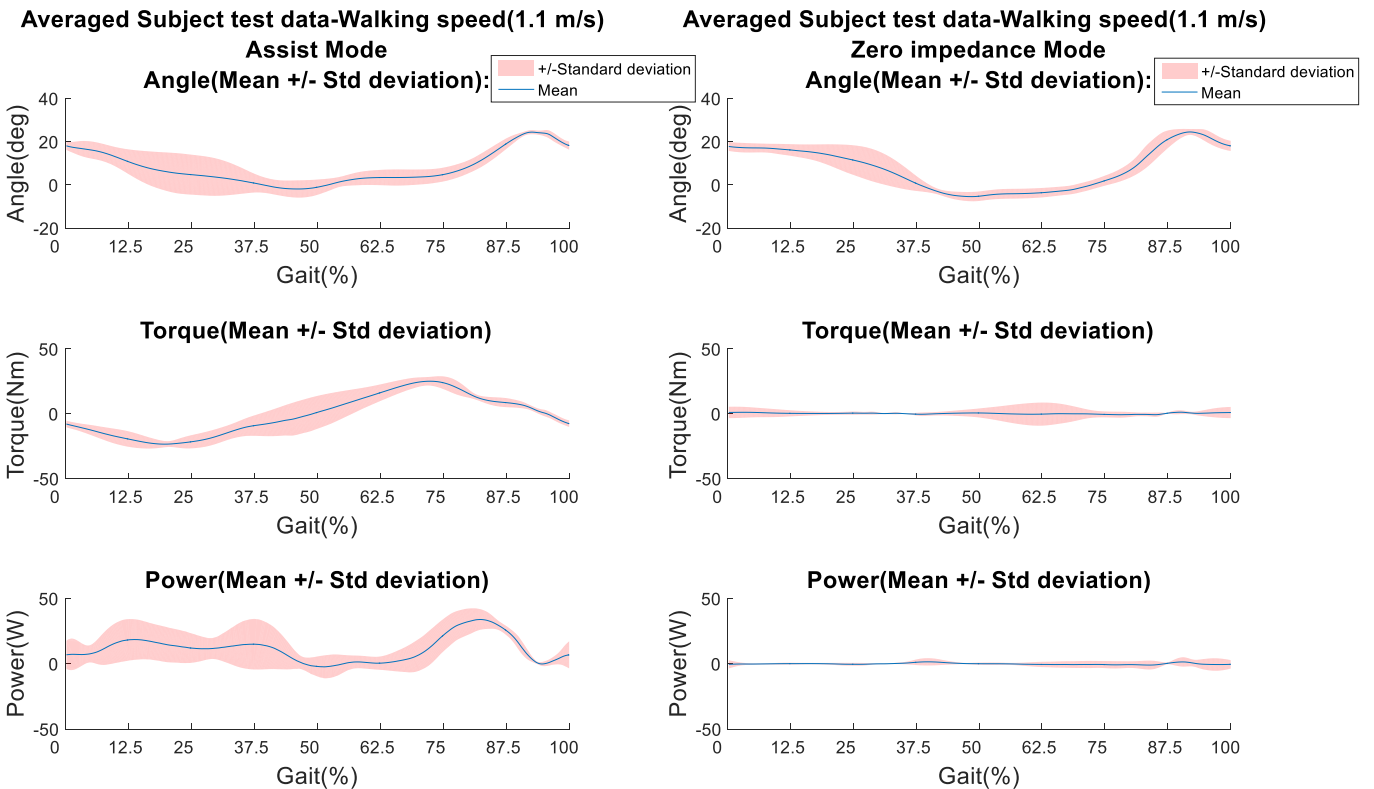


Figure 5.7 Averaged gait data for speed 1.1 m/s - Assist mode and low impedance mode

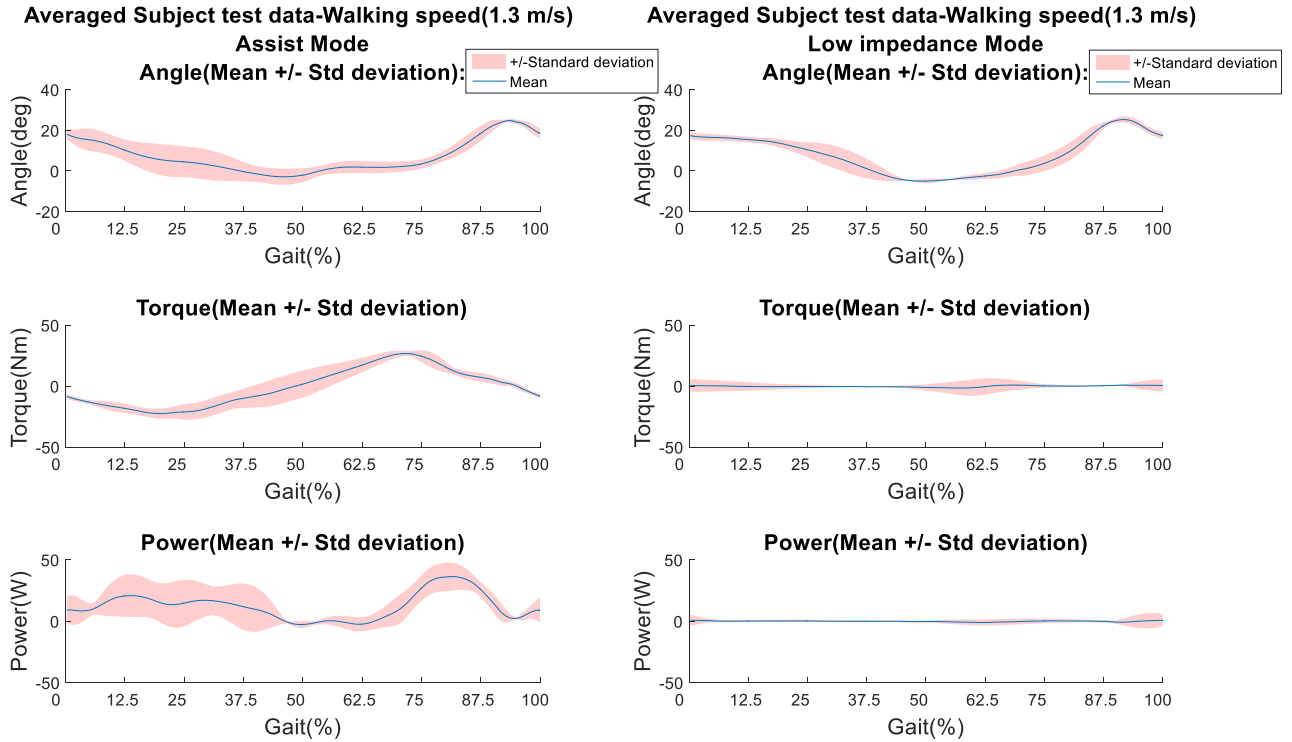


Figure 5.8 Averaged gait data for speed 1.3 m/s - Assist mode and low impedance mode

Figure 5.9 shows the gait trajectory comparison between assist mode and zero impedance mode for all five test subjects. Further, the difference in gait profile of both modes of operation was determined in degrees with respect to gait percentage and plotted. The difference plot shows that the difference between both modes is in the range of ± 10 degrees.

