Chapter 6

An Active Orthosis for Gait Rehabilitation

Gait therapy is vital for restoring neuromuscular control in patients suffering from neurological injuries. Robots can provide prolonged, systematic, and repetitive gait training sessions. Currently available robotic devices use stiff actuators with high end point impedance. This work presents a new compliant robotic gait rehabilitation system. Pneumatic muscle actuators (PMAs) were used for actuation purposes. The robotic device is lightweight and works in perfect alignment with patient's joints. The modeling of robotic device with PMA was performed. Model reference-based adaptive control (MRAC) was used to guide the patient's limbs on physiological gait patterns, and joint torques required to achieve these trajectories were measured. The PMA with the proposed design is capable of providing the required joint torques. Simulation studies are reported.

6.1. Introduction

6.1.1. *Gait rehabilitation*

Neurologic injuries such as stroke and spinal cord injuries (SCI) cause damage to neural system and motor function, which results in lower limb impairment and gait disorders. Patients with gait disorders require specific training to regain functional mobility. Traditionally, manual physical therapy has been used for gait rehabilitation of neurologically impaired patients. Body weight–supported (BWS) manually-assisted treadmill training has been in practice for more than 20 years (Figure 6.1) [FIN 91, HES 95, BEH 00]. It allows the patient to perform a favorable gait for

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greater balance training and longer stance durations compared with over ground gait training [MUR 85, HAS 97, HES 99, PAT 08]. BWS treadmill training has also proven significant improvements in step length, endurance, and walking speed of neurologically impaired patients [HES 95, VIS 98, LAU 01, TEI 01].

The quality of manually-assisted BWS treadmill training is dependent on the therapist's experience and judgment, which varies widely among the therapists. The BWS training requires a team of three therapists to train the patient's limbs and to stabilize the pelvis, which increases the cost of therapy. The training sessions are normally short due to the physical therapist's fatigue. Manually-assisted training also lacks proper methods of recording the patient's progress and recovery.

Figure 6.1. *Manually-assisted BWS treadmill training*

6.1.2. *Rehabilitation robotics*

Automated rehabilitation solutions have been researched lately to overcome the above-mentioned shortcomings of manually-assisted training [COL 00]. Robot-assisted gait training has several advantages over manually-assisted treadmill training. It relieves the physical therapist from the strenuous task of manual assistance and facilitates in delivering well-controlled repetitive and prolonged gait training sessions at a reasonable cost. The physical therapist's role is limited to supervision. The subjectivity of a manual training process is eliminated by providing measurement of interaction forces and limb movements to assess the quantitative level of improvement in gait parameters.

The history of robotic rehabilitation started with the adaptation of industrial robotic manipulators to the field of physical therapy [NAP 89, BOL 95, HOG 00]. Following that trend, various devices have been designed for restoration of upper limb and gait functions. The industrial robotic manipulators are mainly designed for tasks such as pick and place and are inherently stiff and massive. However, robotic rehabilitation devices need compliant and safe human–robot interface [VEN 06, SUG 07]. Subsequently, robots for applying suitable forces and capable of providing a safe interaction with the patients have been developed [COL 00]. Most of these robots are wearable and work in proximity with the patient's limbs. *Active orthosis* is a more common term for these wearable robotic devices. From the studies of human gait biomechanics and manual physical therapy practice, different gait training strategies are incorporated in the robot control schemes to enhance the rate of recovery.

The process for developing, testing, and analyzing the efficacy of robotic gait rehabilitation orthoses involves four stages (Figure 6.2). Stage 1 involves the process of determining kinematic and kinetic constraints for the design of active orthosis. Studies from the fields of clinical gait analysis and human gait biomechanics provide the basic criteria for the design of these robotic devices. Stage 2 is to design the active orthosis, which can be adjusted to patients with different anthropometric parameters. Stage 3 is to select a suitable gait training strategy according to the patient's disability level and phase of rehabilitation. Stage 4 is to evaluate the functional outcomes of robot-assisted gait rehabilitation and adjust the gait training parameters accordingly. The functional outcomes involve the improvement in gait parameters like stride length, stepping frequency, stance duration, and muscle coordination patterns.

The main focus of this work was the development of a compliant active gait training orthosis and a gait training strategy based on adaptive control. The proposed orthosis provides a compliant and safe interaction with the patient, and the adaptive gait training strategy enhances the patient's voluntary participation in the gait training process. The patient's interaction with the active orthosis was estimated using a combined patient-active orthosis dynamic model. Section 6.1 follows with an overview of the biomechanics of human gait to familiarize the reader with the relevant concepts used in the development of active orthosis. A review of existing active gait rehabilitation orthoses and gait training strategies is also provided. Section 6.2 deals with the design of compliant active orthosis. Sections 6.3 and 6.4 present the modeling and controller design of compliant active orthosis, respectively. Section 6.5 presents the simulation results of the proposed orthosis controller. Section 6.6 contains conclusions.

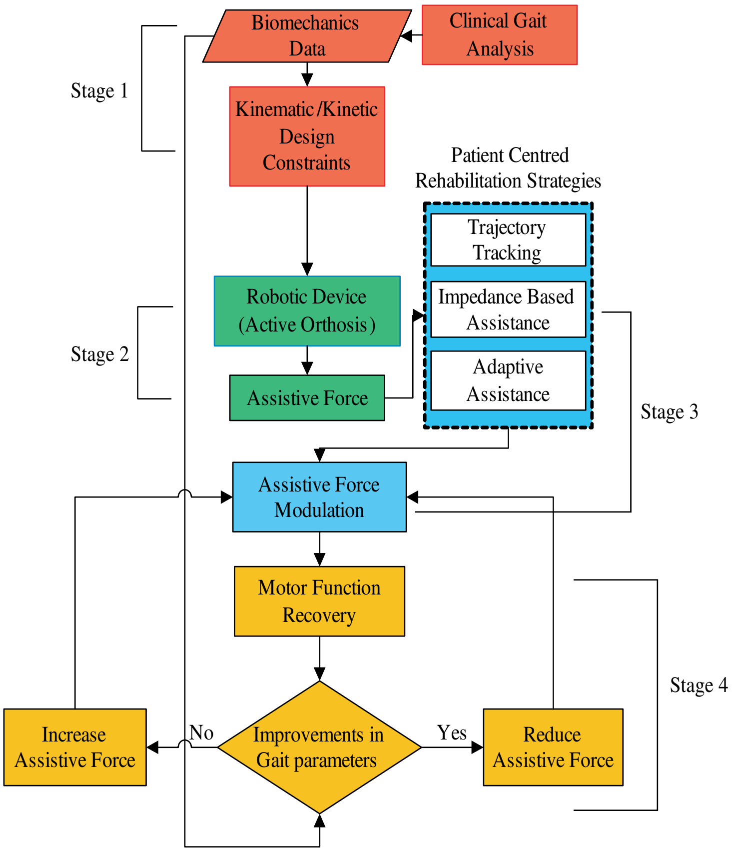

Figure 6.2. *Overview of robotic gait rehabilitation process: Stage 1, Stage 2, Stage 3, and Stage 4*

