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Chapter 7

Design and Motion Control of a Lower Limb Robotic Exoskeleton

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Additional information is available at the end of the chapter

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Abstract

This chapter presents the results of research work on design, actuator selection and motion control of a lower extremity exoskeleton developed to provide legged mobility to spinal cord injured (SCI) individuals. The exoskeleton has two degrees of freedom per leg. Hip and knee joints are actuated in the sagittal plane by using DC servomotors. Additional effort supplied by user's arms through crutches is defined as user support rate (USR). Experimentally determined USR values are considered in actuator torque computations for achieving a realistic actuator selection. A custom-embedded system is used to control exoskeleton. Reference joint trajectories are determined by using clinical gait analysis (CGA). Three-loop cascade controllers with current, velocity and position feedback are designed for controlling the joint motions of the exoskeleton. A non-linear ARX model is used to determine controller parameters. Overall performance and an assistive effect of WSE-2 are experimentally investigated by conducting tests with a paraplegic patient with T10 complete injury.

Keywords: exoskeleton, legged locomotion, motion control, wearable robot

1. Introduction

Paraplegia is impairment in motor or sensory function of the lower extremities. One of the most significant impairments resulting from paraplegia is the loss of mobility. In addition to impaired mobility, the inability to stand and walk entails severe physiological effects, including muscular atrophy, loss of bone mineral content, frequent skin breakdown problems, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation [1]. Spinal cord injury (SCI) is the most important reason of paraplegia and

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commonly referred to as either complete or incomplete. In a "complete" spinal cord injury, all functions below the injured area are lost, and patient has little hope of functional recovery. An "incomplete" spinal cord injury involves preservation of motor or sensory function below the level of injury in the spinal cord [2]. Most patients with incomplete injuries can recover some functions after successful therapies. Using robots in a rehabilitation process is quite reasonable since the physical therapy consists of time-consuming repetitive movements. Robots can record quantitative measures of improvements in addition to great motion repeatability. Using robotic exoskeletons in physical therapy can provide further motivation to patients by offering more realistic movements. Furthermore, powered exoskeletons can be used as a movement assistant for subjects affected by permanent movement disorders such as complete SCI. Several exoskeletons have been developed for rehabilitation of paraplegic patients. In general, they can be divided into two categories: the first type exoskeletons are based on a gait orthosis and a weight support system in combination with a treadmill [3]. The total weight of orthosis and user is carried by a weight support system. Powered orthosis only applies the force required to complete movements of patient on impaired limbs and move the patient's leg in a normal gait pattern. These systems have some drawbacks. They have a limited workspace and can be used only in clinical environments. Furthermore, these systems hold the patient's pelvis fixed, and this causes the changes in gait kinematics. Although these systems are appropriate for strengthening exercises they do not have any contribution to balancing problem. Lokomat [4–7], LOPES [8–10] and ALEX [11–13] all fall into this category. The second type exoskeletons are ambulatory devices. These systems carry their own weight in addition to user's weight. They offer users limitless workspace. Furthermore, they can be used as a movement assistant in daily activities for subjects affected by permanent movement disorders. These exoskeletons can help to patients for improving necessary motions in order to maintain balance besides the strengthening exercises. Actuator selection and controller design of these exoskeletons are more challenging due to compact design, less power consumption and safety requirements. The following literature review focuses on the second type exoskeletons. Hybrid assistive limb (HAL) [14–16] is developed for the purposes of rehabilitation or living support of people who have disorders in the lower limb and whose legs are weakening. Full body versions of HAL are also developed for heavy labour and rescue support. The lower body model of the device weighs about 12 kg and the full body model weighs about 23 kg. The batteries of HAL's rehabilitation purposed model can provide the necessary power for 60–90 min in normal operation. Hip and knee joints of HAL are powered in sagittal plane Actuators are composed of DC servomotors and harmonic drive gears. A hybrid control system of HAL consists of an autonomous posture controller and a power-assisted controller. The intended motion of the user is determined by using electromyography (EMG) sensors and ground reaction force sensors. The drawback of this control system is that it requires a process of adaptation and adjustment to a specific user. ReWalk [3, 17] has two different models developed for rehabilitation and life support of SCI patients. Hip and knee joints of ReWalk are powered in the sagittal plane by using DC motors. It comprises a wearable brace support suit, which integrates DC motors, rechargeable batteries, sensors and a computer control system. The device is about 20 kg weight. Pre-programmed motion control strategy is used to control of ReWalk. Changes in the user's centre of gravity are used to initiate and maintain walking processes. The user also has a remote control placed in his/her arm for

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selecting different tasks, such as sit-to-stand or climbing stairs. The Vanderbilt exoskeleton [18, 19] is developed to provide gait assistance to patients with spinal cord injuries. The device weights about 12 kg. Hip and knee joints are actuated by brushless DC motors. Control is based on postural information measured on the device. A lithium polymer battery of 29.6 V and 3.9 Ah brings 1 h of autonomy for continuous walk with the device at a speed of 0.8 km/h. Ekso is a gait training exoskeleton intended for medically supervised use by individuals with various levels of paralysis or hemiparesis. Ekso weights approximately 20 kg and has a maximum speed of 3.2 km/h with a battery life of 6 h. It can execute sit-to-stand and stand-to-sit operations and walk in a straight line. Ekso uses a gesture-based human-machine interface to determine the user's gestural intentions and then acts accordingly. No studies have been published about Ekso for discussing its efficacy. WSE [20] is developed to support walking of partially or entirely disabled individuals. The total weight of WSE is about 18.5 kg. A 24V DC motors powered by Li-Po battery pack are used in actuation of WSE. The pre-programmed motion control strategy is used to control of WSE. Adaptive network-based fuzzy logic controllers are used to control of joint motions of WSE. Changes in the user's centre of gravity are used to initiate and maintain walking processes. The Li-Po battery pack of WSE can provide the necessary power for about 3 h in normal operation.

In this chapter, mechanical design, actuator selection and controller design of the second generation prototype of WSE (WSE-2) is described. Additional effort supplied by user's arms through crutches is considered in required torque computations in order to realize realistic actuator selection. A non-linear ARX model of the exoskeleton is created and used in order to determine the best controller parameters. The assistive effect of WSE is experimentally investigated. WSE was worn by a paraplegic patient with T10 complete injury during the experiments. The 78 kg weight patient was successfully walked with the speed of 0.5 cycle/s.