Assist mode vs Zero impedance mode gait trajectory

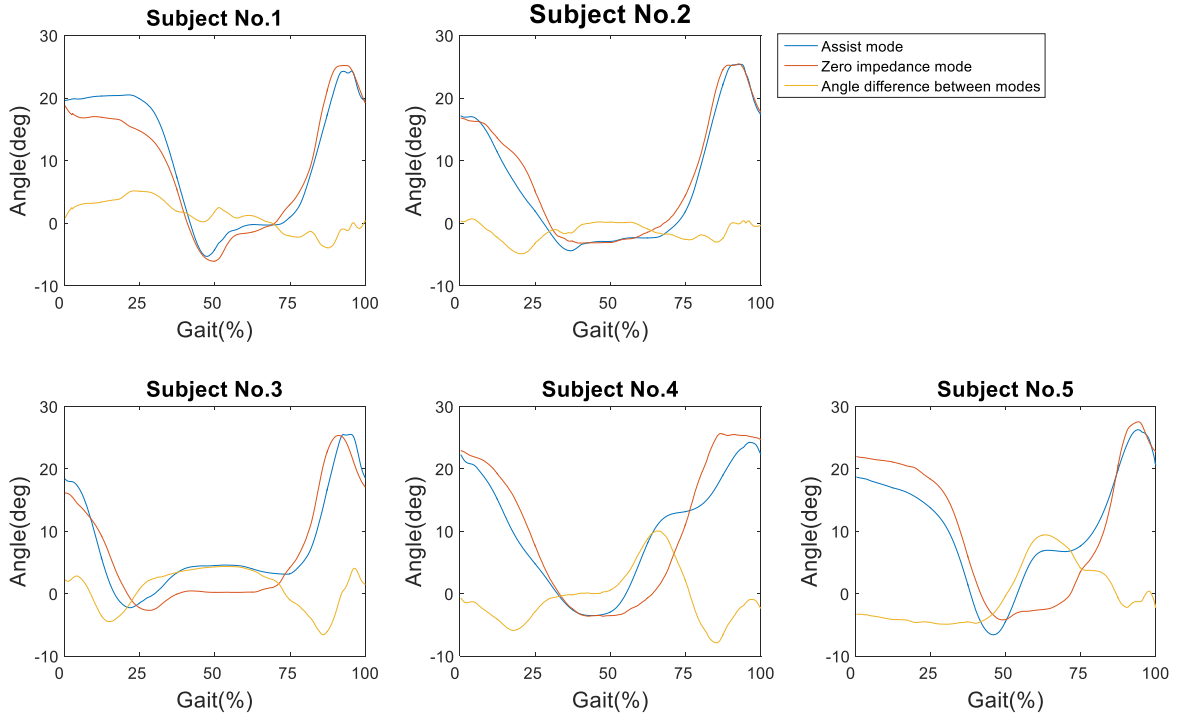


Figure 5.9 Gait trajectory comparison between modes of operation for five subjects

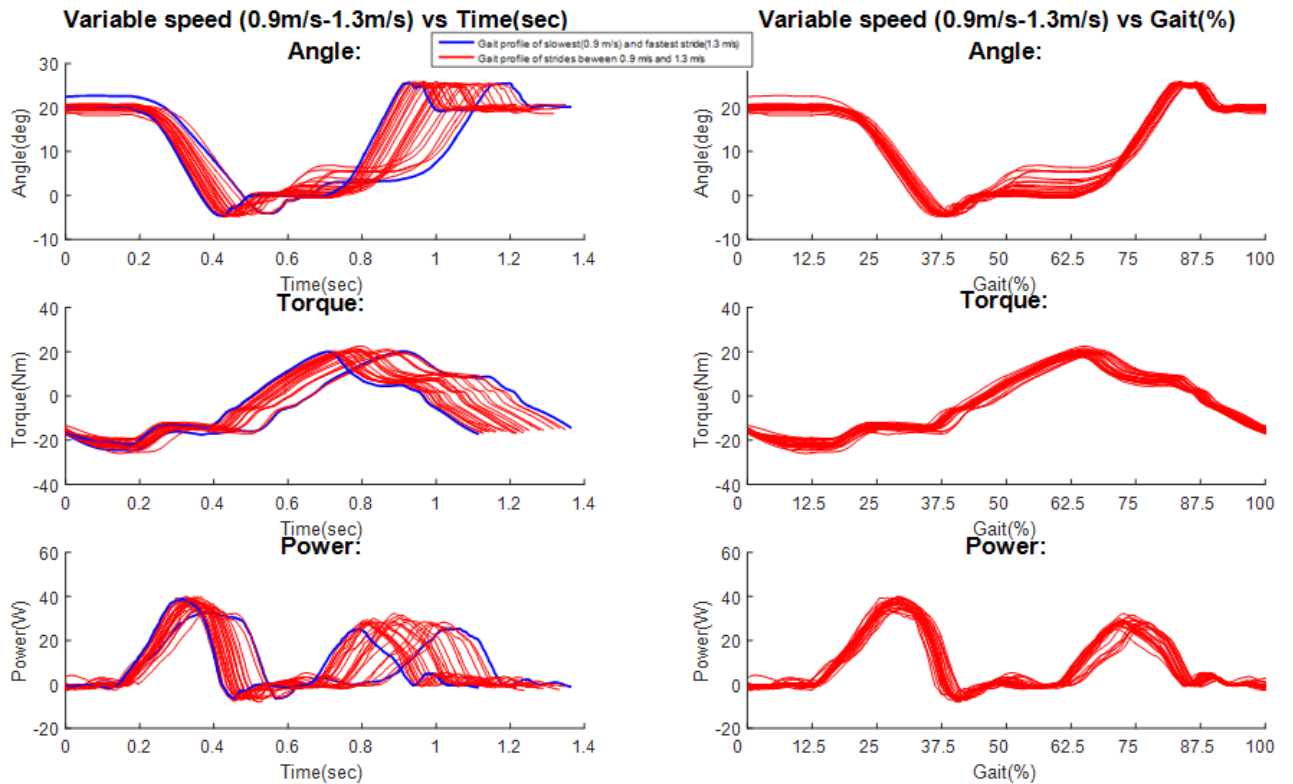


Figure 5.10. Single subject trial with variable speed gait data with respect to time and gait percentage

Figure 5.10 shows the gait data of a single subject walking at variable speed from 0.9 to 1.3 m/s. The gait profile was plotted with respect to time as well as gait percentage. This is to indicate the variability of stride by stride duration when walking from 0.9 to 1.3 m/s. Further, it shows the corresponding torque and power profile with respect to time and gait percentage and shows their variability. The strides for speeds 0.9 and 1.3 m/s are highlighted in blue to indicate a net difference in gait profile between slowest and fastest speed. The gait data was then plotted with respect to gait percentage to show stride by stride deviations and indicate that stride variability with respect time is not a result of stride by stride discrepancies. In the Appendix, Figures 9.1 to 9.5 show stride by stride gait profile of all five test subjects for variable speed walking to indicate gait variations for each subject. These figures show little variations with respect to gait percent indicating gait profile does not significantly change with speed variation. The gait trajectory across the five test subjects show variation with respect to stance time and region of toe off corresponding to gait percent. This arises as a result of having different natural stride length that does not match with the device's range of motion.