6.1.3. *Biomechanics of gait*

A background of gait biomechanics is provided in this section to familiarize the reader with the concepts used in the design of active gait training orthoses and training strategies. During the past 50 years, there have been major advancements in the field of biomechanics particularly associated with kinematic and kinetic analysis of human gait [WIN 91, AND 03, SHE 06, SET 07]. Comprehensive knowledge of physiological gait patterns is now available, which has facilitated researchers to design improved robotic orthoses and training strategies for effective motor function recovery. The knowledge of gait biomechanics is also important to determine the efficacy of robot-assisted gait training in gait analysis laboratories [KAO 10, MUL 10].

A *gait cycle* [WIN 90] (Figure 6.3) is the sequence of events from the heel strike of one foot to the subsequent heel strike of the same foot [WIN 90, WIN 91]. It is defined in terms of time interval and usually expressed as a percentage of gait events taking place. Walking consists of repeated gait cycles [WIN 90]. The gait cycle consists of two phases: stance and swing. The *stance phase* is defined by the percentage of gait cycle when the foot is in contact with the ground and the *swing phase* by the time when the foot is in air and not bearing any load. Approximately 62% of the gait cycle consists of stance and 38% of swing phase.

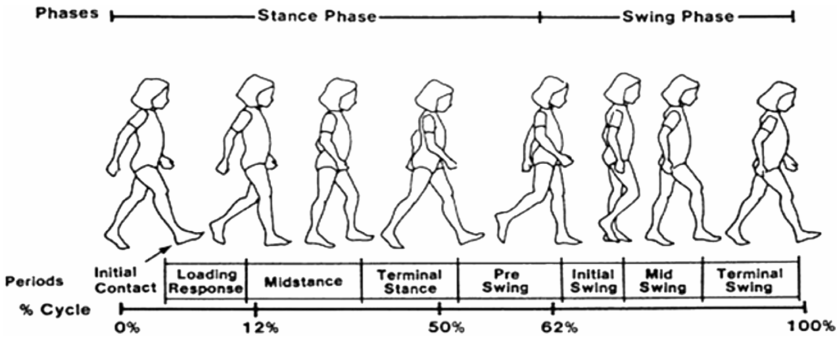

Figure 6.3. *Phases of gait cycle*

Human gait is realized by coordinated inter joint movements of the lower limb. Muscles are used to create moments across these joints. Three planes divide the human body into six parts (Figure 6.4a). The hip joint can provide motions in all the three planes as hip flexion/extension in the sagittal plane, abduction/ adduction in the frontal plane, and rotation in the transverse plane. The knee joint has major rotations in the sagittal plane as flexion/extension and also provides rotations in the transverse plane. The ankle is a complex joint and due to its variable center of rotation, the axes of motion are not simply three Euclidean axes. The important one is plantar/dorsiflexion in the sagittal plane for ground clearance during the swing phase. The sagittal plane joint ranges of motion (Figure 6.4b) and moments contribute most during the gait cycle and are actuated in most of the active gait training orthoses. Gait biomechanics and analysis is an important research area for analyzing the outcomes of robot-assisted gait training [ZIS 07, ALI 09]. The standard procedure involves kinematic data collection (joint angles, velocities, and accelerations) by using reflective markers and motion capture systems, whereas kinetic data (joint moments and power) is obtained by measuring ground reaction forces with the aid of foot plates. Electromyography (EMG) signals are used to judge the activity of various muscle groups in combination with kinematic and kinetic data.

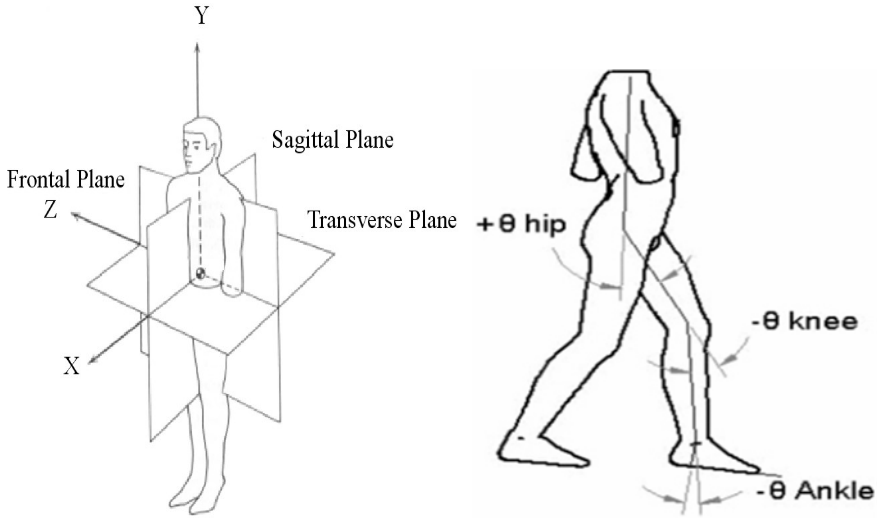

Figure 6.4. *(a) Plane and axis [WIN 91, WIN 90]; (b) joint angles in the sagittal plane [WIN 91]*

6.1.4. *Robot-assisted gait rehabilitation: a review*

Active orthoses are training devices that work in parallel with the human body and have mechanical actuation to apply forces to the human limbs. The history of active orthoses started in late 1970s. Early active orthoses were standard braces with added actuation mechanisms [VUK 74]. Among the first full lower limb active orthosis is the University of Wisconsin prototype [SEI 81]. The orthosis has universal joints at the hip and ankle and provides sagittal plane flexion/extension motions by means of hydraulic cylinders. The remaining degrees-of-freedom (DOFs) are passively held by springs.

