Adaptive Control for Wearable Robots in Human-Centered Rehabilitation Tasks

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Tesis presentada para obtener el grado de
Doctor en Ingeniería Biomédica

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" To raise new questions, new possibilities to regard old problems from a new angle, requires creative imagination and marks real advance in science "

Albert Einstein

To my beloved Amma,

for the eternal support...

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Abstract

Robotic rehabilitation therapies have been improving by providing the needed assistance to the patient and helping the therapist to choose the necessary procedure. Rehabilitation devices are intended to function in a human-centered environment and hence the interaction of the patient is a major research aspect for such assistive devices. A better interaction with the robot/orthosis will encourage the patient to pursue or advance the therapy. Apart from the interaction, the rehabilitation must be adaptive to the specific user needs. Further, the assistance to be delivered must consider achieving both the short and long term goals, such as in gait training. Gait training for individuals with neurological disorders is a topic of interest for both the patients and researchers. This thesis presents an adaptive "Assistance-as-needed" strategy which adheres to the specific needs of the patient and with the inputs from the therapist, whenever needed. The exertion of assistive and responsive behaviour of the lower limb wearable robot is dedicated for the rehabilitation of incomplete spinal cord injury(SCI) patients.

Individuals with SCI demonstrate different levels of lesions and need varied assistance for performing the tasks in rehabilitation. The muscle atrophy in such users may vary from mild to severe depending on the kind of lesion. Complete SCI individuals possess lesser muscle strength and hence a bodyweight support mechanism is needed, to provide complete assistance. In case of incomplete SCI patients, the adaptation may vary depending on their injury level and their adaptation to the joint movement. In some cases, external devices such as muscle stimulation must be used for providing the necessary muscle level assistance.

The main objective of this thesis is to propose and evaluate an adaptive control model on a wearable robot, assisting the user and adhering to their needs, with no or less combination of external devices. The adaptation must be more interactive to understand the user needs and their intentions or volitional orders. Similarly, by using the existing muscular strength, in incomplete SCI patients, as a motivation to pursue the movement and assist them, only when needed.

The adaptive behaviour of the wearable robot is proposed by monitoring the interaction and movement of the user. This adaptation is achieved by modulating the stiffness of the exoskeleton in function of joint parameters, such as positions and interaction torques. These joint parameters are measured from the user independently and then used to update the new stiffness value. The adaptive algorithm performs with no need of external sensors, making it simple in terms of usage. In terms of rehabilitation, it is also desirable to be compatible with combination of assistive devices such as muscle stimulation, neural activity and body balance, to deliver a user friendly and effective therapy.

Combination of two control approaches has been employed, to improve the efficiency of the adaptive control model and was evaluated using a wearable lower limb exoskeleton device, H1. The control approaches, Hierarchical and Task based approach have been used to assist the patient as needed and simultaneously motivate the patient to pursue the therapy. Hierarchical approach facilitates combination of multiple devices to deliver an effective therapy. The control architecture is categorized in two layers, Low level control (LLC) and High level control (HLC). The LLC comprises of the communication with exoskeleton's sensors and provides the HLC with different modes of basic control: position, torque and stiffness. On the other hand, the HLC provides the convenient behaviour to be performed by the exoskeleton in a human-centered scenario.

Multiple assistive solutions can be proposed, considering the different behaviours involved in rehabilitation tasks, with no possible single solution. Task-based approaches engage in each task individually and allow the possibility to combine them at any point of time. Several assistive devices can be used or needed at different point of time in a lower limb rehabilitation therapy. For instance, in Sit-to-stand, the user might need assistance even at the muscle level which can be provided by a muscle stimulation technique, such as to perform the transition. Also, the projection of centre of mass of the user can be used as a trigger to initiate the transition of sit-to-stand . In Balance training, the projection of the centre of mass will help in defining the stability limits of the user, to ensure equilibrium. Similarly, for a gait sequence the gait initiation can be either by the human-orthosis interaction torque or the neural activity of the user. The use of motor related neural activity is widely appreciated as a top-down approach in rehabilitation, which helps in developing the patients' neural activity. This type of user induced initiation or intention based assistance is termed as volitional control.

To achieve a suitable assist-as-needed strategy, the combination of hierarchical and task based approach is needed. It is also necessary to provide an interaction based approach to ensure the complete involvement of the user and for an effective therapy. By means of this dissertation, a task based adaptive control is proposed, in function of human-orthosis interaction, which is applied on a hierarchical control scheme. This control scheme is employed in a wearable robot, with the intention to be applied or accommodated to different pathologies, with its adaptive capabilities.

The adaptive control model for gait assistance provides a comprehensive solution through a single implementation: Adaptation inside a gait cycle, continuous support through gait training and in real time. The performance of this control model has been evaluated with healthy subjects, as a preliminary study, and with paraplegic patients. Results of the healthy subjects showed a significant change in the pattern of the interaction torques, elucidating a change in the effort and adaptation to the user movement. In case of patients, the adaptation showed a significant improvement in the joint performance (flexion/extension range) and change in interaction torques. The active change in interaction torques (positive to negative) reflects the active participation of the patient, which also explained the adaptive performance. The patients also reported that the movement of the exoskeleton is flexible and the walking patterns were similar to their own distinct patterns.

The presented work is performed as part of the project HYPER (Hybrid Neuroprosthetic and Neurorobotic Devices for Functional Compensation and Rehabilitation of Motor disorders) is funded by Ministerio de Ciencia y Innovación (MINECO), Spain.
(CSD2009 - 00067 CONSOLIDER INGENIO 2010).

Highlights

- Building a simulation model to analyse the behaviour of a human-centered strategy
- Handling and recovering the trajectories from the effect of disturbances
- Varying the input speed rate and with fixed stiffness to understand the assistance to be provided
- Task specific adaptation with respect to the human-orthosis interaction
- Task based combination of external devices to provide the sufficient adaptation
- Adaptative control for gait assistance by considering the human movement and human-orthosis interaction
- Initiation of the rehabilitation using the users' volitional orders, such as BMI and mechanical interaction
- Performance evaluation of the adaptive control with incomplete SCI individuals

Keywords

Adaptive control, Assist-as-needed, Exoskeleton, Balance training, Gait training, Hierarchical approach, Human-orthosis interaction, Human-centered, Rehabilitation, Sit-to-Stand, Spinal cord injury, Task-based approach

Resumen

Las terapias de rehabilitación robóticas han sido mejoradas gracias a la inclusion de la asistencia bajo demanda, adaptada a las variaciones de las necesidades del paciente, así como a la inclusión de la ayuda al terapeuta en la elección del procedimiento necesario. Los dispositivos de rehabilitación están destinados a funcionar en un entorno centrado en humanos, por lo tanto la interacción con el paciente es un aspecto de investigación clave para este tipo de dispositivos. Una mejor interacción con el robot/órtesis estimulará al paciente a proseguir y avanzar en la terapia. Además de dicha interacción, la rehabilitación se debe adaptar a las necesidades específicas del usuario. Es más, la asistencia que se suministre debe tener en cuenta la consecución de objetivos a corto y largo plazo, como en el caso de la terapia de rehabilitación de la marcha. El entrenamiento específico de la marcha para las personas con trastornos neurológicos es un tema de interés tanto para los pacientes como para los investigadores.

Esta tesis presenta una estrategia adaptativa de asistencia bajo demanda, la cual se ajusta a las necesidades específicas del paciente junto a las aportaciones del terapeuta siempre que sea necesario. El esfuerzo del comportamiento asistencial y receptivo del robot personal portátil para extremidades inferiores está dedicado a la rehabilitación de pacientes con lesión de la médula espinal (LME) incompleta.

Las personas con LME muestran diferentes niveles de lesiones y diversa necesidad de asistencia para la realización de las tareas de rehabilitación. La atrofia muscular en tales usuarios puede variar de leve a severa, dependiendo del tipo de lesión. Los pacientes con LME completa poseen menor fuerza muscular, por lo tanto es necesario un mecanismo de apoyo del peso corporal para proporcionar una asistencia completa. En el caso de los pacientes con LME incompleta, la adaptación puede variar en función de su nivel de lesión

y de su adaptación al movimiento de la articulación. En algunos casos se deben utilizar dispositivos externos, tales como la estimulación muscular, para proporcionar la asistencia necesaria a nivel muscular.

El objetivo principal de esta tesis es proponer y evaluar un modelo de control adaptativo en un robot portátil, ayudando al usuario y cumpliendo con sus necesidades, en ausencia o con reducción de dispositivos externos. La adaptación debe ser más interactiva para entender las necesidades del usuario y sus intenciones u órdenes volitivas. De modo similar, usando la fuerza muscular existente (en pacientes con LME incompleta) como motivación para lograr el movimiento y asistirles solo cuando sea necesario.

El comportamiento adaptativo del robot portátil se propone mediante la monitorización de la interacción y movimiento del usuario. Esta adaptación conjunta se consigue modulando la rigidez en función de los parámetros de la articulación, tales como posiciones y pares de torsión. Dichos parámetros se miden del usuario de forma independiente y posteriormente se usan para actualizar el nuevo valor de la rigidez. El desempeño del algoritmo adaptativo no requiere de sensores externos, lo que favorece la simplicidad de su uso. Para una adecuada rehabilitación, efectiva y accesible para el usuario, es necesaria la compatibilidad con diversos mecanismos de asistencia tales como estimulación muscular, actividad neuronal y equilibrio corporal.

Para mejorar la eficiencia del modelo de control adaptativo se ha empleado una combinación de dos enfoques de control, y para su evaluación se ha utilizado un exoesqueleto robótico H1. Los enfoques de control Jerárquico y de Tarea se han utilizado para ayudar al usuario según sea necesario, y al mismo tiempo motivarle para continuar el tratamiento. El enfoque jerárquico facilita la combinación de múltiples dispositivos para ofrecer un tratamiento efectivo. La arquitectura de control se clasifica en dos niveles: el control de bajo nivel (LLC) y el control de alto nivel (HLC). El LLC forma parte de la comunicación con los sensores del exoesqueleto y proporciona el HLC con diferentes modos de control básico : posición, par y rigidez . Por otro lado, el HLC proporciona el comportamiento conveniente para ser realizado por el exoesqueleto en un escenario centrado en el hombre

Se pueden proponer múltiples soluciones de asistencia, teniendo en cuenta los distintos comportamientos involucrados en las tareas de rehabilitación, sin existir una única posible solución. Los enfoques basados en tareas involucran a la persona en cada tarea individual, y ofrecen la posibilidad de combinarlas en cualquier momento. En una terapia de rehabilitación del tren inferior es posible utilizar o necesitar en distintos instantes del tiempo de varios dispositivos de ayuda. Por ejemplo, como al ponerse en pie, el usuario puede necesitar de asistencia incluso a nivel del músculo, que puede proporcionarse mediante técnicas de estimulación muscular. Ademas, el centro de masa del usuario se puede utilizar como un disparador para iniciar la asistencia en la transición de estar sentado a ponerse en pie. En ejercicios de equilibrio, la proyección del centro de masa puede ayudar en la definición de los límites de estabilidad del usuario para asegurar el equilibrio. Del mismo modo, la secuencia de iniciación de la marcha puede darse o bien por el par de interacción hombre-órtesis o por la actividad neural del usuario. El uso de la actividad neuronal relacionada con el sistema motor suscita una amplia aprobación como enfoque descendiente en la rehabilitación, el cual ayuda en el desarrollo de la actividad neuronal de los pacientes. Este tipo de iniciación de asistencia inducida por la persona o basada en la intención se denomina control volitivo.

Para conseguir una estrategia adecuada de asistencia bajo demanda, se ha visto necesaria la combinación de los enfoques de control Jerárquico y de Tarea. También es necesario proporcionar un enfoque basado en la interacción con el usuario, para asegurar su participación y lograr así una terapia eficaz. Mediante esta tesis, proponemos un control adaptativo basado en tareas y en función de la interacción persona-ortesis, que se aplica en un esquema de control jerárquico. Este esquema de control se emplea en un robot portátil, con la intención de ser aplicado o acomodado a diferentes patologías, con sus capacidades de adaptación.

El modelo de control adaptativo propuesto proporciona una solución integral a través de una única aplicación: adaptación dentro del ciclo de la marcha y apoyo continuo a través de ejercicios de movilidad en tiempo real. El rendimiento del modelo se ha evaluado en sujetos sanos según un estudio preliminar, y posteriormente tambien en pacientes parapléjicos. Los resultados en sujetos sanos mostraron un cambio significativo en el patrón de los pares de interacción, elucidando un cambio en la energía y la adaptación al movimiento del usuario. En el caso de los pacientes , la adaptación mostró una mejora significativa en la actuación conjunta (rango de flexión / extensión) y el cambio en pares de interacción. El cambio activo en pares de interacción (positivo a negativo) refleja la participación activa del paciente , lo que también explica el comportamiento adaptativo. Los pacientes también informaron de la flexibilidad del movimiento del exoesqueleto y de la similitud de los patrones para caminar a los propios.

Esta tesis se enmarca dentro del proyecto de investigación HYPER (Hybrid Neuroprosthetic and Neurorobotic Devices for Functional Compensation and Rehabilitation of Motor disorders) y está financiada por el Ministerio de Ciencia e Innovación de España.
(CSD2009 - 00067 CONSOLIDER INGENIO 2010).

Resum

Les teràpies de rehabilitació robòtiques han estat millorades gràcies a la inclusió de l'assistència sota demanda, adaptada a les variacions de les necessitats del pacient, així com a la inclusió de l'ajuda al terapeuta en l'elecció del procediment necessari. Els dispositius de rehabilitació estan destinats a funcionar en un entorn centrat en humans, per tant la interacció amb el pacient és un aspecte d'investigació clau per a aquest tipus de dispositius. Una millor interacció amb el robot/ortesi estimularà el pacient a prosseguir i avançar en la teràpia. A més d'aquesta interacció, la rehabilitació s'ha d'adaptar a les necessitats específiques de l'usuari. És més, l'assistència que es subministri de tenir en compte la consecució d'objectius a curt i llarg termini, com en el cas de la teràpia de rehabilitació de la marxa. L'entrenament específic de la marxa per a les persones amb trastorns neurològics és un tema d'interès tant per als pacients com per als investigadors.

Aquesta tesi presenta una estratègia adaptativa d'assistència sota demanda, la qual s'ajusta a les necessitats específiques del pacient al costat de les aportacions del terapeuta sempre que sigui necessari. L'esforç del comportament assistencial i receptiu del robot personal portàtil per extremitats inferiors està dedicat a la rehabilitació de pacients amb lesió de la medul·la espinal (LME) incompleta .

Les persones amb LME mostren diferents nivells de lesions i diversa necessitat d'assistència per a la realització de les tasques de rehabilitació. L'atròfia muscular a tals usuaris pot variar de lleu a severa, depenent del tipus de lesió. Els pacients amb LME completa posseeixen menys força muscular, per tant és necessari un mecanisme de suport del pes corporal per a proporcionar una assistència completa. En el cas dels pacients amb LME incompleta, l'adaptació pot variar en funció del seu nivell de lesió i de la seva adaptació al moviment de l'articulació. En alguns casos s'han d'utilitzar dispositius externs, com ara l'estimulació muscular, per donar el suport necessari a nivell muscular.

L'objectiu principal d'aquesta tesi és proposar i avaluar un model de control adaptatiu en un robot portàtil, ajudant a l'usuari i complint amb les seves necessitats, en absència o amb reducció de dispositius externs. L'adaptació ha de ser més interactiva per entendre les necessitats de l'usuari i les seves intencions o ordres volitives. De manera similar, usant la força muscular existent (en pacients amb LME incompleta) com a motivació per aconseguir el moviment i assistir-los només quan sigui necessari.

El comportament adaptatiu del robot portàtil es proposa mitjançant el monitoratge de la interacció i moviment de l'usuari. Aquesta adaptació conjunta s'aconsegueix modulant la rigidesa en funció dels paràmetres de l'articulació, com ara posicions i parells de torsió. Aquests paràmetres es mesuren l'usuari de forma independent i posteriorment es fan servir per actualitzar el nou valor de la rigidesa. L'acompliment de l'algorisme adaptatiu no requereix de sensors externs, el que afavoreix la simplicitat del seu ús. Per a una adequada rehabilitació, efectiva i accessible per a l'usuari, és necessària la compatibilitat amb diversos mecanismes d'assistència com ara estimulació muscular, activitat neuronal i equilibri corporal.

Per millorar l'eficiència del model de control adaptatiu s'ha emprat una combinació de dos enfocaments de control, i per la seva avaluació s'ha utilitzat un exosquelet robòtic H1. Els enfocaments de control jeràrquic i de Tasca s'han utilitzat per ajudar a l'usuari segons sigui necessari, i al mateix temps motivar per continuar el tractament. L'enfocament jeràrquic facilita la combinació de múltiples dispositius per oferir un tractament efectiu. L'arquitectura de control es classifica en dos nivells: el control de baix nivell (LLC) i el control d'alt nivell (HLC). El LLC forma part de la comunicació amb els sensors de l'exosquelet i proporciona el HLC amb diferents maneres de control bàsic: posició, parell i rigidesa. D'altra banda, el HLC proporciona el comportament convenient per a ser realitzat per l' exosquelet en un escenari centrat en l'home.

Es poden proposar múltiples solucions d'assistència, tenint en compte els diferents comportaments involucrats en les tasques de rehabilitació, sense existir una única possible solució. Els enfocaments basats en tasques involucren a la persona en cada tasca individual, i ofereixen la possibilitat de combinar-les en qualsevol moment. En una teràpia de rehabilitació del tren inferior és possible utilitzar o necessitar en diferents instants del temps de diversos dispositius d'ajuda. Per exemple, com a posar-se dret, l'usuari pot necessitar d'assistència fins i tot a nivell del múscul, que es pot facilitar mitjançant tècniques d'estimulació muscular. A més, el centre de massa de l'usuari es pot utilitzar com un disparador per iniciar l'assistència en la transició d'estar assegut a posar-se dempeus. En exercicis d'equilibri, la projecció del centre de massa pot ajudar en la definició dels límits d'estabilitat de l'usuari per assegurar l'equilibri. De la mateixa manera, la seqüència d'iniciació de la marxa pot donar-se o bé pel parell d'interacció home-ortesis o per l'activitat neural de l'usuari. L'ús de l'activitat neuronal relacionada amb el sistema motor suscita una àmplia aprovació com a enfocament descendent en la rehabilitació, el qual ajuda en el desenvolupament de l'activitat neuronal dels pacients. Aquest tipus d'iniciació d'assistència induïda per la persona o basada en la intenció s'anomena control volitiu.

Per aconseguir una estratègia adequada d'assistència sota demanda, s'ha vist necessària la combinació dels enfocaments de control jeràrquic i de tasca. També cal proporcionar un enfocament basat en la interacció amb l'usuari, per assegurar la seva participació i aconseguir així una teràpia eficaç. Mitjançant aquesta tesi, proposem un control adaptatiu basat en tasques i en funció de la interacció persona - ortesis, que s'aplica en un esquema de control jeràrquic. Aquest esquema de control s'empra en un robot portàtil, amb la intenció de ser aplicat o acomodat a diferents patologies, amb les seves capacitats d'adaptació.

El model de control adaptatiu proposat proporciona una solució integral a través d'una única aplicació: adaptació dins el cicle de la marxa i suport continuo a través d'exercicis de mobilitat en temps real. El rendiment del model s'ha avaluat en subjectes sans segons un estudi preliminar, i posteriorment també en pacients paraplègics. Els resultats en subjectes sans van mostrar un canvi significatiu en el patró dels parells d'interacció, elucidando un canvi en l'energia i l'adaptació al moviment de l'usuari. En el cas dels pacients, l'adaptació va mostrar una millora significativa en l'actuació conjunta (rang de flexió/extensió) i el canvi en parells d'interacció. El canvi actiu en parells d'interacció (positiu a negatiu) reflecteix la participació activa del pacient, el que també explica el comportament adaptatiu. Els pacients també van informar de la flexibilitat del moviment de l'exosquelet i de la similitud dels patrons per caminar als propis.

Aquesta tesi s'emmarca dins del projecte de recerca HYPER (Hybrid Neuroprosthetic and Neurorobotic Devices for Functional Compensation and Rehabilitation of Motor disorders) i està finançada pel Ministeri de Ciència i Innovació d'Espanya.
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Abbreviations

10MWT	10 Metre Walking Test
AAFO	Active Ankle Foot Orthosis
AAN	Assist-As-Needed
ADL	Activity of Daily Living
AFO	Ankle Foot Orthosis
ALEX	Active Leg EXoskeleton
ARTHuR	Ambulation-assisting Robotic Tool for Human Rehabilitation
ASIA	American Spinal Injury Association
BMI	Brain Machine Interface
BWS	Body Weight Support
BWSTT	Body Weight Support Treadmill Training
CAN	Controller Area Network
CLME	Complimentary Limb Motion Estimation
CNS	Central Nervous System
COM	Center of Mass
COM_p	Projection of Center of Mass
COP	Center of Pressure
CPG	Central Pattern Generator
CSIC	Consejo Superior de Investigaciones Cientificas
DoF	Degree of Freedom
EEG	Electroencephalography
EMG	Electromyography
ERD	Event-Related Desynchronization
FES	Functional Electrical Stimulation
FFC	Force Field Controller
FSM	Finite State Machine

FSR	Force Sensitive Resistor
GRF	Ground Reaction Force
HAL	Hybrid Assistive Limb
HLC	High - Level - Controller
HMI	Human-Machine Interface
HNP-T	Hospital Nacional de Parapl\u00e9gicos - Toledo
HYPER	Hybrid Neuroprosthetic and Neurorobotic devices for Functional Compensation and Rehabilitation of Motor disorders
IBEC	Institute for Bioengineering of Catalonia
ILC	Iterative Learning Control
IMU	Inertial Measurement Units
KAFO	Knee Ankle Foot Orthosis
LLC	Low - Level - Controller
LOPES	Lower Extremity Powered Exoskeleton
MGT	Mechanized Gait Trainer
MNP	Motor Neuroprostheses
MPC	Model Predictive Control
MRCP	Motor Related Cortical Potentials
NR	Neuro Robotics
PAM	Pelvic Assist Manipulator
PCI	Peripheral Component Interconnect
PMA	Pneumatic Muscle Actuators
POGO	Pneumatically Operated Gait Orthosis
PID	Proportional - Integral - Derivative
PSFS	Penn Spasm Frequency Scale
QUEST	Quebec User Evaluation of Satisfaction with assistive Technology
RoM	Range of Motion
SCI	Spinal Cord Injury
SCIM III	Spinal Cord Independence Measure version III
SEA	Series Elastic Actuators
WISCI II	Walking Index Spinal Cord Injury version II
ZMP	Zero Moment Point

Symbols

θ	Joint angle	deg
θ_{act}	Actual angle	deg
θ_e	Position error	deg
$\theta_e Th_{lo}$	Lower threshold position error	deg
$\theta_e Th_{up}$	Upper threshold Position error	deg
θ_{ref}	Reference angle	deg
θ_{los}	Joint limit of stability	deg
$\dot{\theta}$	Joint velocity	m/s
$\dot{\theta}_r$	Reference Joint velocity	m/s
$\dot{\theta}_a$	Actual Joint velocity	m/s
$\ddot{\theta}$	Joint acceleration	m/s^2
τ_{act}	Actuator torque	Nm
τ_{int}	Interaction torque	Nm
$\tau_{int} Th_{lo}$	Lower threshold interaction torque	Nm
$\tau_{int} Th_{up}$	Upper threshold interaction torque	Nm
ΔK	Stiffness variation	Nm/deg
$\epsilon(X)$	error in Cartesian space	
$\epsilon(\theta)$	error in Joint space	
B	Damping matrix	
C	Coriolis and Centrifugal torques	
D	vector of dissipative-velocity	
f	frequency	kHz
F	Force	N
F_a	Actuator force	N
F_e	External force	N
F_m	Coriolis and Centrifugal force	N

F_{int}	Interaction force	N
G	Gravitational Torques	Nm
G_{init}	Gait Initiation	
I	Inertia Matrix	
J	Jacobian Matrix	
K	Stiffness	Nm/deg
K_{high}	Maximum Stiffness	Nm/deg
K_{low}	Minimum Stiffness	Nm/deg
K_j	Stiffness matrix	
s	Confidence factor	
T_a	Vector of Actuator torques	
T_e	Vector of Environment torques	

Chapter 1

Introduction

This chapter gives a detailed overview about the neurological disorder and the level of injury which defines the individuals as plegic patients. A detailed description and background study about these clinical aspects of the patient is needed before designing the kind of assistance to be proposed. This chapter will analyse the medical background and the need of an effective therapy to ensure a better outcome of the therapy. These introductory studies about the patients and therapies led the way in defining the objectives of this dissertation. Furthermore, the framework in which this thesis work has been performed is also elaborated in this chapter. The organization of this thesis dissertation is also briefly explained towards the end of this chapter.

1.1 Introduction

Robotic rehabilitation is an emerging technology in the field of Neurorehabilitation, which aims to achieve an effective patient recovery. Human behaviour has been the major topic of research to many scientists focussed on developing human centered strategies. The importance of human walking has helped in developing major humanoid robots such as ASIMO, HRP, KOBIAN, iCub etc. by following a pattern but it's still a mystery or an unsolved problem of mimicking or following the similar pattern as the human. Human locomotion involves a lot of sequential patterns which also creates a major challenge in terms of static and dynamic balance control[1]. Any neuro-musculoskeletal disorder results in the degradation of human balance control. In case of vestibular patients, they have excessive reliance on vision for a stable balance control. Studies on human's static and dynamic stability are used to understand the equilibrium conditions of human body balance and to implement control strategies in humanoids[2, 3] and exoskeletons[4].

Neurological disorders are one of the greatest threats to public health because of the indefinite cause behind its existence[5]. The two major neurological disorder are the hemiparesis and hemiplegic. These disorders are result of several medical conditions, such as trauma, tumours, stroke or congenital causes. Hemiparesis affects the right or left side of the body while hemiplegic or Spinal cord injury (SCI) leads to complete or incomplete paralysis. Stroke is one among the hemiparesis condition, which is observed primarily in adults. Spinal cord injury is characterized as a hemiplegic condition, which also involves different levels of lesions. Stroke survivors have been a main focus for rehabilitation for many decades because of the partial muscle strength possessed[6]. Traditional therapies with few or no need of assistive external devices have proved to be efficient in providing the sufficient therapy for both.

Nearly half of spinal cord injuries are the result of motor vehicle crashes[5]; the other major causes include falls, violence, and sports accidents [7]. Recent reports suggest that the number of new injuries due to violence peaked at 24.8% during the 1990s and has declined to some degree since then. Acts of violence remain the primary cause of SCI among some minority populations. The number stemming from motor vehicle crashes has diminished over a longer period, probably as a result of air bags and other improvements in auto safety. SCI individuals need rehabilitation for a longer period of time, which is also in function of

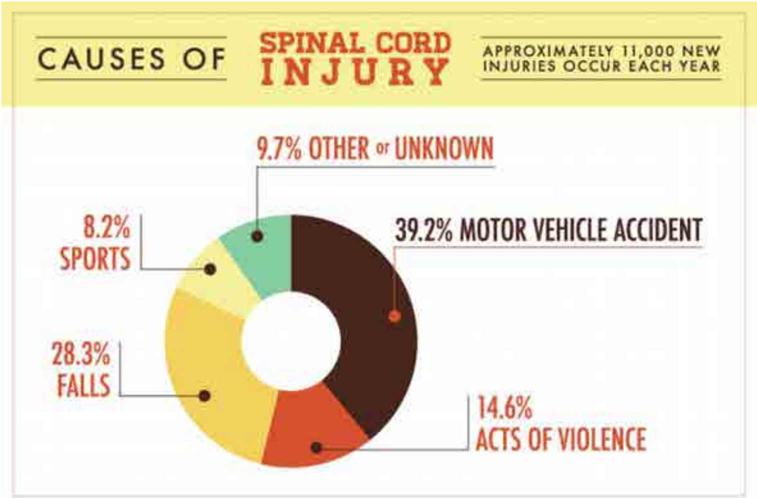

FIGURE 1.1: Basis of Spinal cord injury

the level of injury. In some cases it is impossible to see a significant improvement with the traditional therapy so the need of assistive devices is crucial.

Traditional rehabilitation therapies are laborious and intensive especially for gait rehabilitation, often requiring more than three therapists together to assist manually the legs and torso of the patient to perform training. This fact imposes an enormous economic burden to any country's health care system thus limiting its clinical acceptance. Furthermore, demographic change (aging), expected shortages of health care personnel, and the need for even higher quality care predict an increase in the average cost from first stroke to death in the future. All these factors stimulate innovation in the domain of rehabilitation [8] in such way it becomes more affordable and available for more patients and for a longer period of time. This led the way towards the development of assistive devices in rehabilitation. Robotics based rehabilitation therapy has become a topic of interest for many rehabilitation therapists and for researchers working in the field of medical robotics. Robotic rehabilitation can

1. Replace the physical training effort of a therapist, allowing more intensive repetitive motions and delivering therapy at a reasonable cost
2. Assess quantitatively the level of motor recovery by measuring force and movement patterns[9]

There has been a wide range of research focussed on the effectiveness of therapy and training for stroke patients. In case of SCI, the therapy must vary depending on the lesion level of the user, to demonstrate the assistance as needed strategy. The other challenge is adapting to the user needs based on their performance (movement and interaction). The next section will discuss in detail about the different levels of SCI and the level of impairment caused as a result of it.

1.2 Spinal Cord

Spinal cord is responsible for transferring the nerve impulses from the brain for performing a voluntary motion. The spinal cord is an extension of the brain which spreads to the different parts of the body Fig1.2. A damage in the spinal cord results in the loss of voluntary motion (paralysis) and sensation. The spinal cord, encased in a protective by

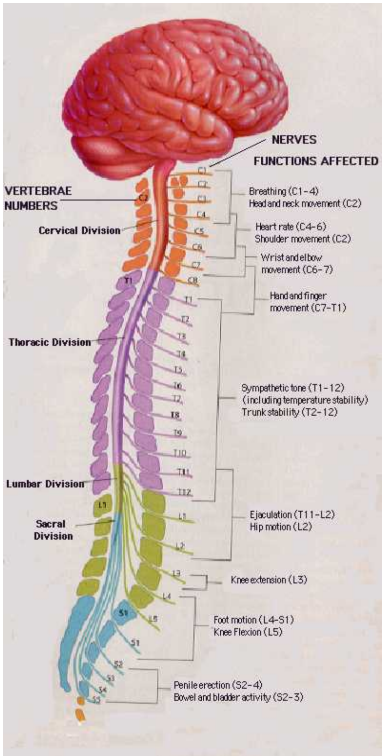

FIGURE 1.2: Illustration of Spinal cord and its functions

the spinal vertebrae, extends from the brain stem to a point in the lower back called conus medularis. At each vertebral junction, a pair of spinal nerves exit from the spinal cord and innervate specific muscles and sensory nerve filaments enter the spinal cord. The vertebrae and nerves are classified into several sections beginning at the neck with seven cervical vertebrae, seven pairs of nerve roots that exit above each of those vertebrae and eighth pair that exits below the seventh vertebrae. Below the cervical vertebrae are 12 thoracic vertebrae and 12 pairs of spinal nerves. In the lower back there are five lumbar vertebrae and nerve roots and five used sacral vertebrae with five nerve roots. The lowest part of the spinal column is a single bone called the coccyx. The vertebrae and nerves are numbered from the top with a letter that corresponds to the spinal section. The first vertebrae below the skull is C-1, whereas T-1 is the first vertebra in the thoracic section [6].

Nerves for the voluntary motor system originate in the motor cortex of the brain and extend down through the basal ganglia to the brain stem. Here they cross over to the opposite side and continue to descend in the spinal cord until they synapse at the point where they are about to exit from the spinal cord. The nerves that originate in the motor cortex of the brain's left hemisphere cross over to innervate the right side of the body, and those from the right hemisphere cross over to the left side. These are known as "upper motor neurons." Beyond the synapse, the "lower motor neuron" exits the cord and extends to its particular muscle destination. Any point in the body, then, is connected to the controlling center in the brain by only two neurons (except in the case of sensory neurons).

When the spinal cord is damaged, communication is disrupted between the brain and parts of the body that are innervated at or below the lesion. The lesion may be complete (no nerve fibers are functioning below the level of injury) or incomplete (one or more nerve fibers is secure). The cord need not be completely severed to result in a complete injury; the nerve cells may be destroyed as a result of pressure,bruising, or loss of blood supply, and if they die they do not have the ability to regenerate. The amount of functional loss depends upon the level of injury (the higher the damage occurs, the more of the body that is affected) and on the neurological completeness of the injury. Individuals with neurologically complete injuries have more severe and more predictable patterns of functional impairment.

1.2.1 Levels of Injury

The level of SCI can be determined in two ways: bony damage and neurological damage. The first method specifies the level of bony damage, observed from the radiological results (x-ray) and the second method is based on neurological damage, measured by careful testing of an individual's ability to perceive pinprick. Skin surface is mapped into segments called dermatomes, where each dermatome is known to be innervated by sensory nerves at a particular spinal level. Testing the skin, therefore, can reveal the level at which the spinal cord has been damaged. Individuals with the cervical level injury will have impaired

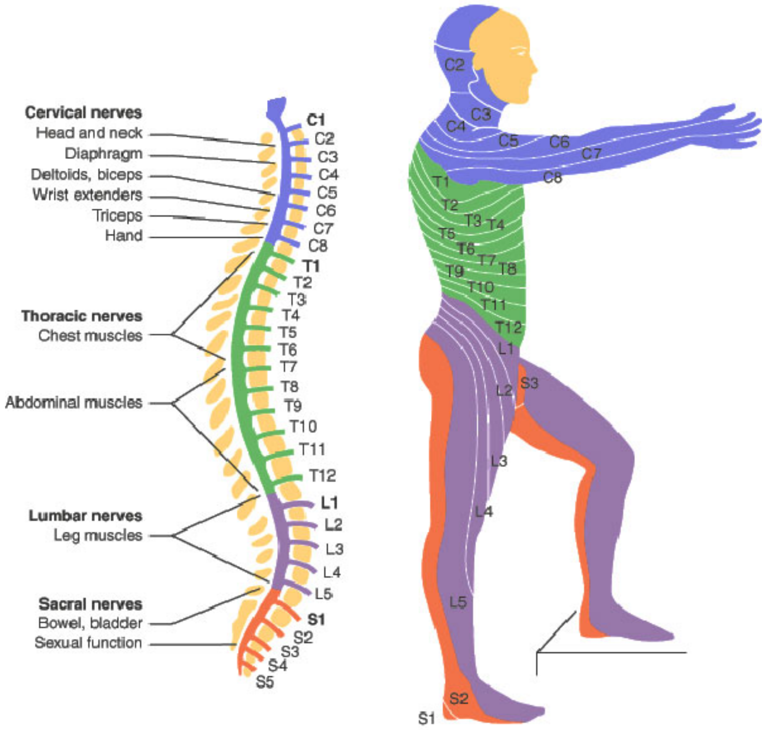

FIGURE 1.3: Different Levels of Spinal cord injury and its functional impairment

function in both their upper and lower extremities, a condition known as tetraplegia [7]. Those who are injured at or below the thoracic level will have paraplegia, with function maintained in their upper extremities but some degree of impairment in the trunk and lower extremities. The American Spinal Injury Association (ASIA) developed a system for describing the severity of injury that is widely used in the medical community using letters that pertain to the extent of injury (usually A through D) [10]. The Table 1.1 explains the different impairment scales and the level of the impairment as per the ASIA scale.

ASIA A injuries are complete, with no motor or sensory function preserved below the neurological level of injury, including the sacral segments S4-S5. ASIA B injuries are incomplete, with sensory but no motor function preserved below the neurological level of injury. ASIA C and D classifications refer to incomplete injuries with increasing degrees of motor function preserved below the neurological level of injury. Certain incomplete spinal cord injuries produce unusual patterns of deficits, depending upon which tracts within the cord are affected. If the damage occurs within the central part of the cervical cord, leaving the outer ring of fibres intact, the individual will have greater weakness in the upper limbs than in the lower limbs, and sacral sensation may be spared. Brown-Sequard syndrome is a lesion that affects only one side of the cord. This causes paralysis on the same side of the body as the lesion, and loss of pain and temperature sensation on the opposite side of the body.

TABLE 1.1: ASIA impairment Scale of different lesions

Grade	Description
A	Complete; no sensory or motor function preserved in the sacral segments S4-S5
B	Incomplete; sensory but not motor function preserved below the neurological level and extending through the sacral segment S4-S5
C	Incomplete; motor function preserved below the neurological level; most key muscles have a grade < 3
D	Incomplete; motor function preserved below the neurological level; most key muscles have a grade > 3
E	Normal motor and sensory function

1.2.2 Functional Expectations

Along with the SCI, a person may have an array of other complications including fractures, internal injuries, and brain injuries, all of which require treatment. When the need for acute medical services has passed, the individual is usually transferred to a rehabilitation unit for multidisciplinary services to help build strength, redevelop skills in activities of daily living, identify and obtain adaptive equipment, and prepare the individual and the family for return to home and community. An SCI rehabilitation team typically includes one or more physicians, nurses, physical therapists, psychologists, occupational therapists, rehabilitation counsellors and social workers.

For individuals with complete lesions, patterns of functional loss and preservation are fairly consistent from person to person [11]. The most dramatic changes in function are apparent between adjacent neurological levels in the cervical area. For example, nerves that innervate the diaphragm are at cervical level 3-4, so many persons with injuries at or above C-3 need ventilator assistance to breathe. Most individuals with C-4 injuries regain breathing capacity, but do not have usable function in their arms and hands. As a result, they need assistance with virtually all activities of daily living (ADLs), including feeding and dressing. Individuals with injuries at the C-5 level usually have function in the deltoid and biceps muscles which gives them the possibility of using an automatic motion known as tenodesis (i.e.,when the wrist is extended, the thumb and index finger come together). This permits them to hold a light object and carry out some self-care activities. Most people with C-5 injuries use power wheelchairs for mobility while individuals with C-6 injuries use manual wheelchairs equipped with plastic wheel rims and projection knobs. Most people with C-6 tetraplegia can move independently with the use of appropriate equipment. People who have C-7 and C-8 levels of injury can use the triceps muscles, which permits them to lock their arms for transfers to and from wheelchair to other mobility devices and are nearly independent in their daily lives.

At the thoracic level, the nerves innervate muscles that provide control of the trunk, so injuries below this level allow for some balance and trunk stability. Control over hip muscles is maintained at the lumbar level, so individuals with injuries below T-12 are sometimes able to ambulate using crutches and braces. This takes a great amount of effort, however, and most people choose to use wheelchairs for mobility, particularly when they are covering anything except short distances within a home or workplace.

Injuries at the lumbar level often results in paraplegia. Persons with L1-L2 injury may be capable of standing and walking with braces. However, this involves a great amount of energy even for short time periods; a wheelchair will be needed for mobility of any significant distance. Those with L3-L4 injury also may be able to walk with orthotic devices. Injuries at the cervical or thoracic level damages all of the upper motor neurons that innervate the body at or below the point of injury, which results in loss of voluntary control of bowel and bladder functioning[12]. Fortunately, an intact lower motor neuron reflex arc may be retained from these muscles to the synapse in the spinal cord and back. This enables the possibility of "retraining" the body to respond to direct stimulation and provides some degree of control over bowel functions. Individuals who have sacral injuries, on the other hand, may have direct injury to the lower motor neurons in this area hence they may not have the ability to develop these reflexive responses.

With these functional impairments in spinal cord injured individuals, there is always a need to provide a sufficient and effective therapy. The therapy must be more patient specific, personalized and capable of combining multiple devices to increase its outcome. The objective of this thesis is defined in the next section and is mainly focussed towards a human-centered rehabilitation for SCI individuals.

1.3 Objectives

Robot based rehabilitation therapies are expanding their levels of assistance, but still many unsolved questions exist. Completely assisted robot therapies can induce slacking to the patient, while an excessive demand on the rehabilitation task or exercise on unassisted robot strategy might cause them some harm, so it is necessary to provide a personalized assistance, dynamically adapted to the changing patient's needs. The objective of this dissertation is **to advance a step-forward in human-centered rehabilitation approaches, by providing the necessary robotic assistance, using a wearable robot and the possibility to combine with subsequent necessary assistive devices**. The following are the intermediate goals which are defined under the global paradigm of 'Assist-as-needed' (AAN), in a human-centered rehabilitation scenario.

1. Identifying an **optimal control scheme** or architecture for assisting the patient-in-need, using an exoskeleton. The performance of the control approach can be studied by means of simulation, to understand the functioning of the orthotic system and to define the assistance to be provided. By means of simulation studies, the assistance or resistance needed for a specific kind of movement can also be identified.

2. **Perform adequate assistive (or resistive) action**, to handle the influence of disturbance in a trajectory, when and as needed. Disturbances can be due to the human or by the effect of external assistive devices. Assistance or resistance must be programmable by therapist for specific joints, time and evolution depending on the user performance.

3. Propose a **control model for human-centered rehabilitation** tasks in lower limb, such as Sit-to-stand, Balance training and Gait, to be handled in a secure framework to assist the patient whenever needed. The level of assistance to the tasks must be in function of the human-orthosis interaction and movement. The therapist must be able to vary the assistance level or confidence, depending on the patients advancement in the therapy.

4. Allow the **combination of assistive devices** based on the therapeutic needs must be possible and the control model must be compatible in accepting multiple inputs. The devices can be used to provide the necessary assistance or to support the movement

of the user. In some cases, the external devices can be just to monitor the user interaction with the orthosis.

5. **Real time adaptation of joint movement** is necessary to handle the equilibrium of the patient and to help in gait assistance. In gait assistance, it is important to follow the gait pattern with real time input and synchronized joint movements. An effect in one joint obviously affects the trajectory of other joints, due to the dynamics and correlation between them.

6. **Short term and long term** adaptation to the user movement must be satisfied. Short term goals such as supporting the user movement within a gait sequence is needed to ensure dynamic stability and provide a sufficient assistance. On a prolonged or continuous gait training, the support must vary depending on the user's physical conditions, fatigue or slacking.

7. Implement **an interactive orthosis system** by considering the human intentions (volitional control) and reactive inputs. Study the human intentions by using the motor-neural activity and mechanical interaction, to provide physiological feedback to the therapy and volitional control to the assistance.

8. **Evaluation of control performance** in a human-centered approach is needed to ensure the adaptation is supportive and enhances the rehabilitation. The assessment is to ensure that the adaptive control is efficient to handle the human-orthosis interaction in real time. In a human-centered environment, the performance can be analysed by the human interaction with the orthosis and the joint parameters obtained.

9. **Comprehensive clinical evaluation** for the assessment of the patient performance is necessary to validate the efficiency of the system. The assessment must include both the physical and clinical attributes of the patient. Physical assessments and statistical assessments such as SCIM, Borg scale, QUEST can be used as evaluation tools.

1.4 HYPER

The presented work is performed as part of the project HYPER (Hybrid Neuroprosthetic and Neurorobotic Devices for Functional Compensation and Rehabilitation of Motor disorders) is funded by Ministerio de Ciencia y Innovación, Spain (CSD2009 - 00067 CONSOLIDER INGENIO 2010). The research in this project is focused for advancing in Neurorobotic (NR) and Neuroprosthetic (MNP) devices in interaction with the human body. The objective is enhancement of both rehabilitation and functional compensation of motor disorders in activities of daily living by using the individual or combined action of neurorobot (exoskeletons) and neuroprosthesis (MNP). These devices, by their combined action, can enhance and help restore the latent capacities of patients suffering from stroke or spinal cord injury, have lower limb motor or higher disorders. The integration of the various components of this project aims to overcome the major limitations of current rehabilitation solutions for particular cases of stroke and spinal cord lesions. It is a combination of artificial and biological structures integrated to restore motor function in patients. The project aims to validate, both clinically and functionally, the concept of developing hybrid

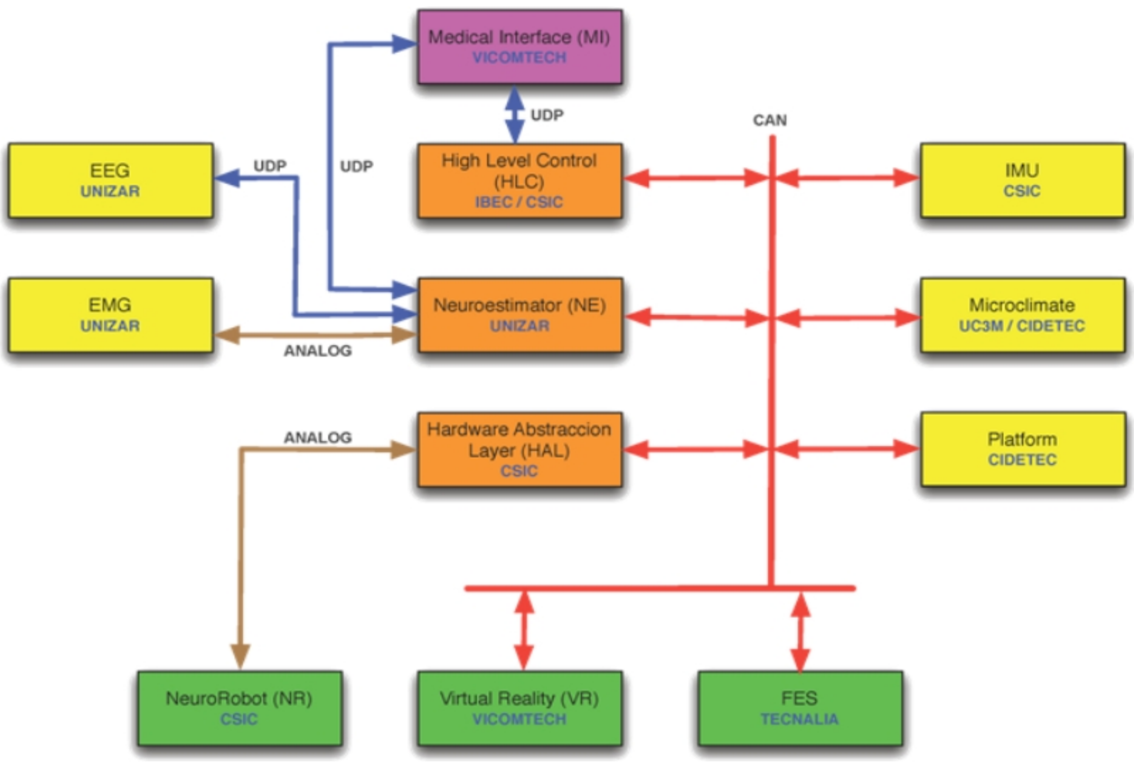

FIGURE 1.4: Consortium of project HYPER

devices for rehabilitation and functional compensation of motor disorders, under an assist-as-needed paradigm. The main challenges are to improve the outcome of therapy and allow an early recovery.