Figure 5.11 shows stride-by-stride power profile for variable speed testing for five test subjects. The subjects were made to walk at a speed of 0.9 to 1.3 m/s and the corresponding gait trajectory, torque, and power recorded. Further, to illustrate stride-by-stride power variance with respect to gait cycle, the power data was plotted with respect to gait percentage. The gait percentage shows heel strike to heel strike corresponding to 0 and 100 as a percent of gait.

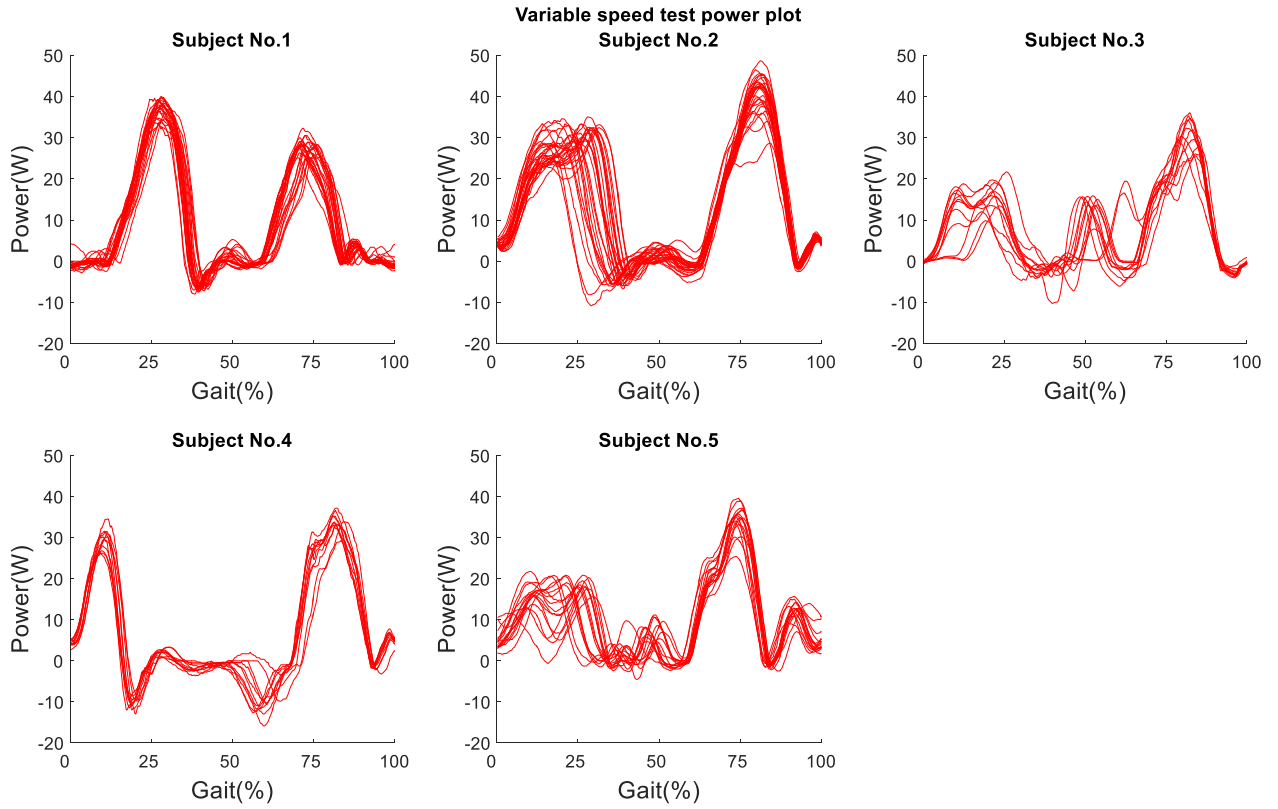


Figure 5.11 Power for variable speed walking from 0.9 m/s to 1.3 m/s.

CHAPTER 6

DISCUSSIONS

The controller adjusts the virtual stiffness and damping components by preset gains which determine the nature or behavior of the exoskeleton. These components determine the impedance of the system. It varies from low impedance mode where the device does not assist or resist the user and simply aims to follow the user's leg with minimum interaction torque, to a high impedance mode where the device has a high stiffness and aims to contribute power to the user's gait cycle.

The bench tests determine the functioning of different aspects of the controller to understand better their impact during gait testing. From the Figure 5.1, it can be inferred that higher stiffness parameters increase the torque output response for the same interaction or displacement error and vice versa for a lower stiffness system. This behavior is as expected from the Eq (5) and can control the system to be more compliant or stiffer as required. Figure 5.2 shows the averaged torque versus displacement curve for two sets of impedance parameters, with a low stiffness parameters and high stiffness parameters. This plot gives an intuitive understanding of how output torque varies with respect to displacement due to the interaction between the user and the device. The controller gains selected for controller response tests are given in Table 5.1. The response shows a greater torque with respect to displacement for the high impedance parameters as opposed to low impedance parameters, which is as desired by the controller.

The safety bench testing demonstrates the compliance of the impedance controller to prevent injury to the user during operation by ensuring the device does not overpower the user and force unintended motion. Figure 5.3 indicates that the user may stop the device without significant resistance as indicated by the low torque in this region and no sudden spike in torque resistance.

The bandwidth of the controller as determined by Figure 5.4 goes up to 3.95 Hz after which magnitude decreases and phase lag increases. Hence, the cut-off frequency for the controller is 5.8 Hz where the phase lag goes beyond 180 degrees, which is expected due to series elastic actuation, slack in steel cables used for force transmission and a backlash region between flexion and extension of the device leg bar.

The plots depicting constant walking speed for the test subjects (Figure 5.6, Figure 5.7 and Figure 5.8) compare the gait profile for low impedance mode and assist mode averaged among the five test subjects, and depict mean and standard deviation for the leg bar angle, interaction torque, and power.

The comparison of averaged gait trajectory between both the modes shows varying deviations for all walking speeds. There may be multiple factors contributing to this discrepancy apart from being an average of five test subjects with different gaits. As the leg brace acts as an intermediary between the exoskeleton device and the user's leg, it plays a great role in the functioning of the device and force transmission. The brace is made up of a thermoplastic material with soft cushion padding that is to be strapped on to the user using Velcro straps. It consists of a back brace and a thigh brace, which must be strapped separately. This brace offers a limited range of adjustability and relies on soft padding for tighter fitting, which may introduce more compliance between the device and the user. Further, differences due to the compliance of the soft tissue are not accounted for which may further decrease the effectiveness of force transmission between device and user and may introduce deviation in gait trajectory due to this soft tissue. The individual's natural stride length may further introduce deviations as the subject is forced to adapt the devices stride length for every walking speed, which may result in subject walking with a different gait trajectory for each mode. This change in gait during different modes of operation may greatly influence this deviation in trajectory and affect the analysis of obtained data.