The first modern automated BWS treadmill training system Lokomat was developed in the late 1990s and is commercially available. The system has a wearable driven gait orthosis (DGO) having mechanical actuation to power hip and knee sagittal plane rotations [COL 00]. Direct current (DC) motors with a ball screw mechanism are used to power these joints. Dorsiflexion to the ankle joint is provided by passive elastic bands, and the hip abduction/adduction is kept free. DGO works on the assumption that the orthosis joints are in perfect alignment with the patient's joints, and the joint positions are measured with encoders built in DC motors. The physical contact between the DGO and the patient is through two force-torque sensors placed in series with DC motors that move orthosis links. DGO is connected to the treadmill by a rotatable parallelogram linkage to stabilize the patient's trunk.

In this manner, the DGO moves only in a vertical direction, avoiding any sideways tilt of trunk. Active leg exoskeleton (ALEX) was developed at the University of Delaware for gait training of stroke survivors [BAN 09]. ALEX uses gravity-balancing orthosis (GBO) [AGR 04] as its foundation. GBO is a passive device without any mechanical actuation and uses the conventional method of fixing the center of mass by means of a spring mechanism. Linear servo drives have been used on the GBO for providing actuation at the hip and knee joints for flexion/extension rotations in the sagittal plane. Hip abduction/adduction and four trunk rotations are held passive by means of springs.

Lower extremity powered exoskeleton (LOPES) uses a Bowden cable-based actuation system [VEN 07]. It is built on the idea of a lightweight exoskeleton system having a pair of springs in series with an electric motor. The electric motor is coupled to the springs via Bowden cables. Due to the cable-based actuation, the electric motor is placed on a remote station and acts as a low weight pure force source. The displacement of springs recorded by linear potentiometers is used as a force measurement. The actuated DOFs include two pelvis rotations: hip sagittal and frontal plane rotations and knee sagittal plane rotations.

Ambulation-assisting robotic tool for human rehabilitation (ARTHUR) was developed to mechanically interact with a single leg during treadmill training. It consists of two moving coil brushless servo motors that drive either end of a two bar linkage [EMK 06]. ARTHUR provides motions to the knee and ankle joints in the sagittal plane. Pelvic Assist Manipulator (PAM) is being developed to allow naturalistic movements to the human pelvis during gait training [AOY 07]. Two 3-DOF robotic arms are used to assist the patient's pelvis during treadmill training. These two robotic arms are placed at an angle to give the therapist access from the side and from behind. PAM uses pneumatic actuators to provide lateral and rotational pelvic movements for the patient. PAM is used in combination with Pneumatically Operated Gait Orthosis (POGO), a device that provides pneumatic actuation for hip and knee sagittal rotations.

Although the robotic orthoses can provide systematic and prolonged treadmill training sessions, there are some drawbacks associated with their designs. Two approaches are seen in actuator placement for powering the active orthoses. In one approach, the actuators are placed on a remote station, and the actuation is transferred to the orthosis via cables, rigid linkages, and pneumatic or hydraulic systems [AOY 07, VEN 07]. The benefit of this method is that there are no limitations on actuator weight and hence the power capacity of the actuators. Inefficient transfer of power, non-durability of actuation transfer mechanism (cables), and lack of precise control are the drawbacks associated with this approach. In another approach, the actuators are directly mounted on the orthosis frame [COL 00, BAN 09]. The main advantage of this approach is the efficient transfer of power and a good alignment of orthosis joints with patient joints. The weight of actuators and gear assembly increases the overall weight of the orthosis. Reduction in weight of actuation mechanism reduces the maximum moments that could be applied to the patient's joints. Gravity balancing techniques have been developed to compensate for the weight of the orthosis by using spring and counter weight mechanisms [AGR 04].

6.1.5. *Gait training strategies: a review*

The goal of robot-assisted gait training is to reinstate neuroplasticity so that the motor function could be improved. Although successful determinants of gait training are largely unknown, repetitive and task-oriented training strategies may result in significant improvements [BAY 05, PAT 07]. These determinants have been formulated by drawing concepts from rehabilitation, neuroscience, and motor learning literature [KWA 97, BAR 06]. Gait training is to be provided according to the level of disability while encouraging the patient's active participation in the training process. Robot-assisted treadmill training utilizes trajectory tracking, impedance, and adaptive control-based training strategies.

Trajectory tracking or position control is widely implemented by robotic training devices. Trajectory tracking works on the principle of guiding the patient's limbs on fixed reference gait trajectories. It mainly consists of proportional feedback position controllers with joint angle gait trajectories as input [LUM 93, LUM 95, LUM 02]. For trajectory tracking, the issue of determining the reference trajectory is important. Mathematical models of normative gait trajectories and pre-recorded trajectories from healthy individuals are commonly used. A *teach and replay* technique has been introduced by the designers of ARTHUR in which a joint angle trajectory is recorded during manual assistance and is then replayed during robotic assistance [EMK 08]. Recently, a reference trajectory generation method has been developed for hemiparetic patients. The desired trajectory for the impaired limb is generated online based on the movement of unimpaired contralateral limb [VAL 09].

Trajectory tracking is suitable for training patients with SCI or acute stroke when they have no muscular strength to move their limbs. A limiting feature of trajectory tracking is the imposition of a predefined trajectory, leaving the system inflexible to considering the patient's intention and capabilities. For patients having some muscular strength, trajectory tracking may cause damage to their neuromuscular system when they try to resist the fixed forces applied by actuators [LUM 06, PAT 06]. This may result in abnormal gait pattern generations and would leave the patient unable to adapt to physiological gait [KAH 06].

The patient's active participation and involvement in the robotic gait training process is important to develop neuroplasticity and motor control [EMK 05, KAE 05].

The terms *patient cooperative*, *assist as needed*, *compliant*, and *interactive robot-assisted gait training* are used in the literature [RIE 05, VEN 07, WOL 08]. Robot-assisted gait training uses impedance and adaptation-based control strategies to actively involve the patient in the training process.