2. Mechanical design

Basic working principle of a lower extremity exoskeleton is transferring the user's weight to the ground by creating a force path between the user and the ground. Thus, an exoskeleton eliminates the effects of gravity on user. So, the body of an exoskeleton must be strong enough to carry both its own weight and the weight of the user. But at the same time, it should be as light as possible due to ergonomics, small actuator usage and low power consumption requirements. In addition, critical parts of an exoskeleton should be easily adjusting for adapting to different sized users. The second critical point in the design of an exoskeleton is the choice of actuators. The selected actuators must be capable of meeting the speed and moment requirements for the targeted motions, as well as being compact and lightweight as possible. In addition, an exoskeleton should be able to perform all the targeted movements in a manner that will provide the least inconvenience to the user. In order to achieve this, degree of freedom and range of motion of the joints must be selected properly. In fact, increasing the joint's degree of freedom and range of motion provides more comfortable using. However, releasing all degrees of freedom (DOF) of joints is not safe for the users who have lost their muscular activity in their legs since it causes involuntary movements. For devices in contact

with the user, safety is a very important criterion. Safety measures are more important for exoskeleton applications where the user does not have sufficient mobility. Therefore, all electrical and mechanical safety precautions should be taken into consideration in exoskeleton design for preventing the user from damaging. In this study, it was tried to design a lightweight, ergonomic and safe exoskeleton in accordance with the above-mentioned design criteria.

Design requirements of an exoskeleton can be determined by a prior analysis of human motion since they are required to perform similar tasks with human body. Clinical gait analysis (CGA) is one of the best tools for determining the required degrees of freedom (DOF), joint motions and joint torques of lower extremity exoskeletons. In clinical gait analysis, motions of specific points on the limbs are collected in the form of CGA data via video motion capture. So, all the joint kinematics during a walking cycle can be obtained in the form of CGA data. Then, joint torques which required for a walking cycle can be determined via dynamical equations that include joint kinematics, limb masses and inertias. In this study, CGA data obtained from CGA normative gait database of Hong Kong Polytechnic University [21] is used in mechanical design and an actuator selection process of WSE.

Degrees of freedoms and motions of human extremities are generally defined in standard anatomical planes, as shown in **Figure 1**. Generally, human lower extremities are modelled with 7 degrees of freedom (DOF) (3 DOF at the hip, 1 DOF at the knee and 3 DOF at the ankle) in standard anatomical planes, as shown in **Figure 2**.

The events occurring in each anatomical plane during the walking cycle can be briefly summarized as follows:

Sagittal plane: Extension of the hip joint ensures to advance the body. Flexion of the knee joint ensures the shock absorption at heel contact. Flexion of the knee joint in swing phase shortens the leg length and allows the foot clearance. Dorsiflexion and plantarflexion of the ankle joint during the stance phase ensure a more comfortable walking by moving the foot pressure point from heel to toe.

Coronal plane: Projection of the body's centre of gravity must be stay inside the footprint for a stable walking. In the coronal plane, abduction and adduction of the hip joint slide the centre of gravity for ensuring body to stay inside the footprint. Abduction and adduction of the ankle joint are necessary for adapting foot to rugged terrains.

Transverse plane: Medial and lateral rotation of the hip and the ankle joints ensure body to change the direction of movement.

In brief, movements in the sagittal plane ensure body to move forward, movements in the coronal plane provide to balance body and movements in the transverse plane ensure to change the direction of motion.

The degrees of freedom of an exoskeleton must be in accordance with human anatomy for ensuring a comfortable use. Increasing the degrees of freedom also increases the comfort. But disabled individuals cannot prevent some unintended motions during walking since they have a weak control on leg muscles. So, some degrees of freedom must be locked for the exoskeletons designed the use of disabled users. Although this approach reduces the users

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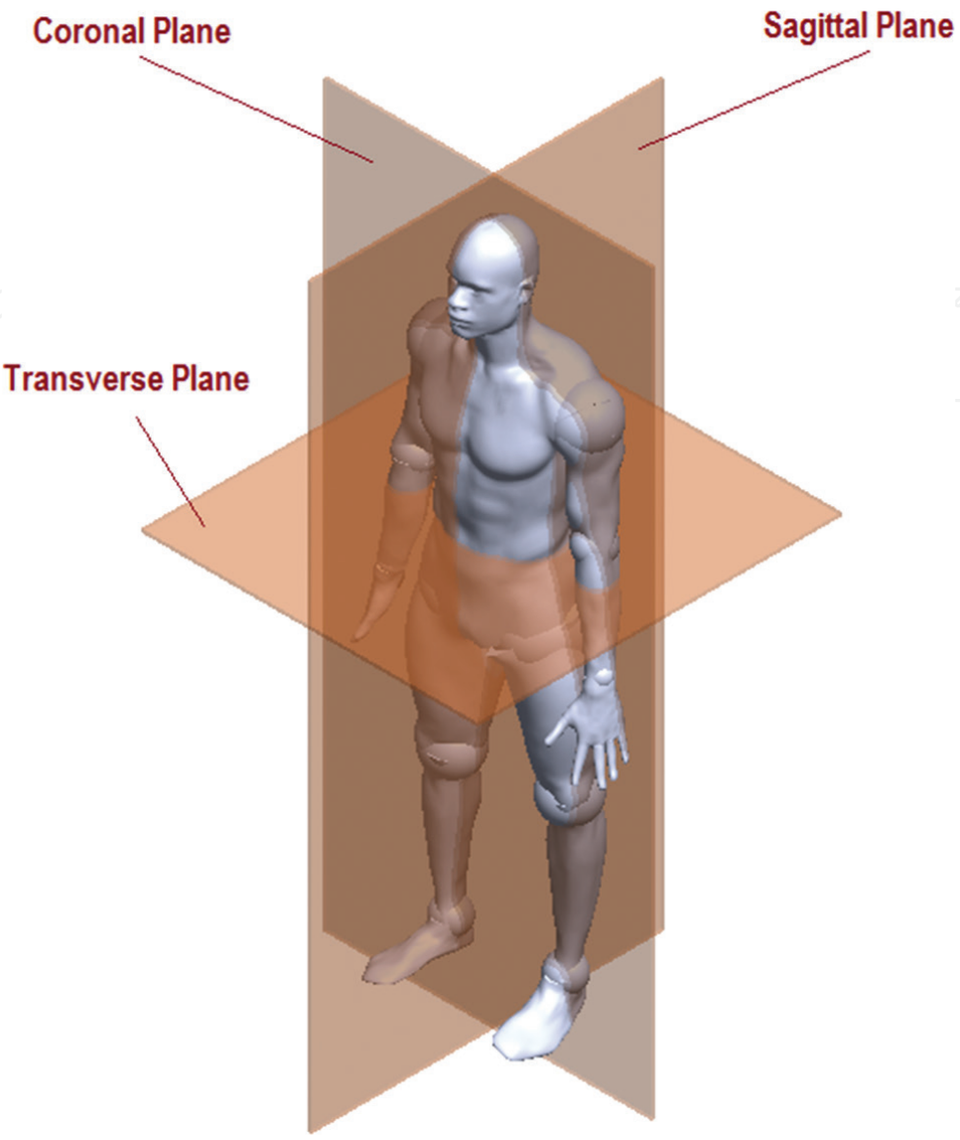


Figure 1. Standard anatomical planes.

comfort, it is necessary to ensure a safe and stable walking. For the user safety, all degrees of freedom of WSE-2 in coronal and transverse planes are locked. But, the degrees of freedom in sagittal plane, which are required to advance the body, are released. In addition, degree of freedom of the ankle joint in the sagittal plane was limited within a certain range. Under these conditions, users are required to use crutches to control the body balance and walking direction since the degrees of freedom of WSE-2 in coronal and transverse planes are locked.