Furthermore, HYPER proposes a multi-modal BMI (Brain Machine Interface) system. Its main objective is to explore different levels of neural activity and characterize the involvement of the user with the rehabilitation process. Based on the BMI information, EMG (Electromyography) signals and a set of sensors and algorithms, a Neuroestimator tries to estimate patient's current capability to perform a task. This information is used to modify in real time the intervention with the hybrid systems. The consortium consisted of a combination of research centres, universities and specialised hospitals which helped in achieving the multidisciplinary aspect of this thesis (Fig 1.4).

1.5 Organization

This thesis dissertation has been developed based on the above mentioned objectives and within the specified framework of project HYPER. The thesis is organized as five chapters; Chapter 2 to Chapter 5 presents the human centered rehabilitation approach and its evaluation with different participants, while Chapter 6 presents the conclusions, contributions and future focus related to the work.

Chapter 2 introduces in brief about the existing methods in robotic rehabilitation models and which have been used by many prominent robotic devices. The assistive robots for rehabilitation can be classified as treadmill based training models and the over-ground or wearable robot models.The rehabilitation robots can be further classified based on the type of actuators which in turn influences the kind of control model to be applied. Traditional control models are used as basic models in combination of hybrid approaches.

Chapter 3 explains the simulation study performed analysing the kind of control model needed to provide the suitable assistance. The simulation model was developed based on the real parameters of the exoskeleton and its efficiency was evaluated by inducing disturbances. Two control methods, Cartesian based and joint space based, were used to recover the trajectory. This study was used to understand the role of interaction in a human-centered strategy.

Chapter 4 presents the adaptive control approach for the three lower-limb rehabilitation tasks, Sit-to-stand, Balance Training and Walking. The combination of external devices was also performed, in case of Sit-to-stand and Balance training, as an external support tool for assistance. In case of gait training, the role of volitional order initiation was considered to provide assistance based on the user. The adaptive control model was influenced by the human-orthosis interaction torques and the confidence level of the therapist, to modify its input joint function.

Chapter 5 elucidates the performance assessment of the proposed control model with healthy subjects and incomplete SCI individuals. The patient performance was evaluated based on the clinical and statistical measures QUEST. The assessment of the control model was deduced based on the patients performance; change in their joint flexion/extension and interaction with the orthosis. Borg scale is used to assess the effectiveness and the repeatability of the clinical intervention.

Chapter 6 presents the contributions which have been achieved as a result of this dissertation and the publications in prominent conferences and high impact journals. The possible future extension of this dissertation work is also expressed in this chapter.

Chapter 2

Control Strategies in Rehabilitation Robotics

This chapter briefly analyse the basic control strategies and their importance in specific to their significant contribution to the technology. The role of such basic control models in a rehabilitation set-up is necessary to understand and model the assistive behaviour. This chapter also presents the existing control approaches, based on the control strategy and the hardware set-up, in rehabilitation scenario. Further, the existing the rehabilitation robotics devices are studied briefly to analyse their advantages and disadvantages in terms of a human-centered strategy. These studies paved the way to understand the aspect of an Assist-as-needed strategy for SCI individuals.

2.1 Introduction

Rehabilitation robotics is intended to function in a human-centered environment with the necessary interaction of the patient. A better interaction with the orthosis encourages the patient to pursue or advance the therapy. The global objective of any researcher focussed in developing a rehabilitation robot, irrespective of upper or lower limb, is to adhere to the paradigm "Assistance-as-needed" (AAN). The concept of assistance varies depending on the human involved and their dynamics. Several researchers have developed different kind of approaches based on the kind of assistance or type of impairment and motivated to propose control strategies to support AAN. An efficient AAN control strategy can be achieved by understanding the human dynamics involved, developing an efficient hardware and an interactive control approach.

Robots in rehabilitation can be classified as assistive and therapeutic robots. Wearable robots are mainly oriented to assist individuals in performing their activities of daily living (ADL)[4], while they have rarely been used as therapeutic robots. The goal of therapeutic robots is training and enhancing the patient capabilities affected by neuro-muscular deterioration, such as: chronic degenerative low back pain, head injury, stroke, peripheral neuropathies and cerebral palsy [13].

Studies about the human control is necessary to develop an efficient human-centered strategy. The dynamics involved in the human system is interesting and challenging to understand the in-built concepts of how the postural and dynamic stability is perceived by human. The dynamics involved in maintaining the body posture to prevent falling is defined as balance and this is related to the inertial forces acting on the body and body segments. The study of human balance is crucial for understanding the concept of maintaining the equilibrium.

Humans use multiple strategies in order to maintain their equilibrium while standing and walking [13, 14]. A study conducted by Kuo and Zajac [15], shows the multi joint strategies involved in maintaining a standing posture in the presence of a constraint like blocking the knee movement. Winter et al. [16] demonstrated that the muscles controlling the sway motion along the medial/lateral direction apply a simple spring stiffness control that maintains balance. The motor mechanisms involved in human balance control have been studied and analysed by Winter et al. [17] . They developed a spring damping balance system the stiffness of which varies according to the oscillations of the inverted pendulum model. The studies carried out on human balance and the motor strategies involved in maintaining postural stability have paved the way to the design of methods that ensure the right robotic AAN. Thus, the goal of assistive robotic systems is not to override the human control, but to involve the user in the control so as to avoid slacking.

These studies on humans, have been used to develop and implement control strategies in robotics. The concept of robots in rehabilitation can be further classified as wearable and therapeutic robots. While, wearable robots are mainly oriented to assist individuals in performing their activity for daily living, the goal of therapeutic robots is training and enhancing the patient's capabilities due to neuro muscular deterioration. To develop a realistic control hypothesis, it is essential to understand how humans perceive balance, as well as the strategies involved in training. This chapter illustrates in detail about the traditional control models, application structure and the existing hardware types which have tried to answer the research paradigm of AAN.

2.2 Control Strategies

Control strategies for rehabilitation robots need to be efficient in understanding the patients mobilities and dependence and their subjective training policies. An efficient policy can be defined as the one which can assist the patient whenever needed. But the term assistance is itself not clearly explained in many cases, hence its hard to define the kind of assistance to be provided. Control strategies intended for the development of efficient robotic therapies have been evolving rapidly. As proposed by Marchal-Crespo and Reinkensmeyer [18], robotic control strategies can be categorized based on their type of application: assistive, resistive and haptic-simulation. In lower limb rehabilitation, robotic control strategies should be assistive or resistive so that a stable therapy can be ensured. It is always important to find the most effective control algorithm in order to produce greater rehabilitation benefits. Force based or impedance based control strategies have proven to be efficient in rehabilitation therapies. With regard to human-robot interaction, force based adaptive control plays an important role in providing an efficient robot behaviour The dynamics of interaction forces are characterized by the joint impedance properties[19].

2.2.1 Position Control

Position control is the most traditionally viable method commonly used in the industrial sector to ensure precision of movements. In contrast the role of position control in rehabilitation is very less, since there is a narrow range of possible improvement in such a strategy. This type of control can be used in exceptional cases, provided the user has very less mobility options and they are completely dependent on the robot. The use of position control is efficient in following a prescribed path which provokes slacking in the user, since there is no room for the user to improve in this mode of therapies[18]. Patient's physical effort is also reduced by trajectory tracking training [18, 20] summarised this phenomenon as "Slacking Hypothesis", which means that a rehabilitation robot could possibly decrease recovery as a decrease in motor output, effort, energy consumption and/or attention.

Trajectory tracking or position control 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. For trajectory tracking, the reference trajectories can be defined based on mathematical models of normative gait trajectories [21–23] or pre-recorded trajectories from healthy individuals [24–27] or by the movement of the unimpaired limb [28].

Reference trajectories adopted from literature have been used [29–31], but customised to patients to provide a suitable user-specific reference trajectory. Moreover, the desired reference trajectory can also be generated online according to the movement trajectory of the unimpaired limb of hemiparetic patients such as in a upper limb robotic device[32]. Even though the reference trajectory generation are customised, trajectory tracking control still has some limitations such as decrease in motor learning [33, 34] due to the dynamics of the task. Hence the trained task is not exactly the target task which does not fulfil the motor learn rationale that training needs to be task specific. Furthermore,this guidance based movement reduces the burdens on the participant's motor system to discover [33] the principles needed to perform the task successfully.

Trajectory tracking is suitable for training of patients with SCI or acute stroke when they have no muscular strength to move their limbs [35]. A potential issue with this method is the imposition of a predefined trajectory which limits kinematic error, an important parameter that drives human motor learning [36–38]. This may result in abnormal gait pattern generations and would leave the patient unable to adapt to physiological gait [32].

Patient's active participation and involvement in the robotic gait training process is important to develop neuro plasticity and motor control [36, 39]. The terms of patient cooperative, assist as needed, adaptive and interactive robot assisted gait training are used in literature [25, 40] which are achieved by impedance and adaptation based control strategies, to actively involve the patient in training process. This resulted in the development of force-based control strategies in rehabilitation.

2.2.2 Force Control

Force control method in robots have expanded widely in the past few decades due to its inevitable property of providing enhanced sensory capabilities. This model enhances the performance of the robot which were limited only to industrialized procedures by using the tactile sensing, visual feedback and more. Hence this control model is also considered efficient for the methods in rehabilitation, with more user adaptive characters[41]. Motion control is deemed to be inadequate because of the unavoidable modelling errors causing a rise in the contact force, leading to an unstable behaviour during the interaction. Force feedback helps in achieving a robust and versatile behaviour of a robotic system, especially in presence of humans. Force based controllers are effective, based on the variables to be modified and the objectives to be achieved, and thus proved to be essential in therapies involving patient cooperative strategies [42]. Shadmehr and Wise [43] applied a force perturbation to the robot, which causes large errors in the trajectory and the subjects must adapt to it. This process of inducing perturbations has enabled scientists to test several hypotheses to study the computational mechanisms of motor control and motor learning.

In force control, the planning errors increase the contact force and moment, which also lead to change in the end-effector position from the desired trajectory. Similarly, the control model applies an opposition force to reduce such deviations in the trajectory. This leads to a build up of contact force until saturation of the joint actuators are reached. A compliant behaviour is necessary to avoid collision with a high stiffness environment. Such interaction model can be classified as passive or active based on their application.

Passive interaction control model alters the movement of the robot in relation to their interaction forces. The compliant mechanism can be achieved as a combination of structural compliance of the joints or links. This type of passivity does not require any force or torque sensors and the trajectory cannot be modified in real time. The major drawback of such a system is their lack of flexibility, handle lower errors and cannot guarantee higher contact forces.

In active interaction control, the interaction forces are feedback to the controller to modify or generate new trajectory positions based on the measurement of the contact force and moment. Active interaction model can be slower and more sophisticated in comparison with the passive model. But the execution speed and disturbance rejection capability can be improved with a combination of some degree of passive compliance, which will also maintain the reaction forces at lower level [44]. These active interaction based approach can be further grouped as direct or indirect force-based algorithms in function of its performance.

2.2.2.1 Direct force control - Hybrid force-motion and Hybrid force-position

2.2.2.2 Indirect force control - Impedance and Admittance

2.2.2.1 Direct Force control

The direct force control is capable of handling the contact force and moment to achieve a desired value, with a force feedback loop [45]. This model of force application needs a specified desired motion and force with respect to the constraints imposed by the environment. Hybrid force-motion control can be grouped in this model with controlled motion along the unconstrained task directions and force along the constrained task directions[44]. In the absence of an accurate model of the environment the force and motion control action can be called hybrid force-position control, where the force control dominates the motion control to tolerate the position error [46]. Force feedback approach reduces the impedance, which is subject to some non-linear friction force. The control law for an augmented force feedback controller is given by the equation,

$$F_a = K(r - x) + B(\dot{r} - \dot{x}) + K_f[F_e + K(r - x) + B(\dot{r} - \dot{x})] \tag{2.1}$$

K and B are the scalar desired stiffness and damping respectively and r represents the virtual position. The force feedback term serves to minimize the deviation of the actual endpoint force from the desired force, which looks like a damped spring characterized by K and B connected to the virtual trajectory.

$$M(\ddot{x}) + B(\dot{x}) + F_m(x, \dot{x}) = F_a + F_e \tag{2.2}$$

M is the mass, B corresponds to damping and F_a, F_e correspond to the actuator force and the external force or disturbance. F_m represents the Coriolis and centrifugal forces acting on the robot. From the above relation, the proportional feedback control reduces the mass and friction by a factor of the force feedback gain plus one, so that arbitrarily large force gains drive the actual impedance to the desired impedance specified by the controller.

2.2.2.2 Indirect force control

Indirect force control do not possess an explicit force feedback to control the contact force and moment to a desired value. Impedance and admittance are the common type of control with the indirect force approach. The deviation from a desired trajectory or motion is in relation with the mechanical impedance and admittance of the robot.

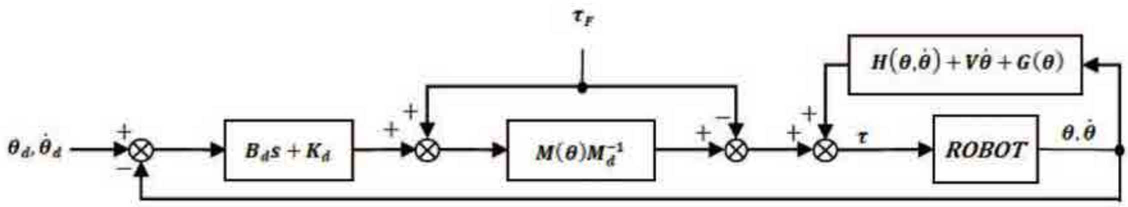

FIGURE 2.1: General scheme of an Impedance control approach [47]

In impedance control, the deviation is corrected by generating forces, characterized by a given mass, damping and stiffness. The acceleration input is computed based on the position and orientation feedback along with the force and moment measurements. Indirect force control with a static relation between the deviation of position and orientation of the robot with the desired motion and contact force being considered: such as Stiffness and Compliance control[48]. The mechanical interaction dynamics may be characterized by mechanical impedance, which can be considered as dynamic stiffness[49]. Low mechanical impedance reduces interaction forces, protecting both robot and any object it manipulates[50]. Vanderborght et al. [51] explains the role of impedance control and its variation in terms of active control models with both fixed and variable impedance in actuators. Unluhisarcikli et al. [52] developed lower limb exoskeleton equipped with force field controller which helps in providing the suitable impedance needed while walking. In therapeutic robots, the interactive behaviour is one of the major objective of control, which requires intimate physical interaction with humans[54, 55]. This type of versatile interaction requires the ability to modulate and control the dynamics of interaction[56, 57]. Impedance in the joint

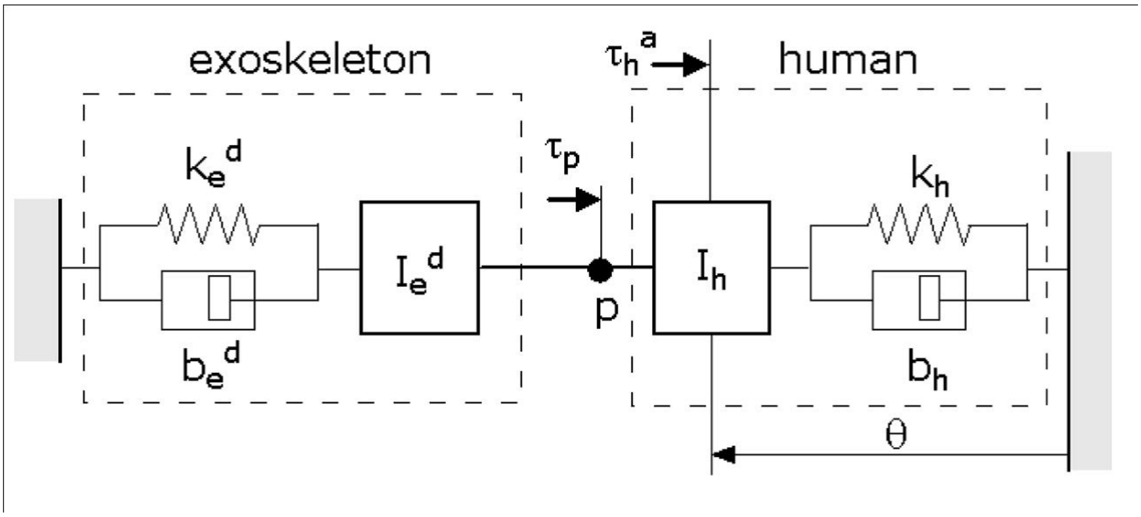

FIGURE 2.2: Impedance control approach on a 1-DoF exoskeleton [53]

level training has also been widely investigated by many researchers [58] to understand the role of ankle joint during stance phase of walking and to elucidate how ankle impedance is regulated during locomotion. Aguirre-Ollinger et al. [53] use an 1-DoF exoskeleton to demonstrate that active impedance control improves the dynamic response of the human limbs without affecting the user-induced control orders.

A primitive approach proposed by Hogan [59], termed as simple impedance control, consists of driving a low friction mechanism with force or torque controlled actuators and using motion feedback to increase the output impedance.

$$I(\theta)\ddot{\theta} + C(\theta, \dot{\theta}) + D(\dot{\theta}) = T_a + T_e \tag{2.3}$$

where, θ is a vector of robot joint variables, I is the inertia matrix, C denotes torques due to Coriolis or centrifugal accelerations, D is a vector of dissipative-velocity dependent torques, T_a is a vector of actuator torques and T_e vector of environmental torques. The impedance target behaviour of a spring with stiffness matrix K_j and a damping matrix of B_j, will be expressed as

$$T_a(\theta, \dot{\theta}) = K_j(\theta_o - \theta) + B_j(\dot{\theta}_o - \dot{\theta}) \tag{2.4}$$

where,θ_o is a virtual joint trajectory in robot joint space. combining the above two equations we get,

$$I(\theta)\ddot{\theta} + C(\theta, \dot{\theta}) + D(\dot{\theta}) + B_j(\dot{\theta}) + K_j(\theta) = K_j(\theta_o - \theta) + B_j(\dot{\theta}_o - \dot{\theta}) + T_e \tag{2.5}$$

This controller implements in robot joint space a dynamic behaviour analogous to that depicted by the spring controller and damper specified towards a virtual trajectory. This control scheme ensures following the desired trajectory, in the absence of interaction forces. In case of contact with the environment, a compliant behaviour is imposed at the expense of a finite position and orientation displacement [60].

In admittance, the control reacts to the interaction forces by imposing a deviation from the desired motion. The motion control is made stiff to enhance disturbance rejection which ensures tracking of a reference position resulting from the impedance control action. A stiffness approach can be realised by assigning a desired position and orientation and a suitable static relationship between the deviation of the end effector position and orientation from the desired motion and the force exerted on the environment. The selection of stiffness parameters strongly relies on the kind of task to be executed. A high value of stiffness results in higher accuracy at the expense of higher interaction forces. In the presence of some physical constraint the stiffness must be reduced to ensure low interaction forces. Hence an adaptive stiffness model is necessary to avoid any discrepancies between the desired and achieved positions due to the constraints imposed by the environment.

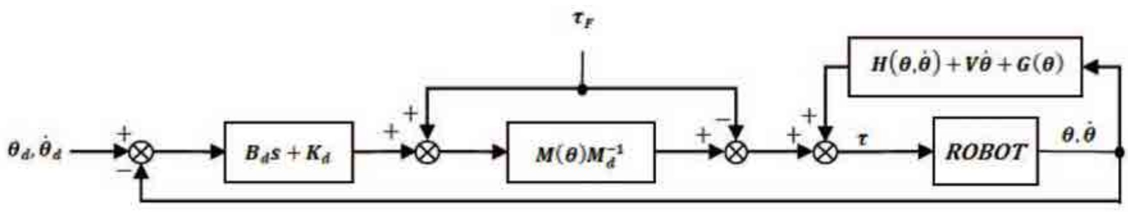

FIGURE 2.3: General scheme of an Admittance control approach [61]

2.2.3 Adaptive Control

The potential issue with trajectory tracking and impedance control based training is the controller parameters cannot be varied based on real time judgement of the patient's abilities [62]. Adaptive assistance is utilized to enhance patient's active participation in the training process [37] and to produce patient desired robot motion.The term adaptation is used for real time tuning of the controllers designed for robotic actuators to match patient's disability level and to actively involve him in the training process. Robot motion is initiated from the physical interaction between the patient and orthosis. As disability level varies from subject to subject, online estimation of patient–orthosis interaction force is the most

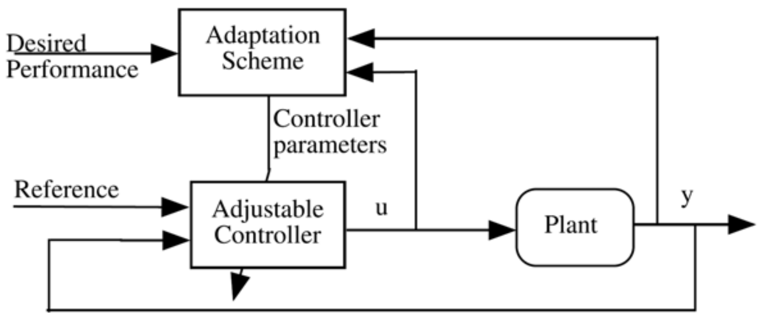

FIGURE 2.4: A simplified model of the Adaptive control system [62]

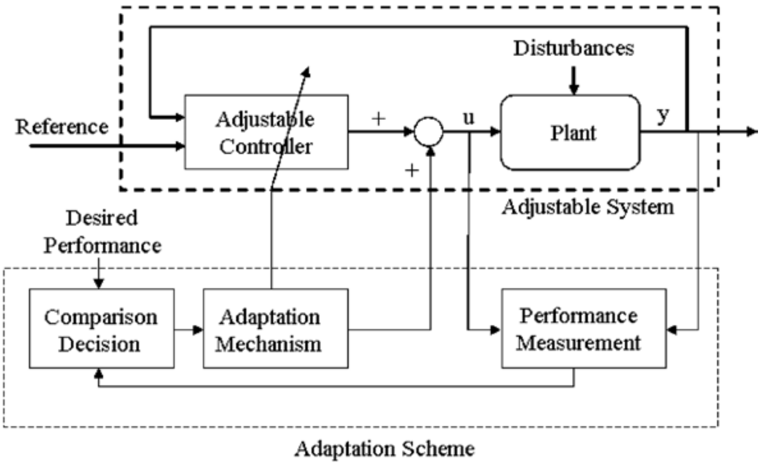

FIGURE 2.5: Basic configuration of the Adaptive control system [62]

crucial task in adaptive assistance paradigm [35]. Figure 2.4 demonstrate a dynamic model where the input/output parameters influence the controller performance in real time.

Adaptive control is capable of handling the disturbances by acting directly on the controlled variables and the disturbance affecting the controller's performance. Figure 2.5 illustrates the configuration of an adaptive control system, in presence of the disturbances acting it. Adaptive approaches have also been used to adjust the desired trajectory, such as based on contact forces between the robot and the limb [25], re-planning the (minimum jerk) desired trajectory at every time sample based on the actual performance of the participant [40] or adjusting the replay-timing for better synchronizing a compliant gait training robot to the participant [31]. The improvised trajectory control for rehabilitation with minimum jerk helps to perform smooth trajectories for safer interaction between human and a haptic interface. An Adaptive trajectory control established by tuning the inertia, damping and stiffness has resulted in a impedance based approach which resulted in a smooth performance of human machine interaction [63, 64].

Mefoued [65] proposed an adaptive control driven by a multilayer perceptron neural network to actuate an orthosis for knee movements. The controller adapts to the different uncertainties and non-linearities based on the interaction of the subject. Physiotherabot adapted to the joint movements of the users based on the feedback data received from the patient with a rule based control architecture [66]. A powered orthosis, with both active and passive based approaches Hasani et al. [67], had the ability to provide the knee joint torque support according to the intention and ability of the wearer, where the intentions are estimated using the muscle model. All these studies imply the need of an efficient and effective therapy by considering the internal and external changes affecting the therapy or movement. The efficiency and effectiveness of a system can be ensured by the reactive and proactive behaviour of the robot with respect to the patients' movement.

The uniformity of the adaptive process is not guaranteed during different training sessions [36]. The adaptive algorithms estimates the human-orthosis interaction force from the combined dynamic model of the human and orthosis mechanism. The quality of interaction force estimation is dependent on accuracy of force and joint position sensors and also on the estimation algorithm [68]. The abrupt forces like muscle spasms arising from patient and resulting actuator non-backdrivability presents a major problem to the interaction torque estimation. Backdrivability is the ability of the robot being moved by the patient with low mechanical impedance to allow patients voluntary movements.

2.3 Control Approaches

Rehabilitation robots can be classified based on their control architecture or the application of the control strategy or by the combination of sources. The control approaches can be classified based on the model of the control to be observed irrespective of any kind of devices. In some cases, a combination of the approaches are also possible which helps in adhering to the general paradigm of providing AAN. The control approaches can be categorized as

2.3.1 Model-based

2.3.2 Hierarchical

2.3.3 Interaction-based

2.3.4 Task-based

2.3.1 Model-based

Model based controllers can be categorised based on the kind of model being used, dynamic and muscle based control model. Most of these model based controls are complicated in terms of establishing a predefined model of the patient and exoskeleton. The robotic actuations are computed considering the inertia compensation, Zero moment point (ZMP) [69, 70], balance criterion and thus relying on series of sensors to recognize the kinematic and dynamic variables [71]. It has been tested with a 1 DoF exoskeleton with an inertia compensation estimated from the angular accelerations [72, 73]. Low et al. [74, 75] used a ZMP based control model, for a payload carrying robot, to adapt the joint evolution and applied a trunk compensation to assist when needed. The assistance was defined when the desired ZMP was varying from the actual ZMP evolution. A similar approach based on the COM projection was used to evaluate the performance of a humanoid model, to perform sit-to-stand movement [76]. Stewart platform type AFO , with 6 linear pneumatic actuators were controlled based on classic forward/inverse kinematic model [77, 78]. Model based controllers have been used in hybrid approaches such as Electromyography (EMG) based assistance, where the control loop estimates the input torque based on the myoelectric signals [79, 80].

2.3.2 Hierarchical

A hierarchical control model has the ability to combine physical level interaction and signal level feedback loops which helps in understanding both the environment and the user level to permit a better assistance. Tucker et al. [81] proposed a generalised hierarchical framework for both orthotic and prosthetic devices. Such hierarchical model has been employed in many exoskeleton structures to ensure safety at every point of the therapy. Lu et al. [82] used a four layer control model to realise an adaptive control scheme, by applying forces and learning the impedance parameters of both the human and the robot. The Hierarchical approach involves multiple layers of master/slave or handshake control models which helps in the better adaptation. The layers in this approach are generally classified as Low-Level Control(LLC), Middle-Level(MLC) and High-Level(HLC). The LLC, generally involves the hardware structure and the communication with the sensors, drivers etc. The MLC is used to combine any external devices or other physiological tools, such as EMG, Electroencephalography(EEG). The HLC manages the input from all these levels and decides the input to the LLC. In some case, there is another layer involved in the HLC, which is ideally an interface to receive the input orders or some considerations from the therapist [23, 81]. The purpose of the HLC is to perceive the locomotive intent of the user through a combination of activity mode detection and direct volitional control. The activity based detection can be performed by a combination of methods, which helps in making the approach more user friendly, such as rule-based [83–85] and automated pattern recognition [86–92].

2.3.3 Interaction-based

Force based interaction controllers in the form of either impedance or admittance controllers are efficient in achieving the active compliance [93]. Impedance based controllers are the typical example of such interaction based controllers, with the hybrid combination of the mechanical interaction and physical interaction [61]. The impedance will be increased when there is deviation in the trajectory which will apply resistive forces on the patient, to guide them towards trajectory. Such systems only based on mechanical interaction can be deemed as force field controller [94]. Impedance based control have been used extensively especially in legged robots because of this interactive property [56]. A cooperative control is efficient to control and encourage patients participation both physiologically and psychologically. The active and high intensity physical participation of patient improves the motor learning process during rehabilitation training [39]. Koenig et al. [95] have reported a closed loop control of patients' physical participation using heart rate or weight sum of interaction torque [29] between a patient and the robot as feedback. These interaction based control approaches are applied based on the detection of specific or pre-defined events. On occurrence of the specific event, the controller varies it's performance such as to compensate or assist the movement of the user. Such an event-based fuzzy control is prominent among the industrial and some rehabilitation devices.

2.3.4 Task-based

This model depends on combining the necessary control in function of the kind of task to be handled, task or challenge based control. The task based control provide the essential training and engage the patients in performing repeated and intense goal directed tasks [96].

These task-oriented practice are important for relearning the motor behaviours, recovery after stroke and reduced learned non-use behaviour. Task-oriented therapies facilitate the practice of a variety of simple and complex functional movements within a real context, with environment feedback to cue task success or failure. In terms of upper limb, virtual reality based controllers play a significant role for guiding the patient to perform a movement [97]. Virtual reality based models can provide therapies within a functional, purposeful and motivating context [98]. In lower limb, gait pattern based control models play a key role in this type of control models. Activity based therapies attempts to restore function via standardized therapeutic principles of exercise physiology, psychology and neuroscience [99].

2.4 Rehabilitation Robots

Robotics based rehabilitation therapy has become a topic of interest for many rehabilitation therapists and also for researchers working in the field of medical robotics. Robotic rehabilitation is not only addressed to patients training, but also to provide gait assistance to individuals with severe neurologic disorders, such as Spinal Cord Injury (SCI) or Stroke. This type of therapeutic assistance can also be extended to patients with muscle disorders or with other post-operative rehabilitation requirements. Robots are suitable for testing and application of motor learning principles to neurorehabilitation. Studies on the effectiveness of robotic neurorehabilitation have proven that robots are beneficial in measuring the patient's impairment level, but they have not demonstrated to be so effective regarding functional outcomes [100]. Robots have also been used extensively to study the process of motor learning in healthy subjects.

The implementations of the control strategies are dependent on the mechanical design of different devices. In order to understand the role of each rehabilitation robotic device, this section has been classified based on the classical Treadmill based and Wearable robots. Considering the advancement in the field of rehabilitation and the kind of tools being developed, it is interesting to classify further based on the type of actuators such as Series elastic actuator and Pneumatic muscle actuator. Further more, the combination of external sensors or devices based on the task or application must also be considered, to deliver an effective therapeutic tool. Hence this section is classified as follows, to adhere all the available rehabilitation devices:

2.4.1 Treadmill-based Robots

Treadmill based training devices are considered to be the pioneer in the robot based rehabilitation therapies. These treadmill based training devices have proved to be both assistive and resistive to the users with the combination of a suitable control model. The treadmill based training models have a body weight support system which manages the body weight and focuses mainly on the walking joint trajectories [101]. Hence this model is highly effective for complete paraplegic patients. Treadmill based or ambulatory robots usually guide the flexion/extension of the hip, knee (and possibly ankle) joints along their individual angular reference trajectories [102–104]. Several treadmill based training robots have proven to be efficient in providing the necessary gait assistance. The availability of an external body support system handles the complications in maintaining the equilibrium [20, 29, 35, 104, 105] for severely impaired stroke subjects and the use of body weight support (BWS) have demonstrated significant improvement in performing the gait and was helpful as a preliminary trial before overground locomotion in the older subjects [106–108].

2.4.1.1 Lokomat

Lokomat is a treadmill-based training device designed and fabricated by HOCOMA technology. Its a BWS training system with 4 DoF, with a lateral bar which supports the upper body weight and individual joint actuations for both hip and knee joints. The walking pattern of the exoskeletons are adjusted based on the interaction with the user and with reference trajectory. An automated gait-pattern algorithm was implemented by Jezernik et al. [109] , which evaluated the performance of the different control strategies: inverse dynamics, direct dynamics and impedance control based algorithms. The use of a patient co-operative control architecture which requires the active force contribution of the patient has shown a significant improvement in the therapy [29, 42]. Lokomat system has a haptic interface based intention model, which permits the user to decide the initiation of the

Figure 2.6: Illustration of treadmill training device- Lokomat; The BWS is variable depend on the user performance while the speed is always fixed with respect to the treadmill. Lokomat is considered more as a stationary device due to the BWS

movement or therapy. This control strategy is based on the patient's movement and helps in accomplishing walking movements at his/her own walking speed and desired gait pattern in order to maximize the recovery progress [110–112].

2.4.1.2 LOPES

Lower Proximity Exoskeleton (LOPES) combines a freely translatable and 2-D actuated pelvis segment with a 3-DoF leg exoskeleton; i.e.,2 in the Hip and 1 in Knee, as shown in fig 2.7. Initially an impedance based approach was applied which allowed a bidirectional mechanical interaction between the robot and patient [105, 113, 114]. A virtual model control (VMC) is employed, which uses simulation of virtual components and generates the desired joint torques [115]. The VMC model incorporates a spring damper system connected to the ankle joint with a fixed stiffness from which the required force is obtained [116]. This VMC based control provides the sufficient virtual support to the ankle joint ensuring an increase in foot clearance for stroke individuals [117]. Vallery et al. [28] demonstrated a

Figure 2.7: Illustration of LOPES - Lower Proximity Exoskeleton; The joint actuation in LOPES is adaptive but apparently huge in size because of the actuator model.

complementary limb motion estimation (CLME) approach to generate the reference motions of individual patient depending on their disability condition and on the movement of unimpaired contralateral limb. With CLME, the trajectory of the disabled leg is generated online based on the healthy leg autonomously resulting in a stable and more natural gait pattern than with a fixed reference trajectory [28, 118].

2.4.1.3 ALEX

Active Leg Exoskeleton (ALEX) is a treadmill based exoskeleton with a single hip and knee joints actuated [24]. The exoskeleton is a type of end-effector based device but the the ability to control more than one joints. The hip ab/adduction movement can also be modified by adjusting the mechanical joints in the transverse plane (fig 2.8). Alex II was developed as an update for its predecessor with some modifications in the manipulator terminal in order to accommodate and provide more effective therapy [119]. The powered hip joint adapts to the specific gait of the user and also demonstrated a reduction in muscle activations [120]. Banala et al. [121] developed a force field controller (FFC) which can make the linear actuators back-drivable. This backdrivability of the actuators allows the

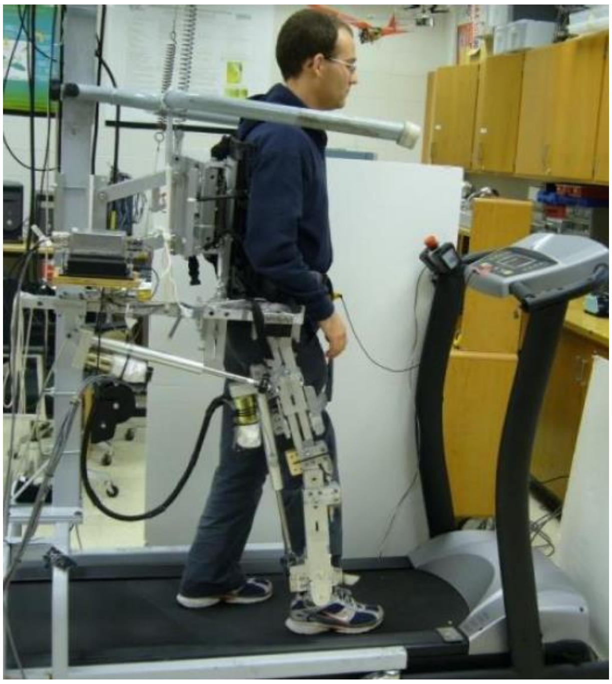

FIGURE 2.8: Illustration of ALEX- Active Lower leg Exoskeleton

exoskeleton respond to the human applied forces by offering less impedance. In other words, the controller applies suitable forces on the leg to assist the execution of a desired trajectory, for safe and effective gait training. The FFC is capable of providing close to zero impedance within a user-defined range of the desired gait pattern and offers high impedance outside the range.

2.4.1.4 End-effector based

End-effector type of robots usually guide the patients feet to trajectories in the para sagittal planes. End effector based devices such as Gait Trainer GT1, Haptic Walker [122] have enhanced the outcome of rehabilitation using stiff, velocity controller. The assistance level is achieved by adjusting the stiffness, which is again moderated by an iterative learning controller (ILC) [123–125]. Active AFO's are the common devices in this type of assistive devices. Some of them use Stewart platform, to ensure dynamic stability with the presence of 6 DoF and specific training. Meng et al. [78] combined a parallel robot with EMG induced system for motion recognition and also to define the adaptive impedance to be exerted.

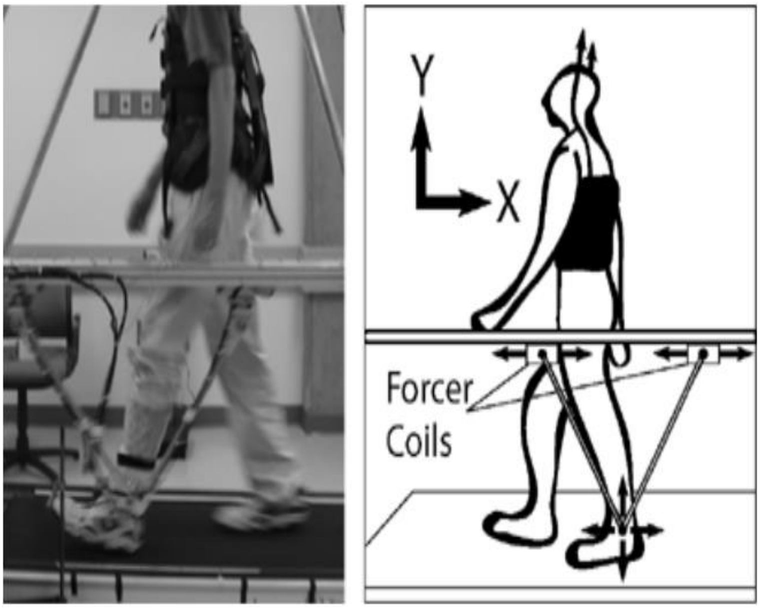

FIGURE 2.9: Illustration of ARTHuR- An end effector based lower limb rehabilitation device

ARTHuR is a 5 DoF end effector-treadmill based walking device (fig 2.9) with a non-linear control approach to precisely apply forces to the leg during stepping [126, 127]. "Teach and reply" option in the ARTHuR permitted the patients to walk on the treadmill with the robot attached. The reference trajectories from the patients were obtained due to the back-drivable property of the robot [128].

Open loop control with pre-defined trajectory was implemented on Gait Trainer due to some DoF linkage driving foot plates. The foot trajectory was defined by its hardware design with 60% stance phase and 40% swing phase in every gait cycle. The orientation of the foot plates also changed as the progression of the gait cycle to simulate the movement of level treadmill walking [129]. As an extended version of Gait Trainer [130], both Haptic Trainer [131] and G-EO [132] both have two programmable footplates. Each of them is actuated in all 3 DoF's along the sagittal plane; so they can be programmed to simulate walking on various terrains or even add perturbations to the training process.

2.4.1.5 Walk Trainer

Walk Trainer is a pelvic orthosis, emphasising the role of the pelvic movement in the gait function and also motivating an over-ground gait modelling. The walk trainer consists of a walking frame, leg orthosis and pelvic orthosis and an active suspension, as shown in

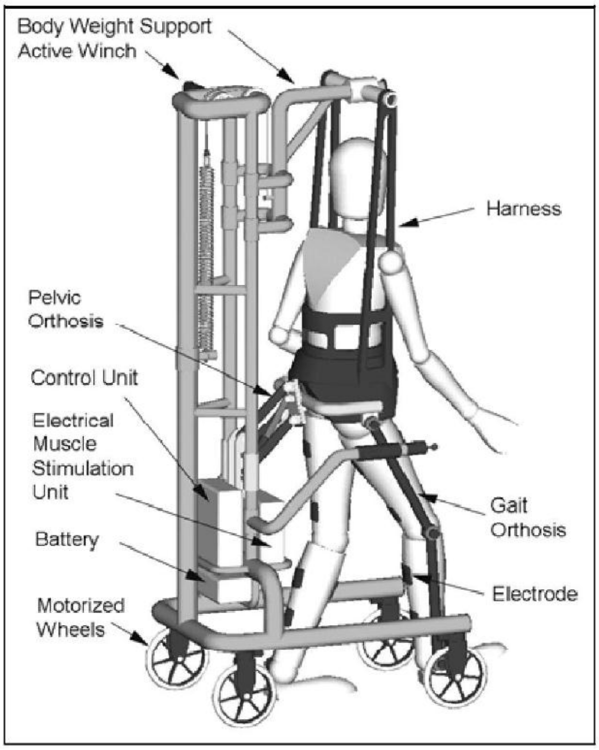

FIGURE 2.10: Illustration of WalkTrainer-A mobile platform with BWS gait training; Patient movement is supported by the BWS and the additional DoF by means of mobile platform. The inertial forces of the mobile platform must be controlled along with the exoskeleton.

fig 2.10. The device works in combination of the muscle stimulation models and robotic assistance [133]. The walking frame helps the patient to move around , like a mobile support, and the harness system supports the trunk of the patient, like a body weight support. The combination of the muscle stimulation helps in serving the severe atrophy condition in paraplegics. The pelvic orthosis permits the user with 6 DoF, allowing free movement along the 3 rotational and translational space. The Walk Trainer operates on a closed-loop electrical muscle stimulation in combination with hybrid force-position control in the pelvic orthosis, to guide the patient.

2.4.2 Wearable Robots

The origin of wearable robots can be tracked back to when General Electric was commissioned by the United States of America (USA) Army/Navy to develop the 'Hardiman' for allowing the wearer to carry a payload of 1500lbs (680 kg) but was never produced due to lack of technological limitations in 1965. The first ever Active exoskeleton was developed by Vukobratovic and Hristic in 1969 and is considered as predecessors of high performance humanoid robots [134, 135]. These devices served mainly to evaluate and develop electromechanical drives for active orthotic devices and to enhance strength or restore movement [136].

An analytical study by Dollar and Herr [137], explains the classification of exoskeletons based on how they are applied: performance based or active exoskeletons. Performance enhancing exoskeletons help in improving locomotion, reducing musculoskeletal forces and muscle fatigue and improving bipedal stability. To the contrary, Active orthosis are capable of augmenting and controlling the power at the joints according to the nature of disability. Many active exoskeletons have been developed for active gait restoration with considerable variations in actuation and sensing technologies and in control strategies. However, there are still some limitations to overcome in providing effective gait compensation [138–141].

Human-centered strategies, such as patient cooperative and support motor function assessment, are essential for ensuring user involvement and oriented to the development of robot behaviour [142, 143]. These strategies reinforce the concept of assist-as-needed by determining the level of robotic assistance provided to the user. In wearable robots, gait assistance is challenging in determining the level of assistance for ensuring dynamic stability, considering the ground reaction forces acting on them. Some of these exoskeletons have proven to be capable in providing assistance on a passive range of motion and using complex systems [71]. The most prominent and significant wearable robots, used for both performance enhancing and therapeutic purposes, are explained in this section.

2.4.2.1 BLEEX

Berkeley Lower Extremity Exoskeleton (BLEEX) (fig 2.11) is an energetically autonomous load carrying exoskeleton used as a performance enhancer to offer robustness to the changing backpack payload (18kg/40 lbs) dynamics. The control model is based on the sensitivity amplification controller, which estimates the force to be applied based on the accelerations and position values of the user. The controller works in combination with position controller, which is ideally used for stance phase, for ensuring complete stability [144]. Since the exoskeleton is intended to assist people to carry heavy payload, the change in the load carrier also affects the movement especially in the swing phase. Hence a hybrid control model was established with position control for stance phase and force based sensitivity

Figure 2.11: Illustration of BLEEX- Berkeley Lower Limb Exoskeleton, a performance enhancing exoskeleton

amplification controller for swing phase. The foot switches help in the transition of the controller, by detecting the different gait states [145] using a load carrying lower extremity exoskeleton has demonstrated A sensitivity amplification controller is used for the swing leg and hybrid control for the stance leg.

2.4.2.2 Vanderbilt

Vanderbilt is a 4 DoF wearable exoskeleton (fig 2.12) used ideally for overground rehabilitation along with an AFO. The exoskeleton has 4 actuated joints along the sagittal-plane and the AFO provides the support for the ankle joints [146]. The high level controller is established using a finite state machine (FSM), which conveys the status information to joint level controller, regarding the different phases in gait and also the transition between Sit-to-stand. The control architecture was developed based on a combination of gravity compensation, feed-forward movement and stability reinforcement.

The gravity compensation component of the controller removes the gravitational influence on the patient which also considers the weight compensation that occurs in the swing phase.

FIGURE 2.12: Illustration of Vanderbilt Exoskeleton; A light weight exoskeleton with hip and knee joints actuated and used in combination with AFO

A feed-forward method is applied during the swing phase followed by a stability reinforcement for the knee joint for stance phase [147]. These control approaches demonstrated an efficient behaviour by enhancing stability without a desired movement path to be followed. The joint controller is fed with a torque which is calculated based on the user movement. Another method of assistance by just applying torques in the hip and knee joint has been demonstrated to provide assistive behaviour for paraplegic individuals, but with a high gain and with the hip and knee trajectories being enforced by the orthosis [148].

2.4.2.3 HAL

Hybrid Assistive Limb (HAL) is a wearable robot developed by Cyberdyne as an assistive device for lower limbs [149]. The Hip and knee joints of the exoskeleton are actuated and the Force Sensitive Resistor (FSR) sensors help in the gait transition. HAL produces muscle contraction torques and control joint viscoelasticity by muscle effort such as co-contraction. The robot suit HAL produces torques corresponding to the muscle contraction based on the user's biological information. The exoskeleton suit also adjusts its viscoelasticity in proportion to the user's information which helps in reducing the user's muscle forces. Kawamoto et al. [150, 151] proposed a symmetry based control model for gait compensation for stroke patients. The control model provides spontaneous symmetric gait assistance based on the unaffected side. Such symmetry based adaptation considers assistance for the swing phase on the basis of the data compensated by the user [151, 152]. A similar kind of approach was also employed with paraplegic patients [153], but with body weight support mechanism and as a treadmill training. The support for the swing phase was provided using the muscle intentions estimated using the human-orthosis interaction.