The relationship between the force exerted by the actuators and the resulting motion is generally known as *Mechanical Impedance.* The concept of impedance control in the field of robotics is first introduced by Hogan [HOG 85]. The impedance controller works on the principle of *force-based impedance control* and is mostly implemented in the form of an outer position feedback loop and inner force feedback loop. Lokomat also uses an impedance controller of the same form [RIE 05]. For gait training purposes, the idea behind impedance control is to allow variable deviation from reference gait trajectory depending on the patient's resistance. As long as the patient is on the reference trajectory with minimum deviations, the robot should not intervene. After a set limit is exceeded, an adjustable moment is applied at each joint to keep the leg within a defined range along the reference trajectory. For higher impedance values, the concept of admittance control is also used by Lokomat [HOO 02]. An admittance controller as opposed to impedance control works on the principle of position-based impedance control. More recent forms of impedance controllers use the concept of viscous force fields [COL 05, CAI 06]. For ALEX, a force field controller is developed for applying tangential and normal forces at patient's ankle. The linear actuators mounted at the hip and knee joints simulate the forces applied at the ankle. Tangential forces help to move the patient along the trajectory, and normal forces simulate virtual walls around the desired ankle trajectory in the plane containing human thigh and shank [HOG 06, BAN 09]. LOPES also uses impedance control for its "patient in charge" and "robot in charge" modes [VEN 07]. For the robot in charge mode, the controller stiffness is increased, so the patient is not in a compliant environment.

The potential issue with trajectory tracking and impedance control-based training is that they do not tune controller parameters based on real-time judgment of the patient's abilities. Adaptive assistance is used to enhance the patient's active participation in the training process [EMK 07]. The basis of adaptive assistance is to modify the robot motion in a way that is desired by the patient.

Adaptive assistance is used for real-time tuning of the controllers designed for stiff robotic actuators to match the patient's disability level and to actively involve him or her in the training process. In adaptive assistance mode, robot motion is initiated from the physical interaction between the patient and the orthosis. As the disability level varies from subject to subject, online estimation of patient–orthosis interaction force is the most crucial task in the adaptive assistance paradigm. In most of the gait training orthoses, this interaction force is estimated from the combined patient–orthosis dynamic model. Different methods are used to estimate the patient–orthosis interaction torque component.

Lokomat uses a moving average-based exponential forgetting technique for interaction torque estimation. After obtaining this estimate, various joint angle adaptation algorithms are formulated to adapt reference gait trajectory parameters by online optimization. These algorithms include inverse dynamics-based joint angle adaptation, direct dynamics-based joint angle adaptation, and impedance control-based joint angle adaptation [JEZ 04]. Later an impedance magnitude adaptation algorithm was formulated for Lokomat [RIE 05]. This algorithm works based on the impedance magnitude adaptation with constant reference joint angle trajectories. When a smaller patient resistance is estimated, the controller impedance is set high to guide the patient's limbs on reference trajectory. Impedance magnitude is reduced in larger estimates, and larger deviations from the reference trajectory are allowed. ARTHUR uses a manual teaching approach of and a replay for robot-assisted gait training. Physical therapists are asked to impart manual gait training to the subjects first, and the kinematic and kinetic gait parameters are recorded. These recorded parameters are then used during robotic gait training to adapt the stiffness and damping of a proportional and derivative (PD) force controller as a function of trajectory tracking error [EMK 08].

The adaptive algorithms discussed above estimate the patient–orthosis interaction force from the combined dynamic model of the patient and orthosis mechanism. The quality of interaction force estimation depends on the accuracy of force and joint position sensors and also on the estimation algorithm [ERD 10]. The abrupt forces like muscle spasms arising from patient and resulting actuator non-backdrivability present a major problem to the interaction torque estimation. *Backdrivability or compliance* is the ability of the robot being moved by the patient with low mechanical impedance to allow the patient's voluntary movements [CAM 09].

6.2. Compliant active orthosis design

The robotic orthoses discussed above are driven by electric motors attached to gear boxes which are highly stiffened and supply very large torques in repose to the patient's clonus and strong spasms. This may result in injury to a patient. Electric motors thus present a mismatch in the compliance of the actuator and limb being assisted. Impedance and adaptive control has had success in addressing this problem but adds another layer of complexity and extra cost. The active orthosis design presented in this study is compliant because the active orthosis behaves softly and gently and reacts to the patient's muscular effort.

6.2.1. *Design criteria*

The first goal was to design an active orthosis for gait rehabilitation. For design purposes, the biomechanics of human gait was studied. To meet the functional and structural requirements of the active orthosis, the orthosis joints should work in perfect alignment with the patient's joints. Also the active orthosis should inhibit excessive knee and hip extension. The actuation system should be powerful enough to guide the patient's limbs on reference trajectories and be able to produce required joint moments (Table 6.1). Actuators should be highly back-drivable with low mechanical impedance to accommodate abrupt forces arising from clonus. The active orthosis must have safety limits at the ends of maximum ranges of joint motion, and the orthosis should return to an anatomical standing position if the actuation mechanism fails. Regarding the cosmetic requirements and ease of use, the active orthosis should be lightweight, easy to wear, and comfortable. The orthosis should also allow fast adjustment to individual patients with different anthropometric parameters. The actuation system should generate no perceivable noise.

Degree-of-freedom	Range of motion	Joint moment
Hip flexion/extension	+60°/–30°	55 Nm
Hip abduction/adduction	+15°/–15°	25 Nm
Knee flexion/extension	+0°/–90°	55 Nm

Table 6.1. *Joint ranges of motion and moments*

6.2.2. *Active orthosis components*

Actuated and free DOFs for the active orthosis were decided based on the joint ranges of motion. The major rotations during gait cycle are in the sagittal plane (Table 6.1). The actuated DOFs were hip and knee sagittal plane rotations. Besides the sagittal plane, hip abduction/adduction provides the second-largest motion. This DOF was kept free. A new type of compliant actuator, PMA (Figure 6.5), was used for providing actuation to active orthosis. Although the design with PMA was a difficult task, as they can provide only unidirectional pulling force, the optimal DOFs necessary to provide physiological gait pattern were chosen. To provide actuation at hip and knee joints, various mechanisms were studied to transfer the actuation from PMA to the orthosis joints. An antagonistic disc-PMA mechanism was selected for actuation purposes. Double groove disks were used at hip and knee joints for sagittal plane motions. The orthosis frame was made from aluminum rectangular tubing to meet the strength requirements for torque transmissions (Figure 6.6). All the orthosis sections were made telescopic so that they could match the anthropomorphic features of a larger patient population.