Another point that should be considered in design of an exoskeleton is the determination of proper motion ranges according to targeted movements. Walking is the fundamental movement selected for WSE-2. In addition, WSE-2 should be proper to sitting and standing up motions for daily use. In accordance with these targeted movements, motion ranges of the hip joints are selected as 100° in flexion and 17° in extension, while the motion ranges of the knee joints are selected as 100° in flexion and 0 in hyperextension.

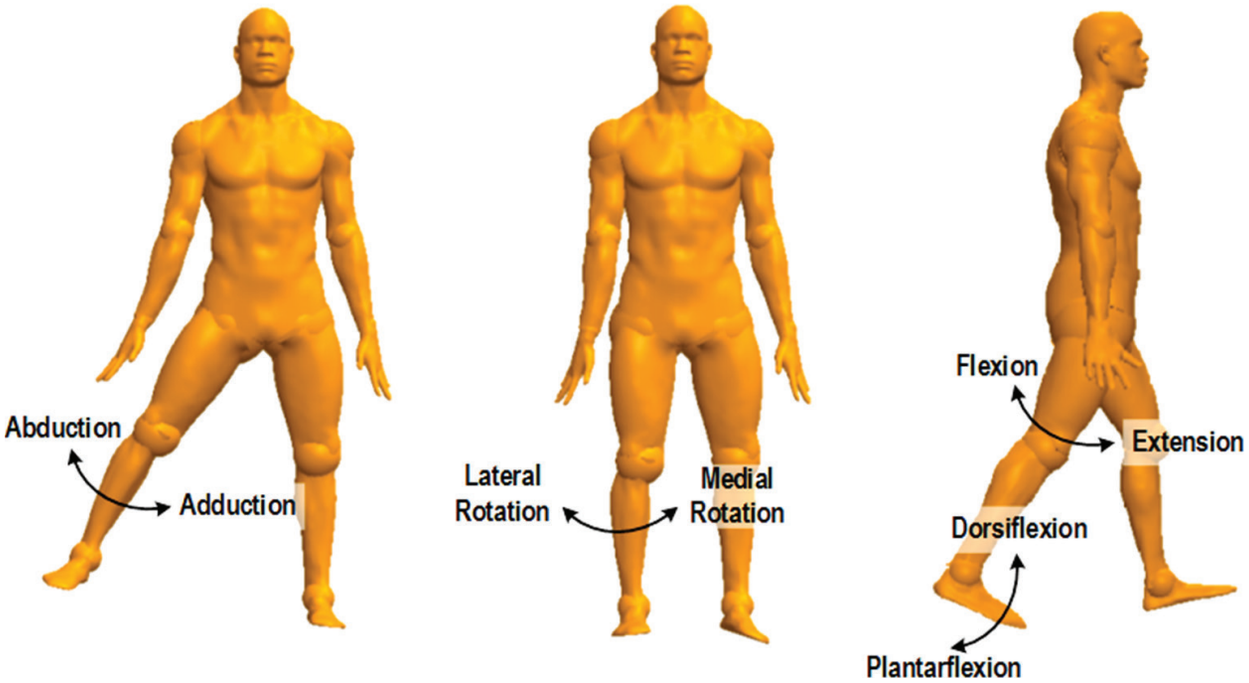


Figure 2. Human degrees of freedom in standard anatomical planes.

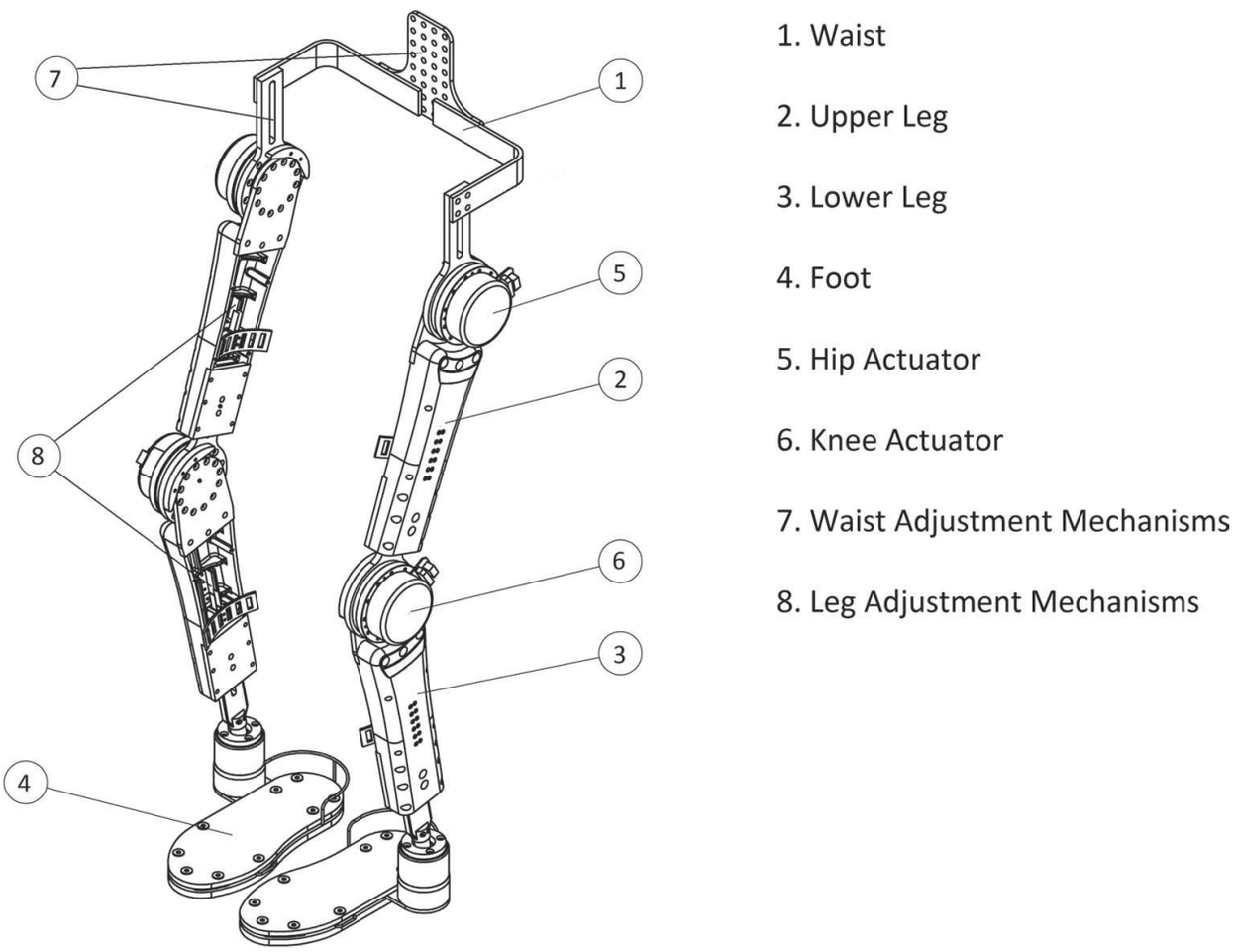


Figure 3. Mechanical design of WSE-2.