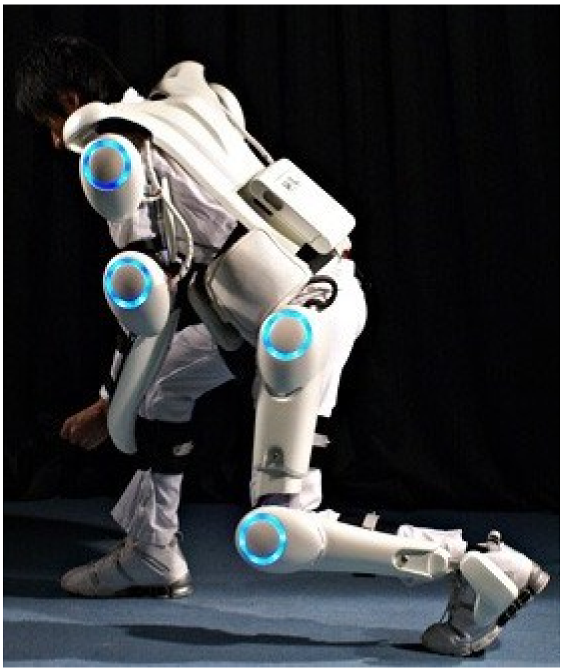

FIGURE 2.13: Illustration of Cyberdyne- HAL:Hybrid Assistive Limb; A lightweight suit initially developed for augmentation and supports the whole body motion of the wearer.

2.4.2.4 Ekso

Ekso, earlier deemed as eLEGS, is a lower extremity powered exoskeleton developed by the University of California, Berkeley and Berkeley Bionics. It has 4 DoF, actuated along in the sagittal plane, with minimal complexity and weight, as shown in fig 2.14. The Range of Motion (RoM) of all the joints are similar to that of a healthy individual. The ankle joints of exoskeleton are passively controlled by a spring mechanism, making it active only in the swing phase and also acts as an energy efficient tool. The exoskeleton is commanded by the user using an interface called Human-Machine Interface (HMI), by sensing the natural gestures of the user. The system uses a arm sensor built by using a gyroscope and accelerometer, attached either to the exoskeleton or crutches, to determine the user intentions and postures [154, 155].

FIGURE 2.14: Representation of Exoskeleton-Ekso

2.4.2.5 ReWalk

ReWalk is a powered exoskeletons developed by Marlborough, MA, USA and Yokneam, Israel Fig 2.15 . Its an assistive device developed for thoracic level motor complete spinal cord injury patients. The system is entirely self-contained and subject-directed, so users can control walking through trunk movements or via wrist pad controller [156, 157]. The tilt sensor helps in determining the trunk angle and this in turn generates the prescribed hip and knee displacement. The walking speed can be regularised in the range of 0.1 to 0.5 m/s, which also depends on the user performance.

2.4.3 Series Elastic Actuator

Active-ankle-foot-orthosis (AAFO) have been developed by many researchers as a preliminary study to analyse the performance of the gait stages in an assistive ankle movements. Most of these AAFO, are developed using compliant series elastic actuators (SEA) which

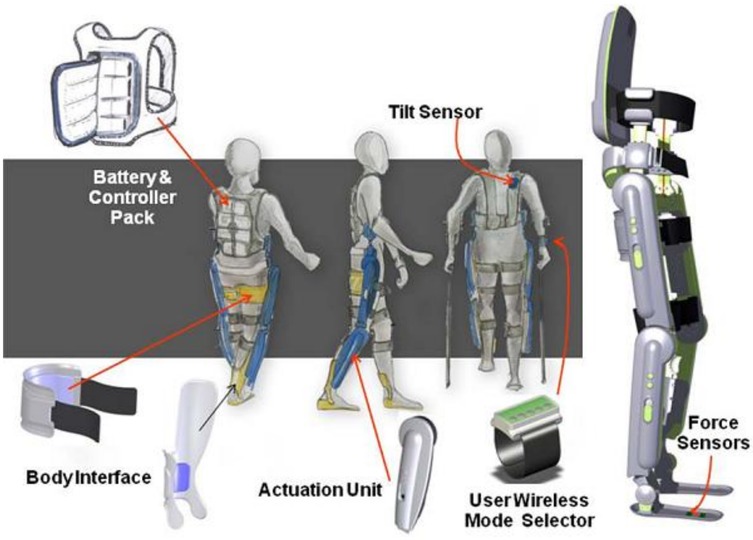

FIGURE 2.15: Representation of ReWalk Exoskeleton

provides the sufficient elasticity in giving the assistance needed [158–160]. These SEA based orthosis are efficient in providing variable impedance based assistance and compliance [51]. The spring mechanism used in these systems, applies the required amount of stiffness which results in providing the sufficient impedance and compliance, in terms of the gait phases or the interaction with the user. These type of actuators have also proven to reduce the energy cost in gait [161]. Blaya and Herr [158] demonstrated that these type of SEA based systems are also effective in reducing the occurrences of drop-foot gait and provides less kinematic difference during swing. RoboKnee [162] is 1 DoF exoskeleton with high level of transparency, detects the user intentions via the joint angles and ground reaction forces and applies torque such that the quadriceps muscles are relax, is intended to achieve low impedance. Roboknee, because of its SEA properties, permits the user to climb stairs and deep knee bends while carrying a heavy load.

GBO (Gravity Balancing Orthosis) is a low-cost, non-motorized orthosis intended for gait training of subjects with stroke and other neuro-motor disorders. The actuators of GBO provides gravity assistance on thigh and shank segments of the human leg. The exoskeleton has 3 DoF and is a treadmill based device with no body weight support, so there is no reduction in weight bearing on the hemiparetic leg [163]. Since, GBO is a passive exoskeleton, the treadmill settings are the main control parameters.

An actively controlled Hip ab/adduction based exoskeleton MINDWALKER [164], equipped with SEA has been developed mainly focussing on assisting SCI patients, Fig 2.16. The hip

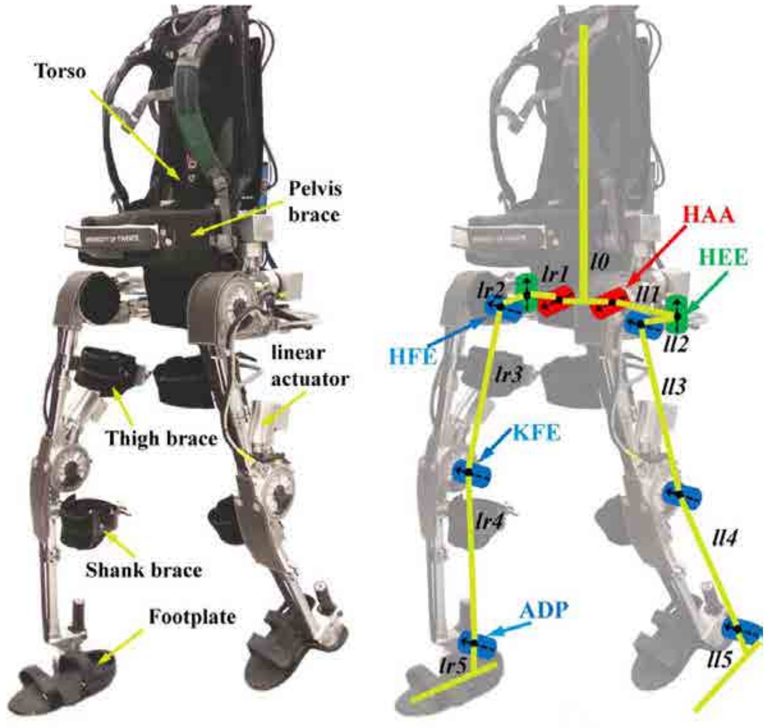

FIGURE 2.16: Illustration of Exoskeleton-MINDWALKER

and knee joints of the exoskeletons are actuated and the ankle joint is given freedom. MASA (Mechanically adjustable stiffness actuator) uses cantilever springs and torque limiter which helps in applying the effective stiffness variation to ensure a safer and more effective human-robot interaction [165]. The voluntary motions of the humans can be be realized by using a variable impedance characteristics in wearable robots [166]. A phase based controller is effective in diagnosing the motion changes and by incorporating a combination of algorithms it can allow a smooth and continuous transition of impedance properties along a trajectory. A compact rotary series elastic actuator (cRSEA) has been used by [19, 166] to demonstrate its efficiency in a gait training.

2.4.4 Pneumatic Muscle Actuators

Pneumatic muscle actuators (PMAs) have the capacity to replicate the function of a natural muscle which inherent's safety because of the soft and biomimetic actuation model. Costa and Caldwell [167] developed a wearable exoskeleton powered by pneumatic muscle actuators, with 10 DoF for assistive walking with more realistic trajectory patterns, producing a more natural muscle like contact. A joint torque control scheme has been employed by applying a stiffness function to the torque realises the amount of pressure to be applied in function of the current pressure in the actuators. This work also explains the developed complex biologically inspired structures, powered by soft actuators [168], which have similar characteristics to those of natural muscles, while still retaining benefits of mechanical structures. Another pneumatically powered ankle joint orthosis is developed to study human locomotion and which can be used for gait rehabilitation after neurological surgery [169, 170].

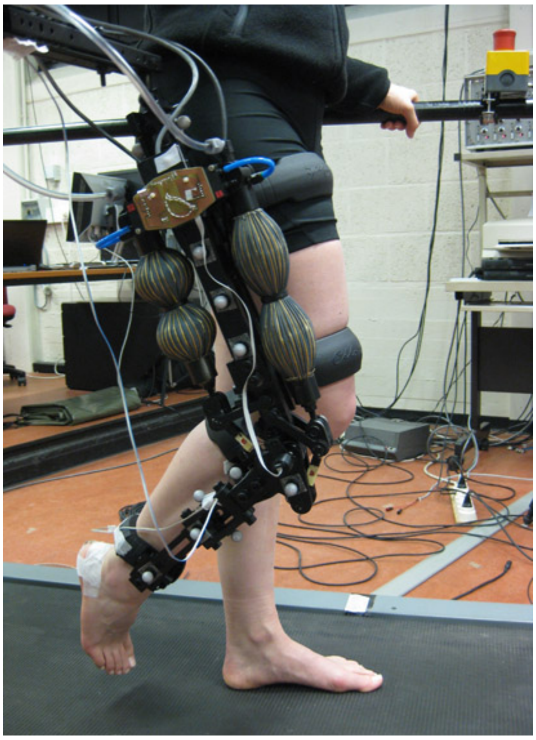

Figure 2.17: Illustration of KNEXO-Knee Exoskeleton

A powered knee exoskeleton KNEXO [171], actuated by two antagonistic PMA's have been used to demonstrate the human-robot interaction in terms of the muscle activity and kinematics. KNEXO (Fig 2.17) was used with a torque based controller and the positions were calculated based on the performance of the user. Another non-linear model based control was developed for the Human inspired Robotic Exoskeleton (HuREx [172]) for gait training. The model based control had a hybrid combination of PMA model and dynamic model of the device. [104] developed a gait orthosis, for one leg, using PMA's with a trajectory tracking controller based on augmented sliding control to guide the subjects on physiological gait trajectories. PAM is used in combination with POGO, resulting in the actuation of hip and knee sagittal rotations[35]. A hybrid approach using both harmonic drive and PMA's has been employed [173] and proved to provide high accuracy position control and compliant behaviour, for hip and knee joints. A variable stiffness adaptation was also evaluated in such a PMA system demonstrating an efficient response in terms adaptation [174]. The recorded movements were used as reference in the actuation mode, which was then used on exoskeletons PAM and POGO [31].

The patient cooperative strategies are also influential in the development of pneumatic based muscle actuators providing a soft and bio-mimetic actuation. This demonstrates another control method based on the impedance properties of the human limbs. The assistive forces from the exoskeleton are used to improve the kinematic response of the user's limbs, which aims to reduce the average muscle torques. This method is an alternative to myoelectric exoskeleton control based on estimating muscle torques from the electromyographic (EMG) activity, which involves modelling the user's musculoskeletal system and requires recalibration.

2.4.5 Hybrid Approach

Hybrid approaches are mainly intended to combine one or more processes to overcome their individual drawbacks and to ensure an effective therapy. The combination of external sensors to the wearable robots is much appreciated to provide an optimal assistance and to respond the changes in user intentions [175]. Reinkensmeyer and Boninger [140] describes the role of technology and combinational therapies in providing the suitable assistance to individuals with neurological disorders. The combination of multiple therapeutic modalities such as muscle stimulation (FES), Brain machine Interface (BMI) may help in enhancing the effect of training.

2.4.5.1 EMG-Electromygraphy

Most common hybrid approaches include using EMG signals to recognise the users muscle activity and to initiate a movement or to define the assistance as a function of the EMG signals. Lloyd and Bessier [176] explained the estimation of torques from EMG as a challenging task because it requires the characterization of several muscles, plus separating extraneous components affecting EMG signals. Ferris et al. [169] applied a low pass filter to EMG signals in order to maintain a proportional force applied to the artificial muscles (i.e., proportional myoelectric control) for an ankle foot orthosis (AFO). Such EMG based adaptive systems are more common in a prostheses model which increase the amputee's residual limb to control the ankle position in an AAFO [177].

A similar approach for upper limbs is used by Rosen et al. [178] in a powered exoskeleton to assist elbow motion, using EMG as the primary command signal. EMG based methods are used as a performance assessment tool of the therapy and patient [105, 179]. In HAL the user intentions are monitored by sensing the muscle synergies which initiates the movement and also varies the level of assistance to the user [180]. In Ekso, an arm sensor is used to detect the posture changes to initiate or modify the walking movement [155]. EMG based motion intention for each joints has been used as to tool to initiate and to compute the torques to be applied for performing the movement [79]. One of the widely used approaches for human centered strategies is to monitor the user muscle synergies to define the degree of assistance [181].

Assistance can be defined using the myoelectric signals to measure the muscle forces and to support voluntary motion of the patient, such as in HAL [150, 182]. These myoelectric signals are also used to determine the joint stiffness to be applied. EMG based learning signal was used to evaluate the necessary feed-forward torque for walking using LOPES [183]. This study helped in understanding influence of the muscle torques which help in developing an adaptive algorithm. This wide range of applications influence to understand the role of force based approaches in different robotic rehabilitation therapies. It is also necessary to understand the role of interaction forces acting between orthosis and patient [184]. These interaction forces are produced as a result of the excitation of myoelectric signals by the patient or by some assistive stimulation devices.

2.4.5.2 Brain Machine Interface

Some of the hybrid approaches are intended to identify the intention of the user to provide a sufficient AAN strategy. One of the widely used approaches for monitoring the human intention relies on the use of brain machine interfaces (BMI) such as in XoR [185]. The detection of the motor-neural activity related to the joint movement helps in providing the top-down approach in rehabilitation. This kind of approach is widely appreciated and demonstrated to be effective in restoring the neural activity [186]. Some of the hybrid approaches are also used as initiation commands or to perform assistance. The combination of muscle synergies and neural signals for evaluating the volitional commands of the patient has gained more importance. This type of volitional inputs motivates the user to initiate the therapy and subsequently improves the ability of intention [187, 188]. These systems are efficient in monitoring the user intentions mainly, because a real displacement of the joint position is not needed always to initiate the gait.

2.4.5.3 Wearable sensors

Wearable sensors, such as IMU, FSR are used for the automatic detection and initiation of the gait sequence and can also be combined with an exoskeleton to detect the user intentions to ensure complete assistance [67, 84, 175]. User intention based rehabilitation can be implemented by different approaches such as using an external input order (joystick) in LokoMat [110]. Gait initiation in MINDWALKER is based on the displacement of the CoM (centre of mass) which is calculated heuristically [164, 189]. Gait phase detection algorithm based on the feedback data from the sensors are used for walking with a robotic prosthesis [83]. The algorithms employs the shoe insoles and IMU sensors attached to the body to heuristically detect the transitions between gait. An instrumented cane, which reflects the motion of the upper limb, is used to initiate the gait movement using the HAL. The can is equipped with wearable IMU and embedded force sensors and acts as an feasible interface between human and the robot [190].

2.4.5.4 Virtual Reality

Virtual reality, as another bio-cooperative control technique, has proven to be useful in terms of motivating and challenging patients for longer training duration and cadence [98, 191, 192], modifying patients' participating level, updating subjects with their training performance and generalising training result to real life scenario. Real-time estimation of cognitive load during Lokomat robot training was developed, as well as a virtual task with varying cognitive difficulty levels. For less impaired and cognitively capable patients, a virtual environment was used on indicating the desired level of participation. The patients were asked to actively conduct training to match the desired level of participation. In case of more impaired patients or patients who cannot understand the virtual environment display, treadmill speed was controlled to match the feedback heart rate to the desired value [95]. The closed loop control was achieved by modifying the task according the estimation to keep the subject reasonably challenged [193, 194].

2.4.5.5 Functional Electrical Stimulation

The combined action of Function Electrical Stimulation(FES) to the exoskeleton devices helps in increasing the muscle power which in turn reduces the energy demand of the exoskeletons and thereby resulting in a lighter system with reduced power in actuators [27, 195]. Moreover, such hybrid exoskeletons should promote more effective neural plasticity than other standard practices like treadmill training, because of the intensive, community-based gait practice involved. This gait practice occurs during daily training, and thus, increased user participation is promoted during walking training. While FES-induced gait has several benefits [196], mainly related to muscle strength and cardio respiratory fitness, it is not so effective in gait restoration and is limited to a therapeutic environment. Furthermore, FES can induce muscle fatigue, leading to interruptions in training. Improved management of muscle stimulation is therefore crucial to the development of successful hybrid exoskeletons that can be used for longer period of time.

Senanayake and Senanayake [197] demonstrated the application of FES as an assistive tool for restoring muscle activity in SCI patients. Several novel closed loop FES-controller methods have been developed which guarantee closed loop stability [198, 199]. The tracking performance of following a trajectory demonstrates that FES based therapies in the case of SCI subject's are useful. The analysis of Jezernik et al. [109] explains the accurate control of FES generated movements in implantable or external neural prostheses. FES based therapies for swing phase estimation proposed by Cikajilo and Bajd [200], is used for gait re-education for incomplete SCI patients. A hybrid approach in the field of neurorehabilitation is the combined effect of FES technology and exoskeletons [201, 202]. These hybrid approaches are used for both gait compensation and rehabilitation, by the combined actuation of muscle generation and reduction in actuator power[187]. Many passive or reciprocating orthoses have improved their performance by combining with FES neverthless there is little improvement in terms of energy cost and gait velocity [201].

2.5 Exoskeleton H1

This thesis dissertation is performed as part of the project HYPER, explained in section 1. As part of the project, a wearable exoskeleton H1, was developed and built in the Neural rehabilitation group, Consejo Superior de Investigaciones Cientificas (CSIC), Madrid. The

FIGURE 2.18: HYPER Exoskeleton - H1,Technaid. S.L

exoskeleton was mainly focussed as a rehabilitation tool for adult people in the range of 1.60m to 1.90m tall and with a maximum body weight of 100kg, with abnormalities in gait due to stroke or Spinal cord injury.

The exoskeleton is a wearable device weighing about 9kg, built with aluminium and stainless steel for mechanical resistance and reduced weight. The exoskeleton (Fig2.18), with 6 DoF's (degrees of freedom) consists of 6 joints: hip, knee and ankle for each leg. The maximum achievable joint torque and angles in each joint is listed in the table. The mechanical joint permits both active and passive movements along the sagittal plane. The joint lengths can be adjusted by a two telescopic bars mechanism that are fixed in different positions by screws. The foot plates are compatible with different types of shoes, hence the patients do no need to change their footwear.

Each joint is driven by a harmonic drive brushless DC motors, because of their higher efficiency, small volume solution, more torque density, reliability, reduced noise and reduction of electromagnetic interference. The exoskeleton is equipped with an encoder to measure the joint angles and the joint links contains a strain gauge to measure the human-orthosis interaction torques. Strain gauges are used as force sensors, to measure the torque produced by the interaction between the subject's limb and the exoskeleton. The footplates are equipped with two force sensing resistors (FSR), each under the heel and toe, that detects the foot contact between the foot and the ground. These resistors are mainly used in the detection of the different gait phases [203, 204].

The control architecture is developed on a *Matlab* kernel inside PC104 computer, with real time capabilities. All control algorithms are implemented using the *xPC*target environment of *Simulink*. The control model of the Exoskeleton is categorised in two distinct levels; Low Level Control (LLC) and High Level Control (HLC). The Low level control involves communication with the sensors and to the basic control modes: position or trajectory control, torque control and stiffness control. The High level control model is were the algorithms and combination of various assistive devices is taken into account. As part of the consortium, the discussions about the essential low level control algorithms were performed by Institute for Bioengineering of Catalonia (IBEC) and CSIC groups. The implementation of the control model was performed by the group in CSIC [205].

The communication channel among the different sensor channels and the controller is established using a CAN (Controller area network) bus communication. Each board containing all the circuitry for the analog filters of the sensors,strain gauges send information on joint position and interaction torques respectively, through CAN bus at a frequency of 1kHZ. The control hardware involves a CAN board and two I/O boards with digital and analog capabilities, communicating with PC104 via PCI bus. The CAN bus has two ports, one connected to the data bus receiving information from the sensor and the other port connected to the High level controller(HLC). All these control hardware, weighing around 2kgs, are mounted in a backpack and can be carried depending on the capacity of the user. For most of the clinical trials performed, this control hardware had been carried away separately to make the patient more comfortable.

This dissertation is intended to develop an efficient algorithm which can work in the above mentioned set-up with minimum delay and maximised performance and reliability. All the

TABLE 2.1: Mechanical limits of the exoskeleton H1

Joints	Joint angles(deg)	Motor Torques(Nm)
Hip	-20 and 100	± 40
Knee	-5 and 100	± 40
Ankle	-15 to +20	± 20

algorithms proposed in this dissertation has been applied in the High level controller with a suitable selection or combination of control modes in the lower level. The stiffness control for each joint is realised in the scale of $1 - 100Nm/deg$

2.6 Conclusions

Efficient control strategy needs a sufficient hardware system which can be viable to provide the necessary assistance and resistance when needed. Treadmill based training devices are efficient in assisting the patients but the impact of slacking reduces the significant effect in terms of therapeutic advancement. Such treadmill based devices can show improvement in case of complete paraplegics and acute stroke patients. Wearable robots need to maintain dynamic stability and the role of different control strategies is to ensure equilibrium along with safe interaction. The hybrid combination of systems, such as EMG, EEG or wearable sensors, ensures that the user initiate the movement, which psychologically motivates the user and ensures involvement in the therapy. These user intention based systems are also practically feasible and viable to adhere the paradigm of 'Assist-as-Needed'

The control model should be hierarchical, to expand its performance by including more devices, if needed. On the other hand, task or challenge based systems are efficient in terms of focussing on one particular movement in a therapy. Model based control are similar in context but it is complicated to have a mathematical control model of the human-exoskeleton. In next chapters, the role of these individual control models will be evaluated to choose the efficient one. Initially simulation studies were performed, Chapter 3, to analyse the effects of human interaction or any assistive devices in the control model of the exoskeleton. These effects, assistive or resistive, can influence the exoskeleton's movement which must be considered while modelling an AAN strategy. The input forces, affecting the movement, is considered as a disturbance for the controller. The role of these disturbances is studied in brief which helps in choosing the optimal control model.

From the detailed analysis in this chapter, it is also understood that Task-based and Interaction-based approaches help in solving the global objective of AAN, by breaking down them to small focus points. The combination of the approaches makes a system susceptible to the changing environment and also in combining with multiple devices. In Chapter 4, different tasks involved in a lower limb rehabilitation therapy such as Sit-to-stand, Balance training and Gait training, are handled individually with an adaptive control model. The adaptive model considers the human-orthosis interaction to analyse the user performance and/or to react to the user needs or to the inputs from the therapist.

Chapter 3

Simulation Analysis

[*] This chapter presents the control strategy for an assistive exoskeleton, to reduce the effects of disturbances on planned trajectories during rehabilitation therapies. Disturbances are mostly caused by muscle synergies or by unpredictable actions produced by functional electrical stimulation. The effect of these disturbances can be either assistive or resistive forces depending on the patient's movement, which increase or decrease the speed of the affected joints by forcing the control unit to act consequently. In some therapies, like gait assistance, it is also essential to maintain synchronization between joint movements, to ensure a dynamic stability. A force control approach is used for all the joints individually, while two control methods are defined to act when disturbances are detected: Cartesian position control (Cartesian level) and Variable execution speed (joint level). The trajectory to be followed by the patient is previously recorded using the active exoskeleton, H1, worn by healthy subjects. A realistic simulation model of the exoskeleton is used for testing the effect of disturbances on the particular joints and on the planned trajectory and for evaluating the performance of the two proposed control methods. The performances of the presented methods are evaluated by comparing the resulting trajectories with respect to those planned. The evaluation of the most suitable method is performed considering the following factors: movement stability, minimum time delay and synchronization of the joints.

[*] This chapter has been written based on the following publications:
"Recovering planned trajectories in robotic rehabilitation therapies under the effect of disturbances",V. Rajasekaran, J. Aranda and A. Casals, International Journal on System Dynamics Applications (IJSDA),2014.
"Handling disturbances on planned trajectories in robotics rehabilitation therapies", V. Rajasekaran, J. Aranda and A. Casals, XIII Mediterranean Conference on Medical and Biological Engineering and Computing (MEDICON) 2013, IFMBE Proceedings

3.1 Introduction

For a suitable AAN strategy, the knowledge about the needs of the user and the movements of the orthosis are needed. The control model should consider both the input parameters of the human and orthosis, to decide the assistance or resistance to be applied. In rehabilitation studies, a force based control approach is influential in providing the required assistance as needed. In order to evaluate the performance of the control model and the parameters, a simulation analysis of the exoskeleton must be performed priori to the real implementation.

This chapter focusses on analysing and reducing the influence of disturbances in therapies, by providing the required assistive and resistive forces to the affected joints. When one of the joint trajectories is affected by a perturbation, it is important to analyse the way to prevent the disturbance from affecting other joints. The response of the exoskeleton to recover the trajectory must be as fast as possible in order to maintain the trajectory tracking error as low as possible. Another important factor is to maintain the synchronized movement in all the joints while following the planned trajectory. Especially in gait assistance, it is critical to maintain both, synchronization of all the joints, such that body balance remains stable. As can be extracted from the literature, it is noticeable that the role of force control is highly essential to perform an effective therapy. The role of interaction forces must also be considered to provide assistance when needed.

3.2 Exoskeleton model

Simulation tools are commonly used in many human-centered studies, to analyse the performance of the system in presence of multiple effects. By means of these tools, the behaviour of the system can be predicted and the kind of action can also be studied. In case of rehabilitation, simulation tools are used as a priori to real implementation to avoid the causing any damage to the human inside the loop. Further more, it also helps in designing the necessary performance by an exoskeleton which can contribute in modelling the system. For this dissertation, the exoskeleton model was used to evaluate the control performance, before pursuing it with real subjects.

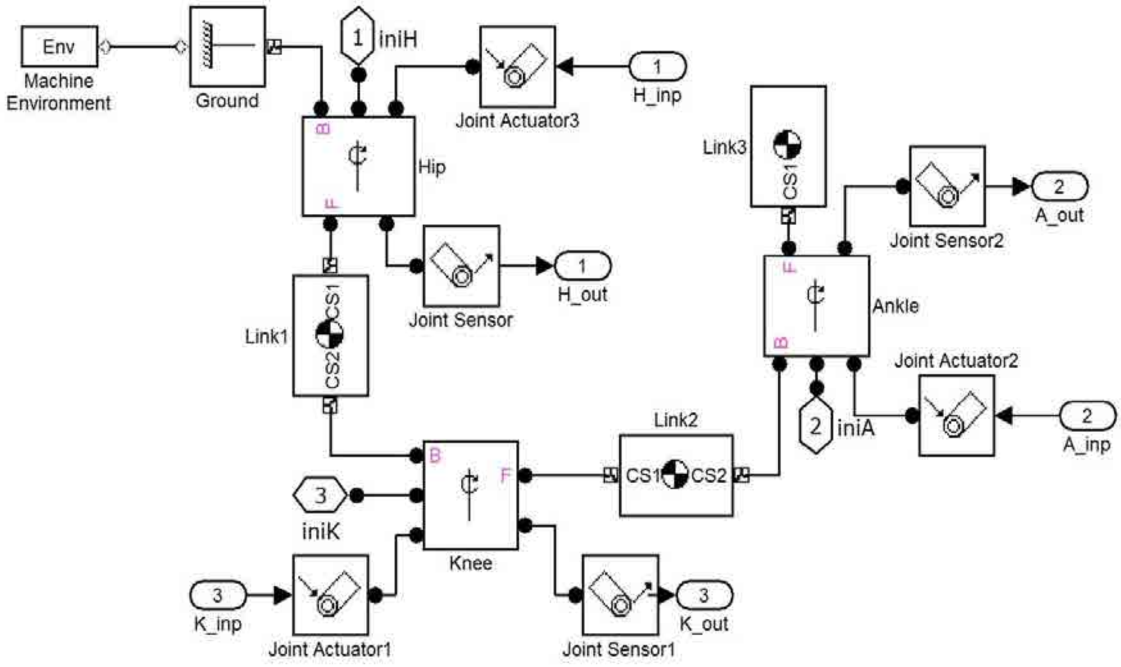

FIGURE 3.1: Simulink model of the exoskeleton with 3 DoF

A simulation model of the exoskeleton was built based on the real parameters of the exoskeleton H1 presented in the section 2.5, using simulink . Each joint is driven by a Proportional-Integrative-Derivative (PID) control which are again modified to realise the various control model parameters. This model is used to verify the suitability of the proposed methods for handling the disturbances efficiently. The simulink model implementation, as shown in fig 3.1, consists of 3 revolute actuators with 1 DoF for each joint. The actuators are driven by an indirect force control approach.

The force control approach in this orthotic system is realized by applying force to each joint actuator, proportional to position error, acting as the proportional component (P) of the controller. The applied forces of the joints are tuned based on the evolution of the error over time. The figure shows the simple simulation model used for testing this force control strategy. The exoskeleton model has three revolute joints which are actuated by force and their positions are evaluated at each time instance. Each joint is controlled and driven by an individual PID source.The learned joint movements from the real exoskeleton are used as input parameters for this analysis. The results indicate that the joints are able to follow the force orders based on the reference signal and the current position response of the system.

A joint trajectory pattern obtained from the exoskeleton is used as the reference pattern in this analysis. These joint trajectories are captured from healthy individuals wearing the exoskeleton. The recorded joint trajectories correspond to a cycling movement, obtained as an example for evaluation. An extracted sequence of five cycles is used as a reference, after performing an optimization in order to define a suitable planned trajectory.

3.3 Handling disturbances

A force control approach is applied to correct the action of an exoskeleton on the joint affected by a disturbance or any unpredictable force. The undisturbed joints, which are not directly affected by the disturbance, are constrained due to their kinematic relationship. Hence it is necessary to help them recover from these indirect effects of disturbances. As explained in the previous Chapter2, force based control maintains the interaction forces between the patient and the robot. Force based patient cooperative control has proven to help in guiding the patient, both in assistance or resistance as needed [42]. Another prominent approach is the impedance based control, which can be defined as a combination of force and position control, acting on the joints, based on the requirement of the therapy [68].

In gait assistance for SCI and stroke individuals, some assistive tools are required in order to aid them in maintaining the equilibrium. FES is one of the most widely used assistive tools for gait assistance or for other therapeutic procedures. However, one of the major drawbacks of using prolonged artificial muscular stimulation is muscle fatigue [206]. FES induced assistance may also cause disturbances like the generation of muscle synergies and residual forces on the patient during the execution of a pre-planned trajectory. An occurrence of these types of disturbances or unpredicted forces becomes a common scenario in most neuro-rehabilitation therapies. These unpredicted forces can be impulsive signals, those caused by FES induced motion in the therapy and sinusoidal or continuous signals produced by muscle synergies. Nef et al. [207] developed a patient-robot cooperative therapy, which aims to improve rehabilitation and therapeutic progress of stroke and SCI patients by encouraging more intense training and increasing the patient's motivation and activity using an upper limb rehabilitation robot ARMin. This work demonstrates a patient-robot cooperative control for a wearable lower limb exoskeleton in the presence of disturbances. A reactive or proactive behaviour of a system can be demonstrated by implementing an efficient control strategy.

Two methods have been identified to handle the effect of these disturbances in a pre-planned trajectory and to study the way to minimize their influence in patients therapies. These methods are chosen based on providing an essential therapy by using a force control and ensuring synchronization of the joints. The two methods, through which the effects of the unpredictable forces can be tested, are: Cartesian position control and Variable execution speed method.

The Cartesian control approach acts similar to force control by acting proportionally to the joint position error in Cartesian space. Variable execution speed approach acts as an impedance based control, considering stiffness and damping in the joints. These two methods handle unpredictable forces or disturbances, which appear along the execution of a learned trajectory pattern. Disturbances can be generated by different sources, but the focus is on the effect of patient's muscle synergies [169, 178], in presence of muscle spasms or spasticity, and on the unpredictable effects of FES. Muscle stimulation techniques aim to assist the patient in achieving the goal [197] but, at some points, these stimulations can cause too short or large forces which affect the pattern to be followed.

Additionally, the prolonged stimulation of muscles leads to muscle fatigue, which is difficult to model and affects the output pattern to be followed [202]. In gait assistance it is important to follow the gait pattern with time constraints and synchronized joint movements. An effect in one joint obviously affects the trajectory of the other joints, due to the dynamics and the correlation between them. Planning and re-planning a trajectory based on the effect of disturbances, plays an important role in rehabilitation studies and especially in gait assistance with time constraints. Time constraints are necessary in order to handle the equilibrium of the patient and at the same time they help in gait assistance.

3.4 Control Approach

Force control approach plays an essential role in order to achieve a robust and versatile behaviour of a robotic system especially while performing dependable operations in the presence of humans [41]. A fundamental requirement for the success of a robotic rehabilitation task is the capability to handle the physical contact forces between robot and patient. The use of a force based control framework for robot assisted rehabilitation has been proposed by [208] for an upper extremity exoskeleton. In this work, the force orders are evaluated based on the error in joints. When using the Variable input rate method, the force to be applied to each actuator is directly proportional to the error in position of each joint. In the case of using Cartesian position control, the applied force is directly proportional to the error in the Cartesian space. The applied forces of the joints are calibrated based on the evolution of the error along time.

In order to evaluate the response of the two methods presented, a sinusoidal signal is introduced as a disturbance to one of the joints. This sinusoidal signal represents muscle synergies within the range of -10N to +10N [209]. A change in behaviour is observed as a result of the dynamic evolution of the disturbances applied to other joints.

When a disturbance is observed in one of the joints, the joint position orders are processed, for each method, for recovering the trajectory. The disturbance signal is applied at the same frequency, in both cases, in order to evaluate their efficiency. The same experimental set-up is also used in the presence of an impulsive signal, acting as a disturbance in one joint. The effect of FES stimulation acting on a muscle can be considered as an impulsive signal [210]. The disturbance is initiated in a joint at a particular time instant t and the effect of this signal is also observed in the other joints. The impulsive disturbance applied is within the range of +50 N to -30 N [202]. The force applied is initiated at the time instant 't', such that the positive force applied refers to electrical stimulation of the muscles. Then it progressively decreases, up to negative values, emulating the muscle fatigue caused by prolonged stimulation.

3.4.1 Cartesian model

Cartesian trajectory control is one of the classical approaches that help in recovering the global trajectory, by monitoring the influence of any joint disturbance to the global frame. Cartesian control has been used in upper limb rehabilitation and it has demonstrated its efficient tracking performance [211]. The major drawback of this method is the need of having available the patient kinematic model. In this approach, the joint control determines the characteristics of the Cartesian control by using the transpose of the Jacobian matrix.

In the present work, the Cartesian position control acts as a force control approach by the transformation of the position errors in Cartesian space into forces. This forces are converted to convenient joint torques by using the transpose of the Jacobian matrix of the exoskeleton kinematics. This Cartesian position control is added in parallel to the exoskeleton's individual joint control consisting of a classical PID, as PID is unable to handle the coupled effect of disturbances. Figure 3.2 shows the schematic representation of the Cartesian position control approach. When a disturbance appears in any joint trajectory, the Jacobian matrix computes the new positions to be achieved by all the joints, preventing the direct effect of disturbances. In this approach, the errors in joint positions are evaluated

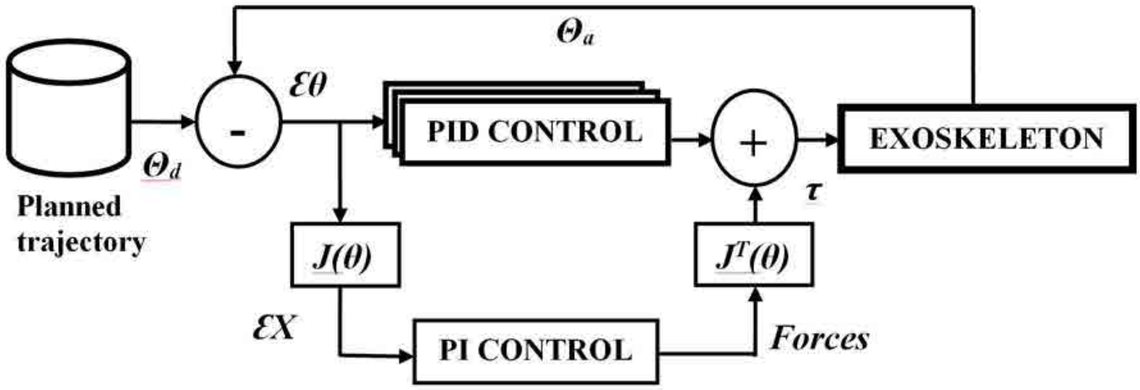

FIGURE 3.2: Cartesian-based control model for handling the disturbances

by continuously monitoring the actual position against those of reference. These errors are then transformed to the Cartesian space by using the direct kinematic transformations, as shown in equation

$$\epsilon(X) = J(\theta) * \epsilon(\theta) \tag{3.1}$$

where $\epsilon(X)$ is the error in Cartesian coordinates,$\epsilon(\theta)$ the error in the Joint space and $J(\theta)$ is the Jacobian matrix of the joint coordinates. In the presence of disturbances, the Jacobian matrix is used again to compute the joint torques by monitoring the deviation of the trajectory in the Cartesian space. At the end, the computed torque is added to the joint torques in order to avoid the effects of disturbances on the other joints and therefore, follow the preplanned trajectory. The input joint torques to be applied to recover the desired trajectory is calculated by the equation,

$$\tau = J^T(\theta) * (K_1\epsilon(X) + K_2 \int \epsilon(X)) \tag{3.2}$$

where, τ is the joint torque, J^T the transposed Jacobian matrix of the joint coordinates, $\epsilon(X)$ the Error in Cartesian coordinates and K_1, K_2 are constants.

The initial approach was to generate forces in the joint space by considering them proportional to the error in the Cartesian space, producing a spring effect that forces the mechanism to follow the programmed trajectory. However, an integral function of the error signal is included in order to avoid the steady state error caused in the output pattern when this strategy is applied.

3.4.2 Variable input model

The Variable execution speed method operates exclusively in the Joint space. This method acts as an impedance based approach, a combination of stiffness and damping acting on the joints. A joint affected by a disturbance is handled by applying a stiffness control, whereas for the other joints the rate of their position inputs to the control unit is varied. The occurrence of a disturbance in a joint changes the expected trajectory and affects the other joints due to the dynamics of the exoskeleton. The damping effect of the impedance approach is realized by increasing or decreasing the frequency of input orders which results in accelerating and decelerating the joint movements. This damping effect helps in maintaining the correlation between the joint movements while the stiffness applied on disturbed joint helps in recovering the trajectory. By using this approach, synchronization of joints is

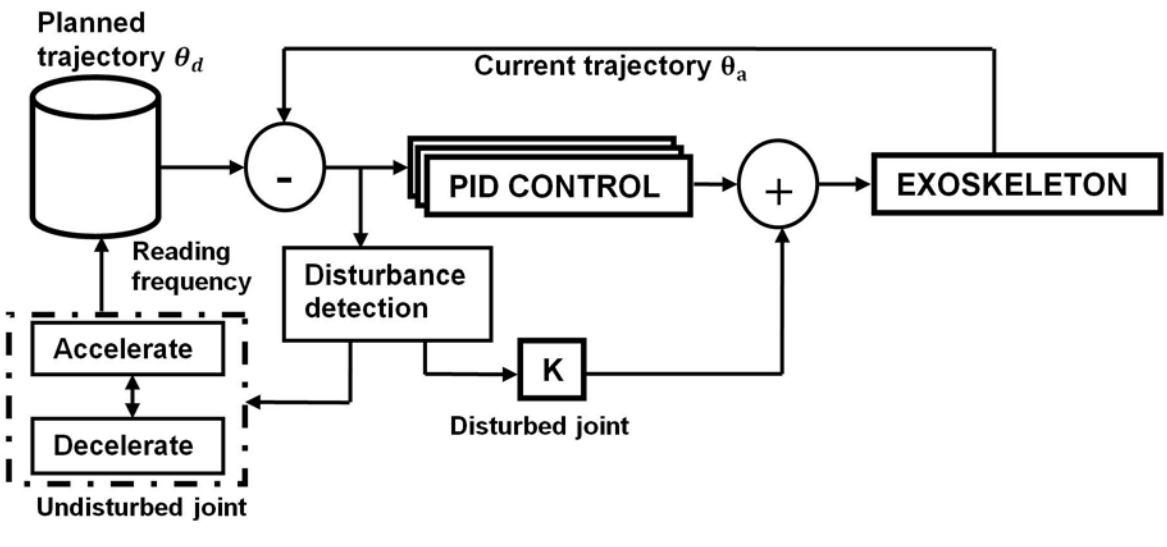

FIGURE 3.3: Variable-input rate approach for recovering trajectories, from the influence of disturbances

maintained while trying to handle the disturbed joint. When a disturbance is detected, a constant stiffness is applied to the disturbed joint while adjusting the execution speed of the other joints, by varying the rate of their programmed input position orders. A disturbance is considered when the difference between actual and reference trajectory is higher than a given threshold. A high stiffness is applied only to the disturbed joint in order to avoid

Table 3.1: Relation between the direction of the movement, interaction forces and disturbances; the resulting action on undisturbed joint and effect on the patient

Movement	Disturbance	Interaction Forces	Action	Effect
Positive	Positive	Resistive	Accelerate inputs	Assistance
Positive	Negative	Assistive	Decelerate inputs	Resistance
Negative	Positive	Assistive	Decelerate inputs	Resistance
Negative	Negative	Resistive	Accelerate inputs	Assistance

the patient being completely assisted to perform the movement, thus a complete assisted therapy would provoke slacking in the patient development. But there is also a need to inhibit these residual forces from affecting the dynamic stability in gait assistance and also the synchronization of the joint movements in following a trajectory.

In this approach, the torque is added on the disturbed joint is directly proportional to the error in position as given by the equation,

$$\tau = K.(E_\theta) \tag{3.3}$$

where, τ is the applied torque in the joint, K is the stiffness constant and E_θ the error in joint position.

For all the other joints, the joint input orders rate to the controller is varied in order to change the execution speed and maintaining the synchronization (Figure 3.3). The input rate is increased or decreased based on the coincidence in direction of a disturbance and a joint movement. The direction of the disturbance is known by sensing the interaction forces between the exoskeleton and the patient. The interaction forces are defined as assistive or resistive depending on the direction of the movement.

Table 3.1 defines the four cases which are considered depending on the direction of the joint movement and the disturbance acting on it. When a disturbance occurs in the same direction of the movement, then a resistive behaviour is applied in order to avoid the uncontrolled movement of the exoskeleton. This resistive behaviour of the system is assumed as a key to accelerate the inputs of the undisturbed joints. This applied resistive force provides the assistance in overcoming the effects of the disturbance on the patient. Similarly, an assistive force is developed when the disturbance acts in the opposite direction of the movement. A force is defined as assistive when an unexpected disturbance is acting against the direction of a joint movement, reducing its execution speed. In order to synchronize the joints movements, the input position orders rate of the undisturbed joints is decreased to reduce the speed of these joints and accordingly maintaining the coordination between them.

The resistive force is characterized by the disturbance being on the same direction of a joint movement, increasing its execution speed. The joint movements are synchronized by increasing or accelerating the input rate to the undisturbed joints, while maintaining the synchronization. The disturbed joint affects the movement of the other joints due to the dynamics between them and also the global trajectory synchronization. By varying the input order rate to other joints, synchronization between these joint movements can be achieved. Increasing the input rate results in the assistance of the robot to the patient and delaying the input rate results in resistance. This results in guiding the patient to follow the recorded trajectory even in the presence of disturbances.