Figure 6.5. *Pneumatic muscle actuator*

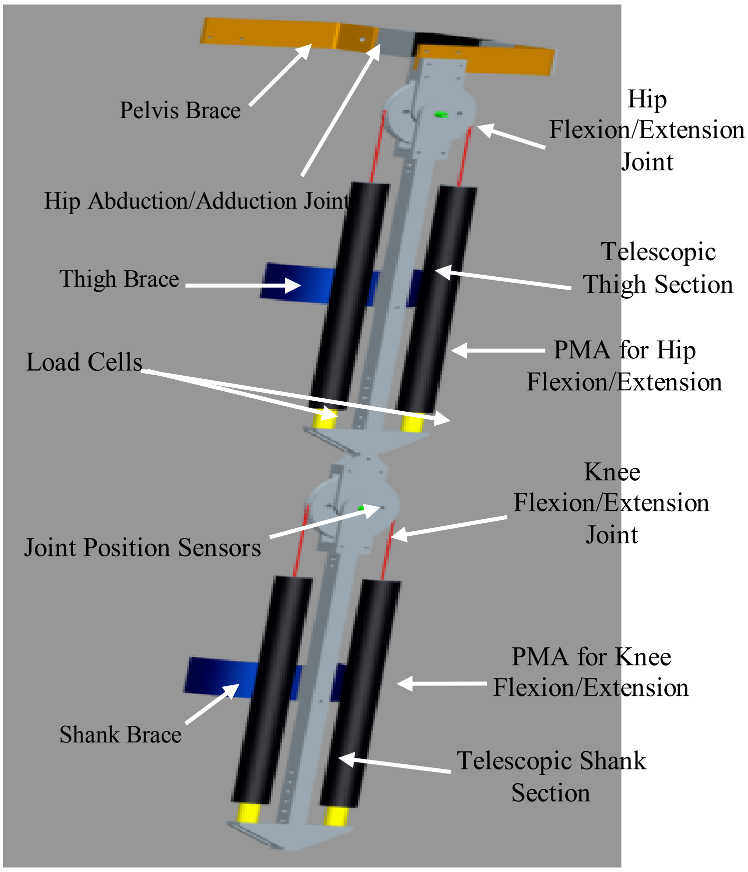

Figure 6.6. *Computer-aided design (CAD) model of active orthosis*

There were two reasons for using PMA for actuation purposes. The first reason is related to the geometrical design of active orthosis. The PMA has a high power to weight ratio, which makes it suitable for the task. The design is made simple and wearable, and the patient's joints will be in perfect alignment with the orthosis joints. The second reason is to introduce intrinsic compliance and back-drivability in the orthosis design. This compliance is beneficial for human–orthosis interaction and provides greater shock tolerance on heel strike, low actuator impedance, and more stable force control.

Absolute joint encoders were used at hip knee and joints to measure the angular positions. Load cells were used in series with each PMA to measure the pulling force generated. The ankle joint was not actuated.

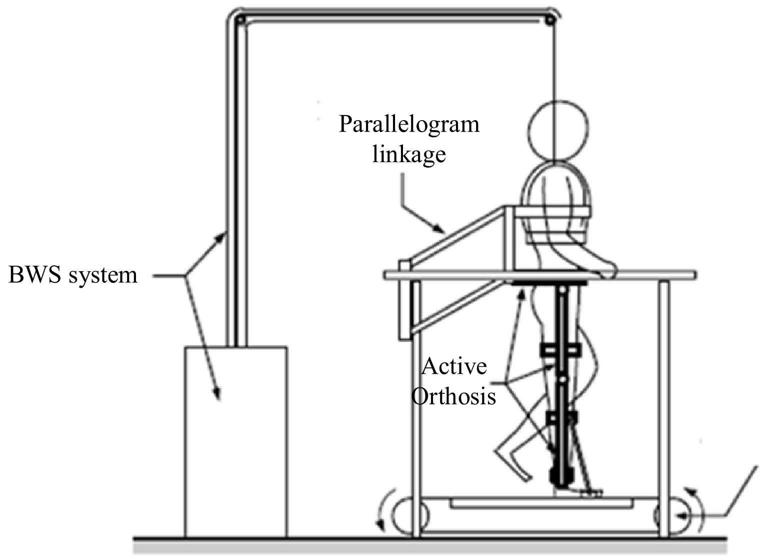

Figure 6.7. *Robot-assisted gait training*

The reason to omit an actuated ankle joint was that the ankle is a complex joint, and axes of motion are not simply three Euclidean axes. It is not necessary to provide an external ankle push off, as it could be done significantly with the aid of a treadmill. Also it is painful to apply an active force at the ankle joint without an individually fit-to-size foot interface. For safe and effective gait training, only ankle plantar/dorsiflexion is necessary for foot clearance during the swing phase. This dorsiflexion could be provided with the aid of some passive mechanisms like elastic straps or springs. Devices like pneumatically-driven ankle orthosis [FER 05] or Anklebot [ROY 09] can be added to the active orthosis if ankle actuation appears to be crucial from a clinical point of view. The schematic sketch of the complete system is shown Figure 6.7. A parallelogram linkage connects the active orthosis with the treadmill and also stabilizes the patient's pelvis in the vertical direction during training. A BWS system compensates for the weight of the patient and helps in foot clearance during the swing phase of gait.

6.3. Modeling

6.3.1. *PMA dynamic modeling*

Modeling of the active orthosis with PMA was a crucial task as they show highly nonlinear force-length characteristics. For this study, we considered the PMA model developed by Reynolds *et al.* [REY 03]. The modeled PMA has been inflated by supplying voltage to a solenoid that controls the flow of pressurized gas into the rubber bladder. It has been deflated by another exciting solenoid venting the contents of the bladder to the atmosphere. When inflated, the PMA shortens via the actions of the braided sheath, exerting a contractile force that is quite large in proportion to the PMAs weight.