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Final design of WSE-2 is consisted of waist, upper and lower legs, hip and knee joints and feet, as shown in **Figure 3**. Length adjustment mechanisms of WSE-2 ensure the adjustment of upper legs, lower legs and waist so as to fit to users with different body sizes. In this way, WSE-2 can be comfortably used by the male users whose height is between 1.70 and 1.86 m and by the female users whose height is between 1.67 and 1.86 m. Furthermore, waist adjustment mechanism makes WSE-2 usable for the users of almost all size whatever their weights are. Parts of WSE-2 exposed to light loads were made of polyamide (P6), while the other parts exposed to high loads were made of aluminium 7075. The total weight of WSE-2 excluding the actuators is 12 kg. WSE-2 is designed to have a two-step safety system, both mechanical and electronic for granted the safety of its users. First, proximity limit switches which cut off the power of the actuators in the case of excessive rotation are placed in the joints. Furthermore, mechanical safety apparatuses are designed and mounted on the joints to prevent excessive rotation, in the event of a malfunction in the limit switches.

3. Actuator selection

Actuators used in exoskeleton applications are required to provide high moments while operating in high speeds. This requires the use of larger and heavier actuators. However, the available area around the joint is too limited for connecting the actuators. So, it is necessary to select the most compact actuators that can provide the required moment-speed values for the targeted movements. Velocity-moment characteristics of the joints for targeted movements should be known for the selection of proper actuators. The moment-velocity values of the joints can be determined from the CGA for walking, which is the fundamental targeted movement of WSE. Variation of joint angles and normalized joint moments during a gait cycle is available in CGA data format. Joint velocities appropriated to targeted walking velocity can be computed from the joint angle variations obtained from CGA. Required joint moments for targeted movements can be calculated from the normalized joint moments obtained from CGA by using the relation

$$T_R = T_N . m_T \tag{1}$$

where T_N is the normalized joint torque and m_T is the total mass (user + exoskeleton) carried by the actuators. In case the user supplies an additional effort by using crutches, Eq. (1) should be modified since the user transfers a certain amount of total weight to the ground through the crutches and the total weight carried by the exoskeleton is reduced by a certain ratio during the stance phase. This ratio was defined as "user support rate" (USR) in Ref. [20]. In the presence of additional effort, Eq. (1) can be rewritten as follows:

$$T_R = T_N(m_T . k_s) \text{ Stance Phase} \tag{2}$$

$$T_R = T_N . m_T \text{ Swing Phase} \tag{3}$$

where k_s stands for USR. An experimental study performed for the determination of a user support rate and the calculation of k_s coefficient is given in Ref. [20]. According to the experimental results, the total load carried by the exoskeleton is reduced about 47.7% in the case of

users use crutches. Comparisons of computed angle, torque and power characteristics of hip and knee joints for a walking cycle in cases of USR = 0 and USR = 0.45 are given in **Figure 4**. Masses of the user and the exoskeleton used in calculations are 78 and 18 kg, respectively, and the normalized joint moments are obtained from CGA.

Actuators of WSE-2 are required to provide moment, velocity and power requirements presented in **Figure 4**. Moreover, it should be as compact and lightweight as possible. Maxon EC 90 flat servomotor coupled by CSD series harmonic reducer is selected for the actuation of WSE-2. Technical specifications of selected actuators are given in **Table 1**.

Moment-velocity characteristics of the actuators are compared with the required moment-velocity characteristics of the targeted movements in order to evaluate the suitability of the selected actuator. Required moment-velocity characteristics for two different USR values (0 and 0.45) are determined by using the angle and moment characteristics given in **Figure 4**. Characteristic curves of the selected actuator are obtained from the manufacturer's catalogue. Comparison of the required moment-velocity curves with the characteristic curve of the