3.5 Results

The performance of each method in the presence of different disturbances or perturbations is presented in this section. The outcome of each method is compared with the planned trajectory pertaining to a cycling movement, to be followed by the patient. The performances of the two methods are evaluated by monitoring the position errors with respect to the planned trajectory. The position errors are calculated between the planned and current trajectories. Firstly, the exoskeleton model is fed with sinusoidal or continuous disturbances and then followed by impulsive disturbances. As explained earlier, the synchronized joint movements and time constraints play an important role in this evaluation. Figures 3.4 and 3.5 shows the results of both methods in presence of sinusoidal and impulsive disturbances respectively along with the dynamic response of each joint trajectory. From both figures, it is noticeable that there is a significant difference between the planned and current trajectories of the hip joint. This difference is due to the application of disturbances in the hip joint of the exoskeleton model. The sinusoidal disturbance acts continuously in the hip joint, both in positive and negative direction of the movement $\pm 10N$, of the planned trajectory. Similarly, an impulsive disturbance is applied to the hip joint (+50N) at the beginning of the trajectory and the progressive decrement of the disturbance, up to -30N, causes a negative effect in the planned movement. The planned trajectory is followed more

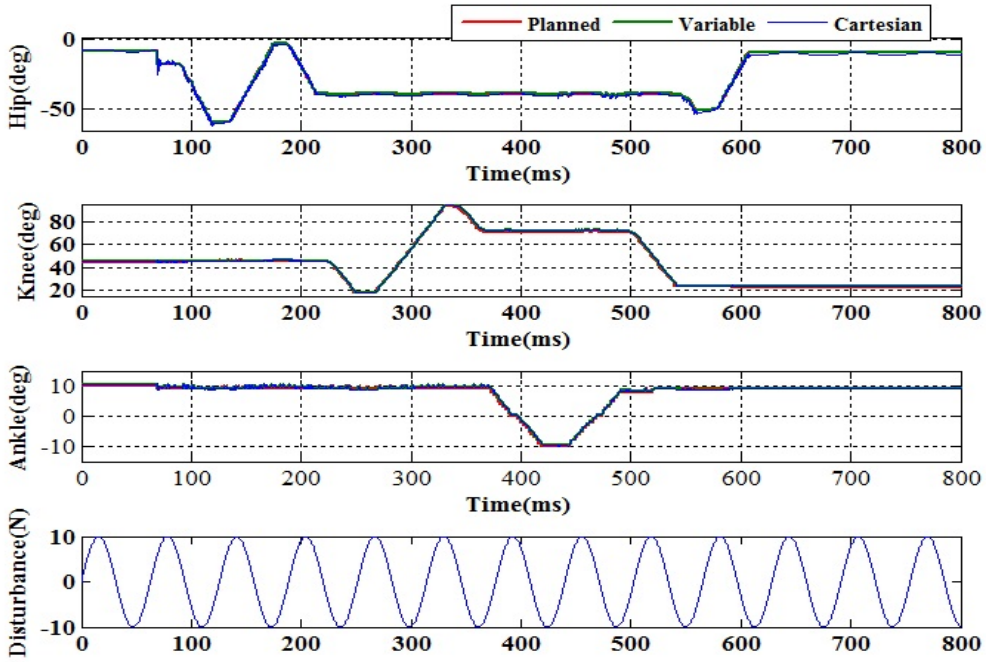

FIGURE 3.4: Comparison between planned trajectory, variable execution speed and cartesian control of hip, knee and ankle joint trajectories in the presence of sinusoidal disturbance

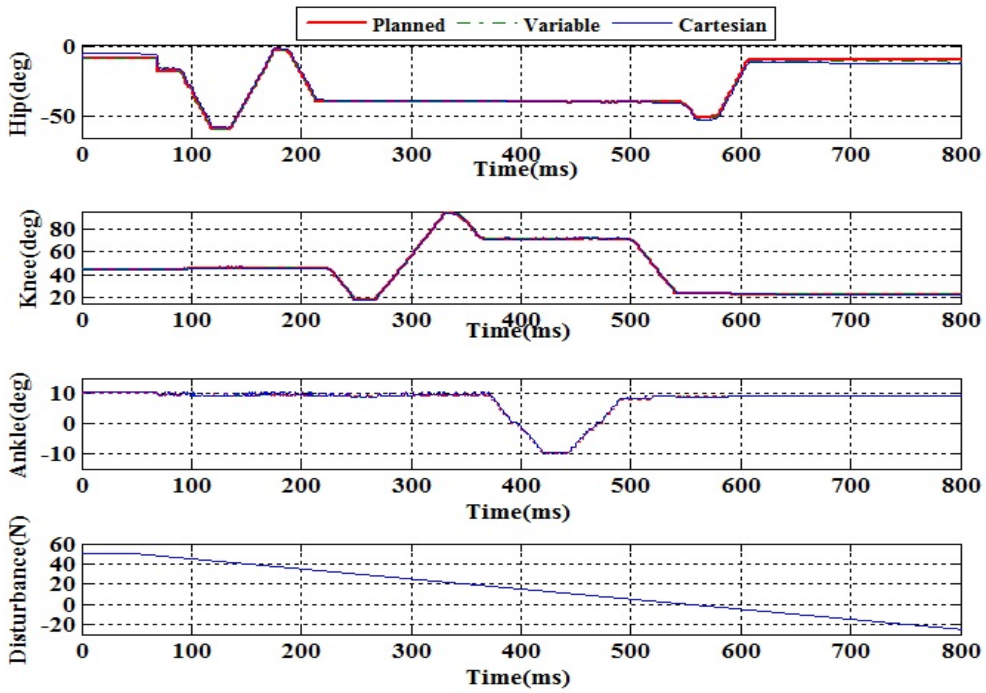

FIGURE 3.5: Comparison between planned trajectory, variable execution speed and cartesian control of hip, knee and ankle joint trajectories in the presence of impulsive disturbance

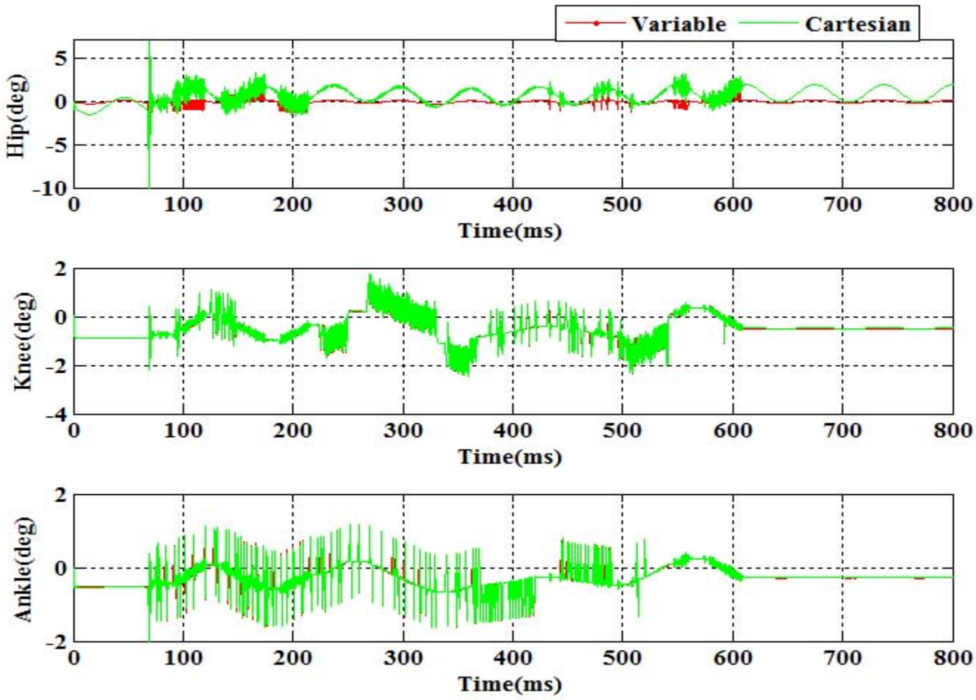

FIGURE 3.6: Trajectory error of hip, knee and ankle joints in the presence of sinusoidal disturbance

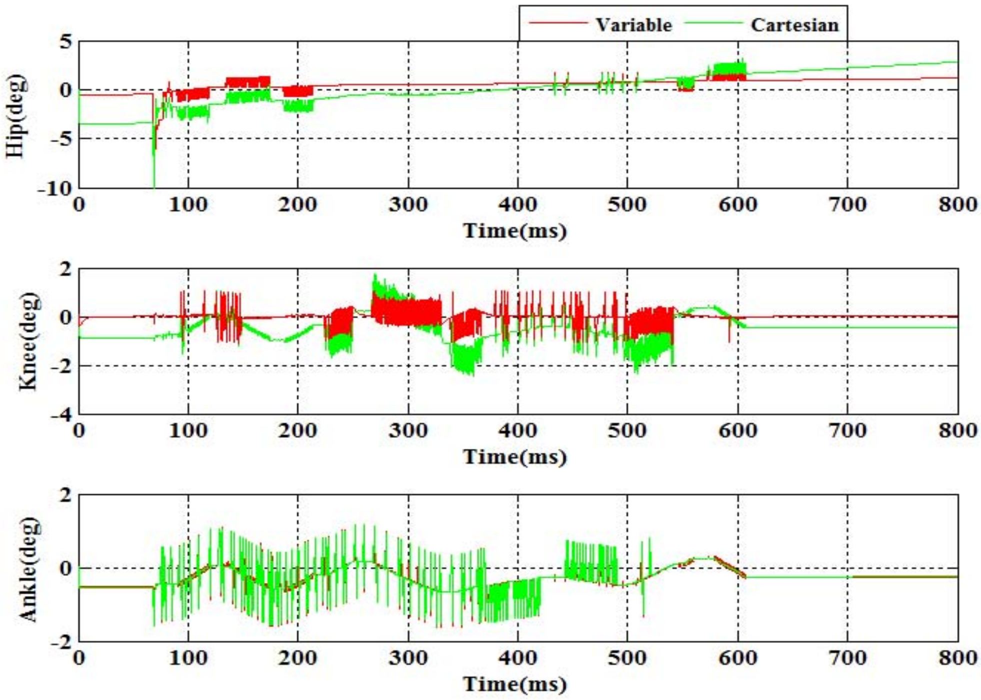

FIGURE 3.7: Trajectory error of hip, knee and ankle joints in the presence of impulsive disturbance

efficiently with the application of the Variable execution speed approach for the two kinds of disturbances considered and this is evident from the minimum position error in all the joints as shown in Figures 3.6 and 3.7. In the sinusoidal disturbance of hip joint, the maximum position error is in the range -2deg to +2deg in Variable execution speed and -10deg to +5deg in Cartesian position control. For impulsive disturbance, the maximum position error is -6 to +2deg and -10 to +3deg for variable execution speed and Cartesian position control respectively. Figure 3.7 depicts that the maximum position error is observed at the time instant on the initial application of the impulsive disturbance.

3.6 Discussion

Simulation studies on robots helps in understanding the behaviour of the mechanism in presence of multiple forces acting on it. These tools are also helpful in understanding the dynamic adaptability of the system and the model to avoid the influences of forces[212]. By means of this chapter, the influence of these external forces, impulsive or sinusoidal, are analysed and the trajectory is recovered by the two methods.

The Cartesian control approach follows the planned trajectory with minimal errors except during the initial and the final stage of the trajectory where a steady state error is visualized. This steady state error is actually found throughout the trajectory pattern but its effect is minimized by including an integral function of the Cartesian error. For Variable execution speed, the minimal error is maintained throughout the hip joint because of the applied stiffness acting on the disturbed joint. In case of the undisturbed joints, knee and ankle joints are affected by the kinematic bonding with the hip joint. The Cartesian control approach is able to control the effect of disturbances in these undisturbed joints but is not effectively following the positions within the minimum error range. In the case of Variable execution speed, the undisturbed joints are able to follow the planned trajectory with minimum error. With Variable execution speed, the frequency of the input orders is varied within 20-50ms range in the undisturbed joints and this helps in maintaining the synchronized movements along the trajectory.

In case of the impulsive disturbance, the Cartesian control follows the planned trajectory until the presence of a negative force which can be assumed as muscle fatigue. In the presence of a negative force, the Cartesian control tries to follow the planned trajectory with the same force and thus leading to increment in the error behaviour at the later part of the movement. A negative force also affects the trajectory of the knee joint but the Cartesian control maintains the synchronization among them by following the trajectory in the new calculated positions. In the Variable execution speed method, the robot follows the trajectory as planned with no effects when disturbances appear.

3.7 Conclusions

The role of unpredictable forces acting on the trajectories are analysed and studied. The common sources of these types of forces or disturbances can be listed as muscle fatigue, muscle synergies, stimulation sources, but not limited to them. The disturbances can be modelled as sinusoidal or continuous and impulsive signals. Other physical constraints which affect the movement of the planned trajectory to be followed can appear. Both assistive and resistive forces acting on the patient were considered in order to evaluate their effects. The Cartesian based and Variable execution speed methods are applied in order to evaluate their performance in presence of different types of disturbances.

Cartesian based position control has shown its capability to follow the trajectory pattern presenting some accumulative errors. These steady state errors have been minimized by including an integral function. However, results have shown a higher tracking error in the presence of both disturbances. The need of a kinematic model depending on the patient's anthropomorphic parameter makes this method difficult to be generalized.

The Variable execution speed approach is able to follow the planned trajectory with minimum position errors. The minimum position errors signify that there is a maximum efficiency in following a therapy. The absence of kinematic model results in less computation time which reduces the reaction time to recover the planned trajectory. Disturbances are defined based on the interaction forces acting between the patient and the robot and hence making it more patients dependent. The experimental results confirmed that the time delay in the undisturbed joint trajectory is minimum, what facilitates the synchronized movements between the joints. The use of Variable execution speed in presence of multiple disturbances would be a challenging task by means of multiple effects acting in a joint at a time instant t. This multiple direct effects on a joint also leads to some unpredictable effects in other joint trajectories, thus leading to the loss of synchronization among the individual joint trajectories.

This study helped in understanding the role of disturbances, both from the human and external source, which can affect the movement of the exoskeleton. Adjusting the input speed alone will not be a possible solution, considering the behaviour of the human inside the control loop. Similarly, a fixed stiffness approach is not suitable in all cases, as it might harm or affect the real potential of the user. Hence an interaction based adaptive approach is necessary in such a human centered strategy. Chapter 4 presents Task specific adaptation based on the human-orthosis interaction. The combination of other assistive devices are also considered, in coherence to the tasks and the needs of the therapist.

Chapter 4

Human-centered strategy for rehabilitation

[*] *Wearable robots are expected to expand the use of robotics in rehabilitation since they can widen the assistance application context. An important aspect of a rehabilitation therapy, in terms of lower extremity assistance, is balance control both in static and dynamic state. In this chapter, task-based control strategies are presented for human-centered robotic rehabilitation therapy to guarantee postural stability and gait assistance, using a wearable robot. The control model has been proposed based on the three main rehabilitation tasks: Sit-to-stand, Balance training and Gait training. The control models proposed use the interaction torques and joint angles for evaluation, irrespective of the tasks. The developed control strategy has been designed to provide the necessary assistance, related to balance recovery or maintain postural stability, under the "Assist-as-needed" paradigm. The user specific adaptation can be defined based on the user's interaction with the orthoses and by monitoring their intentions, depicting an human-centered approach. Further, a volition-based initiation is proposed using BMI and/or mechanical interaction, to ensure an active participation of the user.*

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[*] This chapter has been confined based on the following publications:
" An Adaptive control strategy for postural stability using a wearable exoskeleton ", V.Rajasekaran, J. Aranda, A. Casals and J. L.Pons, Robotics and Autonomous Systems, 2014
" Adaptive walking assistance based on human-orthosis interaction ",V. Rajasekaran, J. Aranda and A. Casals, Proceedings of the 2015 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS 2015)
" Compliant gait assistance triggered by user intention ",V. Rajasekaran, J.Aranda, and A.Casals, Proceedings of 37th International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC 2015)

4.1 Introduction

The studies performed in Chapters 2 and 3 led to the conclusion of having a hierarchical control model, to ensure that the exoskeleton patterns and the human interactions are handled sufficiently and effectively. The control structure is categorized as a Low level controller (LLC) and the High level controller (HLC), as explained in the section 1.4. The safe and suitable management of exoskeleton and human-orthosis interaction will result in delivering an effective therapy in a human-centered rehabilitation scenario. The HLC model involves the decision making process, co-ordination among the joints and combination of essential devices. Task based control models are efficient to answer each task individually, which also permits the choice of combining multiple devices for each model. Since there is no single possible solution for all the tasks involved, considering the human inside the loop, it is essential to handle and solve them independently with respect to the human behaviour. Hence in this dissertation, the control models are classified based on the three major tasks involved in a lower limb rehabilitation therapy;

1. Sit-to-Stand - Transition from wheelchair to stance position
2. Balance Training - Maintaining postural stability
3. Walking - Performing the gait movement

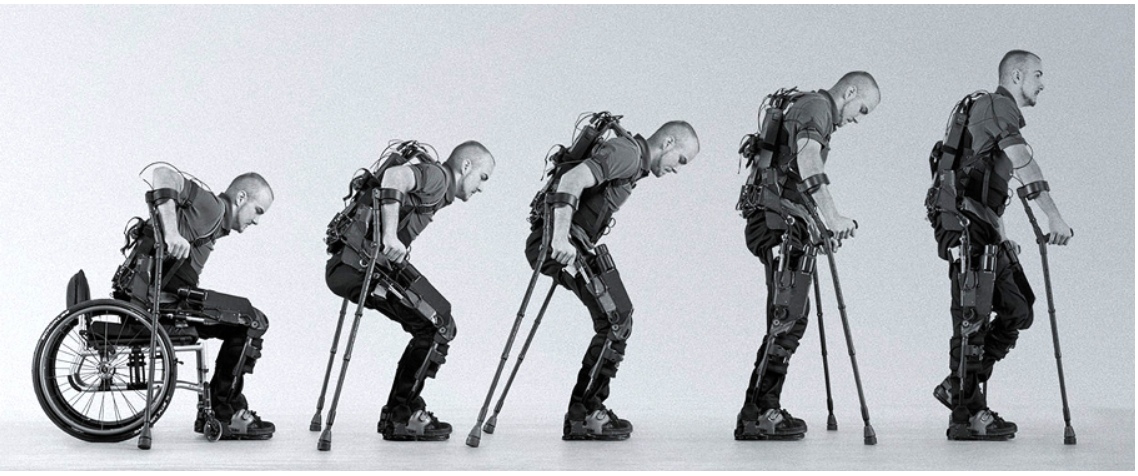

Figure 4.1: Rehabilitation tasks in a lower limb therapy- Sit-to-stand; Balance Training and Walking performed using ReWalk exoskeleton [157]

4.2 Sit-to-Stand

Sit-to-stand is one of the basic movement and essential part in ADL's. This movement also needs the high performance from the patient or subject in lifting up the entire upper body weight and simultaneously maintaining the posture of the body. The muscle power in the knee joint plays a key role in this movement and it is essential to provide adequate force assistance for the knee joint [213]. In some cases, an external support is also needed to help the patient maintain the equilibrium and to make a smooth transition [214]. Wearable sensors based sit-to-stand and stand-to-sit movement was established using the surface EMG signals [215]. The EMG signals are used to monitor the muscle synergies in both the movement and then a fuzzy control model was established to assist the movement. A Smart mobile walker (SMW) based assistance for elderly people has been performed [216, 217] based on the kinematic computations. A compliant actuator exoskeleton MIRAD [218] was developed specifically for performing the Sit-to-stance movement with a total of 6 flexion and extension movements. A COM based sit-to-stand model is also efficient in ensuring stability above the foot while sitting [76], but a complete kinematic model is needed to ensure its performance. Although, the sit-to-stand is a transition phase, it needs to be handled more precisely to ensure a stable posture at the end.

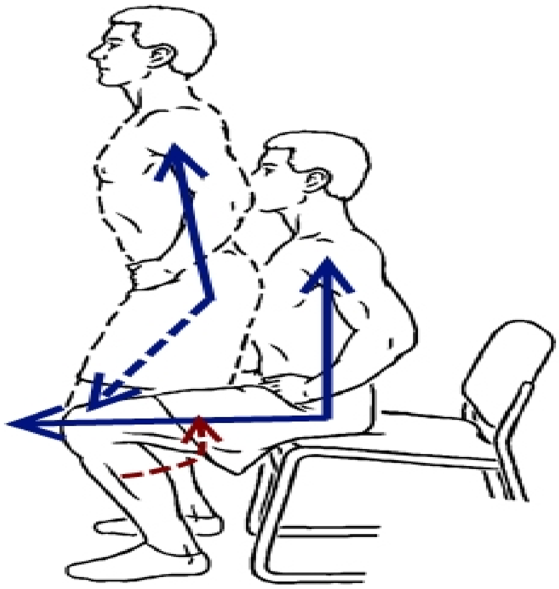

FIGURE 4.2: Transition of Sit-to-stand model as performed by a healthy human; The trunk movement helps in maintaining the COM of the body while the hip joint variation is minimal. The knee joint support varies in a wide range (indicated by red arrow).

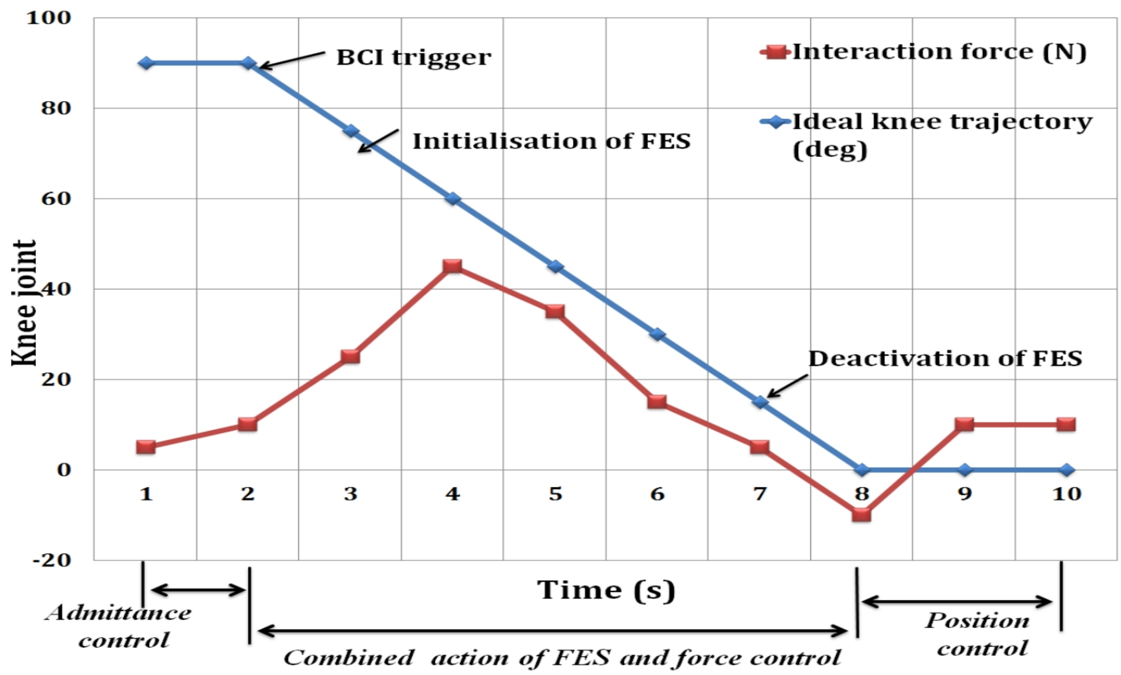

FIGURE 4.3: Velocity-based control scheme for the sit-to-stand task; The switching between the three control models help in ensuring the user induced movement, transition and postural stability respectively. The interaction forces increase and decrease due to the action of FES.

In this work, a velocity based control strategy is applied based on the error in velocity from which we generate the required resistive or assistive forces to achieve the equilibrium position. Figure 4.3, shows the ideal trajectory to be followed by the knee and the possible changes in interaction forces due to FES activation. In order to ensure a stable transition and to initiate based on the user movement, the controller switches between three control modes and this is further combined with assistive devices. The admittance control facilitates the user initiation by giving some freedom in the movement. Similarly, a position or high stiffness control is needed in the end to maintain the stability for keeping the subject in upright position. The joint movement is initiated when the patient tries to move from the idle or sitting stage to a different position, indicating the intention of the movement. When the controller observes the change in the joint positions and simultaneously observe a change in the neuro-motor behaviour, initializes the trigger to actuators and this simultaneously initiates the muscle stimulation by FES. The proposed velocity-based controller acts between the detection of the movement and the final position and this is represented by the following equation,

$$F = K(\dot{\theta}_{ref} - \dot{\theta}act) \tag{4.1}$$

where, F is the required force to assist or resist the joint, K is the stiffness constant for the force acting on the joint. $\dot{\theta}_{ref}$ and $\dot{\theta}act$ are the reference and current joint velocity respectively. In this case, the stiffness K is maintained constant to ensure equilibrium and the velocity varies depending on the user movement. The application of a constant K helps in handling both the FES induced forces and the muscle fatigue (force in negative direction), as described in Chapter 3.

4.3 Balance Training

Balance is a generic term that relies on the dynamics of body posture with the goal of preventing falls. A degradation of balance control in humans as a result of neuro-musculoskeletal disorders is an evident fact, for instance, in patients with spinal cord injury or stroke. Balance training in the presence of external perturbations [219] is considered as one of the important factors in evaluating patient's rehabilitation performance. Balance control can be oriented to achieve either postural or static stability (quiet standing), or dynamic stability in terms of walking.

Researchers have attempted to perturb the human balance system in a large number of ways in order to quantify the human response [220]. Reactive responses can be studied by introducing involuntary or unexpected perturbations. Similarly, to emulate a proactive response of an assistive orthosis we have to deal with voluntarily initiated perturbations. Researchers normally use an inverted pendulum model to validate and analyse strategies for postural stability [14, 221]. Thus referring to the human body, this model acts based on the variable COM (Center of Mass) and the center of pressure (COP). The analytic relation between COP and COM and the horizontal acceleration of COM has been studied using the inverted pendulum model too; i.e. COP-COM is proportional to the horizontal acceleration of the COM in both sagittal and frontal planes. Gage et al. [14] formally evaluated the performance of this model and found that it has similar characteristics to those of the human postural balance taken during a quiet standing posture. Vallery et al. [222] suggested an open loop triggered assistive system to ensure the right robot operation acting only when a loss of equilibrium is detected. This type of open loop assistance avoids instability caused by the lack of synchronization between two active controllers working in combination (human and robot). In this case, they use a variable speed control moment gyroscope to reduce power and torque requirements.

In this work, a control method that assists the patient in balance training based on previous studies performed on humans [221] is used. The projection of the COM (COM_p) of the body plays an important role in the control of human balance. The COM_p helps to detect the loss of stability and to provide assistance to the patient only when needed. We conclude that stiffness based control approaches play an important role in balance control, since a variable stiffness of the joint helps to avoid slacking and an open loop system assures an actuation only when a loss of stability is detected.

Balance control strategies can be analyzed by studying the evolution of the COM and COP of the human body. The efforts required to maintain balance imply the training of muscle activations as well as the coordination of the movement, which also helps the patient to progress towards the walking stage. As mentioned by Winter [13], a control strategy aiming to ensure postural stability can follow three possible strategies: Ankle, Hip and their combined actuation, as illustrated in fig 4.4. These strategies should be adapted according to the magnitude of perturbations. The knee joint is maintained fixed throughout this study, but this does not mean that the knee muscles are inactive. In fact,

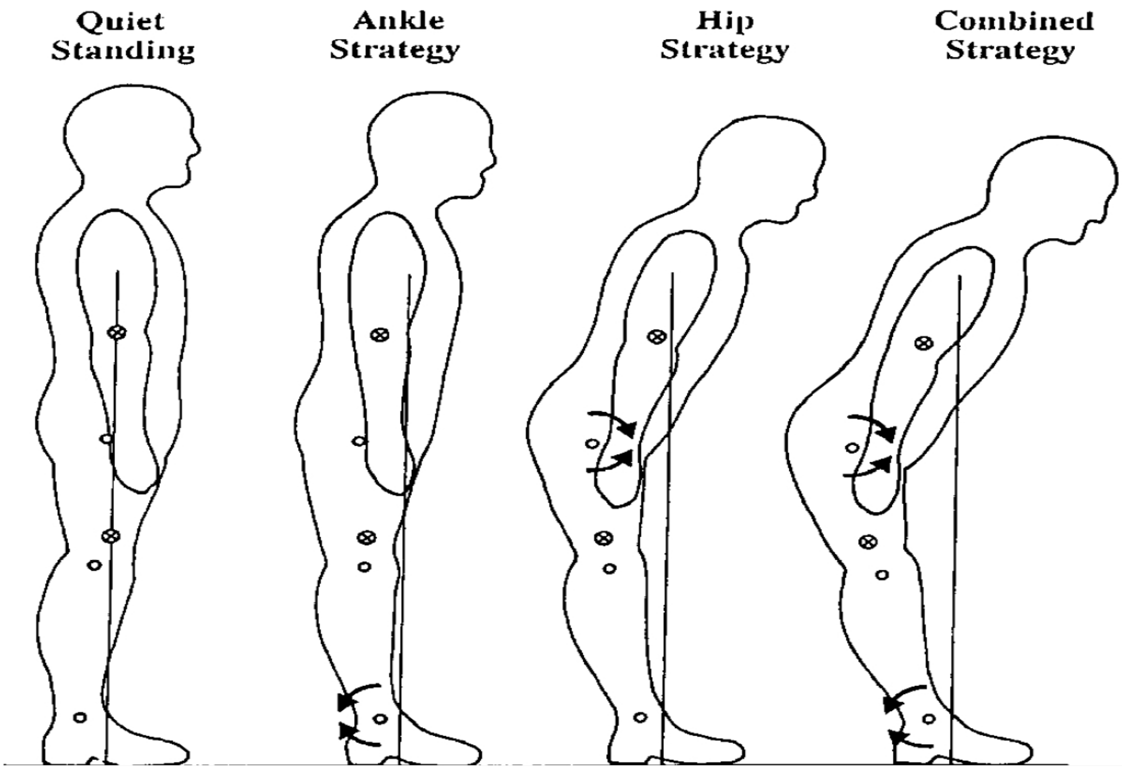

FIGURE 4.4: Postural stability strategies used by human to avoid the effect of perturbations: Ankle, Hip and Combined strategies; Note that irrespective of the strategy the knee joint is maintained constant or rigid and the selection of the strategy also depends on the magnitude of the perturbation; Picture adopted from [13].

a knee interaction force is found in all cases, which means that the patient contributes to maintain the joint rigidity.

The control strategy is based on the assistance and resistance to be applied on the subject in accordance to the monitored COM_p and joint movements. From literature, two types of perturbations are commonly used to study postural stability in humans. The analysis of internal perturbations is used to determine the limits of stability, by requesting the user to bend up to their stability limits. These internal perturbations depend on the user's movements and the human-orthosis interaction. The internal perturbation analysis thus acts as an initial stage for the posterior analysis of the effects of external perturbations, since the controller uses these learned limits to define a threshold that is used to prevent the subject from falling. In humans, the ankle and hip joint strategies vary in function of the magnitude of the perceived disturbance, and their combined effect varies as a result of the coupled action of the joints. In this study, we analyse each strategy individually and compare them amongst each other, in the presence of human-orthosis interaction.

4.3.1 Ankle Strategy

A widely discussed strategy in controlling the movement or maintaining the posture of the body in the anterior/posterior (A/P) direction is the ankle-strategy. This strategy is applied in case of small perturbations affecting posture, due to the action of the ankle muscles to maintain equilibrium and can be evaluated by analyzing the evolution of the center of pressure in postural stability. The role of the ankle joint in postural stability can be studied by maintaining a high stiffness on the hip and knee joints of the exoskeleton. The stiffness of the ankle joint is varied proportionally to its distance to the limits of stability, which have been measured previously for each individual using a Wii platform.

4.3.2 Hip Strategy

In humans, when the ankle strategy fails to control posture due to the action of high perturbations, a hip strategy responds by producing a flexion and extension movement. In this strategy, the upper body moves in the direction opposite to the movement of the lower part, thus changing the angular momentum. In order to make the patient proactive in maintaining their equilibrium, the stiffness of the hip joint is reduced. The flexion and extension of the hip movement cause a displacement of the COM, thus ensuring postural stability. In this study, the hip strategy involves the patient maintaining the COM_p within the limits of stability based on the values obtained from a sensory platform. If the subject is unable to control its postural stability, or if the posture moves beyond the stability limits, the exoskeleton varies the hip joint's stiffness, thus preventing the subject from proceeding further in the same direction. In this method, subjects are not allowed to move their knee and ankle joints and this helps in studying the individual active response of the hip to a perturbation. By maintaining the knee and ankle joint with high stiffness the hip joint is responsible for maintaining the postural stability, in accordance with the magnitude of perturbation.

4.3.3 Combined Strategy

In the combined strategy only the knee joint maintains a high stiffness value and both the hip and ankle are controlled through a variable stiffness, which depends on the stability limits computed in combination with the COM_p. The human-orthosis interaction forces are monitored to evaluate the coupled action of the joints. The combination of hip and ankle actions helps to face large perturbations, which may imply different recovery speeds. This method also helps to maintain the COM_p within the limits by adequately coordinating the movement of the hip and ankle joints. The stiffness values of the hip and ankle joints are varied independently, in accordance with the movement of the subject.

4.3.4 Wii Platform

The Wii platform is one of the most widely known active virtual devices, which is also used to monitor the patient's COM_p in research. The Wii Balance board is a weighing scale that uses Bluetooth technology and contains four pressure sensors which are used to measure the center of balance [223]. In the present work, the Wii platform is used to determine the limits of stability of each individual during the initialization phase. The Wii platform is connected to the high level control to track the COM_p values and the support polygon, by which the limits of stability are determined. In this phase, the High-Level control also monitors the joint angles and interaction forces. The former are used to detect the postural limits and the latter to determine whether an assistive or resistive action is required.

4.3.5 Event-based fuzzy control

An event-based fuzzy control strategy is developed to assist the patient to recover balance when a loss of stability is detected. The exoskeleton provides the joint angles (θ_{act}) and interaction forces (F_{int}) as the measured variables to monitor the subject's condition and the Wii platform gives the COM_p, as shown in fig 4.5. The limits of stability (θ_{los}) are computed based on the joint angle (θ_{act}) and the maximum value from $COM_p(x, y)$ coordinates. This computation is performed in the initialization phase and then the stiffness variation acts based on the joint angles and the defined θ_{los}.

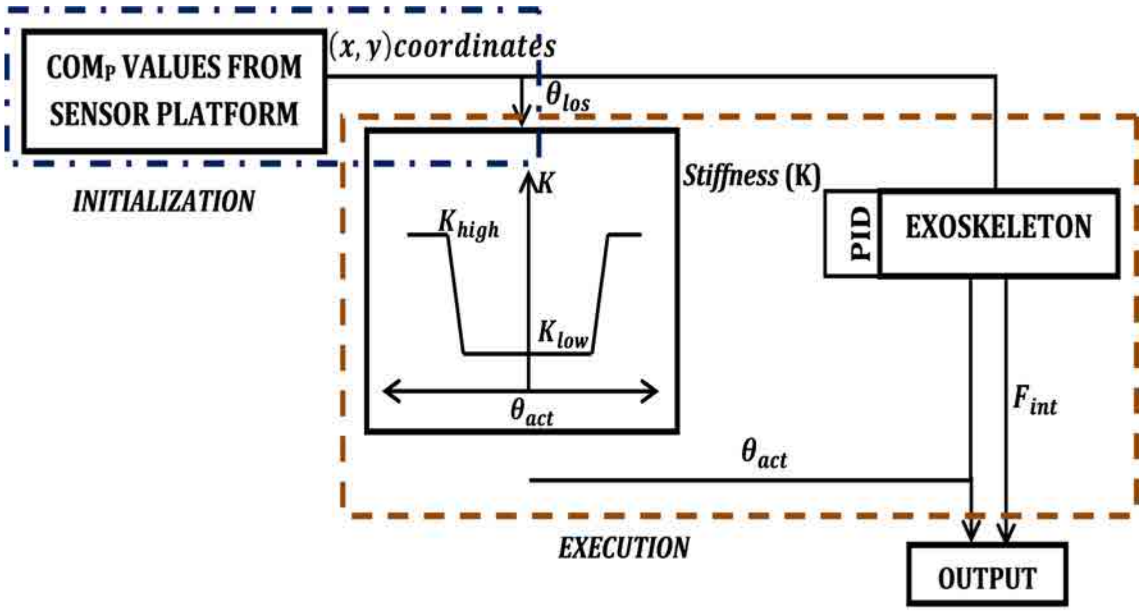

FIGURE 4.5: Event-based fuzzy control model of all the joints for balance training

The controller applies a variable stiffness to the joints, which value depends on their positions with respect to the stability limits and it is bounded, within a predefined range $K_{low} - K_{high}$. To adapt the therapy to the patient conditions, the therapist can define the limits of the free movement of the subject, previous to the robotic intervention, by applying a so called confidence factor (s). This confidence factor, which can be modified based on the subject's recovery progress, reduces the range of movement defined by the joint stability limits, modifying the bounding area defined by K_{low},K_{high} and θ_{act}, as shown in fig 4.6. This resulting bounding area ensures stability by limiting the body movement during the external perturbation analysis. The stiffness of a joint is defined by three intervals determined by the relation between the actual joint position and then resulting joint limits after applying the confidence factor. Its value is calculated as follows:

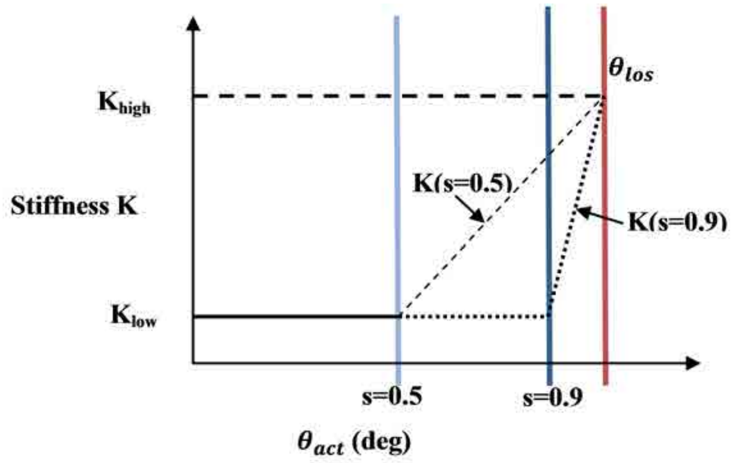

FIGURE 4.6: Relation between the variation of stiffness (K) and confidence factor(s)

$$if, \theta_{act} > \theta_{los}, then K = K_{high} \tag{4.2}$$

$$if, \theta_{act} < s * \theta_{los}, then K = K_{low} \tag{4.3}$$

$$if, s * \theta_{los} < \theta_{act} < \theta_{los}, \tag{4.4}$$

then K value is given by the following equation:

$$K = K_{low} + (K_{high} - K_{low}/\theta_{los} - s * \theta_{los})(\theta_{act} - s * \theta_{los}) \tag{4.5}$$

where K is the stiffness of the joint, s is the confidence factor, θ_{los} is the limit of stability, θ_{act} is the current position of the joint and, K_{low} and K_{high} represent the minimum and maximum stiffness value of the joint respectively.

The terms K_{low} and K_{high} are used to limit the stiffness value of the joints and they are defined heuristically. Since a high stiffness value ($> 90Nm/deg$) would practically be a position control approach, it is essential to limit this high stiffness in such a way that the subject is able to bring himself to a stable position. Similarly a low stiffness value ($< 10Nm/deg$) would leave the patient completely free and this would not ensure any postural stability. Hence, after a series of experiments performed with the subjects, the values of $K_{low} = 20Nm/deg$ and $K_{high} = 75Nm/deg$ have been chosen as the limits of stiffness to be applied to each joint to ensure stability and to prevent the orthosis or the patient from taking over completely. This $K_{high} = 75Nm/deg$ is the stiffness value considered in the knee joint for the three control strategies. For the hip and ankle control strategies the ankle and hip joint stiffness is also maintained at K_{high} respectively.

4.4 Gait Assistance

Gait training is a topic of interest for the patients with neurological disorders and for the researchers involved in developing assistive technologies. The nature and level of disorder challenges the development of a common applicable degree of assistance [137, 138]. Hence a patient specific assistance model is needed to provide the suitable gait compensation and to ensure equilibrium [224–226]. An efficient control strategy must constitute of the physical interaction of the user and signal-level feedbacks during the practical use [71, 81]. The control model should be effective by combining multiple therapeutic modalities to promote an engaging and challenging training along with an enhanced effect, following with more naturalistic movements[140]. Several treadmill-based training robots such as Lokomat[111], LOPES [105, 227], have proven to be efficient in providing the necessary gait assistance to the users especially for paraplegic patients. The availability of an external body support system handles the complications in maintaining the equilibrium [29, 35, 104, 105]. These treadmill-based training devices present the drawback that the user is not actively engaged in the therapy, which frequently leads to slacking in patients [20]. These devices are effective for individuals with complete SCI or acute Stroke, possessing less muscular strength to perform a movement [35].

Assistance in robotic rehabilitation can be achieved using an effective control strategy [18, 93] such as impedance or adaptive control, which acts based on the subjects' performance. Such control strategies operate under the principle of assistance-as-needed, in which assistive forces increase as the participant deviates from the desired trajectory [62]. The deviation of the user can also be used as an input to generate a trigger to initiate the movement or the assistance in accordance to the users' performance [63]. For gait, a personalized assistance in function of the user's intentions and movements is needed to dynamically adapt to the patients' needs. A predefined trajectory pattern based control, without other inputs, imposes a complete assistance which might induce slacking and harm the patient [18]. Thus, it is necessary to measure the human-orthosis interaction torques, to evaluate the user performance and status, in order to design a hybrid combination of force-position control. A similar method by means of the application of torque based on the user interaction was evaluated using a wearable device [67, 202, 228], where the knee joint torque is applied by estimating the required joints' muscle actuations.This type of user-performance based adaptive control can be modelled as a spring-damping coupled system. Several human centered strategies, such as patient cooperative and support motor function assessment, oriented to the development of robot behaviours have been widely studied [71]. These strategies allow the development of patient driven behaviours that rely on the active force contributed by the patient, to achieve the desired trajectories.

Assistive walking using an exoskeleton constitutes a major step forward in rehabilitation robotics. The assist-as-needed concept determines the level of assistance that the robot provides to the patient. Walking based on completely assisted robot therapies induce slacking to the patient, so it is necessary to provide a personalized assistance dynamically adapted to the changing patient's needs. The level of assistance varies with regard to the patient and the therapy, which involves the knowledge of assistance to be perceived. In classic control, the assistance can be either position or force based or a combination of both. These kinds of assistance can be improved with other factors such as the hardware structure, control strategies, combination of assistive devices etc., One of the widely used approaches to detect and evaluate the need of assistance is by evaluating the position errors with respect to a predefined trajectory model. However, the use of a predefined trajectory pattern, without other inputs, imposes a complete assistance without considering the status and progress of the user. It is necessary to measure the human-orthosis interaction forces to design a hybrid force-position control.

This work is intended to advance a step forward in the implementation of an impedance based approach for walking using a wearable robot, without treadmill training and body weight support. The absence of weight compensation carries with it the challenging task of maintaining the equilibrium in presence of ground reaction forces. The goal is to develop an efficient control model for a low-cost wearable robot and to validate the assistive behaviour of the robot for patients with neurological disorders. A hybrid position and interaction forces based control strategy is presented to continuously adapt the user movements to the desired gait pattern in real time. This real time adaptation also ensures the synchronization among the joint trajectories to maintain the dynamic stability.

4.4.1 Adaptive control

Adaptive control can provide the required performance, which is essential in human-centered strategies. The human-orthosis interaction is needed in defining the dynamic analysis of a human centered adaptive control model. Hence, the mathematical model for the dynamic analysis of an exoskeleton can be represented as

$$M_{ort}(q)\ddot{q} + C_{ort}(q,\dot{q}) + G_{ort}(q) = \tau_{act} + \tau_{int} + \tau_d \tag{4.6}$$

where q,$\dot{q}$, $\ddot{q}$ are the vectors of joint positions, velocities and accelerations. $M_{ort}(q)\ddot{q}$ is the inertia matrix, $C_{ort}(q,\dot{q})$ is the centrifugal and Coriolis vector and $G_{ort}(q)$ represents the gravitational torques. τ_{act} and τ_{int} are the orthosis and interaction torques respectively and τ_d correspond to the external disturbances acting on the subject. These actuator and patient torques are influenced by the human-orthosis interaction, while the external disturbances can be due to any assistive or external sources which can affect the dynamic stability of the robot. In the present work, τ_{int} is used and to determine the level of assistance to be exerted by the orthosis.

In the present work, for each joint, the actuator works in collaboration with the patient. The actuator torque can be modified by varying the joint stiffness parameter, which invariably modifies the corresponding joint trajectory and the force compensation. This stiffness variation alters the actuator torque which determines the degree of control transferred

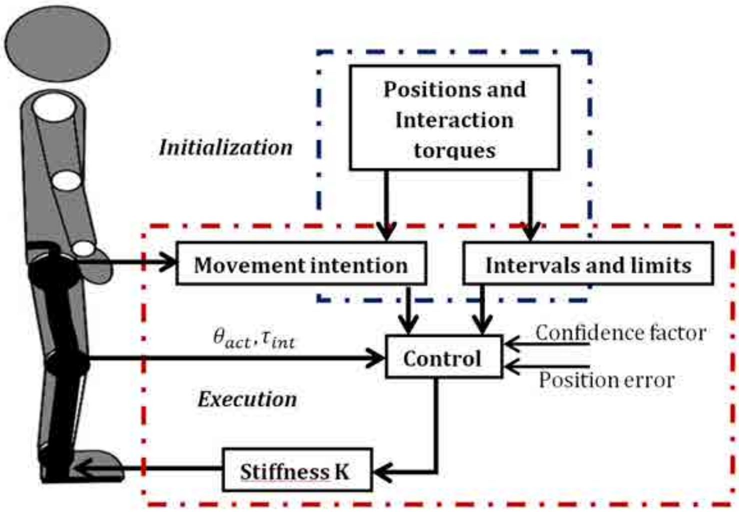

FIGURE 4.7: User interaction-based Adaptive control strategy for Walking; The gait initiation command is received based on the variation in the knee joint movement

from the orthosis to the human or vice versa. Such an impedance control scheme has been widely used for its compliant behavior, which results in an adaptive walking pattern and a more natural interaction between patient and orthosis [4, 133]. Thus the impedance control can be determined by equation,

$$F_j = Ma + Cv + K_j(\theta_{ref} - \theta_{act}) \tag{4.7}$$

where, θ_{ref} and θ_{act} are the reference and actual joint positions respectively, K_j is the stiffness vector parameter of all the joints and F_j represents the applied force to the joint. M represents the mass, C is the damping constant and a and v represent the acceleration and velocity of the joint. Here, the input sample rate to the system is maintained constant and the damping coefficient is kept small (as a result of Chapter 3), therefore the velocity of the joint is not modified by the user and thus, it does not induce any significant effect on the applied force. Hence the force equation, influenced by the position error, is modified as

$$F_j = K_j(\theta_{ref} - \theta_{act}) \tag{4.8}$$

The value K, for any joint, can be determined dynamically based on the performance of the user and the level of assistance to be exerted by the orthosis.

$$K_{t+1} = K_t + \Delta K \tag{4.9}$$

$$\Delta K = abs(\theta_{ref} - \theta_{act}/s * (\tau_{int})) \tag{4.10}$$

where, $\tau_{int}$is the human-orthosis interaction torques and and s is a confidence factor (0.1 - 1) which is used to determine the stiffness to be applied at time t+1. The confidence factor is a variable parameter which shall be defined by the therapist in function of the capabilities of the patient. This confidence factor can be varied according to the progress of the user, in order to modify the time instant at which assistance is to be initiated. A low confidence factor (0.1 to 0.5) means that the assistance should be provided partially or completely to the procedure and a higher confidence factor (0.6 to 1) indicates that the subject is capable of walking without or with little assistance. The inclusion of the τ_{int} as a coefficient establishes a dynamic relation with the stiffness variation. A higher τ_{int} signifies active participation of the user and if that resulted in low position error then the stiffness value will be varying very less or maintained. To the contrary, if τ_{int} is low and the position error is high then the stiffness must be incremented to provide the suitable assistance.

This variation of the K value results in a change in the force acting at the joint level what is perceived as assistance or resistance by the patient. Thus, the K value is incremented as a function of the evolution of the patient and in case of decrement it is fixed to 1Nm/deg, to avoid any uncertainties.

The stiffness variation module, as shown in fig 4.8, is responsible for incrementing, decrementing or maintaining the stiffness parameter of each joint. Within a given range of trajectory errors, stiffness is computed from position errors, but when the defined error thresholds are surpassed, stiffness should be modified according to the measured interaction torques. This condition takes place in general with the presence of external perturbations.