The dynamic behavior of the PMA hanging vertically actuating a mass M has been modeled as a combination of a nonlinear friction, a nonlinear spring, and a nonlinear contractile element. The equation describing the dynamics of this PMA hanging vertically actuating a mass is:

$$M\ddot{x} + B(P)\dot{x} + K(P)x = F(P) - Mg \tag{6.1}$$

where x is the amount of PMA contraction and the coefficients $K(P)$, $B(P)$, and $F(P)$ are given in [REY 03] as:

$$\begin{aligned} K(P) &= K_0 + K_1 P \\ &= 5.71 + 0.0307P \end{aligned} \tag{6.2}$$

$$\begin{aligned} B(P) &= B_{0i} + B_{1i} P \\ &= 1.01 + 0.00691P \end{aligned} \tag{6.3}$$

$$\begin{aligned} B(P) &= B_{0d} + B_{1d} P \\ &= 0.6 - 0.000803P \end{aligned} \tag{6.4}$$

$$\begin{aligned} F(P) &= F_0 + F_1 P \\ &= 179.2 + 1.39P \end{aligned} \tag{6.5}$$

From equation [6.1] the total force exerted by the PMA on the mass is:

$$\alpha = F(P) - B(P)\dot{x} - K(P)x \tag{6.6}$$

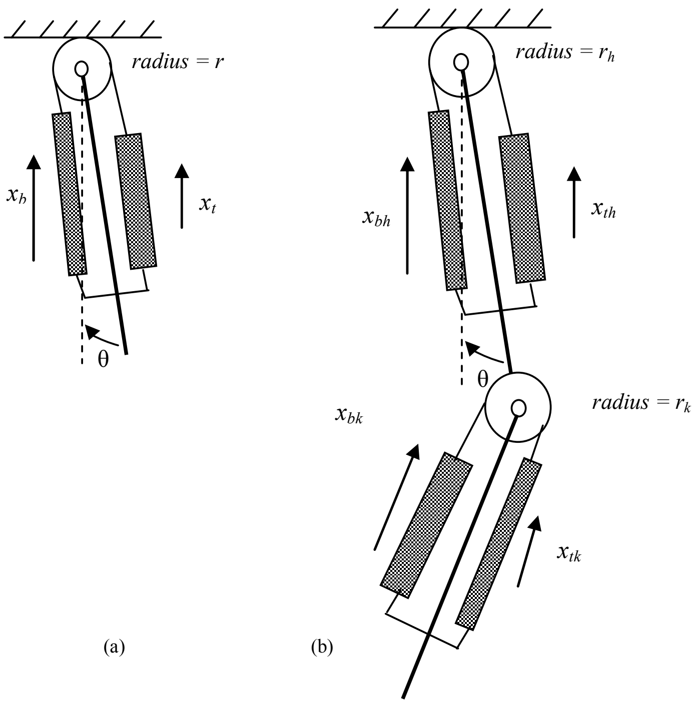

Figure 6.8. *(a) Antagonistic PMA configuration; (b) active orthosis with PMAs*

For the antagonistic configuration of PMA, the torque imparted to the joint by the PMA pair is (Figure 6.8a):

$$T_{\text{total}} = T_{\text{b}} - T_{\text{t}} = (\alpha_{\text{b}} - \alpha_{\text{t}})r \quad [6.7]$$

where T_{b} and T_{t} are the torque due to each of the individual PMA and are given by:

$$T_{\text{b}} = (F_{\text{b}} - K_{\text{b}}x_{\text{b}} - B_{\text{b}}\dot{x}_{\text{b}})r \quad [6.8]$$

$$T_{\text{t}} = (F_{\text{t}} - K_{\text{t}}x_{\text{t}} - B_{\text{t}}\dot{x}_{\text{t}}) \quad [6.9]$$

where x_{b} is the length of PMA b, x_{t} is the length of PMA t, and r is the radius of disc. Thus, the relation for total torque (T_{total}) becomes:

$$T_{\text{total}} = \left(F_{\text{b}} - K_{\text{b}}x_{\text{b}} - B_{\text{b}}\dot{x}_{\text{b}} - F_{\text{t}} + K_{\text{t}}x_{\text{t}} + B_{\text{t}}\dot{x}_{\text{t}}\right)r \quad [6.10]$$

The arrangement of the PMA on the active orthosis is shown in Figure 6.8b. Under these conditions, the hip and knee sagittal plane torques T_{h} and T_{k}, respectively, can be expressed using equation [6.10]

$$T_{\mathrm{h}} = \left(F_{\mathrm{h}} - K_{\mathrm{h}} x_{\mathrm{th}} - B_{\mathrm{th}} \dot{x}_{\mathrm{th}} - F_{\mathrm{h}} + K_{\mathrm{h}} x_{\mathrm{bh}} + B_{\mathrm{bh}} \dot{x}_{\mathrm{bh}}\right) r_{\mathrm{h}} \quad [6.11]$$

$$T_{\mathrm{k}} = \left(F_{\mathrm{k}} - K_{\mathrm{k}} x_{\mathrm{tk}} - B_{\mathrm{tk}} \dot{x}_{\mathrm{tk}} - F_{\mathrm{k}} + K_{\mathrm{k}} x_{\mathrm{bk}} + B_{\mathrm{bk}} \dot{x}_{\mathrm{bk}}\right) r_{\mathrm{k}} \quad [6.12]$$

where the subscripts h and k represent the coefficients for hip and knee joints, respectively.

6.3.2. *Interaction force estimation*

The estimation of patient-active orthosis interaction force requires a combined dynamic model of the patient and the active orthosis. As the anthropometric parameters and disability level vary from subject to subject, online estimation of patient–orthosis interaction force is the most crucial task in adaptive assistance paradigm. A robot dynamic equation was used for estimating the patient–orthosis interaction forces:

$$M(x)\overline{x} + C(x,\dot{x})\dot{x} + G(x) = T_{\mathrm{a}} + T_{\mathrm{p}} - T_{\mathrm{f}} \quad [6.13]$$

where M is the combined patient–orthosis inertia matrix, C is the combined patient–orthosis coriolis and centrifugal torque, G is a term representing gravitational torques, and T_{f} is the joint friction torque. T_{a} is the torque applied by actuator onto the orthosis and is measured by force sensors. T_{p} is the patient–orthosis interaction torque or the resistance offered by patient to applied actuator forces, and x is the generalized position vector representing joint angles. $\dot{x}$ and $\overline{x}$ are joint velocity and acceleration, respectively.