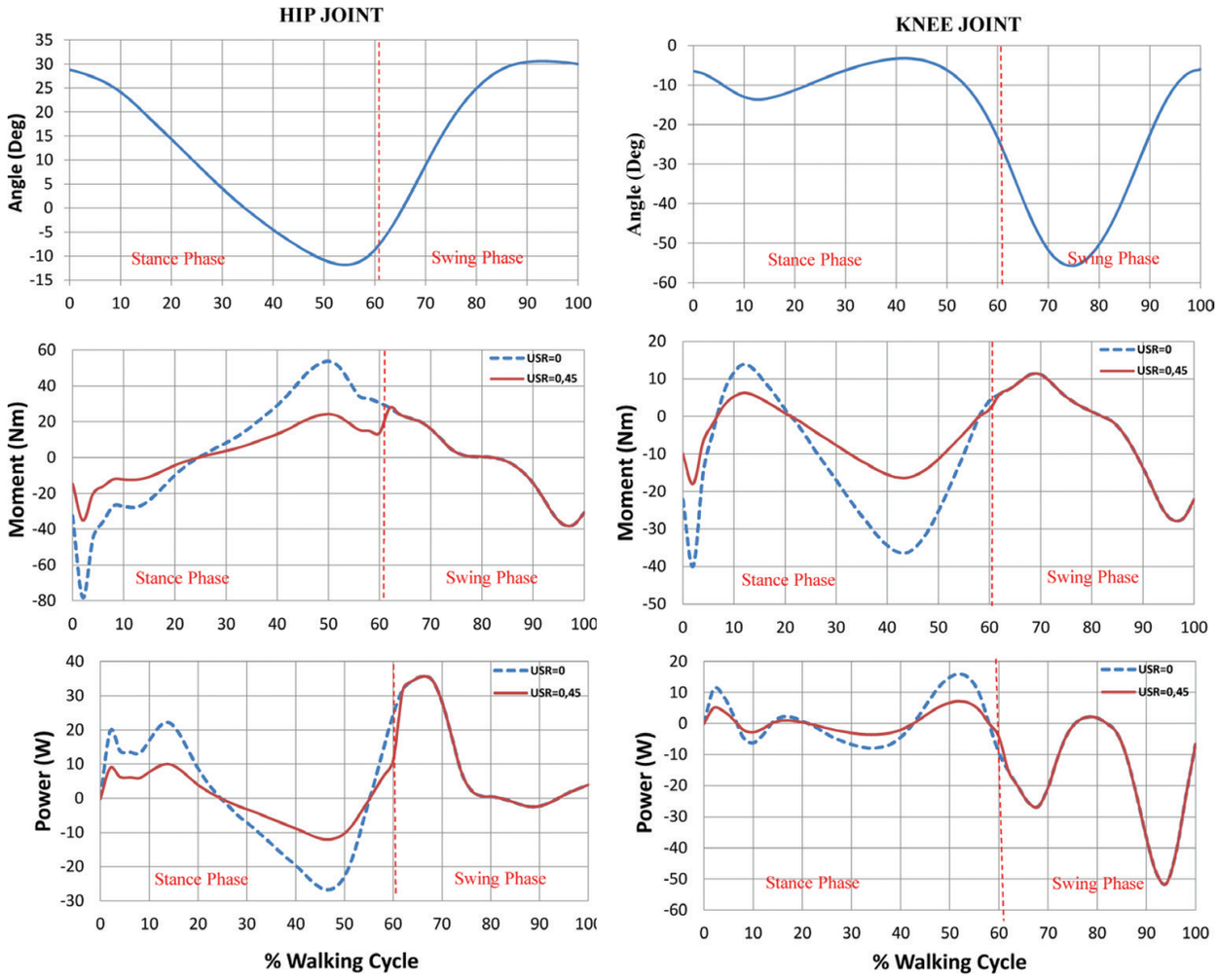


Figure 4. Comparisons of computed angle, torque and power characteristics of hip and knee joints for a walking cycle in the cases of USR = 0 and USR = 0.45.

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Servomotor	
Nominal output power (W)	90
Nominal voltage (V)	24
Nominal current (A)	5.39
Nominal torque (Nm)	0.387
Stall torque (Nm)	4.67
Nominal speed (rpm)	2650
Maximum speed (rpm)	3190
Weight (kg)	0.6
Gear	
Gear ratio	100
Moment of inertia (kg m^2)	0.282 (10^{-4})
Weight (kg)	0.91

Table 1. Specifications of selected actuators.

actuator is given in **Figure 5**. The required moment-velocity characteristics are determined by assuming the total weight carried by WSE is 94 kg and the walking speed is 0.5 cycle/s.

As shown in **Figure 5**, while a required moment-velocity curve for the hip joint exceeds the short-term operation limit of the actuators for USR = 0, it stays in the continuous operation region of the actuators for USR = 0.45. On the other hand, a required moment-velocity curve for the knee joint stays in the continuous operation region of the actuators for both USR = 0 and USR = 0.45. Furthermore, maximum moment requirement is decreased about 45% for hip joints and 30% for knee joints in the case of USR = 0.45. Consequently, it is verified that selected actuators can meet moment-velocity requirements of WSE up to 78 kg user weight and 0.5 cycle/s walking speed.

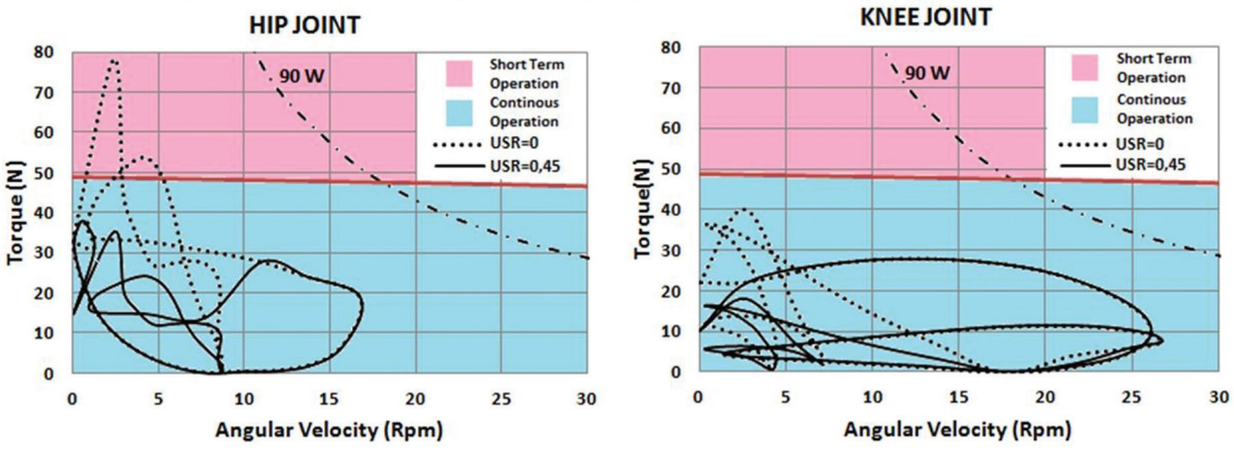


Figure 5. Comparison of the required moment-velocity curves with the characteristic curve of the actuator.

4. Controller design

4.1. Pre-defined motion control (PMC)

Control techniques to be used in control of exoskeletons are generally characterized by the method used in the determination of motion intended by the user. Most of the control techniques used in exoskeletons need interaction signals between the user and device for determining the intention of user. However, majority of paraplegic patients are not capable of generating an effective user-exoskeleton interaction. Another method that can be used in the determination of user's intentions is EMG signals. However, there are technical difficulties in implementing EMG-based control techniques since the EMG signals are extremely noisy and necessitate extensive signal conditioning. A pre-defined motion control (PMC) technique is an alternative method for controlling exoskeletons. In PMC control, the user itself selects the intended motion. The PMC technique bears the advantages of reduced computational complexity, hardware complexity and sensor requirements. So, the PMC technique is selected to be used in control of WSE considering the advantages it offers. In the implementation of PMC, reference motion database which includes the information of sitting, standing up and walking motions is created by using CGA. Cascade PID controllers are designed for motion control and tracking error compensation of hip and knee joints.

4.2. Three-loop cascade control

A control system scheme of WSE-2 is shown in **Figure 6**. Maxon EC 90 flat servomotor coupled by a harmonic drive CSD series harmonic reducer is used in actuation of hip and knee joints. Athena ATHM 800 xPC target compatible PC 104 expandable single board computer combining high integration CPU and high accuracy data acquisition is used in control of WSE. 25.9 V 10 Ah Li-Po battery pack is used as power supply. Limit switches placed at the joints are used to prevent excessive rotations.