From Figure 5.9, we can see the difference between the gait trajectories of assist mode and low impedance mode for all five subjects tested at 1.1 m/s speed. We can observe that the difference with respect to gait percentage is in the range of ± 10 degrees, which can be attributed to the mentioned reasons. Comparing the interaction torque of both the modes with respect to gait percentage for constant walking speeds shows significantly different values. Low impedance mode shows interaction torque in the range of ± 7 Nm as compared to assist mode which has interaction torque in the range of ± 30 Nm. The desired interaction torque for low impedance mode is near zero, however factors such as stride range restriction, the non-backdrivable motor as well as non-optimal control gains for the individual result in a greater interaction torque. This interaction torque is within reasonable limits and does not significantly contribute to the change in the gait of the individual. However, it still does not provide accurate data which may correspond to the user's hip angle trajectory and simply gives the resultant angle of the device upon interaction with the user's leg where the controller aims to minimize the interaction torque between the user and the exoskeleton leg bar. This is an accepted assumption in the data collection process and simply gives us a rough estimate of the user's hip angle trajectory. To obtain accurate data that corresponds to the user's hip angle, a more extensive data collection process such as using marker sets and high-speed cameras for gait analysis may be needed. Observing the interaction torque for assist mode for constant walking speed tests, we see that range of interaction torque is ± 30 Nm. This torque corresponds to impedance gains that result in the highest interaction torque for the device which is determined experimentally. This interaction torque is primarily flexion torque during flexion and

extensive torque during extension (Figure 5.6, Figure 5.7, Figure 5.8). This is a requirement of the device to be assistive or provide assistive torque to the user. However, there is a region of resistive torque characterized by a flexion torque during extension or extensive torque during flexion that occurs in the end regions of the linear slider and is expected from the device. It corresponds to spring compression at end regions or energy storage in the springs at the end regions [2]. Further, the interaction torque gradually changes from flexion to extensive without drastic spikes is a desired feature, considering the safety and comfort factors to the user. The nature of the interaction torque can be further determined by the obtained power plot. A positive power indicates the torque provided to the user is assistive whereas a negative power indicates resistive torque by the exoskeleton. The range for power for the assist mode is 45 W to -12 W and for the low impedance mode is ± 6 W. Further, the power peaks line up with gait trajectory during flexion and extension, or highest power is provided during flexion and extension of the leg bar. The regions between flexion and extension, the end regions of the linear slider, show negative power corresponding to energy storage in the spring. The negative power varies depending on the user's natural stride length and walking speed and was observed in the range of -1.5 W up to -12 W. Further, this device does not use foot sensors for gait phase determination unlike commercially available exoskeletons such as HAL, Ekso, BLEEX, ReWalk, and other similar devices. The controller simply utilizes interaction torque to provide assistance, and the device's mechanism switches direction of motion after device's end angle is traversed. The exoskeleton Lokomat uses interaction torque to assist the user as well and implements an impedance control with force feedback [22], however, it requires a reference angle for the device to set output torque.

The plots for variable walking speeds are plotted with respect to time to indicate a change in gait profile with speed variation. The subject is made to walk at a speed of 0.9 to 1.3 m/s in assist mode. Shown in Figure 5.10, a single subject's data is used to compare stride by stride gait profile plotted with respect to time and gait percentage. This plot indicates stride by stride gait profile deviations as well as a net change in stride duration from walking speed of 0.9 m/s to 1.3 m/s.

The gait trajectory plot shows that stride time gradually decreases from 0.9 m/s to 1.3 m/s and shows a corresponding change in gait trajectory. It shows that gait profile is similar for varying speeds with changing stride duration as inferred from Figure 5.10 and Figure 9.1 to 9.5. The interaction torque profile shows a similar torque profile for all speeds with varying stride duration as well, indicating the device is varying the output torque according to the walking speed of the user to ensure torque provided is primarily assistive during the gait cycle. This adaptability is greatly desirable as it permits the user to walk with varying speeds and ensures the controller adjusts torque to provide assistance without the need for re-tuning controller or greatly compromising on torque assistance.

The Lokomat uses interaction torque to permit the user to change speed as well [22]. However, it adapts gait trajectory using a reference trajectory generator to enable the user to walk with new gait at different walking speed. The HAL exoskeleton uses EMG signals along with foot sensors to determine the change in gait, so as to adapt the gait pattern or walking speed. The power plot is also similar for varying walking speeds with varying stride duration and represents the nature of interaction torque. The power plot shows that the varying interaction torque is assistive in nature during the gait cycle except for the end regions where a negative power or energy absorption takes place with the springs.

Shown in Figure. 5.11 that the range of maximum power is in the range of 35 W to 50 W. Further, the power profile does not significantly vary from stride to stride. It is noted that there are negative power spikes varying from -3 W to -12 W after extension and before flexion. This corresponds to power absorbed by the springs and occurs around the end region of the leg bar range. This varies from person to person depending on natural gait stride as well as gait profile of the individual. Further, the power does not significantly increase with an increase in walking speed as a result of damping factor bringing torque down with an increase in walking speed.

Comparing the obtained data to expected data from a position control strategy, the following may be inferred. Firstly, the controller provides assistive torque to the user irrespective of their inherent gait trajectory and can provide maximum torque in the range of 20 Nm to 30 Nm. We can see this in Figure 5.6, Figure 5.7 and Figure 5.8. These plots show a standard deviation of device angle of up to 10 degrees indicating different gait trajectories for different subjects. The corresponding torques show a mean in the range of ± 25 Nm with a standard deviation of up to 15 Nm. Further, during flexion and extension these torques correspond to positive power indicating assistive torque. Hence, it may be inferred that the controller provides assistive torque to the user irrespective of the user's gait, unlike a position based control that may be tuned to a particular gait. Further, from Figure 5.10 and Figure 5.11, we can infer that the controller used may adapt based on speed, adjusting torque provided to match the gait phase and in turn power. This is visible in Figure 5.9 where stride by stride gait is plotted with respect to time and gait phase.

Qualitative feedback on the subject's perception of the device was collected. Regarding whether the device was indeed assisting the individual, subjects 1 to 5 had a positive response where they stated that the device did assist them during walking however to different degrees for different subjects. Subject 1 had the most positive response, stating that device assisted in gait during flexion and extension when most power injection was observed in the data. Subject 2 and 4 stated the device assisted them, however to a lesser degree and subject 5 stated the devices assisted however not sufficiently enough throughout the gait. This can be attributed to the orthosis fit to the individuals

build, where subject 1 had the best fit and subject 5 had the worst. Comfort wise, subjects 1 to 3 stated it was moderately comfortable once they got used to the range of motion of the device, and subject 4 and 5 conveyed discomfort during extension where the extension range was greater than their natural extension. Further, the weight of the device and the tight fit of the orthosis were sources of discomfort for the users.