The real-time update of this patient resistance component is a challenging task, and the least squares method with exponential forgetting is used to update this component during the training process. Least squares with exponential forgetting is a useful method of dealing with variable patient-active orthosis interaction torque. The intuitive motivation is that the past data is generated by past parameters and should be discontinued for the estimation of current parameters. For more details, refer to [SLO 91].

6.4. Control

The benefits of robot-assisted rehabilitation might be increased by using more advanced robotic systems. Although different robot-assisted gait training strategies are discussed above, it remains to be demonstrated which is the most effective. One way to enhance motor function recovery is to develop a robot control algorithm that seamlessly optimizes the interaction between the active orthosis and the patient to provide as much therapeutic benefit as possible. To promote patient involvement in the rehabilitation process, we hypothesize that an MRAC will be suitable.

The controller estimates the patient–orthosis interaction in real time and modulates the actuator forces accordingly. The MRAC scheme will be helpful to accommodate patients with variable disability levels.

The issue of the justification of the MRAC naturally arises, given the variety of control types available. For the robot-assisted gait rehabilitation where the system dynamics are time and position dependent and where a substantial uncertainty in the system characteristics is produced by the unknown patient resistance torque properties, the model reference approach seems particularly well suited [DUB 79]. The reference model chosen at the discretion of the designer provides a flexible means of specifying the desired closed loop performance characteristics. The use of a model-based controller allows impedance and assistance to be controlled separately so that the orthosis can simultaneously be highly compliant and be able to provide enough assistive force to complete desired spatial movements.

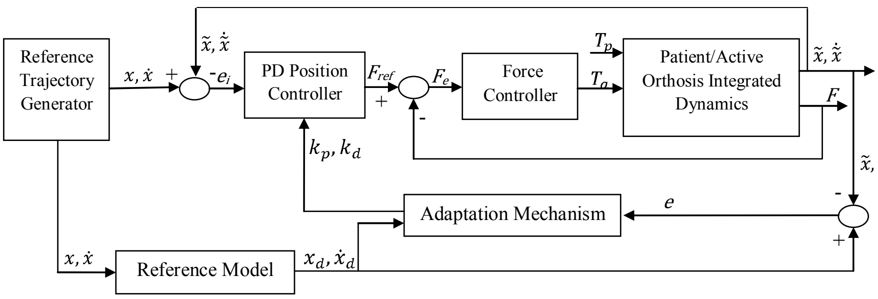

Figure 6.9. *Controller diagram*

The first step in the formation of MRAC is the selection of a reference model (Figure 6.9). The adaptation mechanism is designed to provide PD gains to minimize the trajectory tracking error. The PD position controller works on the principle of impedance control [HOG 85] and generates the reference force based on the trajectory tracking error. The force at the output is measured using a force-sensing system and is fed back to create a resulting error which is processed through a force controller to create torques applied at orthosis joints. The reference model chosen in this case is a linear, second-order, time invariant differential equation:

$$\overline{x}_\mathrm{d} + 2\zeta\omega_\mathrm{n}\dot{x}_\mathrm{d} + \omega_\mathrm{n}^2 = \omega_\mathrm{n}^2 r(t) \qquad [6.14]$$

Rewriting this expression:

$$a\overline{x}_\mathrm{d} + b\dot{x}_\mathrm{d} + x_\mathrm{d} = r(t) \qquad [6.15]$$

where ζ is the damping ratio, ω_n is the natural frequency, and a and b are given as:

$$a = \frac{1}{\omega_n^2} \text{ and } b = \frac{2\zeta}{\omega_n}$$

For the development of the algorithm, the coupling of the system DOFs is neglected, and the nonlinear manipulator dynamics are written assuming:

$$\bar{\ddot{x}} = \frac{ek_m}{m} \qquad [6.16]$$

where k_m is the actuator torque constant and m is the varying effective mass. The orthosis nonlinear dynamic equation can be written as:

$$\frac{m(i)}{k_m k_p}\bar{\ddot{x}} + \frac{k_d}{k_p}\dot{\breve{x}} + \breve{x} = r(t) \qquad [6.17]$$

The equation is of the form:

$$\beta(t)\bar{\ddot{x}} + \gamma\dot{\breve{x}} + \breve{x} = r(t) \qquad [6.18]$$

The adaptation mechanism was designed to provide PD gains to minimize the trajectory tracking error. The PD position controller works on the principle of impedance control [HOG 85] and generates the reference force based on the trajectory tracking error. In this study, it was unnecessary to obtain explicit knowledge of the coefficients β and γ. The force at the output was measured using force-sensing system and was fed back to create a resulting error which was processed through a force controller to create torques applied at orthosis joints. The patient will be able to train in a compliant and comfortable environment compared with the fixed trajectories applied by the existing gait rehabilitation orthoses. $x, \dot{x}$ are reference joint angle and velocity, respectively, $x_d, \dot{x}_d$ are joint position and velocity outputs of the reference model, $\bar{x}, \dot{x}$ are joint variables at output measured by joint sensors, e is the error fed to adaptation mechanism, and e_i is the position error fed to position controller. K_p and K_d are proportional and derivative gain values. F_{ref} is the reference force generated by PD controller; F_e is the error between the reference force and the force measured by force sensors *(F).*

A quadratic function is written in terms of difference between the responses and the actual referenced system as:

$$f(\varepsilon) = \frac{1}{2(q_0\varepsilon + q_1\dot{\varepsilon} + q_2\ddot{\varepsilon})} \qquad [6.19]$$

$$\dot{\beta} = -\frac{\partial f(\varepsilon)}{\partial \beta} = \frac{\partial f(\varepsilon)}{\partial a} \qquad [6.20]$$

$$\dot{\gamma} = -\frac{\partial f(\varepsilon)}{\partial \gamma} = \frac{\partial f(\varepsilon)}{\partial b} \qquad [6.21]$$

where ε is defined as $x_{\mathrm{d}} - \breve{x}$. After algebraic manipulation, the rates of adjustment of β and γ are:

$$\dot{\beta} = (q_0\varepsilon + q_1\dot{\varepsilon} + q_2\overline{\varepsilon})(q_0 u + q_1\dot{u} + q_2\overline{u}) \qquad [6.22]$$