Three-loop cascade control structure used in WSE-2 is shown in **Figure 7**. The cascade controller comprises three feedback loops: a current loop, a velocity loop and a position loop. Both the current loop and the velocity loop are the inner sub-control loops, and the position loop is the primary control loop. The current loop is used to limit the current of actuator by keeping it constant under the maximum allowed value at the start and stop, and it optimizes the variation of current. The velocity loop is used to enhance the ability to resist disturbances in load and to suppress fluctuations in velocity. PI type controllers are used in both the current and the velocity loop. The position loop is used to ensure good dynamic tracking performance and static position accuracy. The PD-type controller is used in the position loop.

The choice of controller parameters for a non-linear system is rather complicated. Linearization of a system is the best way of determining controller parameters. But, it is quite difficult to linearize a highly non-linear system. For this reason, it has been decided to use a linearizable non-linear autoregressive exogenous (NARX) model instead of the

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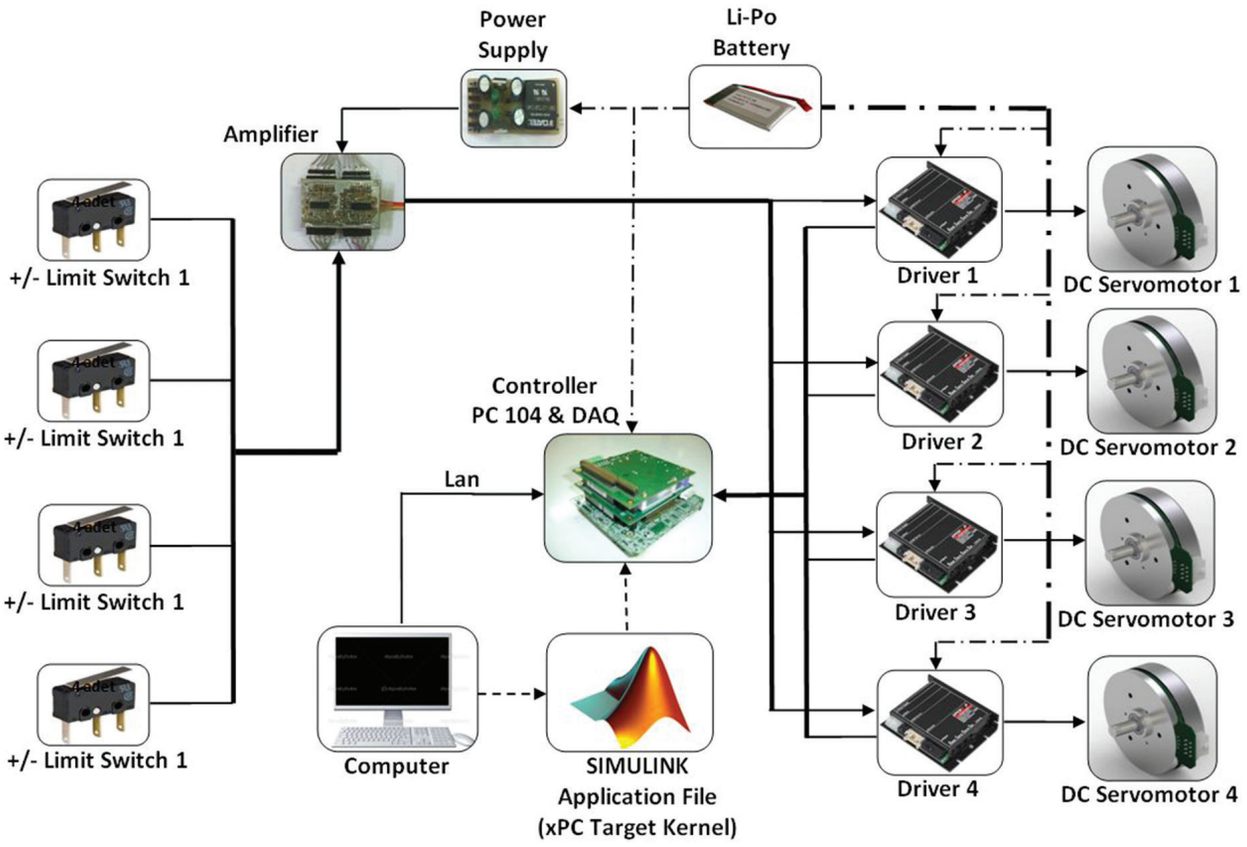


Figure 6. Control scheme of WSE-2.

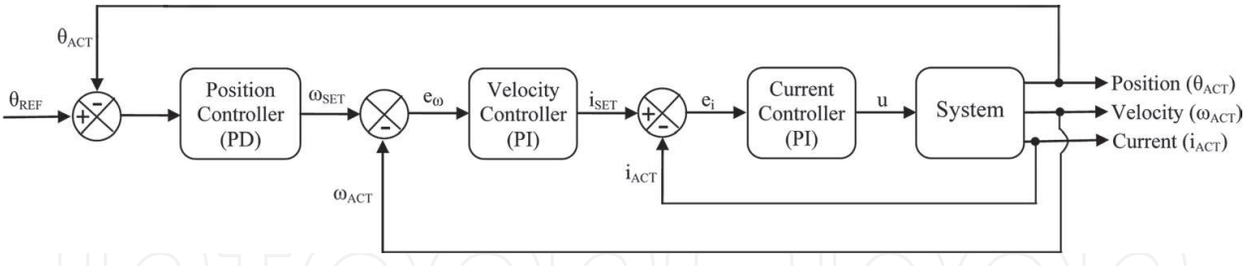


Figure 7. Three-loop cascade control structure of WSE-2.

non-linear model in determination of the controller parameters of WSE-2. Dynamics of hip and knee joints are estimated as an NARX model by using MATLAB system identification toolbox. The NARX model structure given the best performance for system identification of WSE-2 is shown in **Figure 8**. Two-time delayed inputs and outputs are selected as the regressors of NARX models of both hip and knee joints. Sigmoidnet non-linearity estimator with 10 units is used for the hip model while the sigmoidnet non-linearity estimator with six units is used for the knee model. Convergence between the NARX model and the experimental results is 96.58 and 96.78% for hip and knee joints, respectively. Comparisons

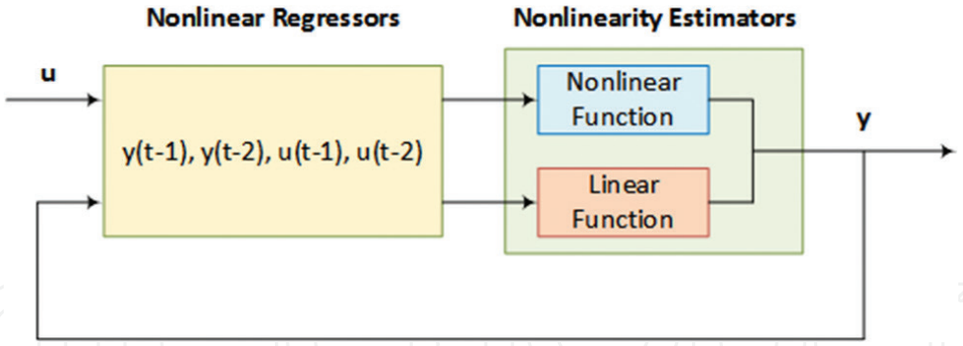


Figure 8. NARX structure of WSE-2.