The following are the parameters involved in defining the stiffness variation module for the gait assistance algorithm:
θ_e - Position error (deg)
$\theta_e Th_{lo}$ - Lower threshold position error (deg)

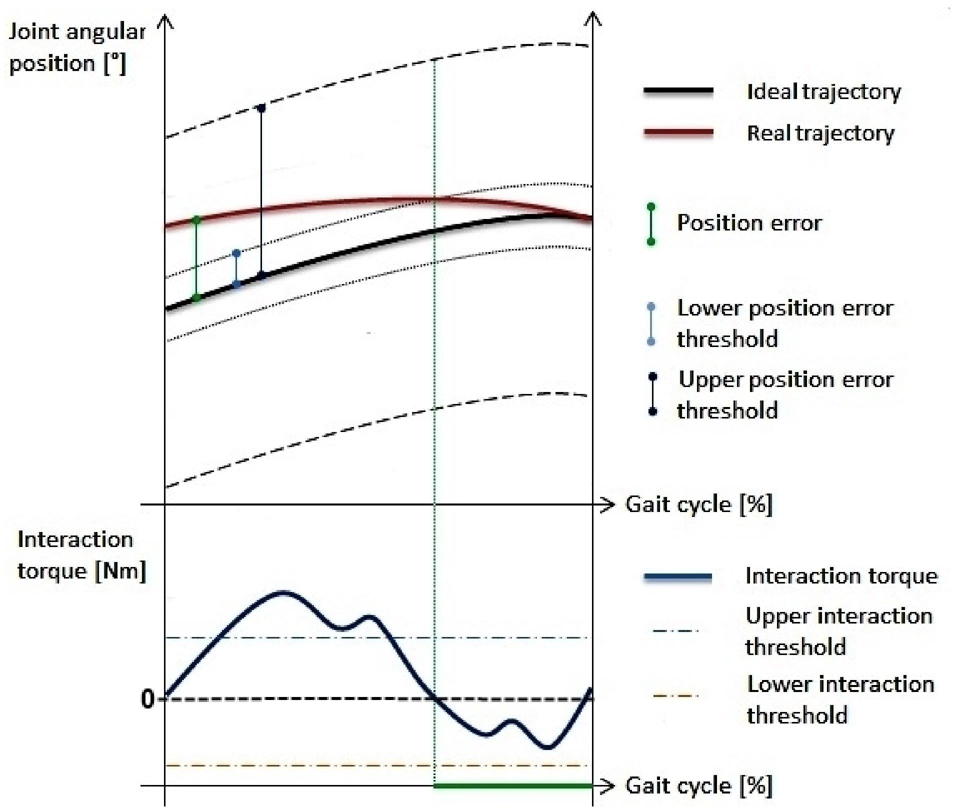

FIGURE 4.8: Relation between the stiffness variation and the joint input parameters: Angles and interaction torques, as used for healthy subjects

$\theta_e Th_{up}$ - Upper threshold position error (deg)

τ_{int} - Human-orthosis interaction torque (Nm)

$\tau_{int} Th_{lo}$ - Lower threshold of interaction torque (negative direction) (Nm)

$\tau_{int} Th_{up}$ - Upper threshold of interaction torque (positive direction) (Nm)

ΔK - Stiffness variation (Nm/deg)

Algorithm 1 Adaptive Gait assistance

Case (1) $|\theta_e| > \theta_e Th_{up}$

$$K = K + \Delta K$$

Case (2) $|\theta_e| < \theta_e Th_{lo}$ && $\tau_{int} Th_{lo} < \tau_{int} < \tau_{int} Th_{up}$

$$K = K - \Delta K$$

Else (3) $\quad MaintainK$

The three cases, depicted in fig 4.9, in terms of the adaptive stiffness variation behaviour, as listed in the algorithm, can be expressed as : (1) When position error is high, K must

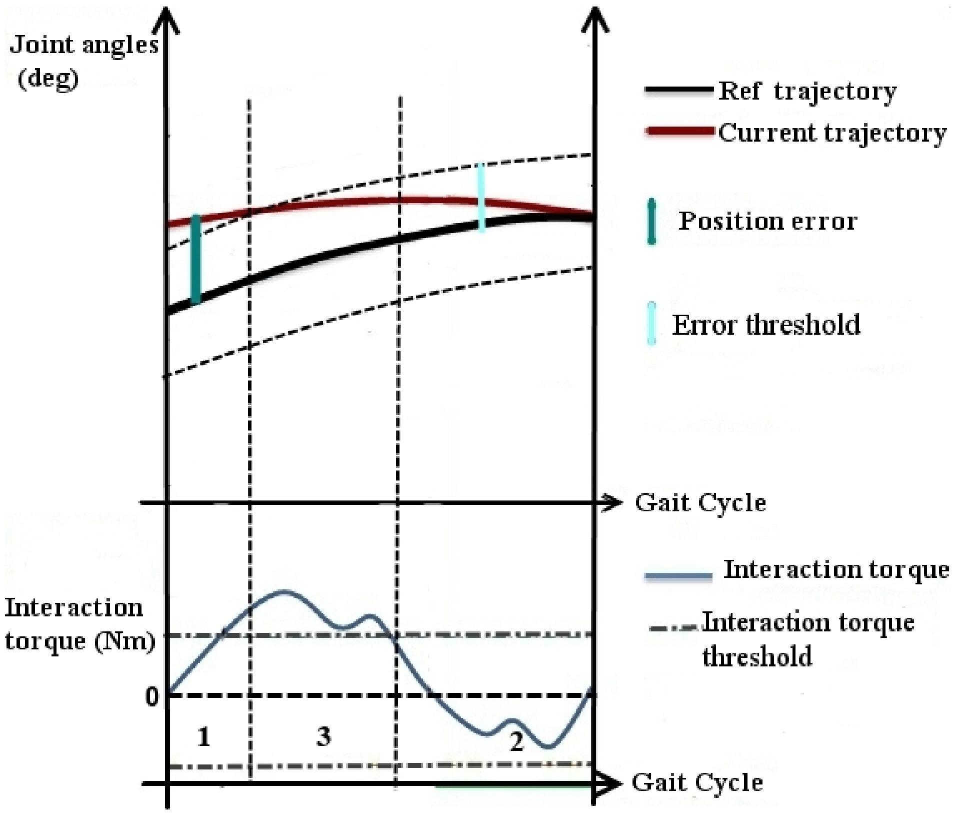

FIGURE 4.9: Relation between the stiffness variation and the joint input parameters: Angles and interaction torques, as used for SCI patients. Note: For patients, both the upper and lower threshold values are same. The three cases (1,2,3) are in coherence with the algorithm for gait assistance.

be incremented; (2) When position error is less than the threshold and interaction torque is low, K must be decreased and (3) In any other case, K must be maintained. In case of the experimentation with incomplete SCI patients, the upper and lower threshold limits are limited to one value and this facilitates in assisting the patient with respect to their movement. This interaction torque limitation helps in identifying the stiffness variation. The upper and lower threshold of interaction torques are necessary to address their non-symmetrical behaviour and variation in function of the movement. In case of the position error, absolute values are observed due to their symmetrical evolution.

4.4.2 Volitional Control

A user specific control model can be realised by identifying the two major actions involved: intentions and movement of the user. The combination of muscle synergies and neural signals for evaluating the volitional commands of the patient has gained more importance in the recent years, as a top-down approach in rehabilitation [229]. The detection of the best instant for gait initiation and termination was performed to develop a volitional control based robotic rehabilitation [67, 84]. This type of volitional input motivate the user to initiate the therapy and subsequently improves the ability of intention [187, 188]. User intention based rehabilitation can be implemented by different approaches such as using an external input order (joystick) [110], displacement of the COM [164, 189] or muscle activity [230]. One of the widely used approaches for monitoring the human intention relies on the use of brain machine interfaces (BMI) such as in XoR [185], Rex [186]. These systems are efficient in monitoring the user intentions mainly, because a real displacement of the joint position is not needed always to initiate the gait. A similar kind of intention based control has been evaluated in a prosthesis, in function of the interaction with an amputee [231]. Neural network based gait phase recognizer and pattern generator has been employed in [232] characterized by the force acting on the user's muscle.

User intentions' can be recognized either by following the user movement in close contact or by detecting the motor-related brain activity of the user. The control model involves monitoring the joint angles and human-orthosis interaction torques, from the exoskeleton and calibrating the input stiffness order to each joint, based on the user interaction. Volitional control is used as a trigger to initiate the joint movement of the exoskeleton, which

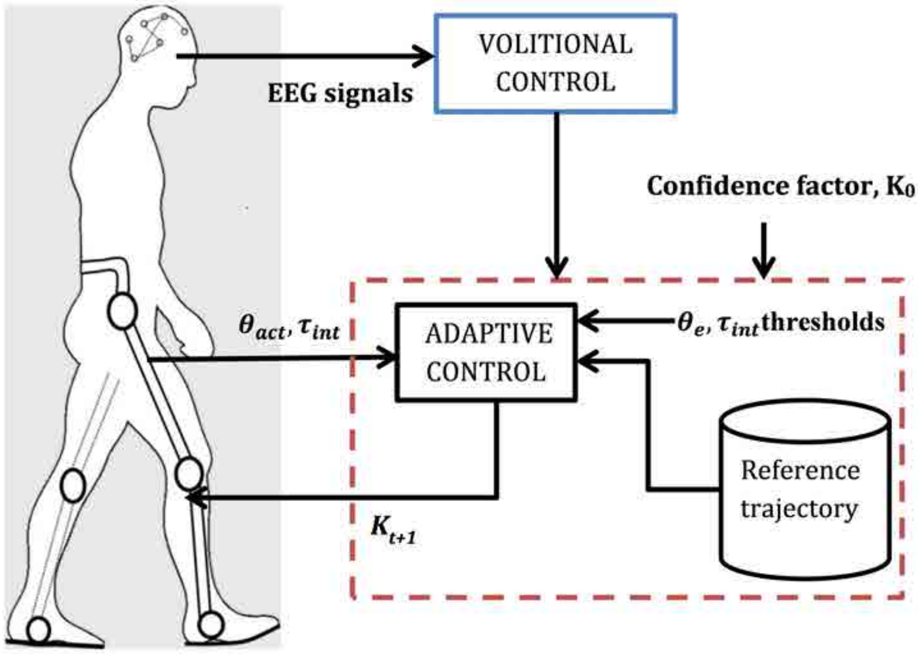

FIGURE 4.10: Schematic representation of the volition based adaptive control with BMI system and exoskeleton. Input command from the BMI system triggers the adaptive control model which initiates the exoskeleton movement

can be performed by using a BMI system or by monitoring the mechanical interaction of the user, as shown in fig 4.10.

4.4.2.1 Mechanical Interaction

The gait initiation can be defined as the time 't' when the user intends to perform their movement and can be determined by using the human-orthosis interaction torques. This type of user dependent initiation is efficient in influencing or motivating the user to provide an input movement. The user motivation is necessary for an assistive rehabilitation procedure such as to avoid slacking. These human-orthosis interaction torques based initiation can be a drawback in differentiating between a tremor [233] and the user intention. Hence the joint position and torques must be considered to identify the gait initiation, to trigger the therapeutic procedure. In other case, this strategy permits the user to initiate the therapy with whichever leg they are more comfortable, as shown in fig 4.11. The gait initiation strategy considers the joint positions and the human-orthosis interaction torques of both legs. This can be represented as

$$G_{init} = Leg_x, \forall x \epsilon [R, L] \tag{4.11}$$

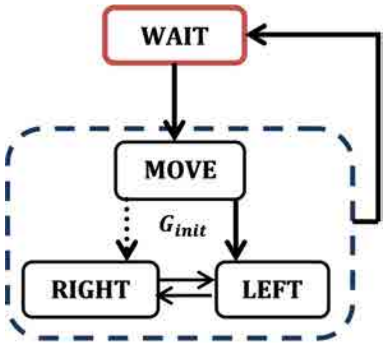

FIGURE 4.11: State transition of the interaction-based gait initiation

where, G_{init} is the gait initiation trigger obtained from the user for Leg_x. The following are the parameters involved in the gait initiation algorithm, based on mechanical interaction:

θ_R - Joint angles of right leg (deg)

θ_L - Joint angles of Left leg (deg)

τ_R - Interaction torques in right leg (Nm)

τ_L - Interaction torques in left leg (Nm)

Algorithm 2 Mechanical interaction based gait initiation

Case $\theta_R(t) \neq \theta_R(t-1), \tau_R(t) \neq \tau_R(t-1) \&\& \theta_L(t) = \theta_L(t-1)$
$x = R$
Else $x = L$

The τ_{int} (τ_R or τ_L) can never be equal to zero and thus it is essential to consider the deviation in terms of joint positions.

4.4.2.2 Brain-Machine Interface

The volitional control was realised by monitoring the motor-related brain activations over the motor cortex region, as triggers for initiating the movement of the orthosis. The brain-machine interface system consisted of a commercial g.Tec system (g.Tec GmbH, Graz, Austria), with 2 amplifiers and 32 EEG channels, placed at AFz, FC3, FCz, FC4, C5, C3, C1, Cz, C2, C4, C6, CP3, CP1, CPz, CP2, CP4, FP1, FP2, F7, F3, Fz, F4, F8, T7, T8, P7, P3, Pz, P4, P8, O1 and O2 (according to the international 10/10 system). The ground and reference electrodes were placed on 'FPz' and on the left earlobe, respectively. The EEG amplifiers were carried by the subjects in a backpack, and connected to a laptop which processed the neural signals. The processing was done with custom-made C++ software, integrated with Matlab scripts. The decoding of movement intention commands was based on the combination of the event-related desynchronization(ERD) of the sensorimotor rhythms [234] and the motor-related cortical potentials(MRCP) [235]. These decoded intention commands were sent to the exoskeleton controller to initiate the gait movement. The BMI system used to decode the motion intention has been explained in detail [236–238], and indeed, it has already been used to decode the motion intention of the upper limb in SCI patients [239]. As part of project HYPER consortium (section1.4), the work related to the detection of neural-motor activity using BMI was performed by another group and has been presented elaborately in [240].

4.5 Summary

In this chapter, task-based control in a human-centered rehabilitation scenario is proposed. The tasks were defined based on the general lower limb rehabilitation therapies for specific users with neurological disorders. The three main tasks involved are: sit-to-stand, balance training and gait training.

The transition from sit-to-stand is a necessary step in every lower limb rehabilitation therapy, for helping the patient to transit from wheel chair to the next level of therapy. A velocity-based adaptation is proposed for this task which will help in reaching the final position with a constant velocity. A combination of control models is proposed to ensure the initiation and the final stage of the transition , such as to ensure stability and to maintain equilibrium. Further, final stage of this transition must ensure keeping or helping the user to maintain the upright position.

Postural balance control is a common rehabilitation therapy for which the use of an exoskeleton offers high benefits. This therapy focuses on moving the ankle or hip separately, fixing the rest of the joints. By using an exoskeleton, the joint assistance can be selected for an individual or in a combined way. An event-based control strategy is proposed to ensure the joint action in function of the users' performance. The performance of the orthosis is compared with three different balance control strategies: ankle, hip or combined. The postural balance study facilitates in moving forward towards proposing an effective model for ensuring the dynamic stability in walking.

An adaptive gait assistance is proposed in function of the joint movement and humn-orthosis interaction. The applied joint stiffness is calibrated based on the user interaction and a confidence factor which limits the level of assistance. The stiffness function must act individually to all the joints and also ensure dynamic stability by means of limiting the joint error. Similarly, the stiffness value of each joint adapts dynamically to the user needs and keeps the joint positions bounded within the limits of the reference gait, in real time.

A volitional-based order is combined as trigger to initiate the movement, thus ensuring the active participation of the user and for an effective therapy. The volitional commands are observed from the BMI system, by monitoring the neural-motor activity of the user and/or by the mechanical interaction of the user. These volitional commands must be determined at the beginning of each gait cycle, such as to ensure the users' complete participation.

In Chapter 5 the evaluation of the presented human-centered approaches are evaluated with healthy subjects, for each task individually. The experimentation with the healthy subjects are deemed as a preliminary study to evaluate the level of assistance. After the conducted preliminary study, the experiments are performed with incomplete SCI patients and this is also presented in the forthcoming chapter.

Chapter 5

Results & Evaluation

[*] This chapter presents and discuss the trials performed with healthy subjects and incomplete SCI individuals, based on the control strategies presented in the previous chapter. The experimental protocol was designed with respect to the abilities of the user's involved in the trials. Results are presented in sequence of the rehabilitation tasks perfromed: Sit-to-stand, Balance training and Gait assistance. Initially the trials were conducted with five healthy subjects as a preliminary study prior to paraplegic patients. Four paraplegic patients participated only in the evaluation of the adaptive gait assistance, with volitional orders through BMI.

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[*]This chapter contents are based on the following publications:
" Gait assistance with volition based adaptive control: A study with incomplete spinal cord injury individuals " ,V. Rajasekaran, E. López-Larraz, F. Trincado-Alonso, J. Aranda, L. Montesano, A J. del-Ama and J L. Pons, Journal of NeuroEngineering and Rehabilitation,(Submitted on 05/10/2015)
" Control of an ambulatory exoskeleton with a brain-machine interface for spinal cord injury gait rehabilitation ", E. López-Larraz, F. Trincado-Alonso, V. Rajasekaran, A J. del-Ama, S. Perez-Nombela, J. Aranda, J. Minguez, Á. Gil-Agudo, L. Montesano, Frontiers in Neuroscience, (In progress)
" Adaptive walking assistance based on human-orthosis interaction ",V. Rajasekaran, J. Aranda and A. Casals, Proceedings of the 2015 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS 2015)
" An Adaptive control strategy for postural stability using a wearable exoskeleton ", V.Rajasekaran, J. Aranda, A. Casals and J. L.Pons, Robotics and Autonomous Systems, 2014

5.1 Introduction

This chapter presents and discusses the evaluation of the human-centered control strategies for rehabilitation tasks, presented in Chapter 4. The performance of the proposed control strategies are evaluated based on the individual tasks and the type of participants. As explained in the previous sections, the joint positions and the interaction torques of all the joints are monitored for evaluating the adaptation of the user. For sit-to-stand and balance training tasks, the joint angles are monitored to demonstrate the user adaptation. The interaction torques will be monitored to analyse the kind of assistance and resistance perceived by the user.

For gait assistance, both the angles and the interaction torques are monitored, to evaluate the adaptive performance of the control model, as explained in section 4.4. The exoskeleton should assist and motivate the patient to pursue their maximum flexion/extension movement in all the joints which also results in a change in behaviour of the interaction torques. Hence, performance in this work is evaluated by

1. Maximum flexion/extension range in the knee joint movement
2. Changes in the interaction torques (negative to positive or vice versa) with respect to the movement
3. Progressive change in the range of motion
4. Real time stiffness adaptation for all the joints, in function of the user needs

In order to evaluate the strategies with incomplete SCI individuals, it is necessary to perform a preliminary evaluation with healthy subjects. Thus this chapter is categorized based on kind of participants in the experiments: healthy subjects and incomplete SCI individuals. For healthy subjects, each task involved and the results obtained by the application of the proposed control models are also discussed elaborately. In case of the incomplete SCI patients, only the performance of the adaptive gait training is evaluated with BMI as a initiation trigger.

5.2 Assessment with Healthy subjects

Evaluation of the adaptive performance of the proposed control approaches is needed to understand the kind of assistance provided. It is also necessary to demonstrate the usability of such control models with healthy subjects before performing it with paraplegic individuals. This section will present and discuss in detail about the results obtained with healthy subjects. The assistance terminology in case of healthy sujects must be validated in terms of performance enhancement or energy efficiency. The joint parameters such as angles and interaction torques are monitored for all the tasks.

5.2.1 Sit-to-Stand

Sit-to-stand is a first step of transition in a rehabilitation task, to help the user to transit from their wheel chair and to progress towards the next step in rehabilitation. This task has been evaluated using a velocity-based control, as presented in the section 4.2 (Fig 5.1). In order to provide the adequate support to the user, this task has been approached with a combination of three different control modes. This test has been performed as a preliminary study with healthy subjects.

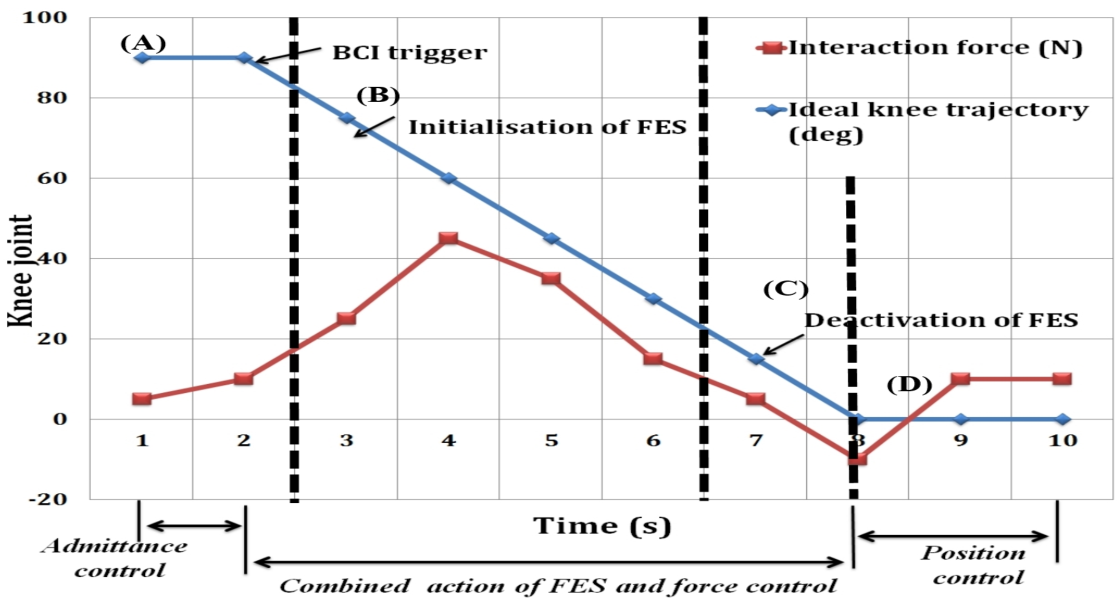

Figure 5.1: Combinational control model with velocity-based approach for the sit-to-stand task; The events A to D are charecterized by the changes in joint positions which in turn signifies the switching between the different control modes.

As shown in fig 5.2, the transition of sit-to-stance can be classified as four events A *to* D. The joint movement is initiated when the change in the joint movement is observed (A). The assistance begins from B to the final D with velocity-based control and a constant stiffness. The constant stiffness value supports the user posture as needed, and ensures stability and assistance throught the transition to standing. This kind of assistance is possible with a fixed stiffness to prevent the user from falling. When the user reaches the standing position D, the control mode shifts to position control which helps in maintaining the equilibrium, with or without any external support.

FIGURE 5.2: Sit-to-stand task performed by a healthy subject with the exoskeleton H1

Fig 5.3 illustrates the variation in the joint angles and interaction torques in the hip, knee and ankle joints. A significant change in the knee joint movement is observed but supported by a large amount of interaction in the hip and ankle joint, with constant position. The movement occurs within a span of 1-3 seconds which is related to the velocity of the movement performed by the user.

For paraplegic patients, a muscle stimulation might be needed to provide both muscle and functional assistance. The following are the possible sequential movements involved in case of experiment with patients, with the combination of FES:

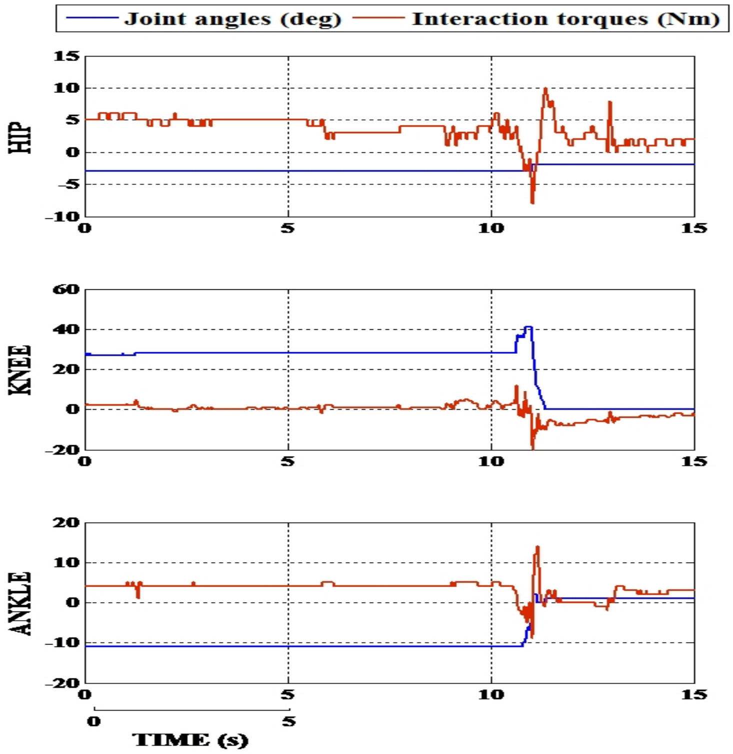

FIGURE 5.3: Changes in the joint angles and interaction torques after sit-to-stand transition performed by a healthy subject

- From stage A to B with admittance control, for all the joints
- At B, FES activation must be initiated while the exoskeleton will maintain with a fixed stiffness. This will help the muscle to move the user while the exoskeleton provides sufficient postural assistance.
- At C, the FES must be deactivated since the joint position is reached the standing phase. The joint stiffness must be maintained constant to help the user maintain the posture.
- At D is the end of the sequence, indicating the switch to position control model.

5.2.2 Balance Training

Balance training is a part of the rehabilitation exercise focused on training the patient to maintain equilibrium while standing. Balance training ideally can define the method to maintain the stability of the body in the presence of any external perturbation. An event-based fuzzy control strategy has been developed to assist the patient to recover balance when a loss of stability is detected.

Postural balance control can be implemented either acting on the hip, on the ankle joint or on both, depending on the kind of perturbation acting on the subject: internal or external. Internal perturbations can be produced by any voluntary movement of the body, such as bending the trunk. External perturbations, in the form of an impact force, are applied by the exoskeleton without any prior notice to observe the proactive response of the subject. In this dissertation, the interaction forces between orthosis and subject are monitored, as they play a relevant role in the definition of assistive and resistive movements to be applied to the joints. The proposed method has been tested with 5 healthy subjects in presence of internal and external disturbances.

A Wii platform is used to obtain the COM_p and to determine the stability limits in the initialization phase. The controller applies a variable stiffness to the joints of the exoskeleton, depending on their position in relation with the stability limits.

5.2.2.1 Protocol

The two types of perturbations were applied to each subject following successively the three balance control strategies, explained in section 2, and their effects were evaluated. Internal perturbations were self-induced by the subjects, making forward and backward movements of the trunk. In the initial phase, the subjects were instructed to reach the maximum flexion angles of their hip and ankle joints, both in the case of individual and their combined effect, by wearing the exoskeleton. These maximum angles were recorded and defined as the limits of stability (θ_{los}) along with the COM_p, obtained using the Wii platform. The interaction forces of each joint were monitored to evaluate the performance of the subject and the exoskeleton. In the execution phase, the exoskeleton generates the external perturbations, by applying an impact torque to the hip joint ($10Nm$) and to the ankle joint ($5Nm$) at random time instants 't' for both the individual and combined strategies. The motor torque

Figure 5.4: A healthy subject training with Exoskeleton H1, on the Wii platform

was applied in the forward direction, i.e., flexion in hip joint and plantar-flexion in the ankle joint, as shown in fig 5.5, emulating the behaviour of an exertive force. The subjects were blinded to the time instant and amount of torque being applied, thus preventing them from applying any internal forces in advance to stabilize themselves. Each subject performed three trials for each strategy and the time instants of torque application were different for each trial. Since this study involves healthy subjects, the confidence factor for limiting the free movement was defined as 0.9, such as to allow a wide range of movement. The goal of this analysis is to evaluate the reactive response of the exoskeleton and the recovery time. The reactive response of the exoskeleton prevented the subject from progressing beyond the stability limits or from falling. This response was evaluated by observing the deviation from the stable position, while maintaining the interaction forces within their predefined limits.

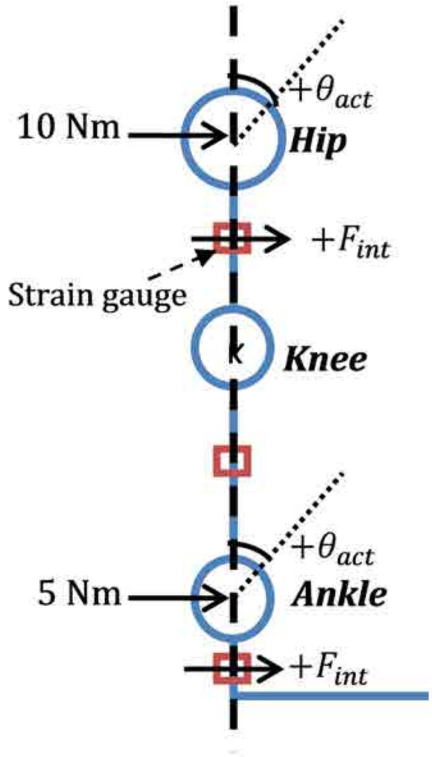

FIGURE 5.5: Experimental model of the external perturbations applied for Balance training; The magnitude of the applied impact torque was assumed based on the human studies

5.2.2.2 Results

Five healthy subjects around the age group of 25-35 (height: $1.71 \pm 0.08m$, weight: $77.5 \pm 5.5kg$) were chosen for this study which involves both male (3) and female (2) candidates. The stability limits for all the subjects were measured individually using the exoskeleton itself and with Wii platform for validation (refer section 4.3). These limits determine the degree of assistance and resistance to be applied to each individual subject. Since the physical parameters of the subjects differed in a narrow range, their average has been taken as the limits of stability. The experimental architecture in this analytic study consists of two levels of control: high and low level control. The high level control involves a pressure sensor platform (Wii) and the monitoring of the positions and interaction forces of all the joints during the initialization phase. In the execution phase, this high level control is responsible for calculating the required stiffness value in order to obtain a desired behavior. The low level control, embedded in the exoskeleton controller, is responsible for applying the suitable torque to the joints in proportion to the position error and the input stiffness.

The performance of each strategy, as explained in the previous sections, is evaluated here.

TABLE 5.1: Stability limits obtained from healthy subjects as a result of internal perturbation

Strategy	Limits of Stability (θ_{los} deg)	Interaction Forces ($F_{int}(N)$)
Hip	-20 and +30	-10 and +10
Ankle	-5 and +15	-5 and +20
Combined	Hip -20 and +20 ;	Hip -5 and +10;
	Ankle 0 and +15	Ankle 0 and +15

The internal perturbations, caused by the subject with the upper body motion, alter the inertial parameters between the different links of the body. This change in motion becomes evident as they produce sudden changes of COM_p values, which in turn affects the postural stability. The measured change of interaction forces is used to evaluate the existence of perturbations perceived when a loss of stability is produced. Table 5.1 shows the average limit of stability obtained as a result of the internal perturbation analysis and the resulting interaction forces performed by the subjects.

The relation between angles and interaction forces in the presence of internal perturbations in the ankle and hip joints can be studied from figures 5.6and 5.7, respectively for a specific

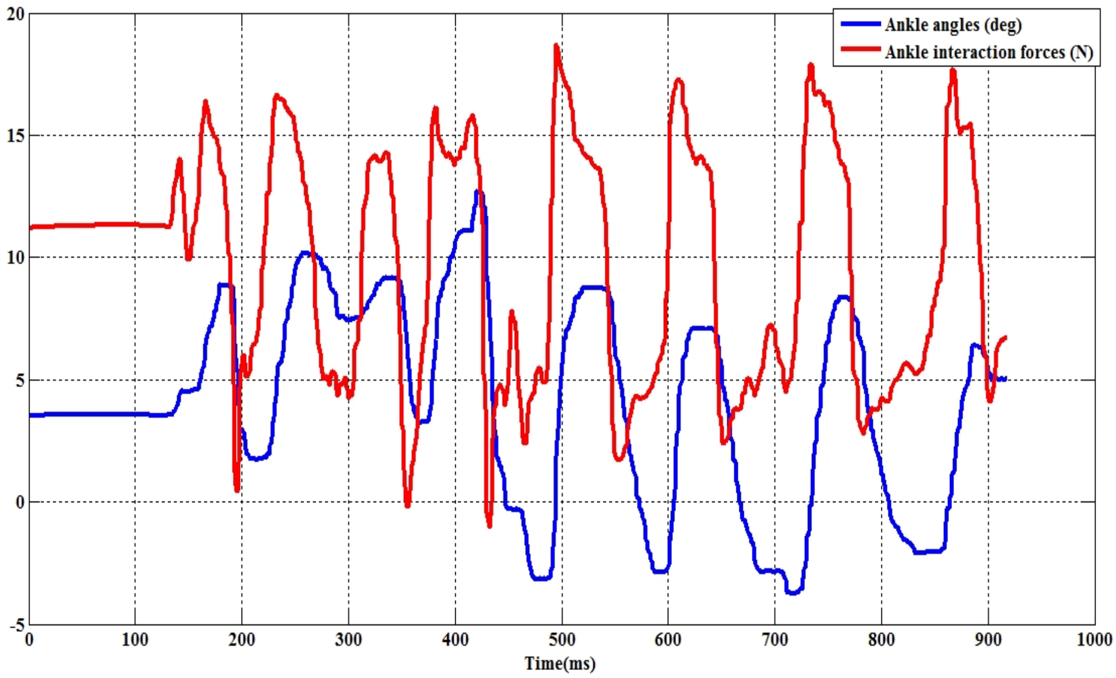

FIGURE 5.6: Ankle strategy: Angles (deg) and interaction forces (N) of the ankle joint in the presence of internal perturbations

individual and session. The interaction forces in the hip joint are found to be lower than those measured in the ankle joint. This difference in magnitude can be due to the effect of the body kinematics, as explained in Chapter 4, since the ankle joint applies higher forces in order to maintain posture, as shown in fig 5.6. The angles of the ankle joint are quite limited in movement, contrary to what happens in the hip joint, where angles have a wider range of freedom.

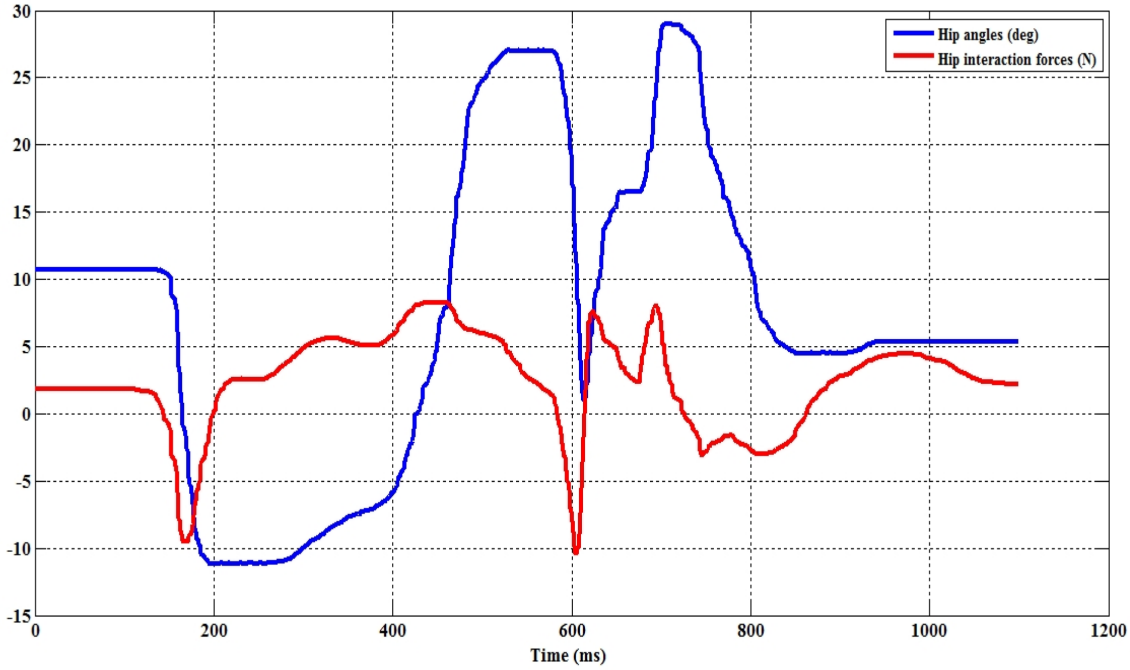

FIGURE 5.7: Hip strategy: Angles (deg) and interaction forces (N) of the hip joint in the presence of internal perturbations

The combined effect of both the hip and ankle joints acting in the presence of internal perturbations proves to be an efficient method to achieve postural stability. The interaction forces of each joint show that their movements complement the negative aspects of individual ankle and hip strategies, and this helps to maintain postural stability. In fig 5.8, the hip joint takes over the control when the ankle joint is approaching its upper limit (i.e. approx. 300ms). This can be seen by the variation of the hip angles and by the shift of the interaction forces.

With reference to the external perturbations, since the subjects did not know the instant at which the impact torques were applied, they produced an unstable movement. The exoskeleton detects the loss of stability and reacts when necessary. The squared region in figures 5.9, 5.10 and 5.11 indicates the onset of torque application on the joint and the

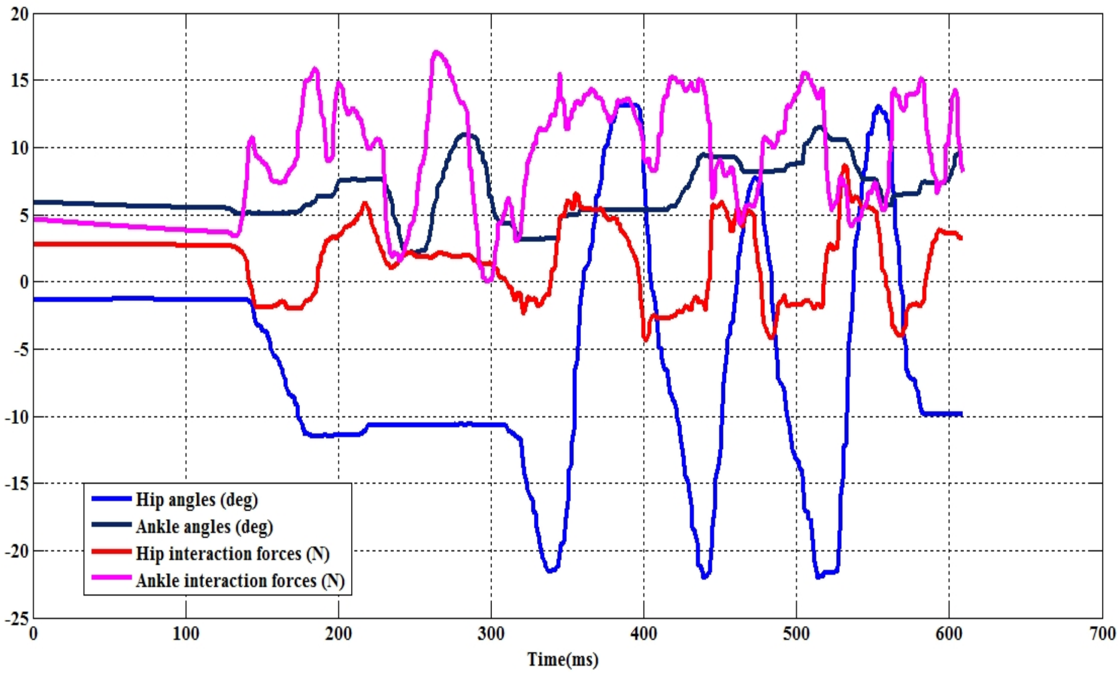

FIGURE 5.8: Combined strategy: Angles (deg) and interaction forces (N) of the hip and the ankle joints in presence of internal perturbations. The flexion movement of the hip joint is observed when the ankle joint reaches its limit (approx. 250ms) followed by the limited movement in the ankle

recovery time period. The recovery time for each joint varies depending on the perturbation and also on the subject's movement. In order to explain the context of recovery period and onset of the applied perturbations, one such trial of a subject is presented in which the perturbation was applied. The trajectory of the ankle strategy, in fig 5.9, demonstrates an unstable movement (1100-1400ms) which is a response to the perturbation (5Nm) applied at time 1100ms. The recovery stage appears after 200ms which prevents the subject from losing stability by increasing stiffness. As explained in the previous sections, the ankle joint is suitable only for small perturbations and this is evident from the big oscillations resulting from the ankle joint trajectories. The measured interaction forces are in a similar range for both the internal and external perturbation analysis, which indicates the response of the ankle joint for the applied impact torque. Fig 5.10 shows the hip trajectory movement in the presence of the applied external perturbation (10Nm) at instant 3900ms. The reactive response of the exoskeleton prevented the subject from reaching their stability limits and the recovery time for this subject was found to be 125ms. The average recovery time in the ankle strategy was found to be 250ms ($\sigma = 6.2$), whereas the hip joint was able to stabilize itself in a shorter time, 150ms ($\sigma = 4.08$).

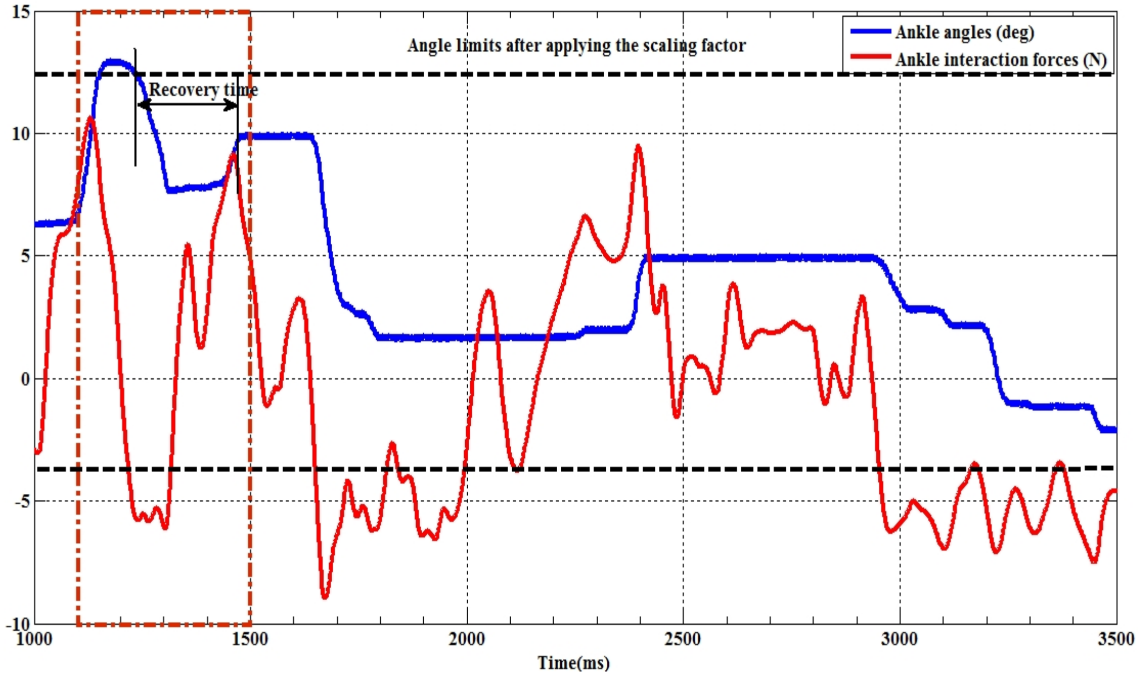

FIGURE 5.9: Ankle strategy: Angles (deg) and interaction forces (N) of the ankle joint in the presence of an external perturbation. An impact torque of 5Nm is applied in the ankle joint at 100ms and the exoskeleton assists the subject by not surpassing the limit.

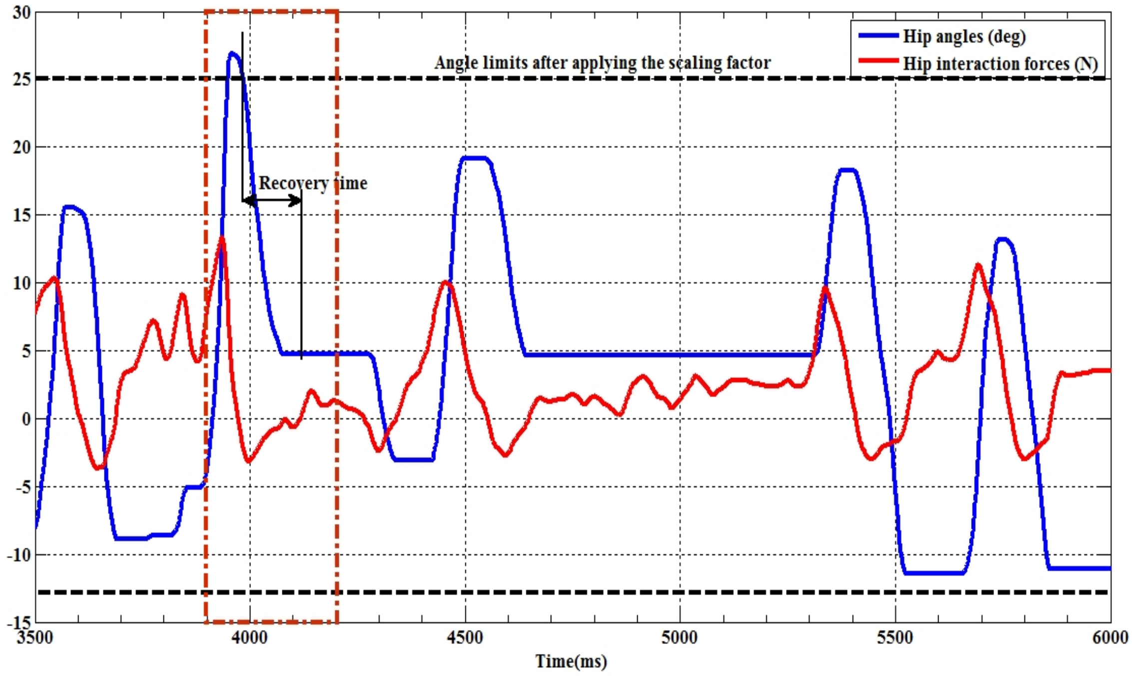

FIGURE 5.10: Hip strategy: Angles (deg) and interaction forces (N) of the hip joint in the presence of an external perturbation. At 3900ms, an impact torque of 10Nm is applied in the hip joint. The recovery time for this subject is found to be 125ms, indicating the onset of stabilization.

In the combined strategy, the hip and ankle joints are perturbed at the same time (400ms). Since both hip and ankle are in action, the combined effect helps to recover the postural stability. In fig 5.11, the behaviour of the interaction forces at 500ms shows big oscillations in the ankle joint due to the impact torque, whereas the hip joint performs with limited oscillations. After the series of experiments performed, the recovery time is found to be approx. 200ms ($\sigma = 10.6$) when using a combined strategy , thus being able to ensure a better postural stability even in the presence of combined perturbation (hip-10Nm; ankle-5Nm). The exoskeleton is able to regulate the stiffness of the joints in accordance with the learned stability limits. From the joint limits attained for each strategy, as shown in Table 5.2, it is evident that the proposed joint stiffness control of the orthosis prevents the patient from surpassing the stability limits (shown in Table 1), thus ensuring postural stability. In comparison with Table 5.1, it is noticeable that the interaction forces obtained from the combined strategy are in a similar range. This ensures that no extra efforts are required from the human to maintain postural stability. This comparison demonstrates the exoskeleton's adaptation to the movement and it's assistive behaviour.