$$\dot{\gamma} = (q_0\varepsilon + q_1\dot{\varepsilon} + q_2\overline{\varepsilon})(q_0 w + q_1\dot{w} + q_2\overline{w}) \qquad [6.23]$$

where the values of u and w and their derivatives are obtained from the solutions of the following differential equations:

$$a\overline{u} + b\dot{u} + u = -\overline{x}d \qquad [6.24]$$

$$a\overline{w} + b\dot{w} + w = -\dot{x}_{\mathrm{d}} \qquad [6.25]$$

After determining the values of γ and β, the rates of adjustment of the feedback gains of the system can be calculated by differentiating the definitions of γ (t) and β(t) obtained by comparing equations [6.17] and [6.18]. M is assumed to change slowly compared to the adaptation mechanism and during adaptive tracking β can be approximated by a. The result is:

$$\dot{k}_{\mathrm{p}} = -\frac{\dot{\beta} k_{\mathrm{p}}}{a} \qquad [6.26]$$

$$\dot{k}_{\mathrm{d}} = k_{\mathrm{d}}\dot{\gamma} - \frac{k_{\mathrm{d}}\dot{\beta}}{a} \qquad [6.27]$$

Explicit expressions for M are not required by the algorithm.

6.5. Simulation results

The active orthosis prototype leg with an antagonistic pair of PMA, actuating the hip and knee sagittal plane joints, is shown in Figure 6.8b. The simulation was performed using a fourth-order Runge–Kutta algorithm with a step size of 0.01 s.

The duration of the simulation was set to complete five gait cycles. The nominal joint angle trajectories of natural gait are reported in literature [WIN 91] and were used as reference trajectories for guiding the patient's limbs. The subject was considered to be completely passive, offering no resistance to the actuator torques. First stride from initial standing posture was eliminated, and the remaining four gait cycles were used for analysis purpose. Data from the right gait cycle was formatted for presentation purpose. The simulation was performed for a subject having a mass of 76.8 kg. Lengths of shank and thigh segment were 0.44 m and 0.43 m, respectively. The weight of the active orthosis (9.8 kg) was added to the subject's weight to form a mass matrix, *M*. A BWS of 40% was used.

The cadence was 85 steps/min. The reference joint angle trajectories were tracked with a maximum error of 1° and are shown as a percentage of gait cycle in Figure 6.10. The maximum hip angle achieved during flexion and extension was 16° and 14°, respectively. A knee flexion of 60° was achieved during mid-swing period. The torques required to track these trajectories as a percentage of gait cycle are shown in Figure 6.11. A peak hip torque of 50 Nm and knee torque of 90 Nm was achieved. The PMA can provide a peak force of 700 N satisfying the desired ranges of peak joint torques. The extent of actuator force adaptation also needs to be determined so that the gait patterns remain purely physiological. The interaction force feedback also needs to be evaluated as it may contain abrupt forces arising from patient's abnormal muscle functions like clonus or spasms.

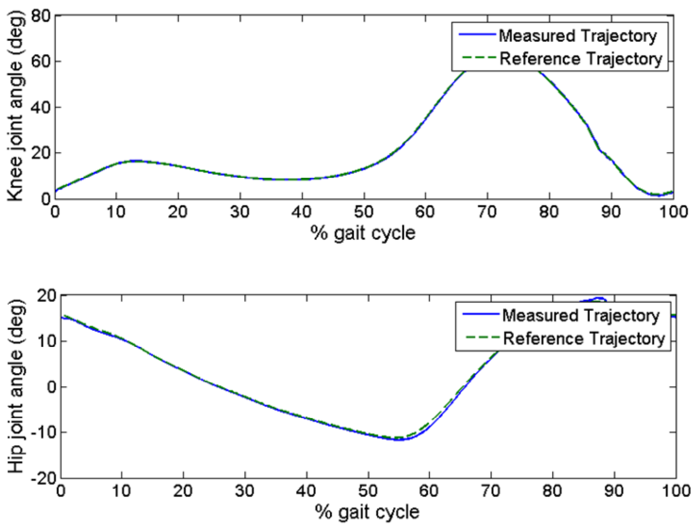

Figure 6.10. *Hip and knee joint angle trajectories as a percentage of gait cycle*

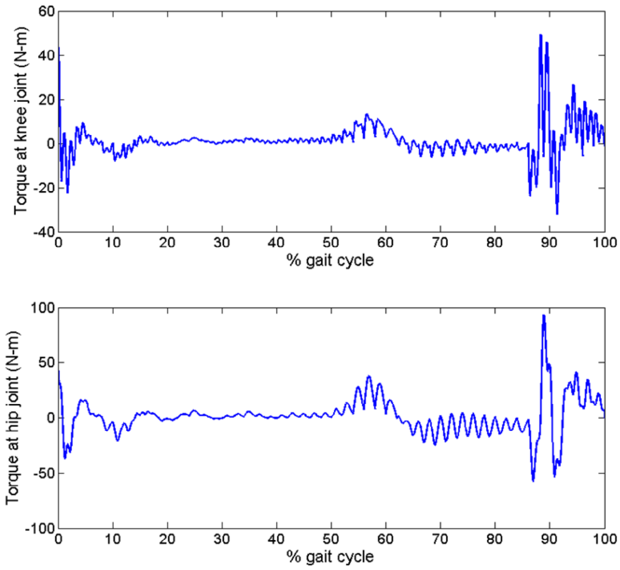

Figure 6.11. *Torque at hip and knee joints as a percentage of gait cycle*

6.6. Conclusions

Robotic orthosis can provide repetitive, prolonged, and systematic gait training sessions. A compliant gait rehabilitation orthosis has been designed. The variable level of disability among different patients represents a problem for devising a suitable gait training strategy. Also the nonlinear actuator dynamics make the control system design difficult. To overcome these problems, a dynamic model of the PMA actuators for the actuated DOFs was developed and was found to be suitable for the desired ranges of operation. To accommodate and train patients with variable disability, an MRAC algorithm was designed. The algorithm estimates patient-active orthosis interaction forces and adjusts the applied actuator forces accordingly. The performance of the MRAC was tested in computer simulations and was capable of guiding the patient's limbs on the physiological gait patterns. Future work involves the evaluation of the dynamic PMA model and MRAC on active orthosis prototype followed by clinical evaluations on the healthy and neurologically impaired subjects.

6.7. Acknowledgment

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