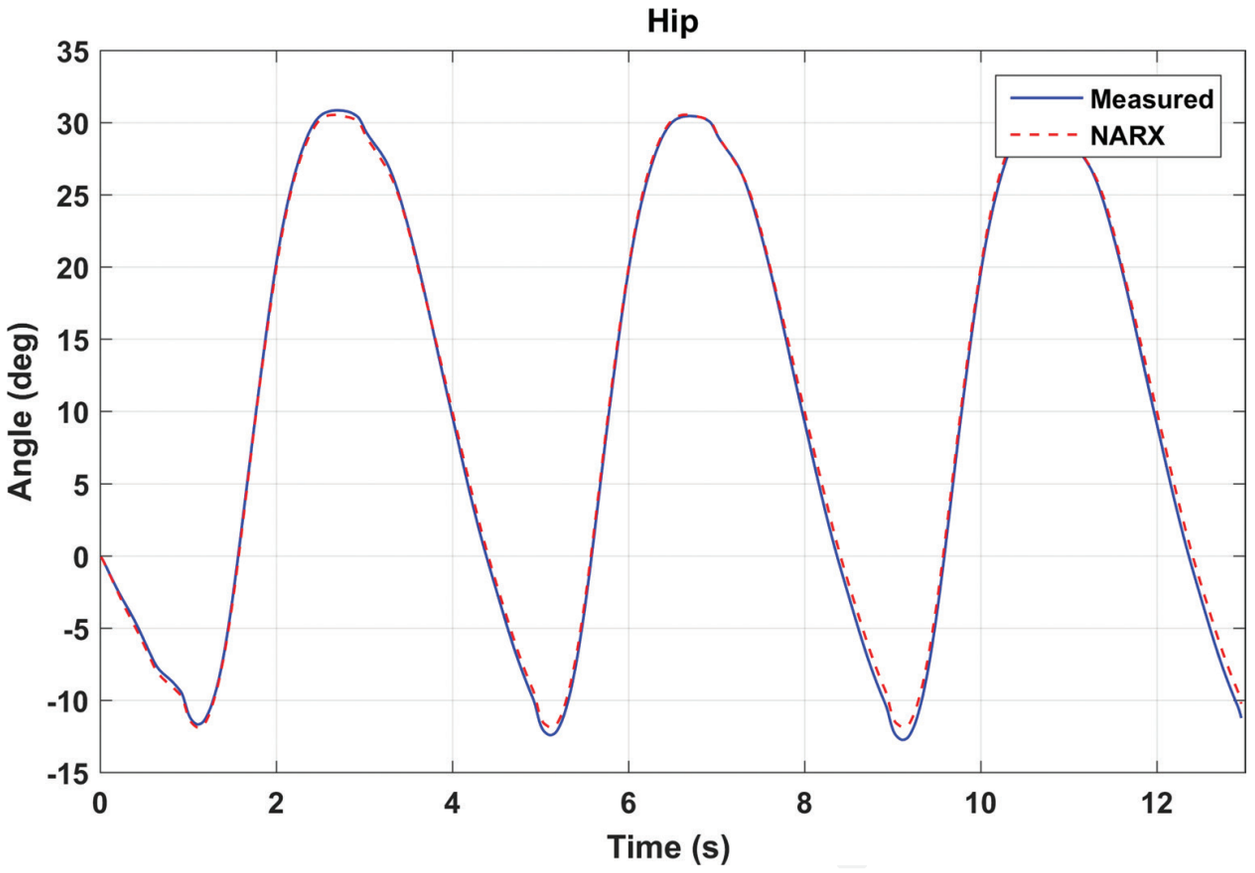


Figure 9. Comparison of the experimental results with the results of NARX model of hip.

of the experimental results with the results of NARX models created for hip and knee joints are given in **Figures 9** and **10**.

Created NARX models are used to determine the parameters of cascade controllers. Controller parameters which given the best simulation results for the hip and knee controllers are presented in **Tables 2** and **3**, respectively.

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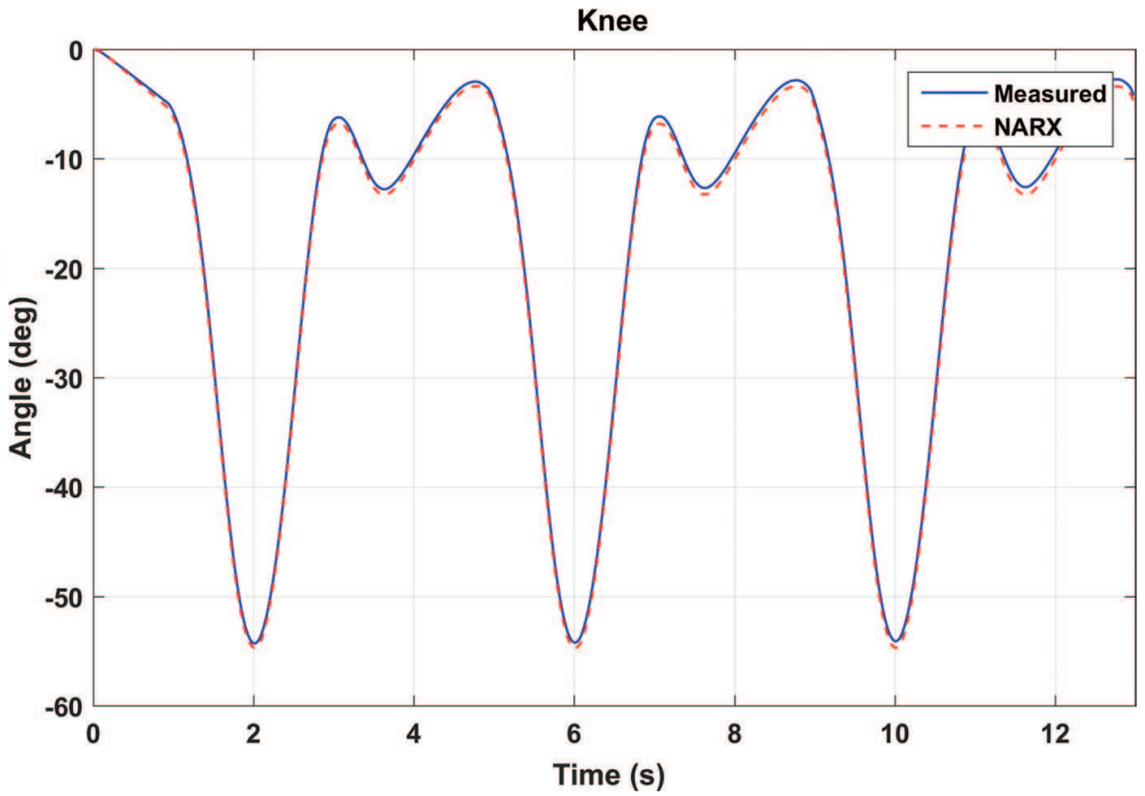


Figure 10. Comparison of the experimental results with the results of NARX model of knee.