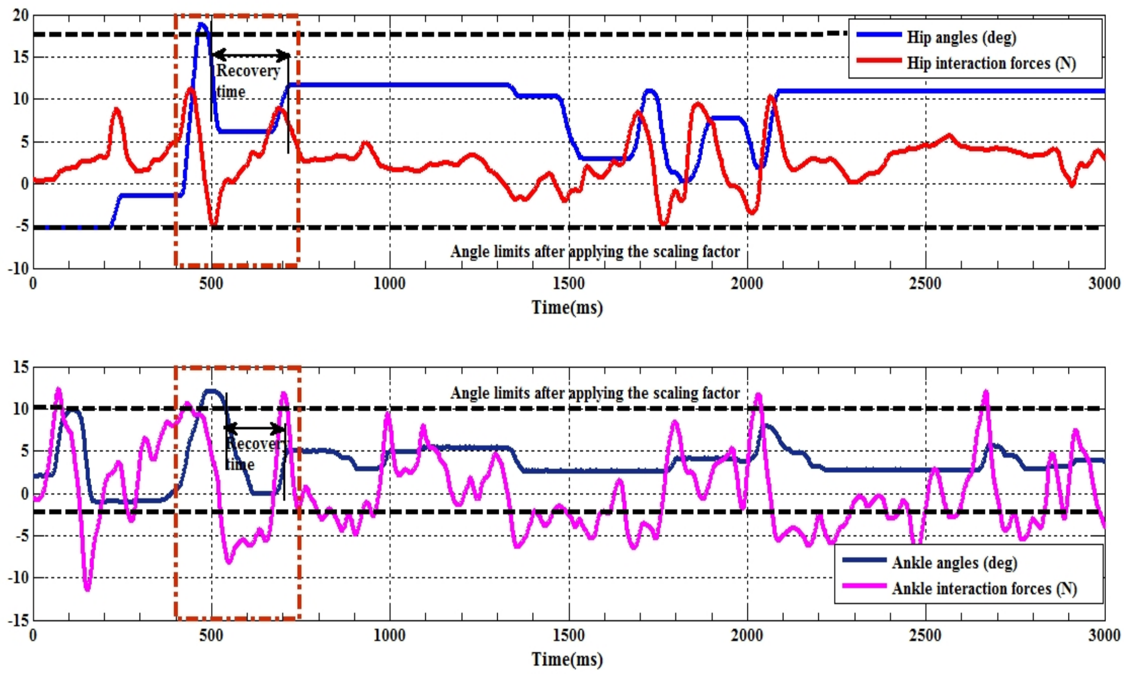

FIGURE 5.11: Combined strategy: Angles (deg) and interaction forces (N) of the hip and ankle joints in the presence of an external perturbation during the same experiment. At 400ms, an impact torque is applied on both the hip (10Nm) and ankle (5Nm) joints. The flexion/extension movement of the hip joint in combination with the ankle flexion contributes to the stabilization at 700ms.

Table 5.2: Position and interaction forces obtained after the action of external perturbation

Strategy	Joint angles (θ_{act} deg)	Interaction Forces ($F_{int}(N)$)
Hip	-15 and +25	-5 and +10
Ankle	0 and +15	-10 and +10
Combined	Hip -5 and +20 ;	Hip -5 and +10;
	Ankle 0 and +10	Ankle -10 and +15

5.2.2.3 Discussion

Maintaining the postural stability is necessary to perform normal activities for daily living (ADL's) and also to progress towards the dynamic stability. Postural stability is a major concern for aging people and for patients with neurological disorders [13]. In rehabilitation, balance training is used as a preliminary step towards the stance-phase stability in gait training [220, 241].

The individual joint action of hip-ankle joint helps in avoiding the effect of disturbances and to maintain the stability, in sagittal plane [242]. In some cases, the joint action is in function of the intensity of the applied force or disturbances. The combination of ankle-hip accelerations in flexion/extension movement reduces and signifies that it requires less neural effort [15]. Maintaining the upright-position with fixed knee joint is not equivalent of inactive knee muscles or joint movement. Indeed, there is an active participation of knee joint which helps in maintaining the posture.

In studies with humans [13, 15, 221], the postural stability was analysed by applying disturbance to accelerate the body away from the upright position, requiring muscles to limit the motion. A similar study is performed in this task, with the exoskeleton arresting or limiting the motion of the body. The exoskeleton helps in maintaining the body sway and to prevent the subject from falling. The subjects must be capable of getting back to the comfortable or stable position, with the support of the exoskeleton. The tracking of the COM helps in preventing the loss of stability by restricting or incrementing the joint stiffness. The joint stiffness incrementation occurs when the user approaches towards their own limits of stability (5.1). Such a feed-forward assistance model has been widely appreciated for its efficiency in preventing the loss of stability [123, 222].

5.2.3 Gait Assistance

The proposed intention driven adaptive strategy is based on the position error and interaction torques. The strategy needs an initial study about the user adaptation to determine the gait initiation and assistance scenario. Hence, the experimentation consists of two phases: initialization and execution. The initialization phase involves monitoring the interaction torques and joint positions with no-assistance provided by the orthosis in order to be able to define the confidence factor s parameter. This initialization phase is used to parameterize the user intentions and adaptations to the movement. In the execution phase, the changes in movement and interaction torques are used as a trigger for gait initiation. The interaction torques, limited by the s parameter (confidence factor), are used to determine the time instants of stiffness variation. Both these phases are performed and evaluated using a lower-limb exoskeleton.

5.2.3.1 Protocol

Initially the walking pattern is tested on subjects applying a low stiffness value ($20Nm/deg$), which is used to obtain the pattern of interaction forces and to adapt themselves to the orthosis. The evolution of these interaction forces are used to determine the initiation of stiffness variation, determining the adaptive behaviour to be exerted by changing the stiffness. This variable stiffness results in either an assistive or resistive behavior, based on the movement deviation. The joint stiffness values vary according to the position error and the trend of the interaction forces produced. The confidence factor is used to determine the time instants of actuating the stiffness function of the joint gradually. This assists in achieving a smooth performance without affecting the joint trajectory. The gradual increase or decrease of the stiffness value indicates the difference in interaction torques. Fig 5.12 illustrates the joint error limits, as explained in the section 4.4. There is an undefined error limit, above the upper error threshold, which is the maximum permissible limit with the incrementation of the stiffness.

The setup includes a recorded gait pattern performed by healthy users and optimized after some repetitions of gait cycles. The values of initial stiffness and confidence factor are defined based on the subject's health condition. Table 5.3 lists the initial parameters used in this experimental procedure. Since the strategy is tested with healthy individuals, the initial stiffness and confidence factor (s) are assumed to be $50Nm/deg$ and 0.9 respectively.

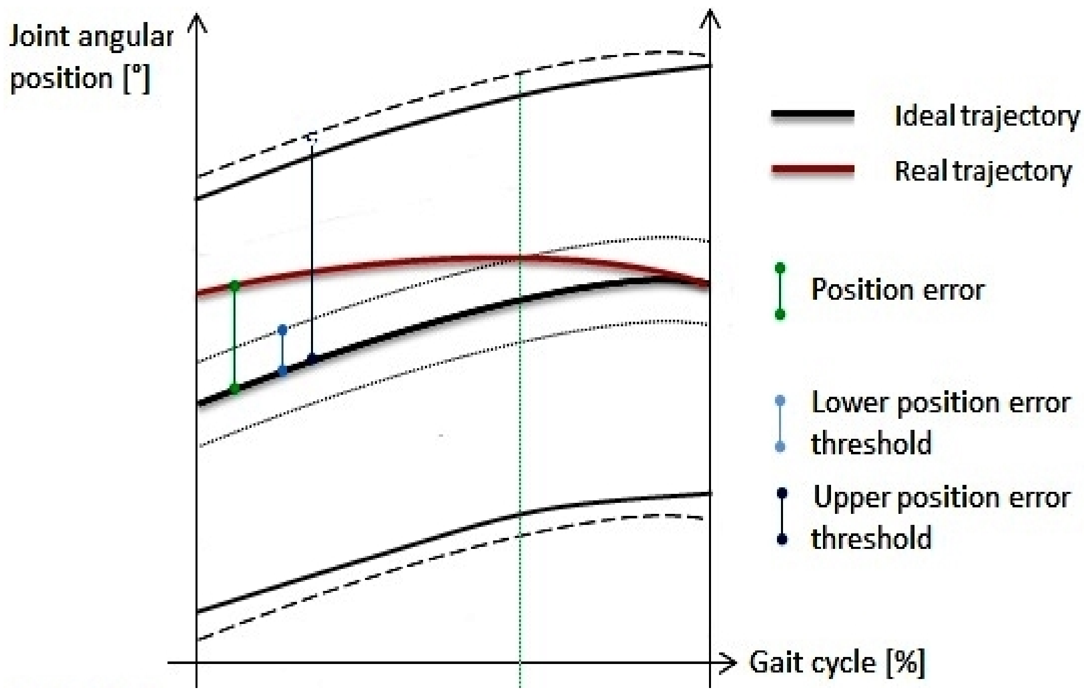

FIGURE 5.12: Error limits for evaluating the stiffness adaptation; The upper and lower error threshold limits are predefined joint limits. The dotted line indicates the higher error limit to evaluate the stiffness increment function.

TABLE 5.3: Initial parameters for the trials with healthy subjects

Gait velocity	5sec/cycle
Initial Stiffness value	50 Nm/deg
Maximum position error	7 deg
Upper position error	5 deg
Lower position error	1 deg
Stiffness decrement	1 Nm/deg
Confidence factor	0.9

High interaction forces are found in healthy subjects, so a higher confidence factor is needed to define their thresholds. This study is performed as a preliminary evaluation of the control strategy, prior to clinical trials with patients suffering from SCI. Thus an intermediate pause must be assumed in between the trials to ensure the active participation of the user, comfort in walking and to avoid muscle fatigue. In case of SCI patients, therapy procedures consider a pause of 1minute at the end of every 2-3 minutes of walking, considering their fatigue. Since this study involves healthy subjects the experimental time can be higher and the pause time can be high in a 2: 1 ratio. Hence an intermediate pause time of 2-3minutes is considered at the end of every 10minutes walking test. This walking experiment is performed for 30 minutes, i.e. 3 sets of 10 minutes walking test. . An interval of 10-20 seconds is introduced at the beginning of each trial, for gait initiation algorithm, with an auditory cue to notify the subject to initiate the movement.

5.2.3.2 Results

The proposed adaptive control strategy using the human-orthosis interaction forces has been tested and evaluated with five healthy individuals of the age group 30 ± 5, weight $80 \pm 8kg$, and height $1.75 \pm 0.05m$.

Gait initiation

In a gait cycle, the knee joint plays a key role for both the initialization of the movement and the swing state. Hence, the gait initiation strategy is evaluated by monitoring the deviation in the knee joint movement with respect to the expected pattern, along with the interaction torques. The flexion and extension movement of the knee joint is monitored to differentiate between the user's intention and tremor movement. As shown in fig 5.13, the right leg of the user showed gait movement initiation in most of the trials. This can be seen from the shift in the interaction torques of the right and left leg, both in hip and

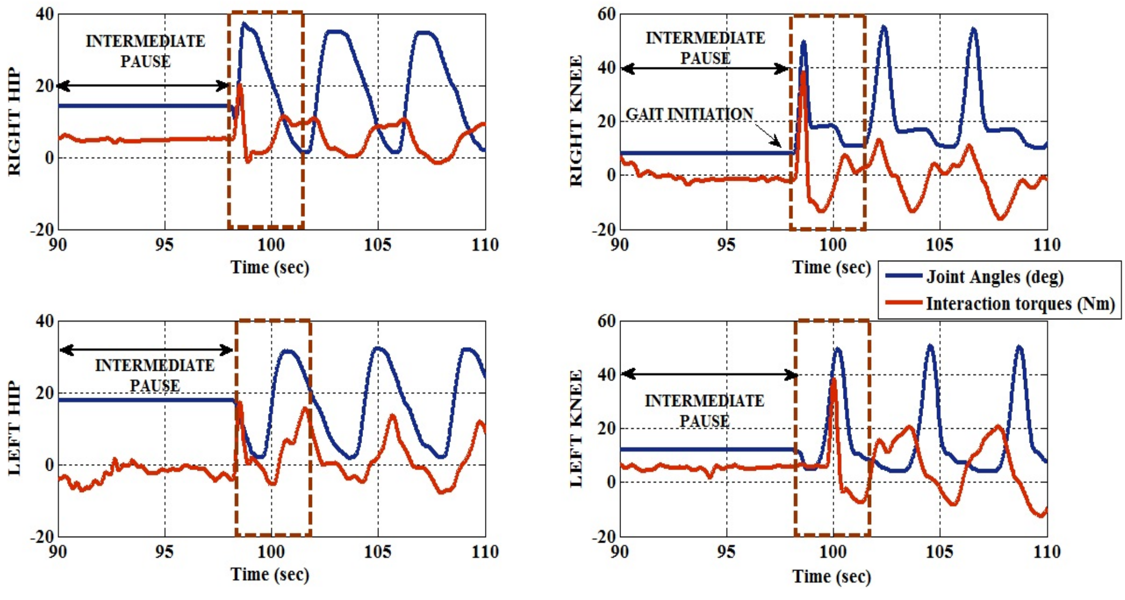

FIGURE 5.13: Mechanical interaction based gait initiation with a healthy subject; Detection of movement intention by observing the change in the knee joint angle and interaction torques

knee joint. The initiation of the gait is characterized by the flexion movement of the knee joint. For instance, at time 97 seconds the right knee joint shows a little displacement in the movement which initiates the gait cycle. The hip joint trajectory appears after a few seconds, immediately followed by the transition to the left leg.

Gait Assistance

The reference gait pattern and the resulting mean gait cycles of the subjects are presented in figures 5.14 and 5.15. The subjects performed a free normal walking movement with no restrictions on angle positions. The deviation from the desired trajectory was found to be high in case of the free walking. After a series of trials (10) this error decreased gradually due to the effect of the stiffness acting on the joints. The stiffness variation helped to maintain this error within a specified range and following a similar pattern of incrementing and decrementing K at every joint. This stiffness variation of the joint results in exerting an assistive or resistive behaviour, based on the direction of the movement. The results of one of the subjects are used to show the response of the control strategy. A detailed results about the performance of the healthy subjects is presented in Appendix A

The gait performance of a healthy user, as shown in figures 5.16 and 5.17, demonstrates the influence of the stiffness variation proposed in this work. The initial walking with low

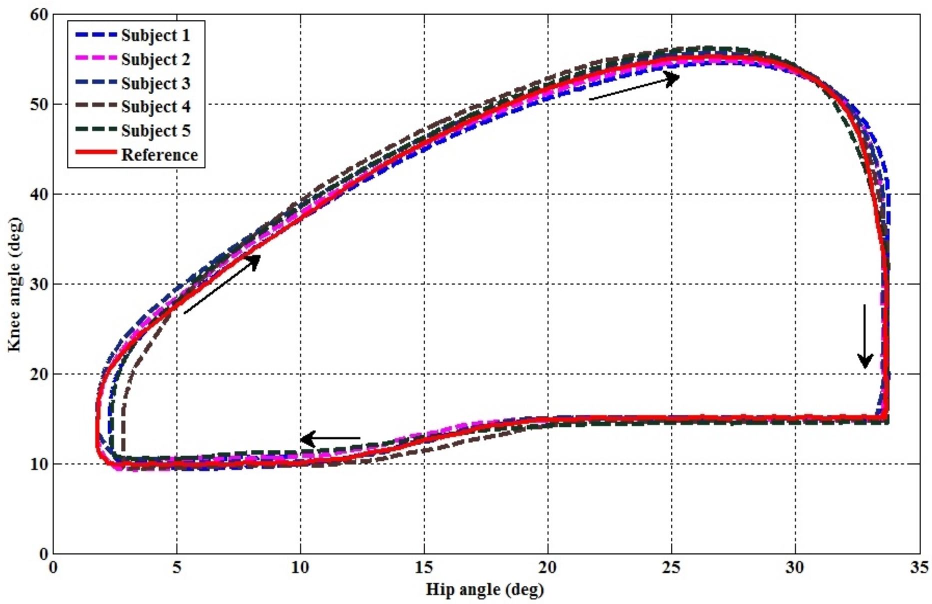

FIGURE 5.14: Reference gait pattern and the resulting mean gait pattern of each subject: Hip (deg)- Knee (deg)

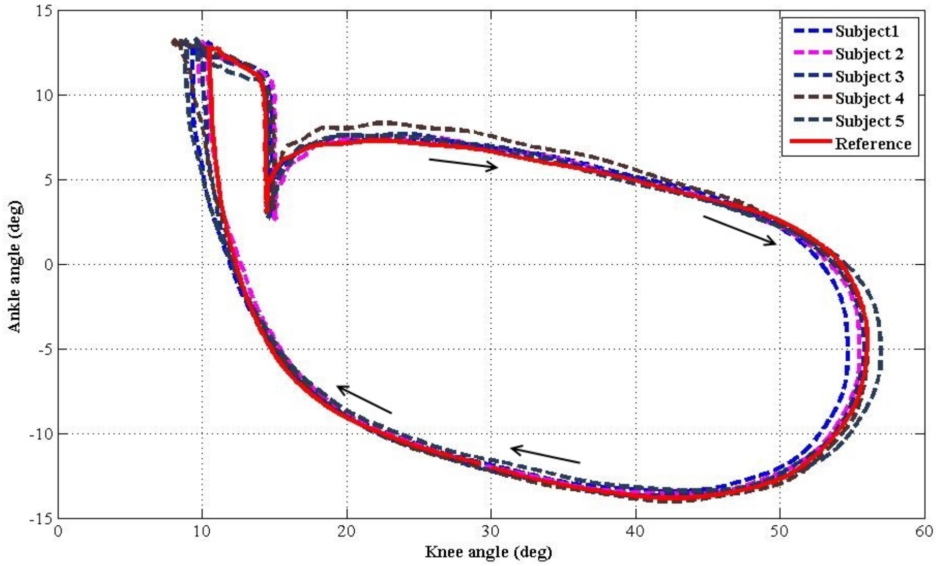

FIGURE 5.15: Reference gait pattern and the resulting mean gait pattern of each subject: Knee (deg)- Ankle (deg)

stiffness value is presented as the 'no-assistance' mode. In comparison with the reference pattern, the no assisted walking is found to produce a maximum deviation. After the application of a variable stiffness, the user is able to walk within a predefined error limits. The stiffness variation also converges with respect to the movement at the end of 10 trials. At the end of 20 trials, the user is following a movement which is quite similar to the reference pattern.

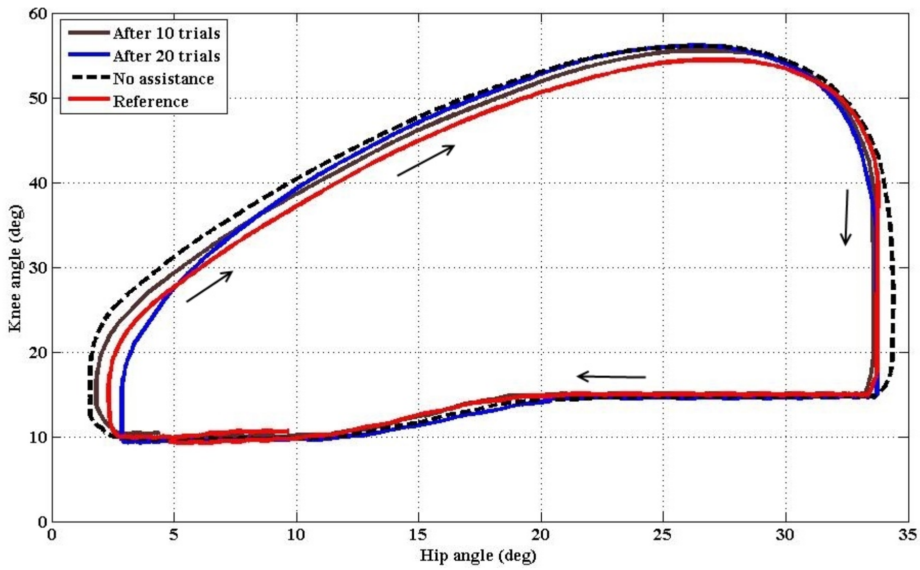

FIGURE 5.16: Changes in gait pattern of healthy subject S1, due to the effect of stiffness: Hip (deg)- Knee (deg)

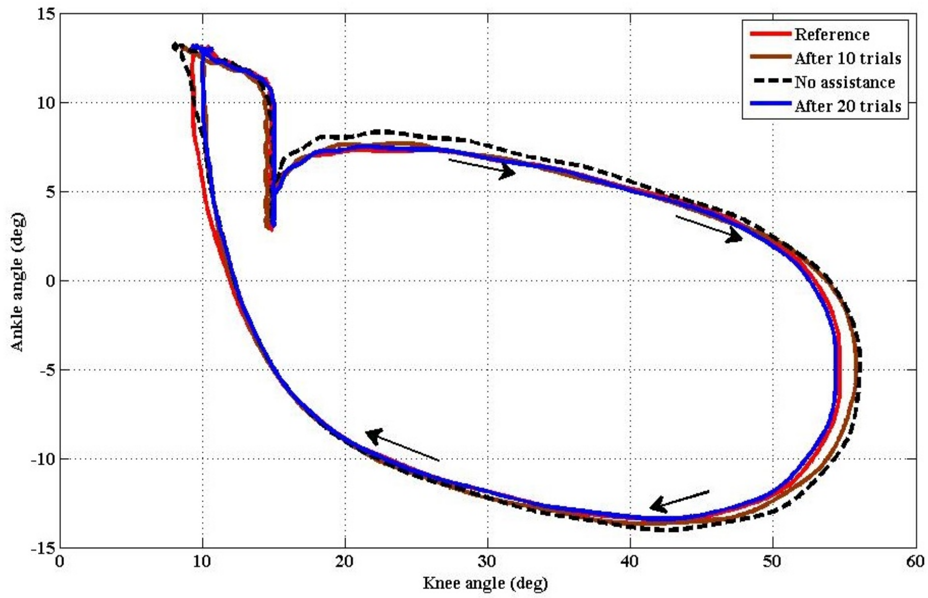

FIGURE 5.17: Changes in gait pattern of healthy subject S1 due to the effect of stiffness: Knee (deg) - Ankle (deg)

The hip joint showed a little variation and more adaptable behavior in terms of stiffness changes in real time. Since the exoskeleton is a planar robot, the lateral hip movement

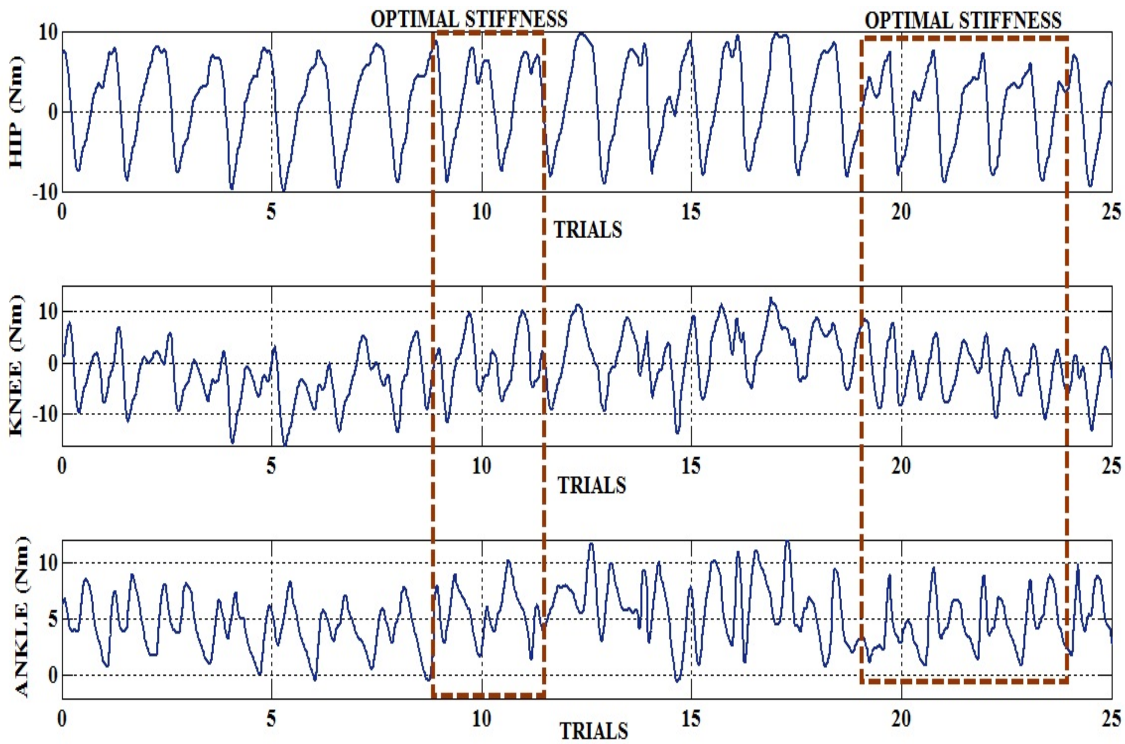

FIGURE 5.18: Interaction torques of each joint showing the change in behavior while stiffness converges to an optimal value

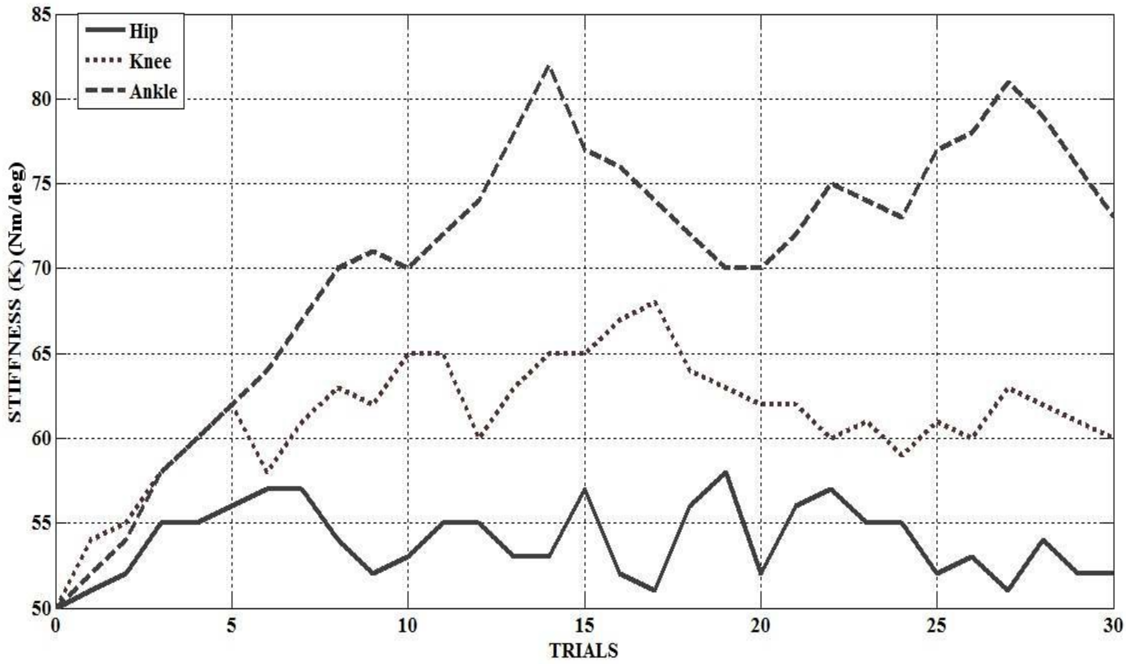

FIGURE 5.19: Averaged stiffness variation in the joint trajectory for healthy subject S1

cannot be monitored. However, this orthosis limitation does not affect the proposed control strategy. A significant variation of the stiffness is found in both ankle and knee joints. The hip joint stiffness varied in a short range which is evident from the interaction forces in fig 5.18. This can be due to the lateral movement of the user's hip joint which compensates the joint trajectory. The interaction force of the ankle joint is in the limits of 12 Nm to -3 Nm, as shown in fig 5.18, and with the application of confidence factor the threshold is limited to 10.8 Nm to -2.7 Nm. This threshold limit is used to initiate the stiffness increment when the position error threshold is reached. Similarly in the knee joint, the interaction forces are in the limit of 14 Nm to -14 Nm and after the application of the confidence factor the threshold is limited between 12 Nm and -12 Nm. The interaction forces are bounded within the limits even in the presence of maximum stiffness.

The knee joint's flexion and extension movement plays an essential role in walking by maintaining the time instants in the transition between gait phases. Thus the stiffness variation for the knee joint was observed to converge, because of the repetitive movements, after a few gait trials, as shown in fig 5.19. Fig 5.20 shows that in knee joint, the trajectory deviation is maintained within a small range, but with a delay in the movement.

In case of the ankle joint the stiffness behavior was observed to follow a different pattern, as shown in fig 5.19. This stiffness behavior is due to the compensation of ground reaction forces acting on the body. The stiffness value changes with respect to the trajectory deviation and the trend of the interaction forces. In the ankle joint, the deviation from the reference position is found to be higher, which explains the pattern of stiffness variation.

5.2.3.3 Discussion

The trials with healthy subjects are needed to understand and evaluate the efficiency of the system, before testing with paraplegic patients. For healthy subjects, The effectiveness of the control model has to be evaluated based on the energy performance and assistance in gait sequence, which can be related to effort applied by the user[243]. In terms of the proposed gait assistance, the stiffness adaptation of the control model helps in maximising the users' gait and reduces their effort. The results of the healthy subjects are evaluated by comparing with an optimized reference trajectory, obtained as an average from 5 healthy subjects.

Gait analysis of healthy users are not similar in many cases depending on the morphological parameters of each individual. The subjects' weight has an impact in the amount of joint torque applied and the height in relation to the flexion, step length involved. The step length and the flexion angles have a correlation which can be again related to the walking

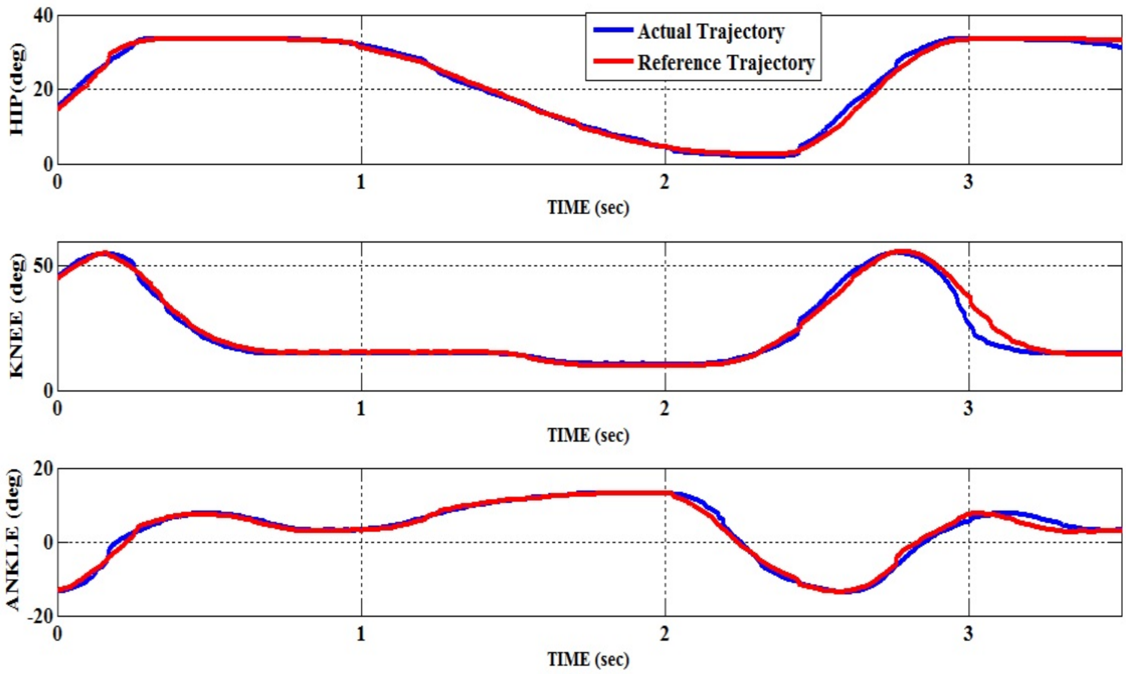

FIGURE 5.20: Trajectory deviation after the application of the variable stiffness model

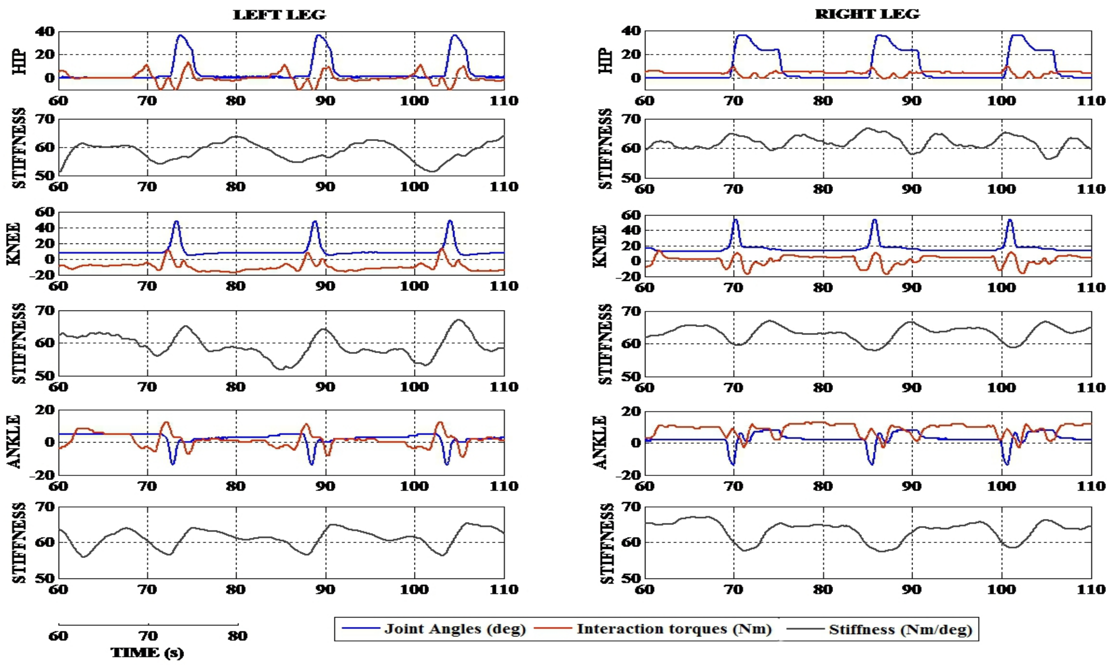

FIGURE 5.21: Stiffness adaptation in all the joints for healthy subject S1 during the last series of gait trials

speed of the user. The reduction in energy performance in human walking was studied using an elastic device with passive stiffness control [161]. A reduction in metabolic cost of walking was observed, indicating the energy efficiency of such a system in healthy subjects. But there is no evidence of a similar approach for all the joints. A similar study with the application of assistive torques on the joints for the gait transfer and control was studied using a soft exosuit [244]. The exosuit is capable of generating moments upto 18% and 30% of those generated by the ankle and hip joints, while walking. From fig 5.18, the interaction torque behaviour in gait training of healthy subjects can be elucidated. The high interaction torques in the ankle joint are observed as a result of the joint moments and the weight of the user acting on the exoskeleton. There is a reduction in the interaction torque in ankle joint while the hip joint continues with a similar pattern throughout the training. In terms of active joint support, the interaction torques are observed to be less which signifies the optimal action and support of the stiffness parameter. A higher stiffness value would increase the interaction torque forcing them to pursue a the reference trajectory, similar to trajectory tracking.

The studies with healthy subjects performed on a treadmill based robot [160] has demonstrated that a low additional torques is needed to enable the backdrivability feature of the robot and during free overground walking. The relation between the assistive torques and stiffness is not linear due to the adaptation of the user performance. It was also observed that in case of the healthy subjects only 12% of the peak torques are needed from the actuators and the remaining is achieved through physical assistance. The confidence factor is used to parametrize the joint torque to be applied, with high confidence value meaning less peak torques. in case of healthy subjects, the higher confidence value resulted in the minimal peak torques which also explains the participation of the user and their effort in gait.

The hip joint actuation reduces the input torque from the user which significantly reduces the necessary muscle activations. Lenzi et al. [120] performed a study on powered orthosis using ALEX-II with healthy subjects. In this work, the input torque is limited by the stiffness parameter and the variation is modified based on the confidence factor, which acts on the joint stiffness gradually. The consequence is the relax intervals that appear as negative slope (decreasing stiffness), which results in achieving a smooth behavior of the system. Lower confidence factors will result in few and shorter steps of stiffness variation, so the increment will be faster. On the contrary, a higher confidence factor will limit the increase of stiffness. The gradual increase in the stiffness value is due to the permanent difference in position error. The error in position of the joint in combination with the change in interaction forces results in a higher stiffness value. Since this study is performed as a preliminary evaluation, the gait speed was maintained at a constant rate to ensure the time span of the gait cycle and the duration of the gait phases: swing and stance. Further, this gait training provide the necessary cycle duration time and intermediate pause between the gait steps to ensure that the user is accustomed to the movement.

5.3 Assessment with incomplete SCI patients

In this section, the performance of the gait adaptation strategy, presented in the previous chapter is evaluated with incomplete SCI patients. Since gait training is one of the widely appreciated and accredited task in a lower limb rehabilitation, it was directly performed with the patients. The gait initiation was handled by the patients' volitional orders by monitoring the motor neural activity, which provides a top-down approach in rehabilitation. The movement initiation command from the BMI system through the motor related activity of the patient initiates therapy.

The gait adaptation strategy presented in the section 4.4, is evaluated by comparing the evolution of the user performance against the application of constant stiffness values. An adaptive assistance in the joint should assist and motivate the patient to pursue their maximum flexion/extension movement in the joint which also results in a change in behaviour of the interaction torques. Hence, performance in this work is evaluated by monitoring the maximum joint angles and interaction torques. An optimal performance is assumed to produce maximum movement range (flexion/extension) and an increase of the interaction torques in the knee joint. The trajectories of the patients are also recorded to analyse the pattern and freedom of the joint movement. The stiffness variation is also monitored to demonstrate the adaptability of the patient with the orthosis. High values of the stiffness are expected for joints that need assistance and low values for the other joints.

The inclusion criteria for the patients included:

1. SCI with any level of lesion, ASIA C or D with gait prognosis;
2. Patients at the early stages of walking rehabilitation;
3. Patient's can maintain balance by standing between parallel bars;
4. No orthostatic complications during standing;
5. Normal range of motion in upper and lower-limb joints;
6. Upper-limb strength to manage a walker or crutches, and to transfer from the wheelchair to a chair;
7. Age between 18 and 60 years; and
8. Height 1.50-1.95 m, and weight up to 90 kg.

FIGURE 5.22: Experimental setup with an incomplete SCI patient wearing the Exoskeleton and BMI system

The exclusion criteria included:

1. Inability to stand in upright position for at least 15 minutes;
2. Any surgery in the previous 3 months;
3. Spasticity higher than 3 in the Modified Ashworth Scale (Bohannon and Smith [245]);
4. Previous/current lower-limb bone fracture;
5. Ulcers or sores in areas of contact with the exoskeleton and/or electrodes;
6. Previous/current history of cardiovascular disease of any kind or exercise contraindications;
7. Upper-limb pain that limits weight bearing on crutches/walker/parallel bars;
8. Significant upper/lower extremity discrepancies;

9. Uncontrolled autonomic dysreflexia;
10. Pregnancy; and
11. Cognitive impairment of any kind.

Demographic data of the four incomplete SCI patients can be seen in Table 5.4. The patients are identified as P01, P02, P03 and P04 respectively. The SCI patients were chosen from the Hospital Nacional de Parapléjicos, in Toledo (Spain), where all the experimentation sessions took place.

Table 5.4: Demographic details about the paraplegic patients; *P04 presented some zones of partial preservation below the level of injury which caused some degree of sensitivity below both knee joints.

Patient	Age	Gender	Height (m)	Weight (kg)	Level of lesion	ASIA scale	Time since lesion (months)	Etiology
P01	30	M	1.85	90	L1	C	12	Traumatic
P02	24	M	1.92	57	L1	C	24	Traumatic
P03	49	F	1.6	76	T12	C	5	Traumatic
P04	21	M	1.8	57	T11	A*	11	Traumatic

The selected patients met all the above mentioned inclusion and no exclusion criteria. All the subjects were duly informed about the study and all of them gave written informed consent before the first session. To perform the trials with incomplete SCI individuals,it is necessary to have a well defined clinical protocol such as to ensure an effective and less tiring therapeutic approach. In this section, the clinical protocol defined as a part of the study is explained in detail. The protocol and the experimental study has been approved by the Ethical Committee of the Hospital Complex of Toledo (Spain) (C.E.I.C. 31/02-2014).

5.3.1 Clinical Protocol

The clinical protocol was defined based on the abilities of the patient and their conditions. The subjects performed an initial walking trial with a fixed stiffness, to analyse the necessary walking assistance and to ensure that the patients get accustomed to the orthosis. An optimized gait pattern, obtained from the average of five healthy subjects, was used as a reference in the fixed stiffness mode. In the following sessions, the patients were assisted based on the adaptive stiffness variation with respect to their interaction. In this case, the reference trajectory pattern is used only when an abnormal movement is detected. In other cases, the exoskeleton adapts to the patients' pattern and varies the stiffness based on the interaction torques. Hence, the walking experiment was performed in 3 sessions, performed in 3 consecutive days: Fixed stiffness (first session) and variable stiffness (next 2 sessions). Each session lasted for a time span of 60-90 minutes, which includes the time for wearing the exoskeleton and the BMI calibration. Since the patients were not able to maintain their equilibrium for a long period of time, the joints of the exoskeleton were maintained rigid ($90N/m$ stiffness) during the BMI calibration. The fixed stiffness values were chosen heuristically based on the studies performed with healthy subjects. The initial stiffness, error threshold and the confidence factor for all the patients and joints, for the adaptive stiffness study, were assumed to be $80N/m$, $\pm 5\,\text{deg}$ and 0.6 respectively, considering the patients' health condition. We imposed equal initial values, for all the joints and all the patients, to evaluate the adaptability of the algorithm independent of patients' specific health condition.

The experimental protocol consisted of familiarization sessions and BMI sessions. The familiarization and the BMI session were performed on one day, while the SCI patients performed one familiarization session and two BMI sessions on three separate days [238]. The setup included: the EEG equipment (only for the BMI sessions), with 2 the amplifiers in a backpack carried by the subject; the exoskeleton attached to the subject's legs; and a walking aid to help keeping balance. As walking aids, we evaluated the use of crutches, a walker and parallel bars. Crutches were discarded as they did not provide enough balance control for the patients. The walker was used by the healthy subjects, but was also discarded for the patients, since they had difficulties to move it while walking with the exoskeleton. Hence, all the SCI patients performed the BMI sessions using parallel bars.

The familiarization sessions were performed in order to allow the subjects getting used to the protocol timings and the exoskeleton movements. On these sessions, one experimenter commanded the exoskeleton by manual triggers, warning the subject before every movement. For the healthy subjects, these sessions consisted of 5 to 10 minutes walking with the exoskeleton. For the patients, the sessions took between 20 and 30 minutes, in which several clinicians monitored every movement and taught the patient how to interact with the exoskeleton. If required, these sessions were repeated until both the patient and the clinicians confirmed that the patient was accustomed to the system, and ready for the first BMI session.

The BMI sessions consisted of screening blocks and closed-loop feedback blocks. The participants performed 3 or 4 screening blocks, each of which included 20 trials, which were used to calibrate the BMI decoder. During these blocks, the participants were standing, wearing the exoskeleton with the joints blocked, and holding the corresponding walking aid (i.e., the walker for the healthy subjects, and the parallel bars for the patients). Hence, the exoskeleton sustained the subjects' weight, and impeded movements on their legs. Given that neither healthy subjects nor the patients could actually move the legs with the joint blocked, in both cases we consider the action as a movement attempt. The screening blocks were composed of rest and movement attempt (MA) intervals. The rest intervals had a random duration between 4 an 7 seconds. Subsequently, an audio cue indicated the start of the MA interval, which lasted 3 seconds. The participants were instructed to attempt to move their right leg, as if they started walking, when they heard the audio cue. The rest of the time, they were asked to stay relaxed and minimize movements.

The closed-loop feedback blocks were composed of trials with four intervals: (i) ”Rest”, (ii) ”Preparation”, (iii) ”Movement Attempt”, and (iv) ”Movement”. The healthy subjects performed 3 blocks of 20 trials each (totaling 60 trials), while the SCI patients performed a variable number of trials, until reaching a distance of 10 meters (i.e., the length of the parallel bars), which corresponds to around 25 gait cycles. During ”Rest” state (5 seconds), the subjects were not required to perform any task, but just to relax after the previous trial. Subsequently, a low tone sounded and indicated them the beginning of the ”Preparation” interval (3 seconds), during which they were instructed to relax and be prepared for the upcoming cue. A high tone denoted the start of the ”Movement Attempt” interval (3

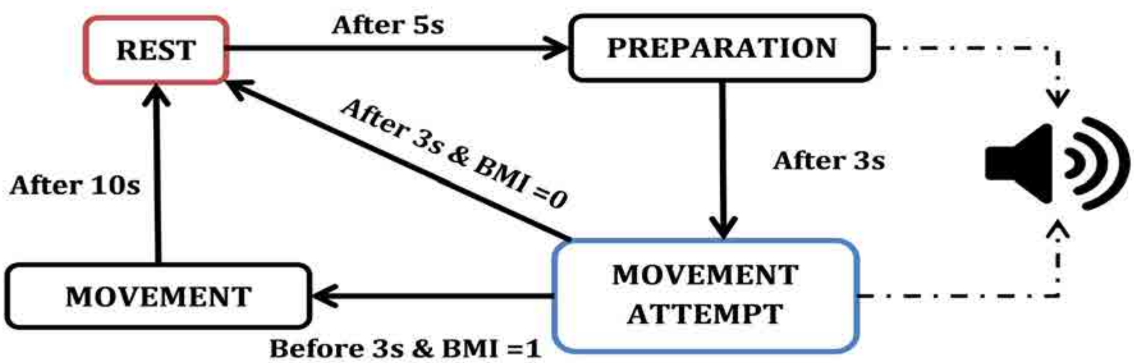

Figure 5.23: Schematic representation of the FSM-based clinical protocol used for incomplete SCI paients; The audio signal is indicated in the preparation and Movement attempt state to alert the patient.

seconds), in which they were asked to attempt to move their right leg in the same way they had done in the screening blocks. If the BMI detected the intention to move during these 3 seconds, the system started the "Movement" interval, in which the exoskeleton walked for one gait cycle: one step with right leg and one with left leg (6 seconds); otherwise, a new trial started in rest state 3 (as shown in fig 5.23).