CHAPTER 7

CONCLUSIONS AND FUTURE WORK

The objective of the developed impedance controller is to leverage the novel design of the exoskeleton to provide assistance during walking. The controller uses the series elastic actuation of the device to determine the necessary torque that needs to be provided to the user to assist with walking. Further, the controller takes into consideration the mechanism that converts continuous unidirectional rotation of the motor into oscillatory motion of the leg bar that is directly connected to the thigh brace. In an attempt to minimize loss of energy which would otherwise be dissipated if the motor had to reverse direction every cycle to accommodate the user's oscillatory motion of the leg [2], the controller ensures that the motor rotates in a single direction. The limitation with this method of not reversing motor direction is a fixed range of hip motion that constrained the user. The user has to go through the entire range of the leg bar motion before reversing the direction of motion every cycle. However, the series elastic elements comprising of the springs at the end of the slider and the flexible steel cables connecting the leg bar permit the user to deviate from this fixed range of motion, introducing compliance and comfort. The series elastic element permits deviation of the range of motion by ± 3 degrees. This increases the range of motion of the leg bar from 29 degrees without spring displacement by 6 degrees. The angle of extension and flexion are determined by the zero position of the device which is preset by adjusting to the upright standing position of the user. Typically, it gives the user around 5 degrees extension and 24 degrees flexion.

The implemented controller and exoskeleton design are closely linked to each other where the controller is designed while considering the specifications of the device's design. Factors such as unidirectional rotation of the motor or fixed leg bar angle range of device are incorporated into the controller. However, the controller may be extended to a generic group of exoskeleton devices rather than a single specific design. The primary requirements for the exoskeleton device to implement the controller are the ability to measure interaction torque or force between the user and the device and series elastic actuation incorporated in the design. Further, a means to determine gait phase of the individual is essential for the device to be able to provide assistance when needed. The exoskeleton in this thesis does not require a separate means to determine gait phase as the stride range is assumed constant and direction of motion of the leg bar switches after the entire range has been traversed once. This is a great limitation in the design and needs to be addressed to function on a wide range of builds as well as walking speeds [33]. On devices with a variable range of motion, the controller needs to be updated so that it takes into account the gait

phase and direction of intended motion. From the obtained data, we may conclude that the controller provides assistive torque during flexion and extension in the range of ± 30 Nm. This falls in the range of 0.3 to 0.5 Nm/kg for the selected range of test subjects. Considering average hip torque requirements during a gait cycle goes up to 1.25 Nm/kg [2] [31], that would indicate that the device is capable of providing 24% to 40 % of hip torque requirements. Further, the controller is capable of permitting the user to walk with assisted gait over a wide range of walking speeds. This would imply that the user could walk for greater distance with less muscular effort exerted, provided the weight of device does not impact muscular effort.

Figures 5.6, Figure 5.7 and Figure 5.8 show that the interaction torque during low impedance mode is in the range of ± 8 Nm, and the desired interaction torque in this mode is around zero. Factors such as limited range of motion of device and non-backdrivable of the device affect the interaction torque. However, this range is not significant enough to affect subject's gait and can still be used as a means for collecting subjects gait data for comparison purpose.

With regards to improving the design of the exoskeleton, a method to adjust stride length needs to be implemented where the user does not to restrict themselves to the fixed leg bar range for all walking speeds allowing them to walk with their natural gait. Further, torque measurement in the current design may not be accurate as it depends on a mechanical plunger compressing a spring to determine the hip torque. This spring displacement separates the magnet and Hall effect sensor which determine the net displacement of the spring. Deviations from actual torque can occur due to factors such as spring plunger not returning to the initial position after compression. Also, deviations can occur due to a change in tension in steel cables transmitting force and inaccurate curve fitting of cable tension, approximations in linearizing Hall effect sensors values, and loss in magnetic strength of the magnet and interference with metal components affecting magnetic flux. Hence a more reliable and repeatable means of torque measurement is needed for accurate data representation and optimal controller functionality.

To improve the speed of the implemented algorithm and data logging, a computation device with a higher processing speed should be substituted as opposed to the 16 bit PIC microcontroller running the device at 500 Hz with data logging at 250 Hz.

The device currently runs with an external DC power supply that is wired to the motor driver and sensor circuit and logs data with UART serial communication. This makes it necessary for the user to be within a certain distance from the DC power supply and computer. However, by using an external battery for power supply and using wireless data transfer for logging, the device may be made wireless permitting a user to walk around freely without the need to be near the workstation. Further, this device has a range of motion of 29 degrees in the sagittal

plane (flexion/extension) and does not provide any freedom of motion in the frontal plane (adduction/abduction) and relies on soft padding and looseness of fit to accommodate any motion in the frontal plane. This may affect the natural gait of the individual as it forces them to adjust their gait in the frontal plane. Providing a passive mechanism such as an elastic element to permit motion of leg bar in the frontal plane may not only permit a user to walk closer to their natural gait but also for better fitting of the device due to variations in individuals builds and hip-thigh fitting. Further, the hip brace may be made adjustable to a wider range of body builds with different femur lengths, thigh and hip sizes and overall height. The results provide an estimate of the level of assistance the device is providing and corresponding changes in gait. However, it does not quantify the muscular effort change that the user is undergoing while using the exoskeleton. Hence, a sequence of tests such as measurement of EMG signals from hip and leg muscles that are responsible for flexion and extension in the sagittal plane with and without the exoskeleton device assisting would help better understand and quantify muscular effort. This would give us an estimate as to how much less or more the muscles are working during the gait cycle and could be vital for therapists to analyze the patient's rehabilitation.

APPENDIX

SUBJECT TEST DATA

Figures 9.1 to 9.5 depict the gait profile for all five test subjects during variable speed testing in assist mode, showing stride by stride profiles. The gait angle, torque and power are plotted with respect to percent of gait where 0% and 100% correspond to heel strike. Figure 9.6 shows a side by side comparison of averaged gait profile for all speeds of constant walking speed testing.