Current controller P-gain	215
Current controller I-gain	43
Velocity controller P-gain	13,317
Velocity controller I-gain	1174
Position controller P-gain	0.48553
Current controller sampling period	10^{-6} s
Velocity controller sampling period	10^{-3} s
Position controller sampling period	10^{-3} s

Table 2. Controller parameters for hip actuators.

Current controller P-gain	239
Current controller I-gain	42
Velocity controller P-gain	14,986
Velocity controller I-gain	1217
Position controller P-gain	0.50786
Current controller sampling period	10^{-6} s
Velocity controller sampling period	10^{-3} s
Position controller sampling period	10^{-3} s

Table 3. Controller parameters for knee actuators.

5. Experimental results

Overall performances of WSE-2 and cascade controllers are investigated experimentally as shown in **Figure 11**. WSE-2 is worn by paraplegic patient with a T10 complete injury user. During the experiments, a 78-kg weight paraplegic user is walked with a velocity of 0.5 cycle/s by using underarm crutches.

Comparisons of reference and actual joint angles for hip and knee joints are presented in **Figure 12**. Only two strides are given in the figures since the walking cycles repeated itself in the same manner. Measured joint angles are found to be in good conformance with the reference joint angles of corresponding joints. Designed cascade controllers are evaluated to ensure smooth and stable motion despite the disturbances induced by the user.



Figure 11. Performance tests of WSE-2.

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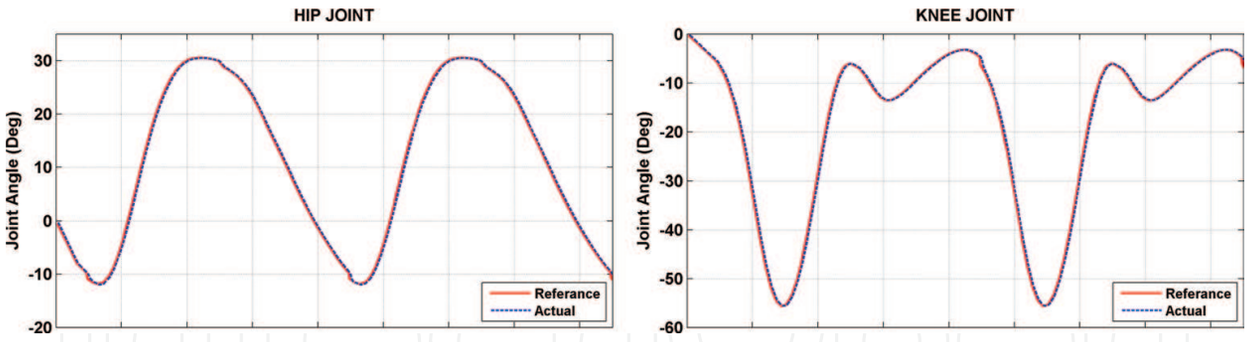


Figure 12. Comparison of reference and actual joint angles.

The experimental results imply that (1) selected actuators are capable of providing the necessary torque and velocity required for a 78 kg weighing complete SCI user walking with a velocity of 0.5 cycle/s, (2) designed cascade controllers provide satisfactory performance in joint tracking control and (3) the assumption that kinematics and dynamics of the exoskeleton is analogous to that of human leg works for WSE-2. Power consumptions of actuators in hip and knee joints are measured during the experiments; power signal is filtered by using a zero-phase digital filter in order to discard possible noise. Plots of filtered and unfiltered actuator power consumption values versus time are presented in **Figure 13**. As seen from figure, the maximum power consumption for hip and knee joints are about 30 and 33 W, respectively. The results show that selected actuators are capable of providing the power requirement for a 78-kg weight paraplegic user walking with a velocity of 0.5 cycle/s.

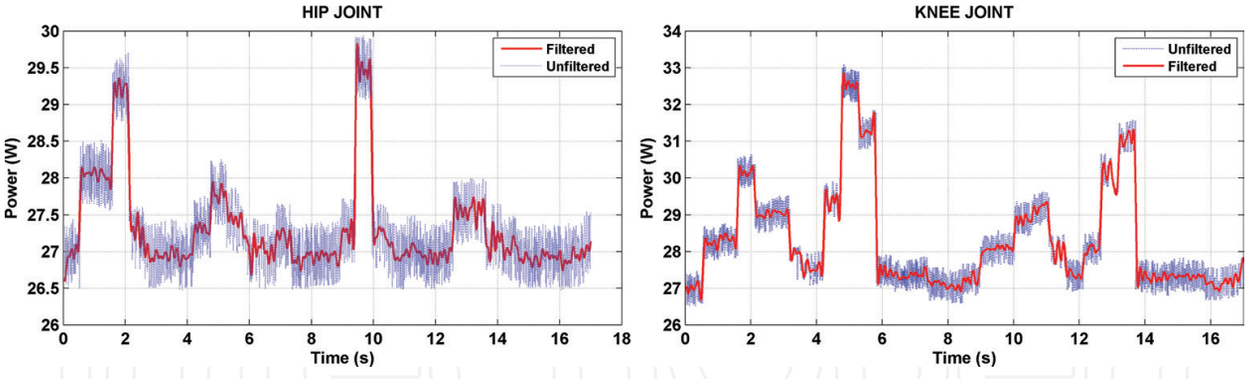


Figure 13. Power consumptions of the hip and knee actuators.

6. Conclusion

Majority of complete SCI patients are not capable of generating sufficient user-exoskeleton interaction required for control action. Subsequently, many control techniques that require user-exoskeleton interaction forces or displacements cannot be used in control of exoskeletons developed for paraplegic individuals. A pre-defined motion control architecture is selected for controlling WSE-2 since it does not require interaction forces or displacements between the user and the exoskeleton. Three-loop cascade-type controllers are designed and used in joint motion control of WSE-2. A non-linear ARX model of WSE-2 is constructed and used to

determine controller parameters. Designed controllers provided good joint angle tracking performance despite disturbances. Consequently, experimental studies conducted with the second generation WSE prototype used by a 78-kg weighing paraplegic user with T10 complete injury showed satisfactory performance in hip and knee joint angle tracking.

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