For safety reasons, every trial required that the experimenter explicitly pressed an activation button during the "Rest" or "Preparation" intervals. If that button was pressed, the BMI decoder started sending its outputs to the exoskeleton controller, which, if detected a movement command during the "Movement Attempt" interval, started the gait cycle. If the button was not pressed, the exoskeleton did not move although the participant had attempted to move his/her legs. This mechanism was added in order to avoid starting a movement with the patient being in an unsafe position after the previous gait cycle, and to skip trials to regularly ask the patients about their fatigue levels. When required, the patients could sit for a few minutes to rest, and the trials continued when they confirmed that they were ready.

5.3.2 Clinical Assessment

The performance of the exoskeleton, for SCI patients, is evaluated based on the standard clinical assessment scales. The assessment of the patients were performed both pre and post sessions, for analysing the change in their performance. All the patients, performed the trials along with their regular therapeutic exercises. The patients performed these assessments using their standard assistive tools such as walkers, crutches and other passive AFO's. The following are the standard clinical assessments performed in SCI individuals, for evaluating their performance. In this dissertation, these outcome measures are considered as an evaluation of the control model and its adaptation to the patient. The assessments can be classified as: Physical and Subjective, based on their significance and the kind of measurements used.

5.3.2.1 Physical Assessment

The physical assessment scales involve measures which use the real performance of the patient to categorise. In terms of gait analysis, the walking performance and muscle spasticity are used widely to analyse the clinical outcomes. The following are the commonly used physical assessment measures for SCI individuals,

Manual Muscle Test (MMT)

The Manual muscle Test (MMT) is a standardized assessment to measure the strength of peripheral skeletal muscle groups [246, 247], developed by the Medical Research Council(MRC). The scores are evaluated individually for each side (i.e. right and left) and are indicated as right and left muscle test respectively. The higher muscle test value indicates strength of the muscle which is further represented based on the manual tests performed.

Spinal Cord Independence Measure (SCIM III)

The Spinal cord independence measure uses 19 daily tasks, categorized in 3 subscales, to monitor the ability of the SCI individuals. The three sub-scales are categorized as (Table 5.5), Self-care - 6 tasks, Respiration and Sphincter - 4 tasks and Mobility - 9 tasks. The tasks are weighted based on their clinical impact and are graded for increasing difficulty which requires higher ability of the patient [248, 249]. Each tasks are scaled from 0 to 9 and the total SCIM ranges from 0 to 100, with higher scores representing the high level performance or independence of a person. The scores of Mobility and Sphincter management is assumed in this study, as an evaluation tool.

TABLE 5.5: Scales and score of Spinal Cord Independence Measure (SCIM); The domains respiration management and mobility are important for the assessment of the control model

Domain	Number of tasks	Score range
Self-care	6	0 - 20
Respiration and Sphincter management	4	0 - 40
Mobility	9	0 - 40

Table 5.6: WISCI II scale of functional mobility assessment of SCI patients for 10 meters walking; The scale description has been limited from 8 to 14 detailing only the levels of the patients involved in this study

Level	Description
8	Ambulates with walker, no braces and physical assistance of one person
9	Ambulates with walker, with braces and no physical assistance
10	Ambulates with one cane/crutch, with braces and physical assistance of one person
11	Ambulates with one cane/crutch, with braces and physical assistance of one person
12	Ambulates with two crutches, with braces and no physical assistance
13	Ambulates with walker, no braces and no physical assistance
14	Ambulates with one cane/crutch, no braces and physical assistance of one person

Walking Index for Spinal Cord Injury (WISCI II)

The walking index assessment is performed using an ordinal scale with 21 levels [250]. This measure represents a gradation of gait performance with respect to the need for physical assistance, braces and walking aids. The performance is represented with a gauged hierarchical scale from 0 to 20, where lower values indicating higher impairments [251, 252]. Table 5.6 explains the levels and description of the WISCI II standards for SCI individuals. The functional levels of the SCI patients involved in this study are listed in the table, for more detailed list of the revised scale please refer [250, 251].

Penn Spasm Frequency Scale (PSFS)

The spasm frequency occurrence in the SCI individuals is measured using this scale. The scale comprises of two parts (Table5.7): First is a self report measure to augment the clinical ratings of spasticity (0 to 4), Second is a 3 point scale assessing the severity of spasms (1 to 3). The second part is indicated only if the spasms are observed [253, 254].

10 Metre Walking Test (10MWT)

This test measures the gait performance of the SCI individuals, bases on the time measures of overground gait speed [255, 256]. The patients can perform this test with the help of standard assistive devices such as crutches, bars or walkers. Three trials of walking is performed and the average of three trials are considered to calculate the gait speed of the

Table 5.7: Penn Spasm Frequency Scale (PSFS) and spasm severity for SCI patients

Spasticity rating	Description
0	No spasm
1	Mild spasms induced by stimulation
2	Infrequent full spasms ocurring less than once per hour
3	Spasms ocurring more than once per hour
4	Spasms ocurring more than 10 times per hour
Spasticity severity	**Description**
1	Mild
2	Moderate
3	Severe

patient. The walking speed depends on the capability of the patient so they can walk at their preferred or maximum speed.

5.3.2.2 Subjective Assessment

The subjective assessment scale involves measuring the adaptability and comfortability of the patient in performing the task. These measures are calibrated based on a questionnaire kind of response from the user and the statistical value is indicated based on their responses. Hence, this scaling can also be deemed as the satisfaction assessment in terms of exertion and usability. The following are the commonly used subjective assessment measures for SCI individuals,

Borg Scale

Borg scale defines the level of patient perceived exertion. The scale analysis the perceptual effort of the patient, to understand their behavioural and physiological performance [257, 258]. The perceptual estimation helps in obtaining the perceptual variation using the psychophysical ratio-scaling methods. The values of the scale range from 6 to 20, as presented in the Table 5.8.

Table 5.8: Borg Scale assessment of perceived exertion; Borg scale value of 17 is considered as a limit for maximal exertion for this study

Scale	Description
6	No exertion
7	Extremely light
8	
9	Very light
10	
11	Fairly light
12	
13	Somewhat hard
14	
15	Hard
16	
17	Very hard
18	
19	Extremely hard
20	Maximal exertion

Quebec User Evaluation of Satisfaction with assistive Technology (QUEST)

Quest is mainly used as a measure of satisfaction for the evaluation of assistive technology devices. The measure can be applied on a broad range of variety of devices in a structures and standardized format. The scale involved some questions relating to the dimensions, ease-of-use, confortability and durability of the device [259]. For the evaluation of the control strategy, it is important to know the effectiveness, patient satisfaction and comfortability are evaluated. Safety and security of the device is also taken into account which gives the validation for usability of such models for gait training. Each question is ranked on a scale of 1 to 5, as presented in Table 5.9

Results on BMI calibration are detailed in [238, 240] and a preliminary diagnosis of 3 healthy subjects and 4 incomplete SCI patients are presented in Appendix C. Results about the gait adaptation is presented briefly in this section and more elaborate results are presented in the Appendix B

TABLE 5.9: Modified QUEST scale assessment used in this study]

1	Not satisfied at all
2	Not very satisfied
3	More or less satisfied
4	Satisfied
5	Very Satisfied

5.3.3 Case Study 1

The Patient P01 is a male with height 1.85m and weighs 90kg and was observed with a lesion L1, impairment scale C in ASIA. The left leg of the user was observed weaker and this inhibits from maintaining the balance while standing. Although the exoskeleton was providing assistance, the patient reported of being tired at the end of every 3-5 gait sequences. Hence the intermediate pause time was observed more often. The ERD of the user was effective at some instants and he performed a maximum of 15 gait trials at the end of each session. Figure 5.24 illustrates the gait pattern of the patient resulting due to the influence of stiffness adaptation. The stiffness variation in each joint was different and especially in case of the ankle joints the walking pattern was adapted more towards the ideal because of the stiffness adaptation.

Figure 5.25 illustrates the maximum flexion of the left knee joint observed in all the 3 sessions. Although there is a slight change in the knee joint, it was followed by a change in the hip joint movement. This variation in the hip joint is considered as a compensatory movement to increase the knee joint flexion, which also explains the adaptation performed in multiple joints at the same time. Table 5.11 illustrates the clinical assessment of the

TABLE 5.10: Results of adaptive gait training with patient P01; The knee flexion of both legs, for each session are represented in $mean \pm \sigma$ and the adaptive stiffness provided in $median \pm range$, Note: Session 1 was with a fixed stiffness value 80 Nm/deg.

Session	Left Knee flexion(deg)	Stiffness (Nm/deg)	Right Knee flexion (deg)	Stiffness (Nm/deg)
S1	49.23 ± 2.94	80	53.3 ± 0.7	80
S2	48 ± 1	83 ± 3	53.55 ± 1.5	82 ± 3
S3	49.25 ± 1	81 ± 5	54.5 ± 1.55	81 ± 2

patient, pre and post training with the exoskeleton. The patients indicated no signs of

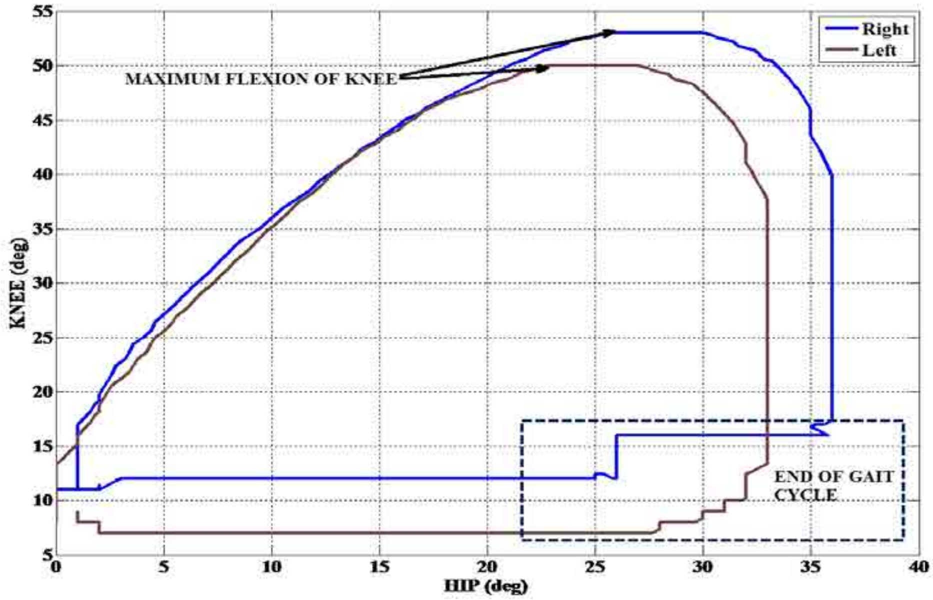

Figure 5.24: Average gait pattern of the patient P01 with variable stiffness at the end of session 3; Maximum knee flexion is found different because of the affected leg movement. The end of gait cycle indicates the onset of maintaining equilibrium position with 90N/m stiffness

spasms and the walking index level indicates that the patient can walk for 10 meters with walkers and no physical assistance.The SCIM of the patient had a significant difference before and after the gait training. In terms of the adaptive property of the exoskeleton, the two domains of the SCIM: mobility and breath sphincter management are considered in brief. The participant showed a progressive change in terms of breath sphincter as pre-29 and post-33, while in mobility it was pre- 20 post - 18. For the 10 MWT, there was an increase in the time span of completing the 10 meter walking without any therapist assistance and only with the use of walker (indicated by the WISCI scale).

Table 5.11: Clinical assessment measures of patient P01

Clinical outcomes	Pre-intervention	Post-intervention
Left muscle test	8	4
Right muscle test	12	12
PSFS	0	0
WISCI II	9	9
SCIM III	65	66
10 MWT (s)	41.53 ± 0.441	48.21 ± 1.314

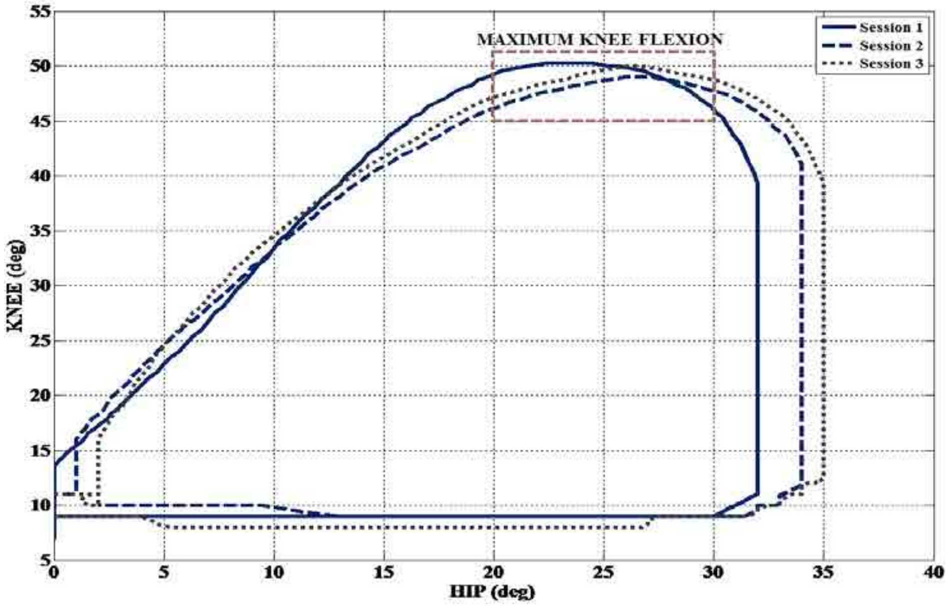

FIGURE 5.25: Maximum flexion of Left knee of the patient P01, at the end of each session; the patient applied a compensatory movement in the hip joint which helped the user to achieve a maximum knee flexion in adaptive stiffness mode.

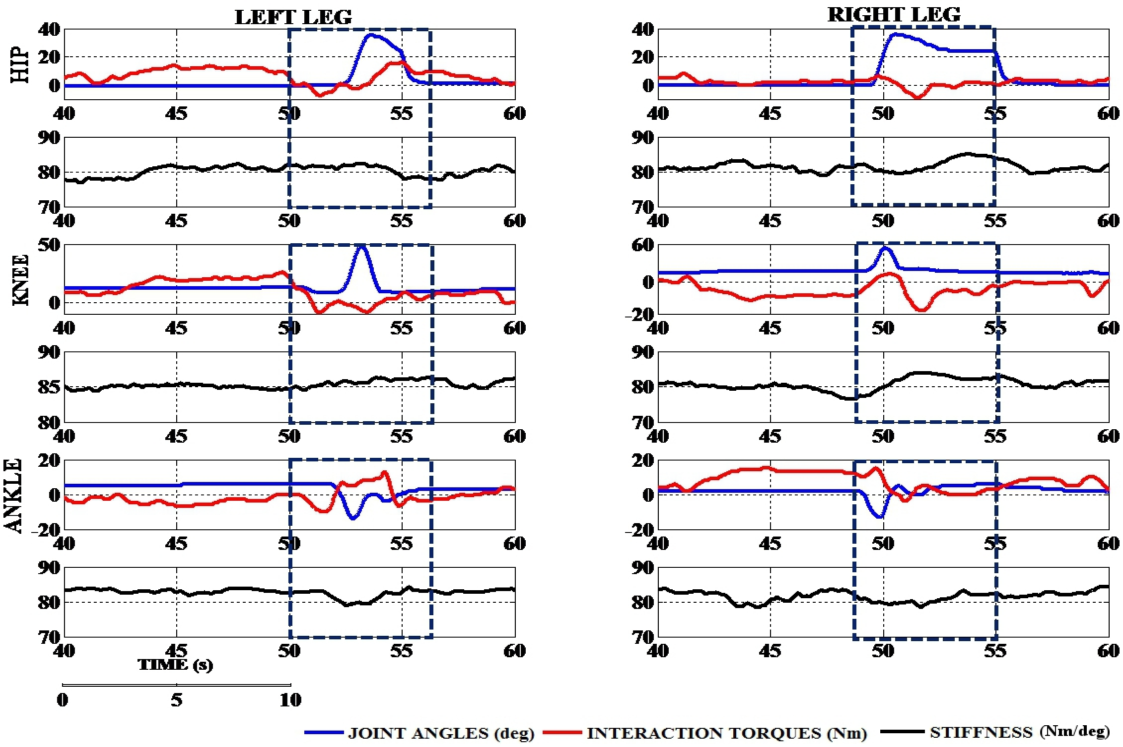

FIGURE 5.26: Joint angles, interaction torques and stiffness adaptation of patient P01; the influence of the adaptive stiffness is in function of the change in interaction torques and vice versa. The hip and ankle joint showed a minimum stiffness variation in comparison with knee joint, because of its biomechanical properties.

5.3.4 Case Study 2

Patient P02 is a male with height 1.92m, weight 57kg and has a Lesion L1 (ASIA- C). The patient uses Knee-Ankle-Foot Orthosis (KAFO) for supporting both the knee and ankle joints of left leg during the normal therapies. The left leg is considerably weaker than the right leg thus the best flexion angles are observed in the right leg, as shown in fig 5.27. Patient also indicated some special zones of sensitiveness in the hip and ankle joints of the right leg. Irrespective of all these, the patient demonstrated higher level of motivation which helped in pursuing walking for more than 20 gait cycles in the first session. In the following sessions, the patient performed comparatively less number of trials (10-15). In the fixed stiffness strategy the patient performed 25 gait cycles without any intermediate pause. In variable stiffness the patient demonstrated the similar level of efficiency and performed 10-15 gait trials. The height of the patient permitted him to pursue longer footsteps, so the stride length was also supported by the adaptive controller.

Since the left knee and ankle joints were both affected, the stiffness variation in these joints were observed to be higher compared to the hip joint's adaptation (fig 5.28). Furthermore, the joint adaptation in between the right and left leg joints are observed to be different but in synchronization with the joint movement. There is an in progressive change in the left

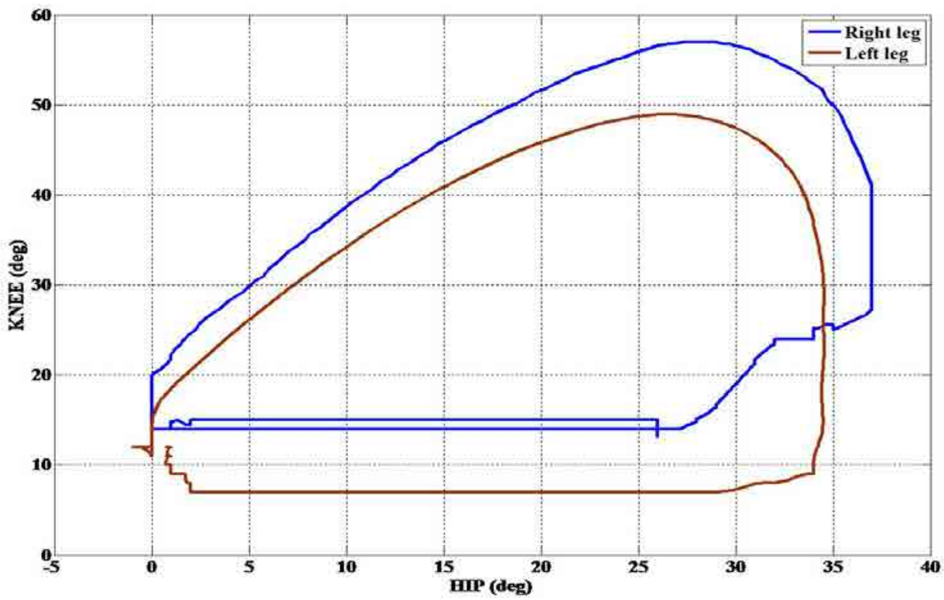

FIGURE 5.27: Average gait pattern of the patient P02 with variable stiffness in session 3; Maximum knee flexion is found different because of the affected leg movement. The end of gait cycle indicates the onset of maintaining equilibrium position with 90N/m stiffness

knee interaction torque, while the stiffness is limiting the maximum value gradually. The

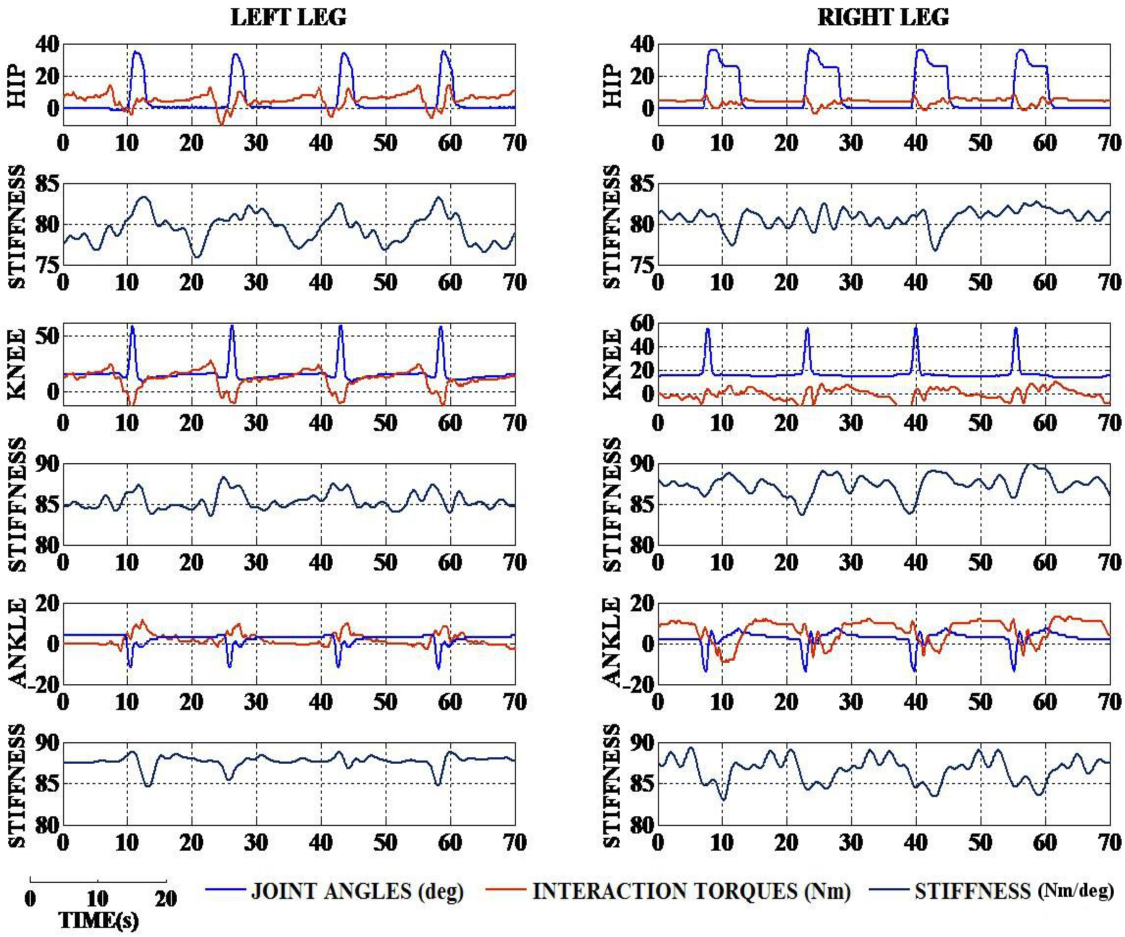

FIGURE 5.28: Joint angles, interaction torques and stiffness adaptation of patient P02; the influence of the adaptive stiffness is in function of the change in interaction torques and vice versa. The hip and ankle joint showed a minimum stiffness variation in comparison with knee joint, because of its biomechanical properties.

TABLE 5.12: Results of adaptive gait training with patient P02; The knee flexion of both legs, for each session are represented in $mean \pm \sigma$ and the adaptive stiffness provided in $median \pm range$, Note: Session 1 was with a fixed stiffness value 80 Nm/deg.

Session	Left Knee flexion(deg)	Stiffness (Nm/deg)	Right Knee flexion (deg)	Stiffness (Nm/deg)
S1	49.8 ± 2.01	80	54.72 ± 2.97	80
S2	47.9 ± 1.55	83 ± 3	56.6 ± 3.62	84 ± 4
S3	49.35 ± 1.19	81 ± 6	55.6 ± 2.63	82 ± 4

pre and post clinical assessment of the patient P02 is illustrated in the Table 5.13. The PSFS demonstrates that the patient has no signs of spasms and WISCI indicate his ability

TABLE 5.13: Clinical assessment measures of Patient P02

Clinical outcomes	Pre-intervention	Post-intervention
Left muscle test	13	14
Right muscle test	8	8
PSFS	0	0
WISCI II	12	12
SCIM III	68	70
10 MWT (s)	20.933 ± 1.026	17.643 ± 2.3

to walk for 10 meters with crutches and no physical assistance. For the SCIM, the patient had a significant difference in the mobility domain: pre - 19 and post - 21, while the breath sphincter management was maintained at the similar level (34). The gait training of 10 MWT was observed to be the least of all the users, because of the height of the patient. There was an increase in the performance of the user, which resulted in the reduced time span in completing the walking without any physical assistance.

5.3.5 Case Study 3

Patient P03 is a female of age 49yrs with height 1.60m and weight 76kgs. The spinal lesion was observed as T12 and categorized as C with ASIA scale. The time of lesion was reported as 11 months. Her right ankle was weaker and also reported some special sensitive zones around the left knee. The adaptation of the walking helped her to pursue this movement in the right ankle joint and also improved the flexion in the left leg gradually. The adopted confidence factor was 0.6 which helped the patient to deliver the maximum potential in the flexion/extension movement. In terms of ERD, this patient demonstrated more efficient motor activity in initiation of the movement and also the walking pattern was more susceptible.

Fig 5.29 illustrates the maximum knee flexion angles of both legs obtained as a result of the stiffness adaptation. The maximum adaptation is observed in a similar range 50 ± 2 for both the knee joints. In case of hip joints, the joint position is different in the initial condition, because of the postural adaptation of the patient. The clinical assessment of the patient, pre and post training (Table 5.15), illustrates her performance. The patient exhibited no signs of spasms (PSFS - 0) and the WISCI - 12 demonstrates here capability

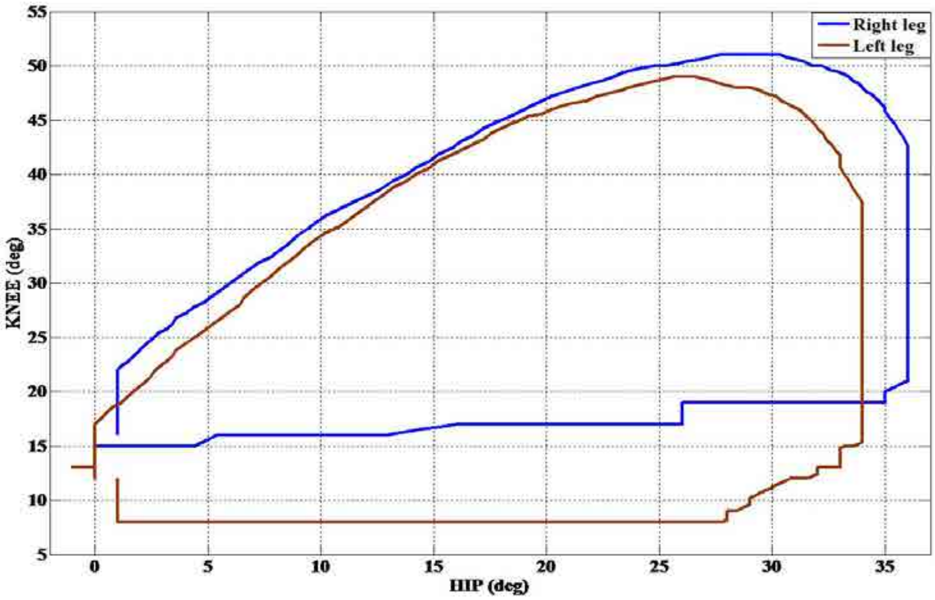

FIGURE 5.29: Average gait pattern of the patient P03 with variable stiffness after 3 sessions; Maximum knee flexion is found different because of the affected leg movement. The end of gait cycle indicates the onset of maintaining equilibrium position with 90N/m stiffness

TABLE 5.14: Results of adaptive gait training with patient P03; The knee flexion of both legs, for each session are represented in $mean \pm \sigma$ and the adaptive stiffness provided in $median \pm range$, Note: Session 1 was with a fixed stiffness value 80 Nm/deg.

Session	Left Knee flexion(deg)	Stiffness (Nm/deg)	Right Knee flexion (deg)	Stiffness (Nm/deg)
S1	49 ± 1	80	50 ± 2.6	80
S2	47 ± 1.58	83 ± 2	50 ± 1.4	82 ± 3
S3	49 ± 1	80 ± 3	50.5 ± 1.5	81 ± 2

TABLE 5.15: Clinical assessment measures of Patient P03

Clinical outcomes	Pre-intervention	Post-intervention
Left muscle test	6	6
Right muscle test	11	10
PSFS	0	0
WISCI II	12	12
SCIM III	56	56
10 MWT (s)	36.983 ± 4.22	27.863 ± 0.928

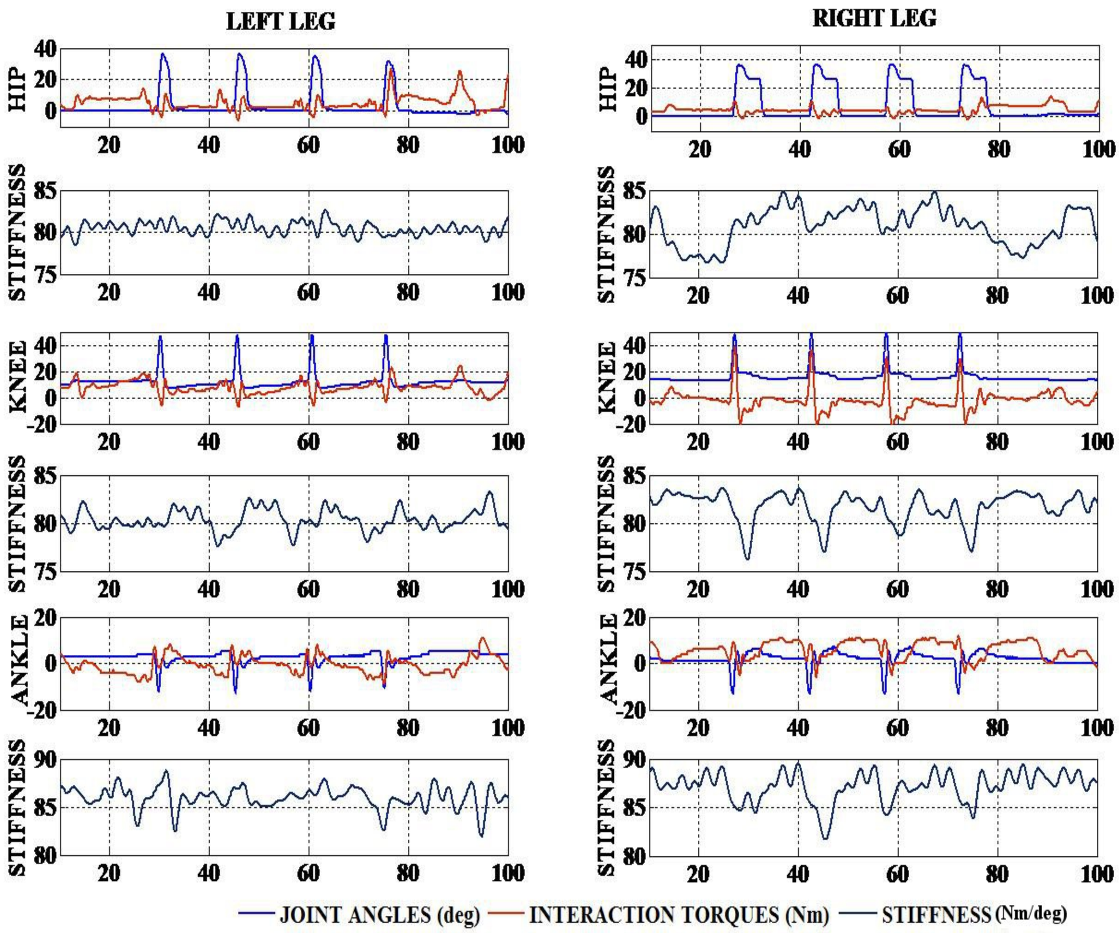

FIGURE 5.30: Joint angles, interaction torques and stiffness adaptation of patient P03; the influence of the adaptive stiffness is in function of the change in interaction torques and vice versa. The hip and ankle joint showed a minimum stiffness variation in comparison with knee joint, because of its biomechanical properties.

to walk with crutches and no physical assistance for 10 meters. In terms of SCIM, the mobility domain level increased from pre to post as 16 to 17 respectively. Further, in breathe sphincter management there was a reduction in the value from 28 to 27, so it is impossible to conclude any positive development or influence by the training. The patient showed a progressive change in the walking speed, by completing the 10 meter walking without any therapist assistance, at 27.863 ± 0.928. This progressive change is an influence of the additional training provided by the exoskeleton by the improvement in the flexion angles.

5.3.6 Case Study 4

Patient P04 was reported with a severe lesions in comparison with the other three patients. The patient was chosen because of the reported zone of partial sensory and motor preservation between T11 and L3. He was categorized as ASIA A and was normally using the Ankle-Foot-Orthosis (AFO) in both legs. The patient was able to maintain the equilibrium for a considerable amount of time.

The average gait pattern, at the end of session 3, in presence of the stiffness adaptation is presented in the fig 5.31. The maximum knee flexion angles in both legs is observed to be different and in a high variance. The hip joint trajectory was also observed to be different in comparison with other patients, due to special zones motor preservation. Since the patient was dependent on the assistance the maximum movement was also observed as 49 deg and 54 deg in the left and right leg respectively.

The walking pattern of the patient was more adaptable to the reference trajectory but due to the limited interaction, there was no significant change in the maximum flexion observed in the patient. The walking pattern was observed in series because of the confidence factor value. The confidence factor (0.6) in this case helped the subject to perform the movement since the user interaction was minimum in most of the gait phase. The stiffness adaptation

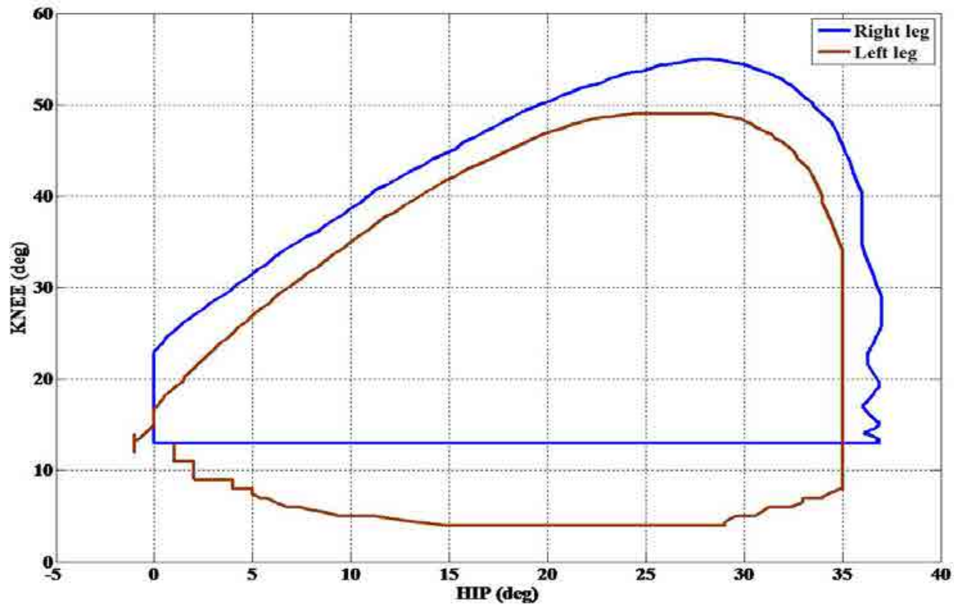

FIGURE 5.31: Average gait pattern of the patient P04 after session 3 with variable stiffness; Maximum knee flexion is found different because of the affected leg movement. The end of gait cycle indicates the onset of maintaining equilibrium position with 90N/m stiffness

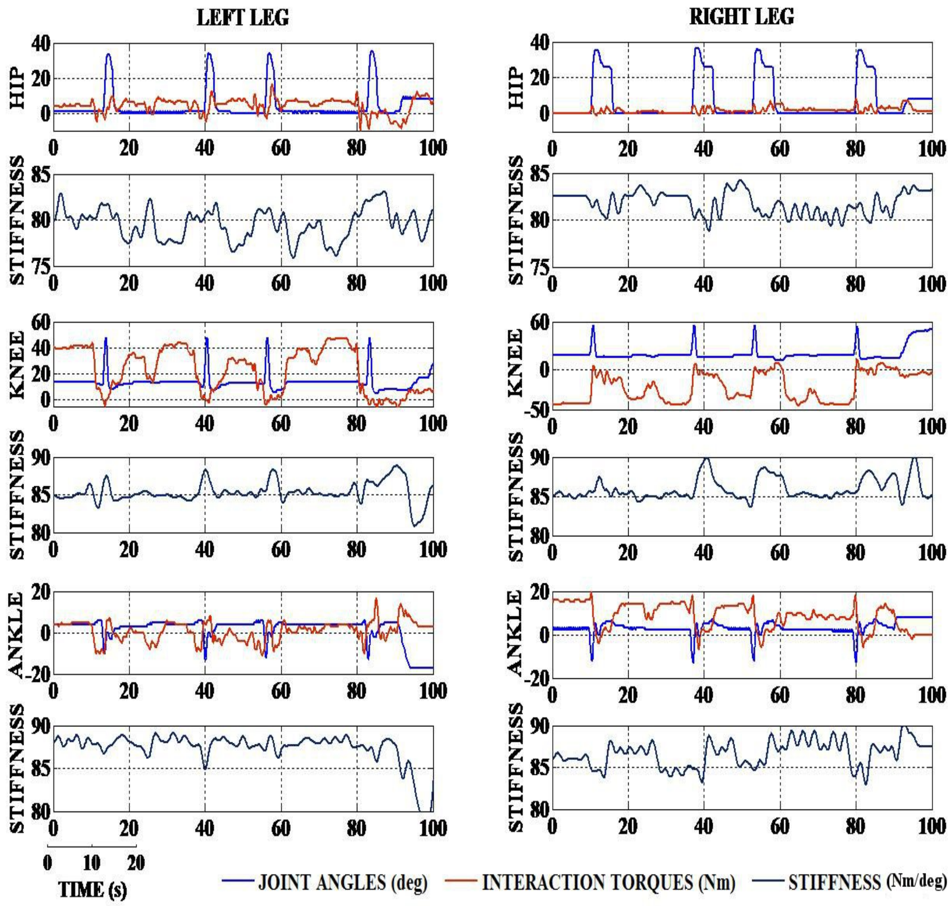

FIGURE 5.32: Joint angles, interaction torques and stiffness adaptation of patient P04; the influence of the adaptive stiffness is in function of the change in interaction torques and vice versa. The hip and ankle joint showed a minimum stiffness variation in comparison with knee joint, because of its biomechanical properties.

for both the knee and ankle were comparatively high than hip joint, as shown in fig 5.32.

TABLE 5.16: Results of adaptive gait training with patient P04; The knee flexion of both legs, for each session are represented in $mean \pm \sigma$ and the adaptive stiffness provided in $median \pm range$, Note: Session 1 was with a fixed stiffness value 80 Nm/deg.

Session	Left Knee flexion(deg)	Stiffness (Nm/deg)	Right Knee flexion (deg)	Stiffness (Nm/deg)
S1	49.3 ± 2.02	80	53.07 ± 2.01	80
S2	48 ± 1	83 ± 4	53.5 ± 2.06	84 ± 4
S3	49.21 ± 1.52	83 ± 4	54.5 ± 2.06	83 ± 4

The pre and post clinical assessment of the patient P04 is illustrated in the Table 5.17. The PSFS demonstrates that the patient initially had mild spasms by stimulation but after the

Table 5.17: Clinical assessment measures of Patient P04

Clinical outcomes	Pre-intervention	Post intervention
Left muscle test	9	4
Right muscle test	19	22
PSFS	1	0
WISCI II	12	12
SCIM III	71	60
10 MWT (s)	73.993 ± 5.676	93.323 ± 4.584

training no signs of spasms was detected. The patient is able to walk for 10 meters with crutches and no physical assistance. In terms of SCIM, the patient showed a change in both the mobility domain: pre - 21 and post - 19, and breath sphincter management: pre - 33 and post - 28. The patient was also reported with other medical conditions, not related to the training, and hence couldnt perform well enough in the gait training of 10 MWT. The complete perfromance analysis of this patient was not viable and so its impossible to conclude if there was any significant progress due to the training.

5.3.7 Discussion

The effectiveness of a rehabilitation therapy relies mainly on the active participation of the user [143, 260, 261]. The user participation has been validated and supported such as to initiate the movement on demand by the user. This type of locomotor training in SCI individuals may help in improving the movement of the user [262]. The role of combination therapies to monitor the neural activity and joint activity motivates the patient to pursue with high diligence[187] and this explains the high interaction torques pertaining to each joint demonstrating the participants' efforts in initiating a movement. The movement initiation command from the BMI system through the motor related activity of the patient initiates the exoskeleton or therapy. The assistance strategy is applied in function of both the human-orthosis interaction torques and deviation in the trajectory.

The efficiency of the adaptive assistance provided by the control model is evaluated for every patient in comparison with their individual gait pattern, obtained by applying the fixed stiffness. The maximum and minimum knee flexion angles obtained at the end of each session is taken and the mean value of n gait cycles of each session is considered for the analysis of patient evolution, as illustrated in Fig 5.33. These maximum and minimum flexion angles evaluate the adaptation performed by considering the patients movement both in their optimal conditions and in presence of physical tiredness. The stiffness adaptation performed for each user, for knee joint, was observed within a range of $80 \pm 5N/m$, as shown in Fig 5.34. The stiffness relaxation is observed more in the session 3, indicating the exoskeleton's adaptation to the users' movement. Similarly for the hip joint $80 \pm 2N/m$ and the ankle joint $80 \pm 7N/m$, Fig 5.35.

Human centered strategies are essential for ensuring user involvement and oriented to the development of robot behaviours[142, 143]. Several Assistance-As-Needed (AAN) strategies have been developed by monitoring the users' muscle synergies to define the degree of assistance and to support voluntary motion of the patient, such as in HAL [150, 182]. Another method of assistance by just applying torques in the hip and knee joints (Vanderbilt[147]) and with the trajectories being enforced by means of a virtual force field (H2[263]) has demonstrated an assistive behaviour for stroke individuals. A similar study, enforcing the symmetry based gait compensation in stroke individuals has been performed using HAL,

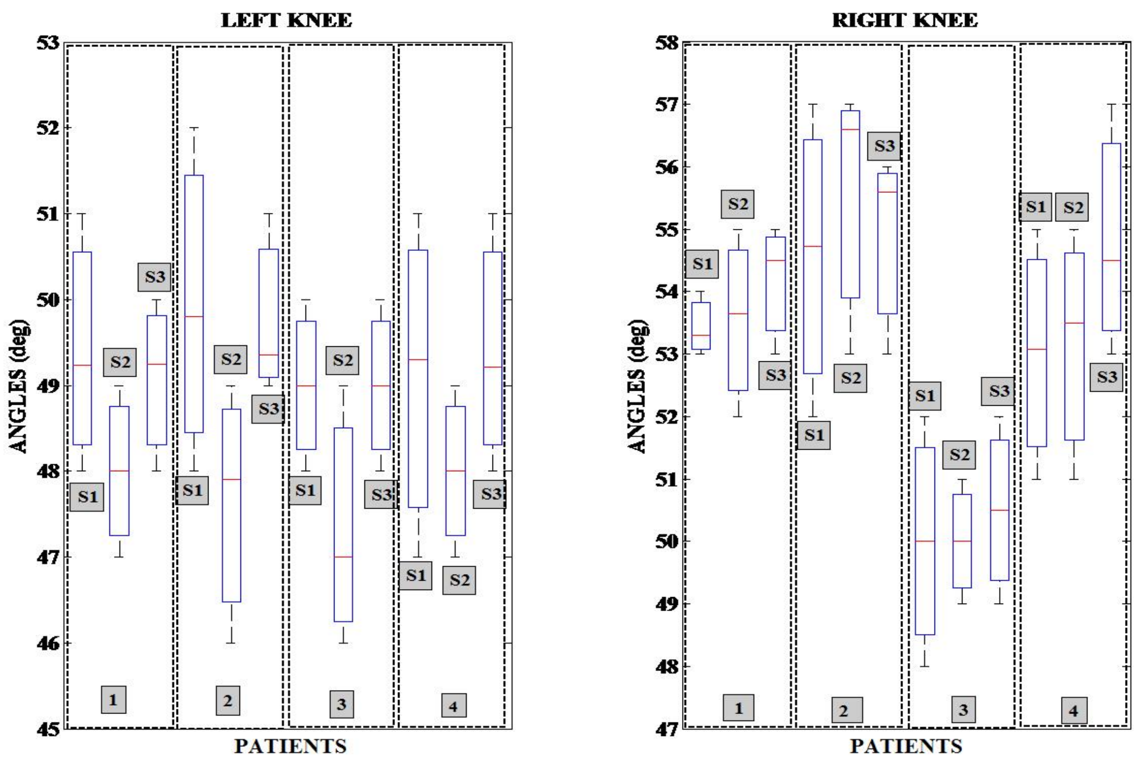

FIGURE 5.33: Maximum, minimum and the mean knee flexion angles of all the patients for each session (S1, S2 and S3); Patient P04 had a high sensibility in both the legs which resulted in a similar range of assistance after both sessions. Note that the S1 was performed with a high fixed stiffness.