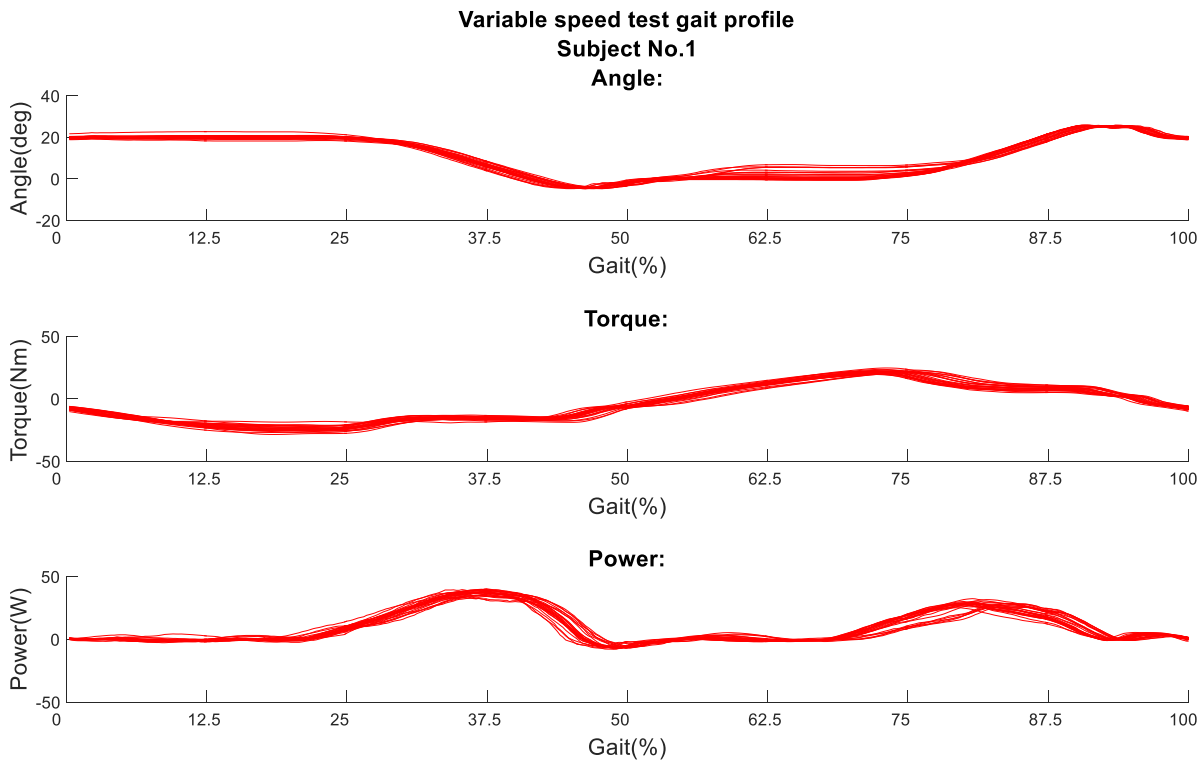


Figure 8.1 Variable speed testing gait profile- Subject 1

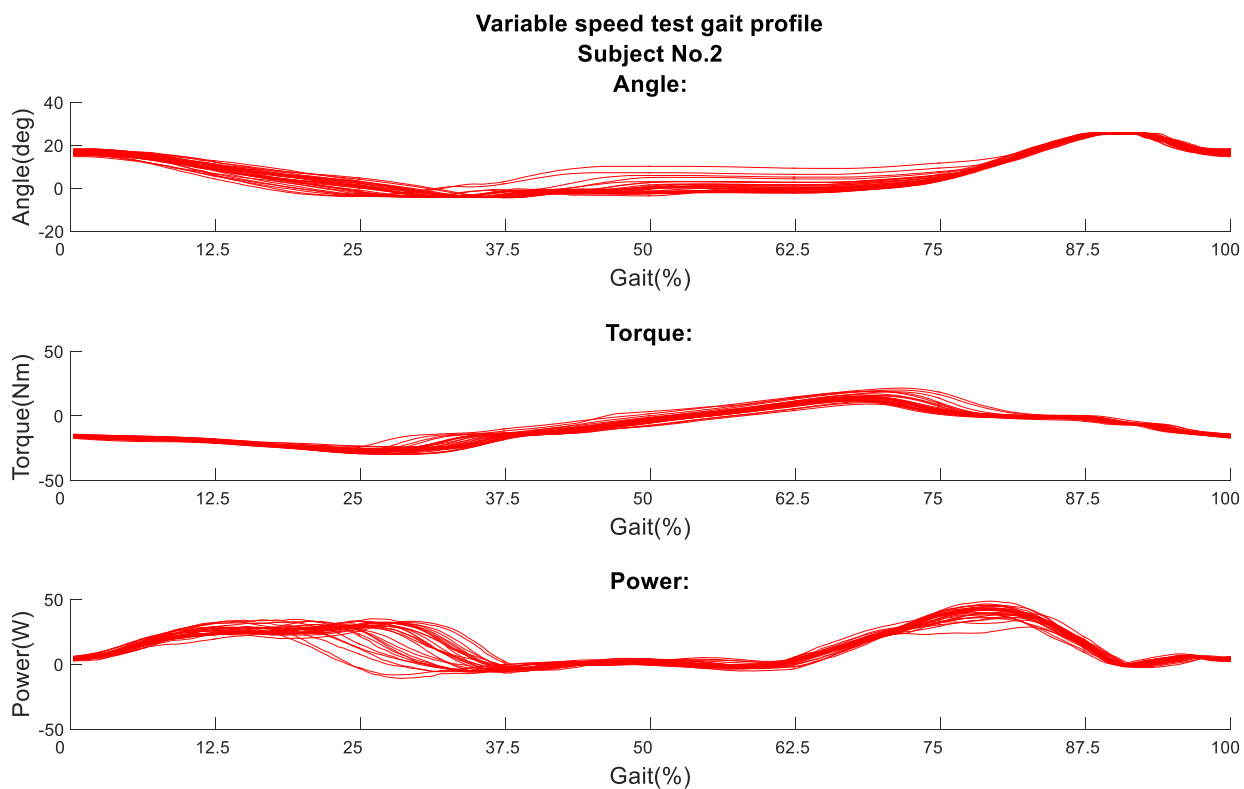


Figure 8.2 Variable speed testing gait profile- Subject 2

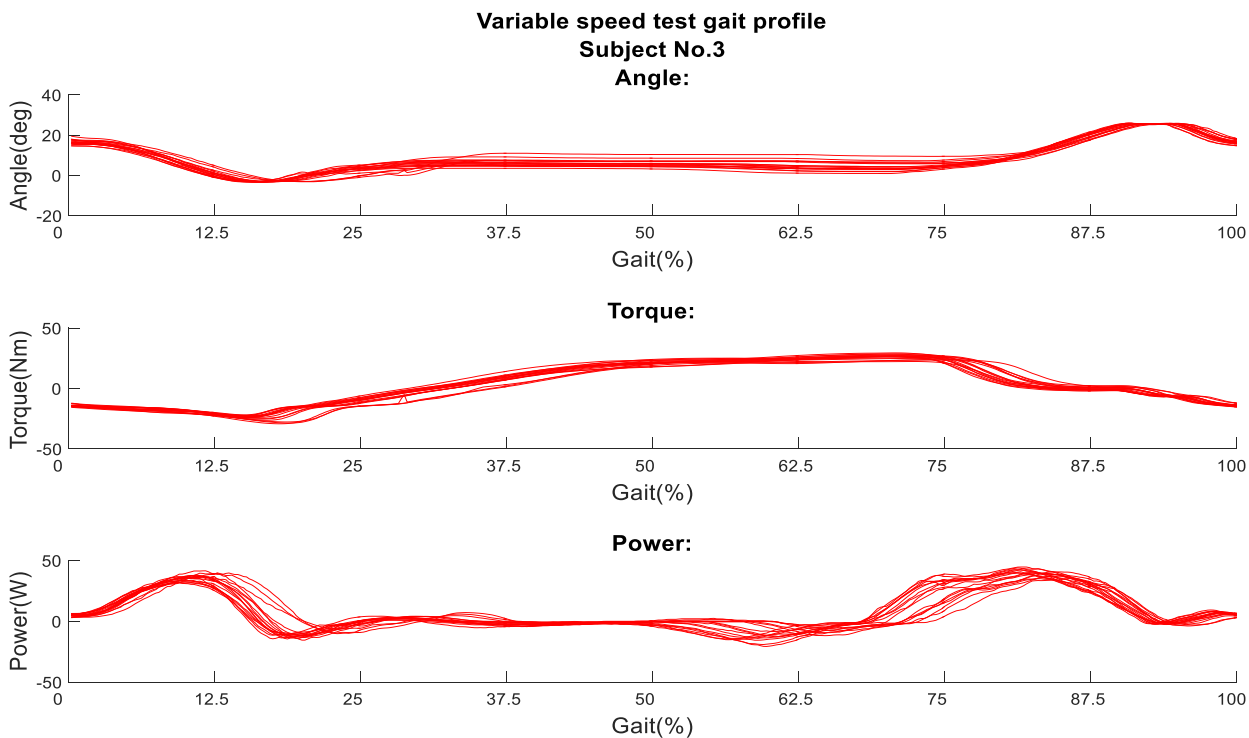


Figure 8.3 Variable speed testing gait profile- Subject 3

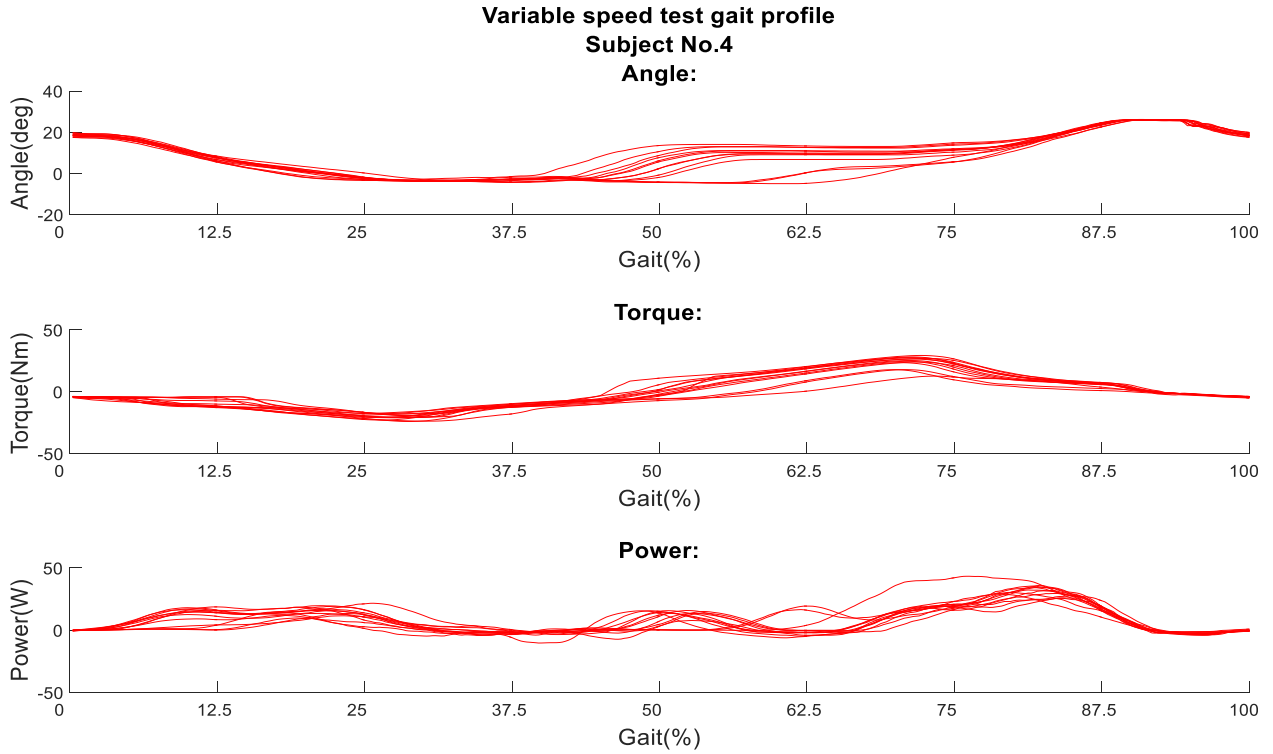


Figure 8.4 Variable speed testing gait profile- Subject 4

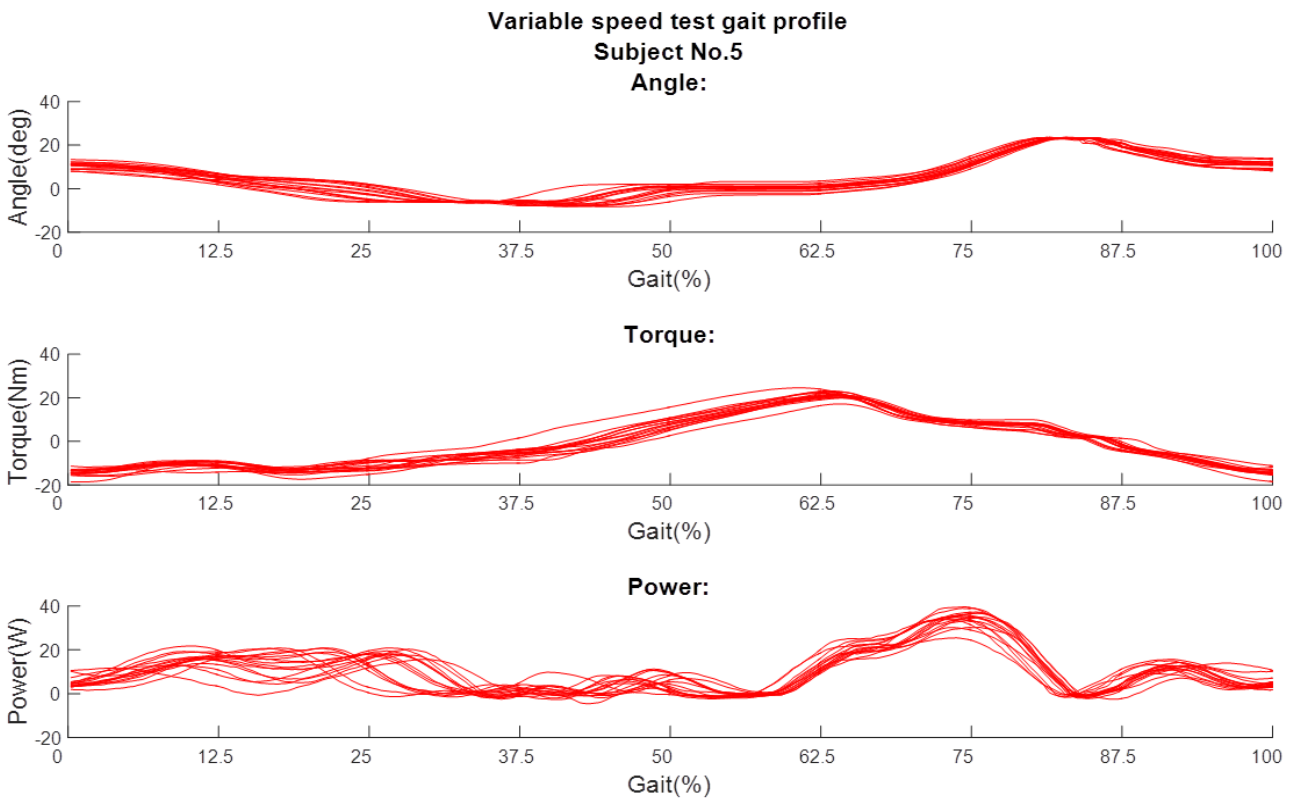


Figure 8.5 Variable speed testing gait profile- Subject 5

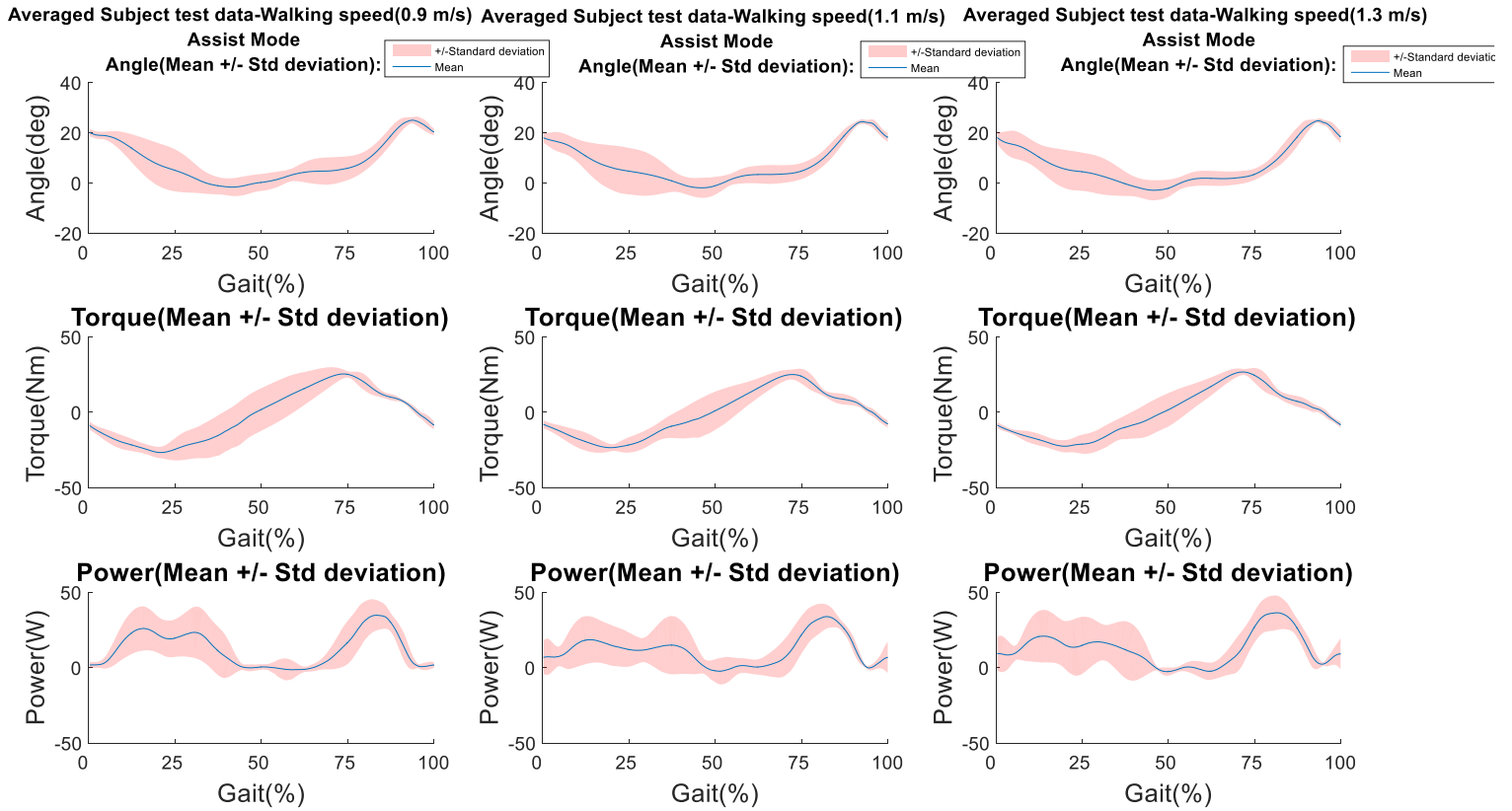


Figure 8.6 Averaged constant speed testing- Speed comparison (0.9 m/s, 1.1 m/s, 1.3 m/s)

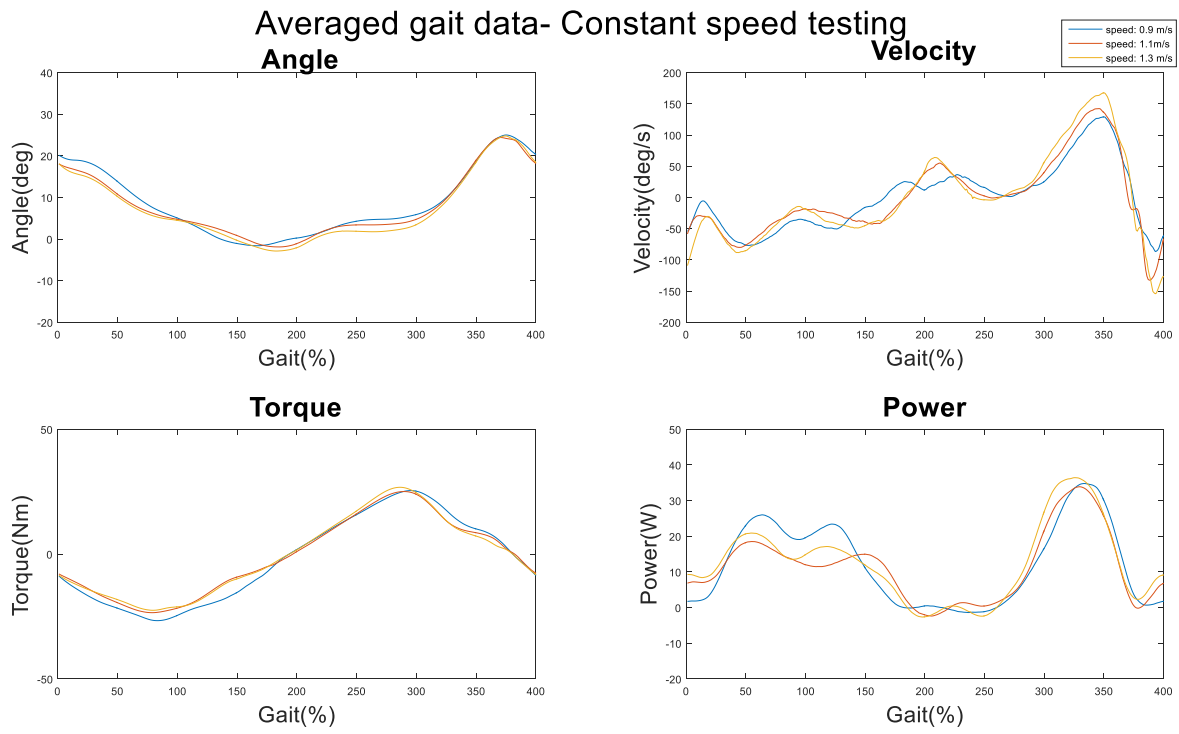


Figure 8.7 Averaged constant speed testing- Speed comparison overlay (0.9 m/s, 1.1 m/s, 1.3 m/s)

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