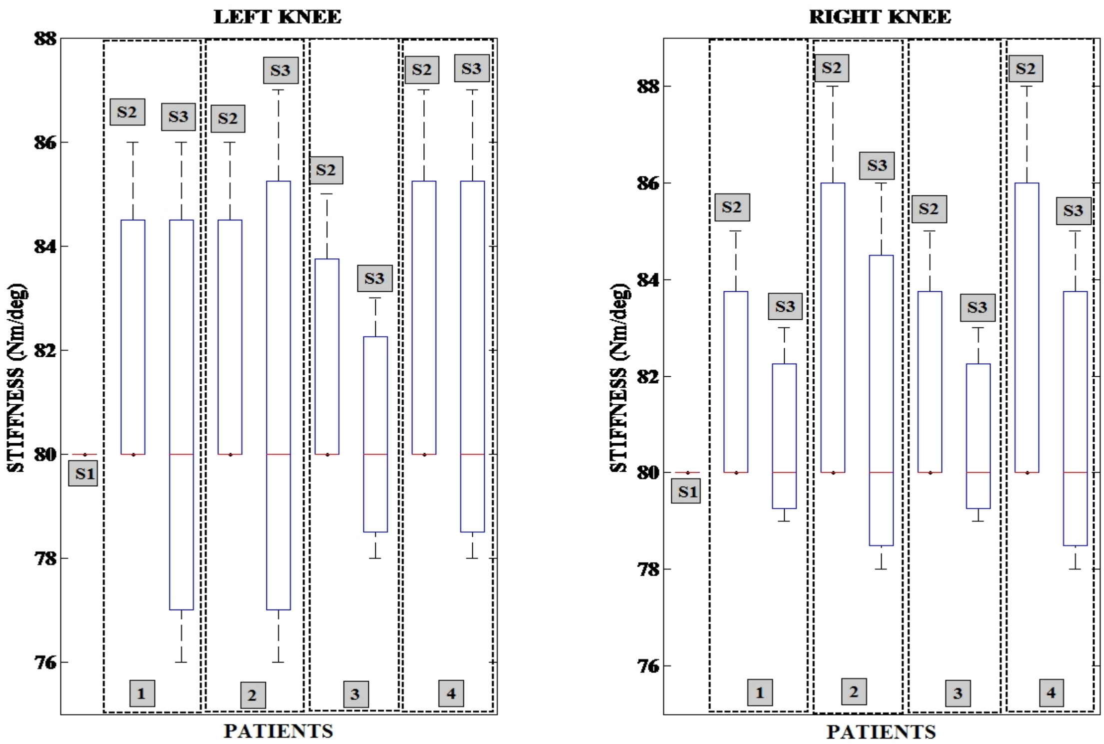

FIGURE 5.34: Maximum, minimum and the initial stiffness value of all the patients for each session (S2 and S3); Note that in session 1 (S1), a fixed stiffness of 80N/m was used for all the patients and also used as the initial value for sessions 2 and 3 (indicated by red line).

based on the movement of the unaffected limb. Such symmetry based adaptation considers assistance for the swing phase on the basis of the data compensated by the user [151]. In case of incomplete SCI individuals, this type of assistance cannot be viable, considering that the major concern is in maintaining balance [4]. Hence in case of Vanderbilt [148, 264], a Proportional-Derivative control loop enforced by high gain and with preprogrammed trajectories obtained from healthy subjects has been used for the training with paraplegic individuals.

A control model based on fixed stiffness value provides no adequate assistance or resistance with respect to the movement. In case of a fixed high stiffness, the function of the exoskeleton is almost similar to that of position control; thus the error with respect to the reference trajectory will be the minimum [228]. Similarly, with high fixed stiffness, the patients demonstrate a high interaction torques, which can be a result of the opposition force applied by the patient or due to some uncomfortable movement performed by the patient, out of their capabilities. Such trajectory based control approaches have proven to be efficient in terms of providing assistance to complete SCI and acute stroke individuals,

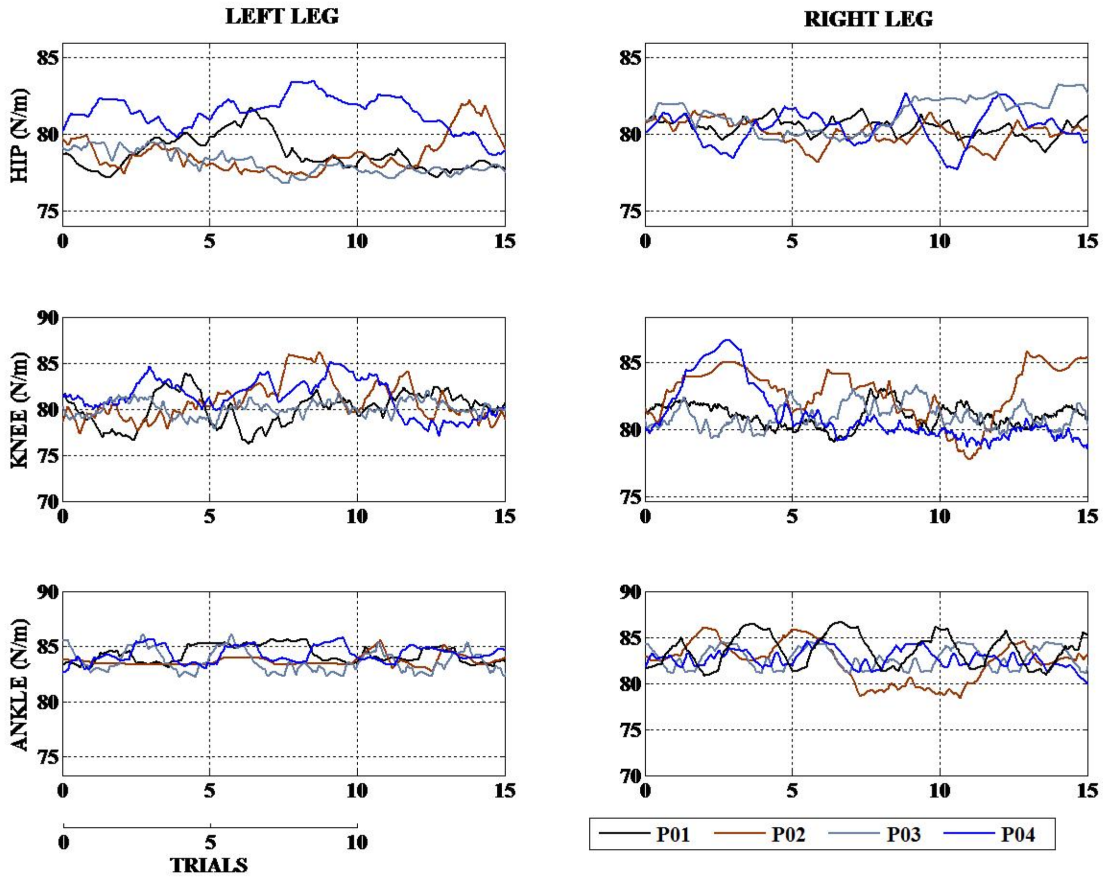

Figure 5.35: Adaptive Stiffness of each patient over the course of last 15 gait trial of session 3. The hip joints showed minimum variations in comparison with the knee and ankle joints, because of the lateral compensation. The stiffness values for the knee and ankle joint differed in high range but were observed to stabilize at the end of 10 to 15 trials.

but also resulting in slacking and inhibits motor learning [36]. For instance, the left leg of the patient P01 is comparatively weak and is completely assisted by the exoskeleton's movement. This assistance helps the patient in achieving the necessary flexion angles but with the negative interaction torques. The fixed stiffness assists or forces the patient to follow the reference trajectory, but the behaviour of the interaction torques explains the kind of movement pursued by the patient especially in the weaker leg. The maximum knee flexion angle, obtained by applying the fixed stiffness, is used to evaluate the performance of the variable stiffness approach. Simultaneously, the interaction torques are monitored to analyse the efficiency of the real time stiffness adaptation provided by the proposed control strategy.

In the variable stiffness, there is a change in the behaviour of the interaction torques as the user was not being forced, similar to a fixed stiffness, to perform the reference trajectory

[228]. The walking trajectory of the patient is not adapted to the unaffected leg, which helps in providing assistance independently, as needed in the case of stroke patients [147]. In some cases, the patient applied a similar force but couldn't reach a flexion angle similar to the fixed stiffness mode. The patients applied a compensatory walking by modifying the hip movement with respect to the knee flexion, which challenge the stiffness adaptation to be performed in real time.

The increment in the interaction torques [263] demonstrate the efficiency of the adaptive stiffness approach, which helps the patient to perform a movement in the weaker leg . This joint stiffness variation results in exerting an assistive or resistive behaviour, depending on the direction of the movement, similar to the flexion-extension movement of the joint [122, 166, 227]. All the patients, except for P04, demonstrated an increase in the knee flexion movement, at the end of two variable stiffness sessions. The patient P04 was observed with a high sensibility in both the legs, which might be a reason for the no-significant change in behaviour of the joint flexion limits.

In this work, the hip joint of all the patients, showed a little variation in the joint stiffness and more adaptable behaviour in real time without imposing a trajectory. In most cases the subjects tried to compensate the movement by modifying the trajectory of the hip joint which helped in realizing increased knee flexion movement [265]. The stiffness variation for the hip joint is observed lesser than the variation in the knee and ankle joint. This might be due to the lateral movement of the user's hip joint which compensates the joint trajectory [243, 266]. Since the exoskeleton H1 is a planar robot, the lateral hip movement cannot be monitored; however this orthosis limitation does not affect the proposed control strategy. The interaction torques, in case of the variable stiffness, show a major variance in comparison with the fixed stiffness output. This is obvious due to the less assistance provided to the weaker legs, compared to the leg with better mobility. The leg trajectory showed variation as a result of the increase in the interaction torques. The flexion and extension movement of the knee joint is essential in walking for maintaining the transition between gait phases. The stiffness variation for the knee joint is observed to stabilize, because of the repetitive movements, after a few gait trials. The trajectory deviation is found to be within a small range, but with a delay, in the knee joint.

An adaptive control strategy for walking, initiated by volitional orders, has been evaluated in function of the position deviation and human-orthosis interaction torques, thus ensuring an effective and safe therapy. The strategy has been evaluated and tested with four incomplete SCI individuals. The volitional commands from the BMI system, by monitoring the motor activity of the patient, are determined at the beginning of each gait cycle. Similarly, the stiffness value of each joint adapts dynamically to the user needs and keeps the joint positions bounded within the limits of the reference gait, in real time. The wearable robot was tested with no body weight compensation, which shows the reliability of the control strategy for ensuring dynamic stability in presence of ground reaction forces. The performance of the proposed control method was evaluated by comparing the trajectories obtained by applying the variable against fixed stiffness strategies.

The experimental results showed that the user's gait intention was recognized effectively with minimum delay and followed by the leg movement. The interaction torques of the weaker leg gradually increased over the course of the trials also maintaining the same joint positions limits, evidence of the user motivation. An increase in the maximum flexion movement of the user was observed in both legs, which invariably signify the effectiveness of the strategy. The results exhibit that the evolution of the stiffness value does not follow a similar pattern for all the joints. The stiffness value converges to be within a narrow range after a series of trials. The stiffness variation was in coordination with the flexion and extension movements, especially with the knee joint. This demonstrates the efficiency of the proposed method for a real time process involving multiple joints.

Usability measures

The usability measures of the adaptive control model can be discussed based on the satisfaction or statistical assessment obtained from the user. Since this thesis is not focussed on the hardware structure and the technical issues occurred due to it, the statistical measurements involves the exertion perceived by the patient (Table 5.8). The patients felt that the training was hard to some extent, especially in the case of patient P04. In general, all the patients felt comfortable and was not forced to perform a movement which reflected in the lower scales (less than 17) of Borg value.

The patients safety, comfortability and the effectiveness of the training is important to evaluate the control models performance and this is pursued by using QUEST (refer Table 5.9). The patient P02 and P03 were not very satisfied in terms of the comfortability, weight and dimensions of the exoskeleton, while patients P01 and P04 reported to be satisfied with

Table 5.18: Borg scale evaluation of exertion perceived by all the patients pre and post session

Patients	Pre-session	Post-session
P01	6	12
P02	6	11
P03	6	11
P04	6	15

Table 5.19: Evaluation of the usability and satisfaction of the experiments using the modified QUEST scale

Question	P01	P02	P03	P04
Dimension	4	2	1	4
Weight	3	3	1	5
Ease in adjusting	2	4	2	4
Safety and security	5	4	3	5
Durability	3	3	4	4
Ease of use	5	3	4	5
Comfortability	3	2	1	4
Effectiveness	4	4	3	4
Overall Satisfaction	5	2	3	4
Total	34	27	22	39

the same. In terms of performance of the effectiveness of the control model, all the patients reported to be satisfied with the performance or assistance provided to them. Furthermore, all the patients felt safe and secure in performing the movement and felt its easy to walk with the kind of assistance provided. Some of the patients were kind enough to comment about their perceived positive impacts such as, "The walking was more comfortable and adaptable to my walking pattern"; "The walking pattern is similar to what I pursue in a traditional training".

5.4 Conclusions

The transition of sit-to-stand is performed as an initial step of the therapy. A velocity-based control was applied as a solution to this task. The application of fixed stiffness, followed by the position control, helps in maintaining the postural stability. In case of paraplegic patients, the muscle stimulation by FES must also be included to provide assistance in the transition.

In balance training, the results demonstrate that the combined action of ankle and hip, that humans use for posture stabilization, can be also applied to an assistive exoskeleton. An event-based fuzzy control was applied to deal with any kind of perturbation, irrespective of its effect on individual or combined joint actions. The evaluation of the strategy takes into account the interaction forces between the orthosis and the subject. The assistance is provided with a decoupled control mode in order to ensure stability throughout the therapy by acting on both hip and ankle. The interaction forces are smaller in the case of the combined effect and this is due to the coupling effect of the hip and ankle joints. A reduction in the interaction forces also shows a reduction in recovery time. This method is efficient in handling the individual and combined effect of external perturbations acting on any joint movements. The exoskeleton maintained the equilibrium by providing suitable assistance throughout the therapy.

The adaptive strategy for gait training based on position error and the human-orthosis interaction torques acting on the system, was initially evaluated with healthy subjects. The stiffness value, for any joint, adapts dynamically to the user needs and keeps the position error bounded in the specified limits in real time. The wearable robot was tested with no body weight compensation which demonstrates the reliability of the control strategy in terms of ensuring dynamic stability in presence of ground reaction torques. The results of the proposed method were verified in comparison with a predefined gait trajectory. The experimental results showed that the evolution of the stiffness value cannot follow a similar pattern for all the joints. Similarly the stiffness value converges to a value within a given narrow range after a series of trials. The stiffness variation was in coordination with the flexion and extension movements. This demonstrates the efficiency of the proposed method for a real time process involving multiple joints.

A volitional control based gait initiation is developed and evaluated with both healthy subjects and SCI patients. Volitional orders are identified using the mechanical interaction or BMI system. In case of the healthy subjects both the methods are used, individually, to trigger the movement of the exoskeleton. For SCI patients the BMI based gait initiation is used, to provide a complete rehabilitation model in terms of top-down approach. In both gait initiation models, results showed that the user's gait intention was recognized effectively with minimum to no delay and followed by the leg movement.

For SCI patients, the interaction torques of the weaker leg gradually increased over the course of the trials also maintaining the same joint positions limits, evidence of the user motivation. An increase in the maximum flexion movement of the user was observed in both legs, which invariably signify the effectiveness of the strategy. The results exhibit that even for SCI patients, the evolution of the stiffness value does not follow a similar pattern for all the joints. A progressive change in the maximum flexion of the knee joint was observed at the end of each session which shows improvement in the patient performance. Results of the adaptive impedance were evaluated by comparing with the application of a constant impedance value, due to lack of breakthrough. The participants reported that the movement of the exoskeleton was flexible and the walking patterns are similar to their own distinct patterns.

Chapter 6

Conclusion, Contributions & Future works

This chapter summarizes the major contributions of the dissertation and highlighting its importance in the rehabilitation therapy. The chapter explains in brief about the contributions and the publications which was achieved as a result of the research work performed. Finally, the prospective future works are presented which can be considered as a continuation of this thesis work and their significant importance.

6.1 Conclusion

The work presented in this dissertation was aimed at developing an user dependent adaptive control depending on the human-orthosis interaction. The global objective of developing a Assist-as-needed strategy was achieved by passing through the intermediate goals. This work permitted achieving the objective established at the beginning of the dissertation. It has resulted in various scientific contributions with significant importance for both technological and rehabilitation aspects. The following are the major contributions achieved as a result of the presented work.

- Development of a simulation tool with the similar configuration of the exoskeleton to validate the control models and to analyse the joint trajectory behaviour in presence of external disturbances. The simulation model movement was also modelled by applying the calculated human-orthosis interaction torques as the input along with the joint angles.
- Evaluation of the model of cascaded control to perform task based modeling. The high level control architecture was modelled based on the interactive inputs from the human and orthosis and to switch between the different control models. The transition between the different models helps in developing the sufficient strategy.
- Assisting the users' transition from sit-to-stand by using a velocity-based control with a fixed stiffness. The fixed stiffness ensured the safe and stable standing position and to maintain equilibrium.
- Evaluating the balance stability of the user an event-based fuzzy control was developed. The strategy was implemented and evaluated in combination with a Wii platform which monitors the COM_p of the user and then define the assistance to be provided.
- Adaptive gait assistance based on the deviation in the trajectory and the interaction torques. The stiffness parameter of each joint were modified, in real time, with respect to the performance and characteristics of the user.
- Initiation of the therapy or walking movement based on the voluntary information from the user was realised using the volition-adaptive control model. The volitional input model is considered based on the mechanical interaction with the exoskeleton

and by detecting the neural signals around the motor cortex region. The gait initiation directly controlled by the brain allows synchronizing the user's intention with the afferent stimulus provided by the movement of the exoskeleton, which maximizes the potentiality of the system in neuro-rehabilitative therapies.

- Evaluation of the performance of the adaptive strategy, for gait assistance, with the help of healthy subjects. Results with healthy subjects showed a significant change in the pattern of the interaction torques, elucidating a change in the effort and adaptation to the user movement. This study was performed as a preliminary study to evaluate the gait adaptation, in real time, before conducting the trials with paraplegic individuals.

- Performance evaluation of the volition-adaptive gait strategy with incomplete SCI individuals. In case of patients, the adaptation showed a progressive change in the joint performance (flexion/extension range) and change in interaction torques. The active change in interaction torques (positive to negative) reflects the active participation of the patient, which also explained the adaptive performance.

6.2 Contributions

The work explained in this dissertation has resulted in a number of publications both in high impact journals and major conferences. Out of the four journal articles, 2 are in the review process while 2 of them have been already published.

6.2.1 Journals

- E. Lopez-Larraz, F. Trincado-Alonso, **V. Rajasekaran**, S. Perez-Nombela, A.J. del-Ama, J. Aranda, J. Minguez, A. Gil-Agudo and L. Montesano, " Control of an ambulatory exoskeleton with a brain-machine interface for spinal cord injury gait rehabilitation" *Frontiers in Neuroscience.* (In progress)

- **V.Rajasekaran**,E. Lopez-Larraz, F. Trincado-Alonso, J. Aranda, L. Montesano, A.J. del-Ama and J L Pons, " Gait assistance with volition-adaptive control using wearable exoskeleton: A study with incomplete spinal cord injury individuals ", *Journal of NeuroEngineering and Rehabilitation.* (Submitted on 5/10/2015)

- **V.Rajasekaran**, J. Aranda, A. Casals and J. L.Pons, "An Adaptive control strategy for postural stability using a wearable exoskeleton." Special Issue: "Wearable robotics for motion assistance ", *Robotics and Autonomous Systems.* Vol.73, pp 16-25, November 2015.
  http://dx.doi.org/10.1016/j.robot.2014.11.014

- **V. Rajasekaran**, J. Aranda and A. Casals,"Recovering planned trajectories in robotic rehabilitation therapies under the effect of disturbances." *International Journal on System Dynamics Applications (IJSDA)*, Vol.3, No.2, pp 34-49, September 2014.
  DOI: 10.4018/ijsda.2014040103.
  http://services.igi-global.com/resolvedoi/resolve.aspx?doi=10.4018/ijsda.2014040103

6.2.2 Conference Proceedings

- **V. Rajasekaran**, J. Aranda and A. Casals,"User intention driven adaptive walking using wearable exoskeleton ", *Iberian Conference on Robotics, (ROBOT)*, Portugal, 2015.

- **V. Rajasekaran**, J. Aranda and A. Casals,“Adaptive walking assistance based on human-orthosis interaction ”, *Proceedings of the 2015 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, Germany, 2015.

- **V. Rajasekaran**, J.Aranda and A.Casals,“Compliant gait assistance triggered by user intention.” *Engineering in Medicine and Biology Society (EMBC), 37th Annual International Conference of the IEEE*, pp 3885-3888, Milano, Italy, 2015.
  doi: 10.1109/EMBC.2015.7319242

- **V. Rajasekaran**, J. Aranda and A. Casals,“Handling disturbances on planned trajectories in robotics rehabilitation therapies.” *XIII Mediterranean Conference on Medical and Biological Engineering and Computing (MEDICON) 2013*, IFMBE Proceedings Vol. 41, 2014, pp 85-88, Seville, Spain.
  http://link.springer.com/chapter/10.1007/2F978-3-319-00846-2_21

6.2.3 Communications

- **V. Rajasekaran**, J. Aranda and A. Casals,“Adaptive Walking using a wearable exoskeleton”, *7th IBEC Symposium on Bioengineering for Future Medicine*, Barcelona, 29th September, 2014.

- **V. Rajasekaran**, J. Aranda and A. Casals,“Interaction based adaptive walking using a wearable exoskeleton.” *WeRob’14- International Workshop on Wearable Robotics*, Baiona, 14th – 19th September, 2014.

- **V. Rajasekaran**, J. Aranda and A. Casals,“Control strategies in robotic neurorehabilitation therapies compatible with FES.” *6th IBEC Symposium on Bioengineering and Nanomedicine*, 8th May 2013, Barcelona.

- **V. Rajasekaran** and J. Aranda, “Implicit force control approach for an Orthotic System.” *Summer School on Neurorehabilitation- Emerging therapies*, 16-21 September 2012, Nuévalos, Zaragoza, Spain.

- **V. Rajasekaran** and J. Aranda,“Modelling an Orthotic system for its Volitional Control.” *5th IBEC Symposium on Bioengineering and Nanomedicine*, 11th June 2012, Barcelona.

6.2.4 Book Chapter

- A.J. Del-Ama, A. Cuesta, **V. Rajasekaran**, F. Trincado, H. In and D.J. Reinkensmeyer. "Robotic Rehabilitation: Ten Critical Questions about Current Status and Future Prospects Answered by Emerging Researchers". In Jose L. Pons and Diego Torricelli, editors, *Emerging Therapies in Neurorehabilitation*, Vol.4 of Biosystems & Biorobotics, pages 189-205. Springer Berlin Heidelberg, 2014. ISBN 978-3-642-38555-1. doi: 10.1007/978-3-642-38556-8_10.
  http://link.springer.com/chapter/10.1007/978-3-642-38556-8_10

6.3 Future Work

This dissertation was the result of my personal motivation in developing a sufficient AAN strategy, specifically focussed on paraplegic individuals. On the course of developing the adaptive model, I have encountered some of the possible future continuations which could lead in developing more AAN strategies or answering more questions to the research community.

Following are some of the promising future works encountered as a result of this thesis,

1. Expanding the role of disturbances to multiple joints by the action of interaction forces, which varies depending on the effect of rehabilitation therapies. The use of Variable execution speed in presence of multiple disturbances would be a challenging task by means of multiple effects acting in a joint at a time instant t. These multiple direct effects on a joint also lead to some unpredictable effects in other joint trajectories, thus leading to the loss of synchronization among the individual joint trajectories.

2. Evaluating the performance of the Gait assistance-adaptive control in a larger number of patients and sessions. This will help in defining a general framework or a scheme for the usage of "confidence factor", depending on the lesion level. The study will also give an insight idea about the change in adaptation to the movement, over a group of sessions.

3. It would be interesting to test the efficiency of assistance in the presence of muscle stimulation (FES), which will act as external disturbances from the control perspective.

4. To iteratively adapt the stiffness value after a set of trials and to modify the initial value in function of the specific user performance and movement.

Appendix A

Trials with Healthy Subjects

Results of the preliminary study conducted with healthy subjects are presented in this section. The performance evaluation of the healthy subject is perfromed based on the joint angle movement and the change in interaction torques. In case of the volitional orders, the movement detection is performed using the BMI or mechanical interaction.

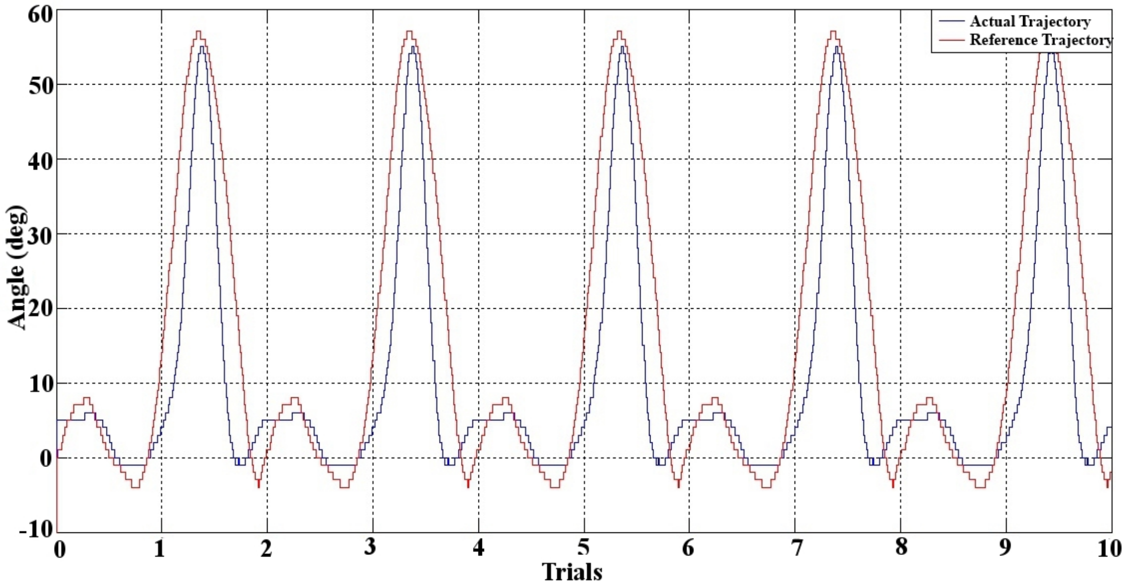

FIGURE A.1: Knee joint trajectories of the healthy subject S1; The resulting trajectrory after the application of adaptive stiffness is compared against the reference trajectrory

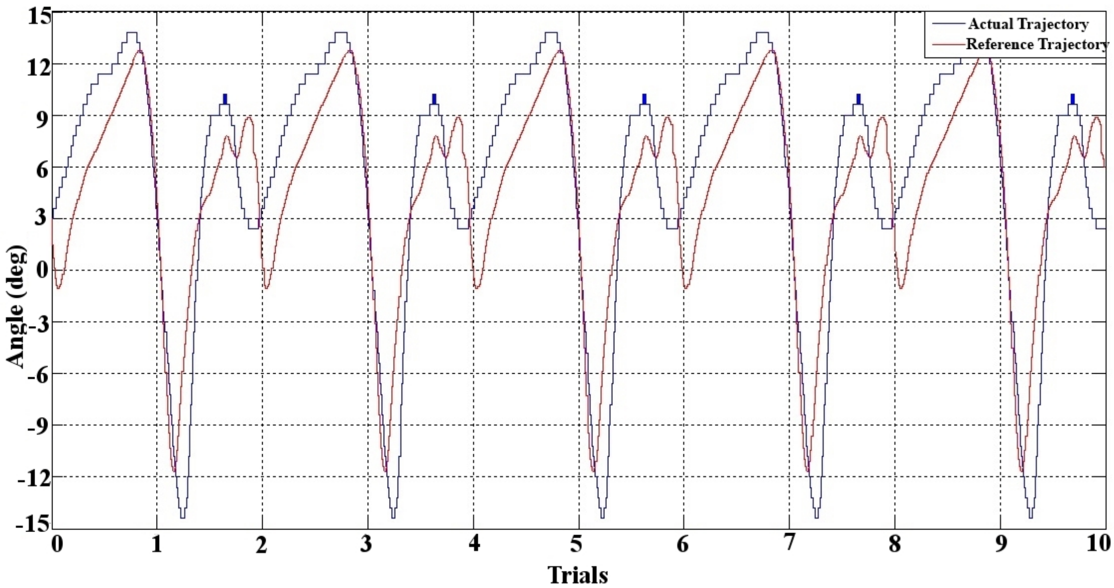

FIGURE A.2: Ankle joint trajectories of the healthy subject S1; The resulting trajectrory after the application of adaptive stiffness is compared against the reference trajectrory

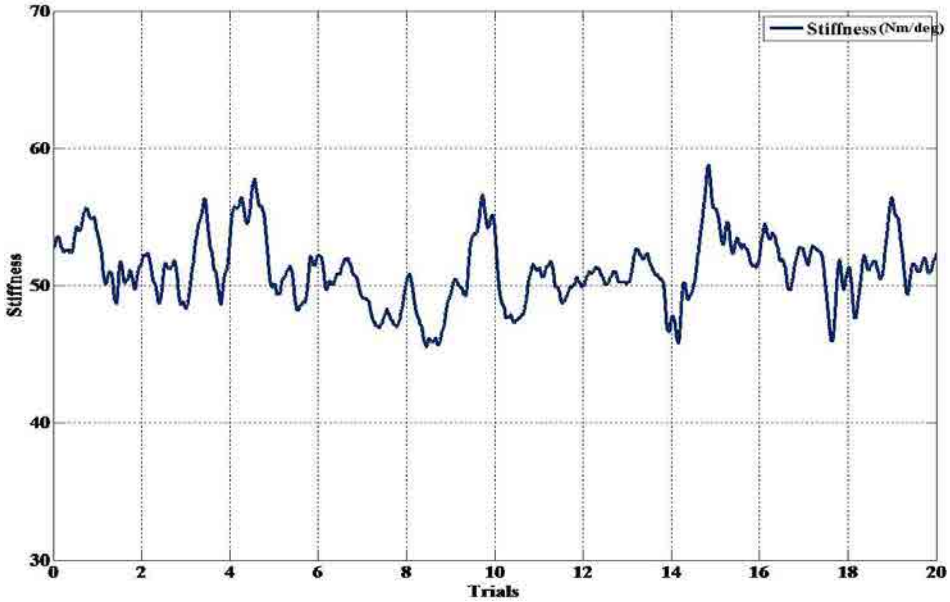

Figure A.3: Stiffness variation in the knee joint of the healthy subject

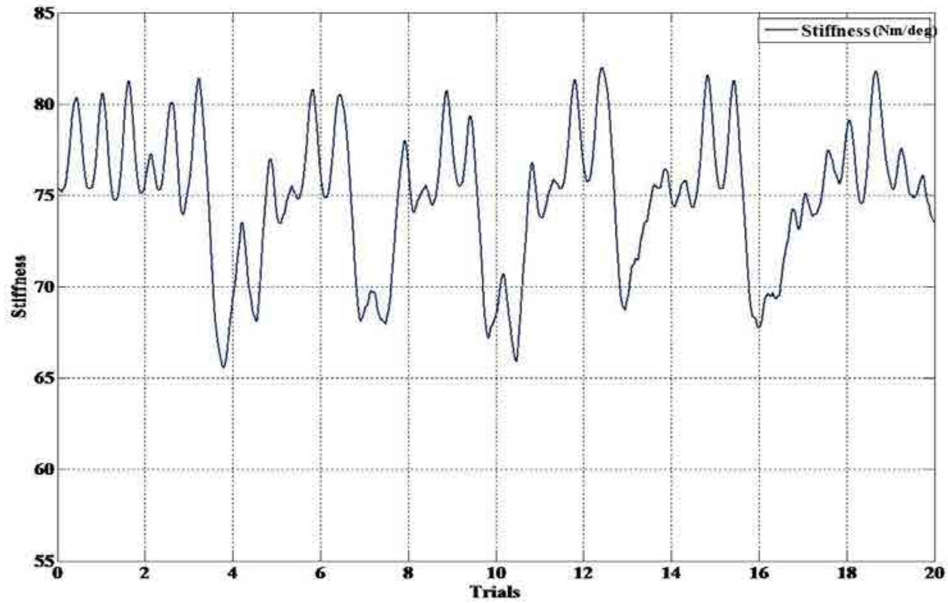

Figure A.4: Stiffness variation in the ankle joint of the healthy subject

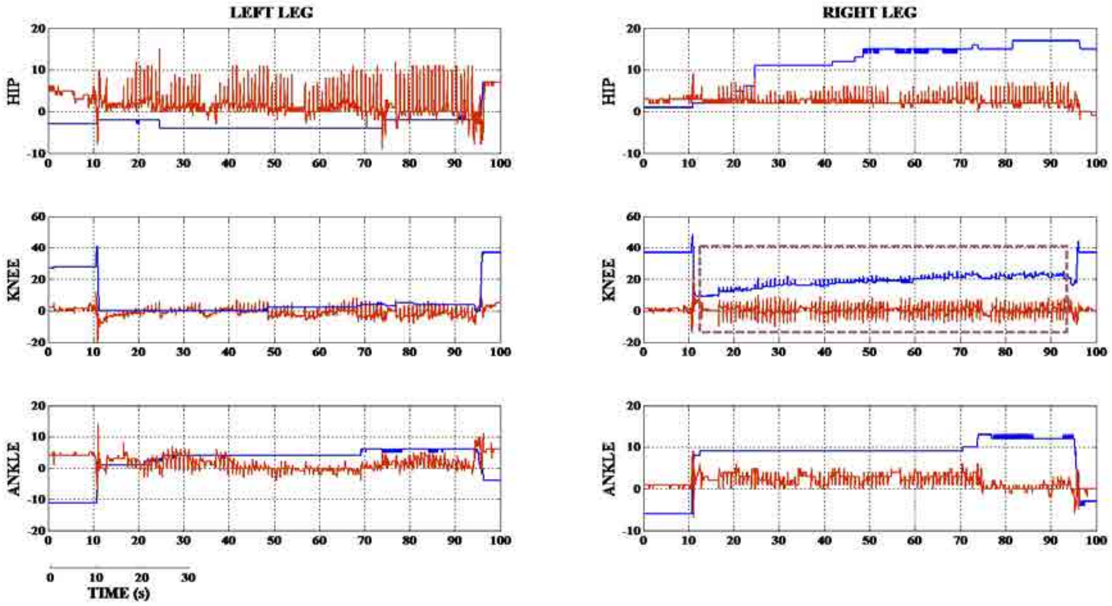

FIGURE A.5: Detection of movement intention by monitoring the joint angles and interaction torques of healthy subject H1; Changes in the joint angles, interaction torques and movements, of right leg, are observed while the motor activity is monitored using the neural signals through the BMI system (highlighted region). The joint links were maintained rigid with 90N/m stiffness, similar to the case of incomplete SCI patients

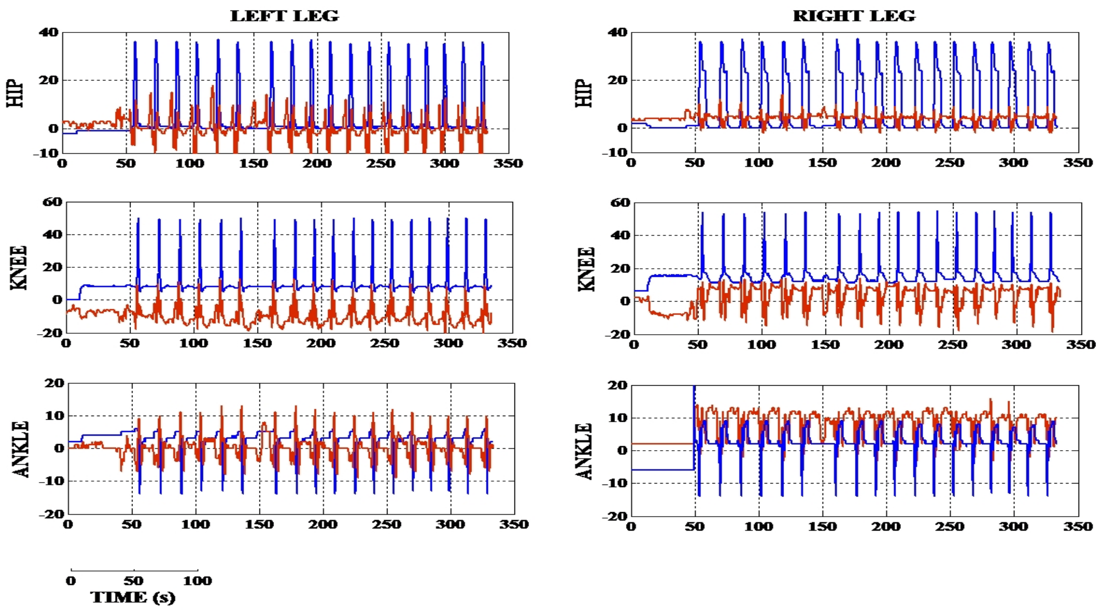

FIGURE A.6: Joint trajectories and interaction torques with adaptive stiffness of healthy subject H1; The maximum joint flexion is observed in most of the gait sequences and the interaction torques follow a similar pattern throught the gait training, indicating that the sufficient joint adaptation is provided

Appendix B

Trials with incomplete SCI Subjects

Results of the incomplete SCI patients are presented in brief in this section. The performance evaluation of each patient considers the joint angle's flexion/extension and the variance in interaction torques. A progressive change in the flexion movement of the knee joint is observed in all the subjects, along with the possible prediction of volitional commands.

B.1 Patient P01

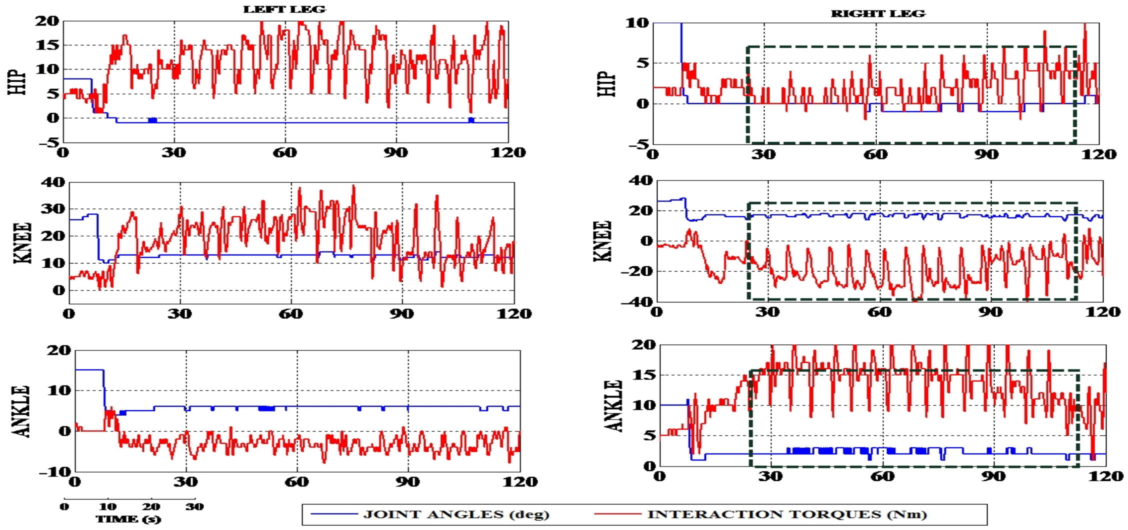

FIGURE B.1: Detection of movement intention by monitoring the joint angles and interaction torques of patient P01; Changes in the joint angles interaction torques and movements are observed while the motor activity is monitored using the neural signals through the BMI system (highlighted region). The joint links were maintained rigid with 90N/m stiffness, to maintain the knee and ankle joints in upright position.

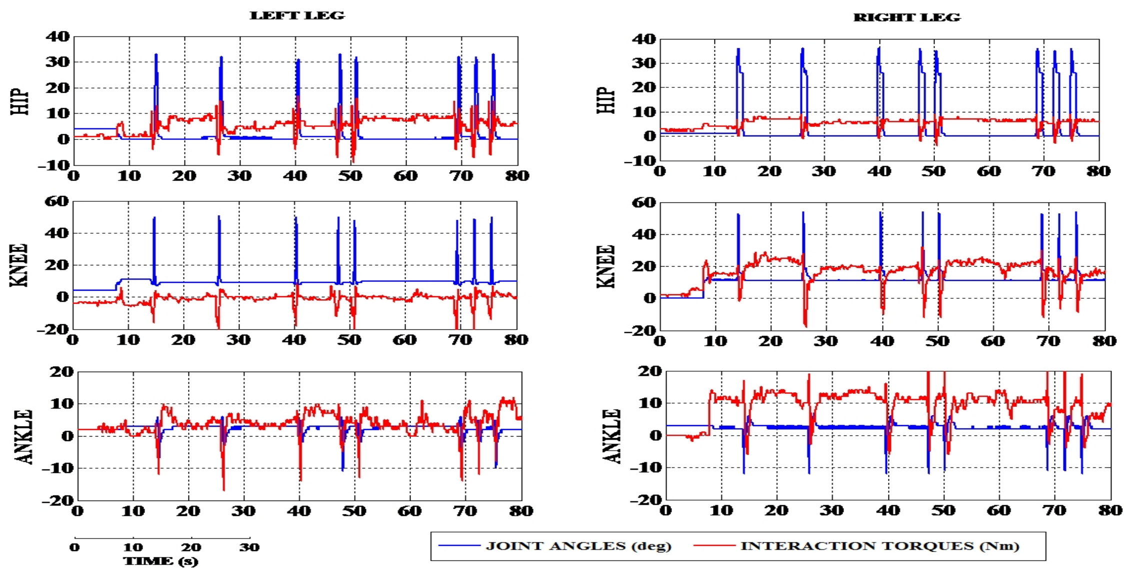

FIGURE B.2: Joint Trajectories of the patient P01 in presence of fixed stiffness (80N/m); The participants' left leg was weaker than the right explaining the difference in movement. Left knee flexion movement is completely assisted by the exoskeleton indicated by the negative interaction torques

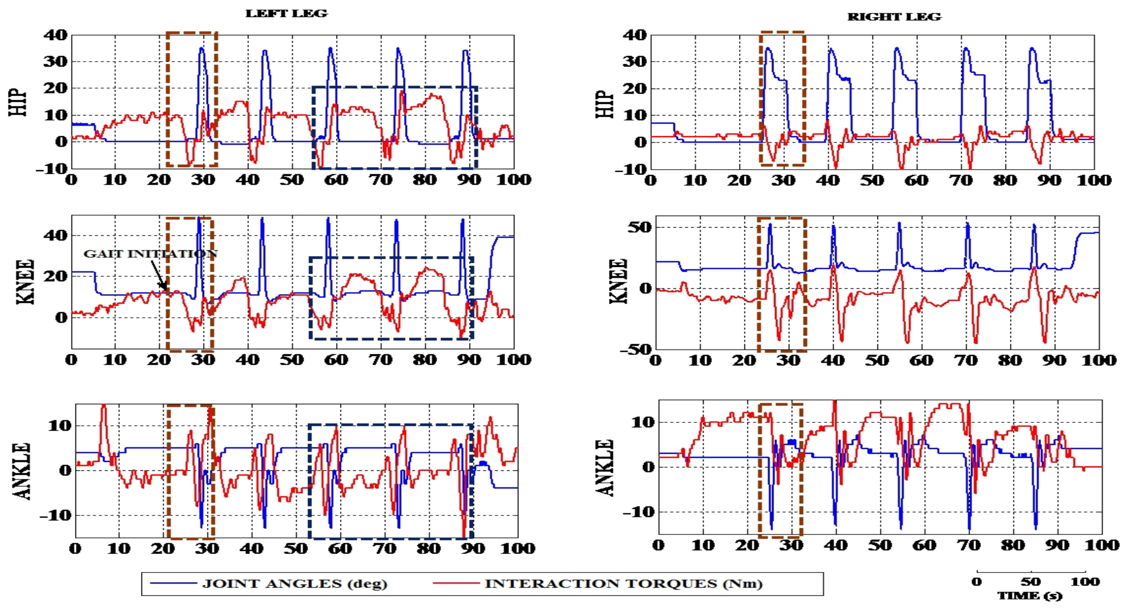

FIGURE B.3: Joint trajectories and interaction torques of the patient P01 with adaptive stiffness; Left leg of the patient was weaker but gradually there was a increase in the interaction torques (highlighted) due to the influence of the adaptive stiffness.

B.2 Patient P02

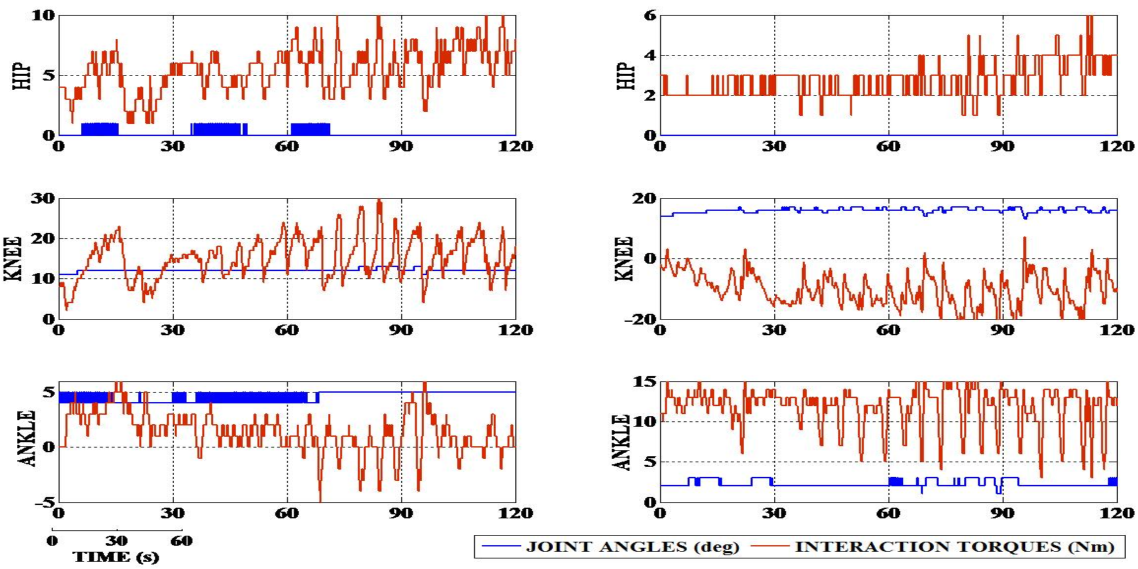

FIGURE B.4: Detection of movement intention by monitoring the joint angles and interaction torques of patient P02; Changes in the joint angles interaction torques and movements are observed while the motor activity is monitored using the neural signals through BMI. The joint links were maintained rigid with 90N/m stiffness, to maintain the patient in upright position.

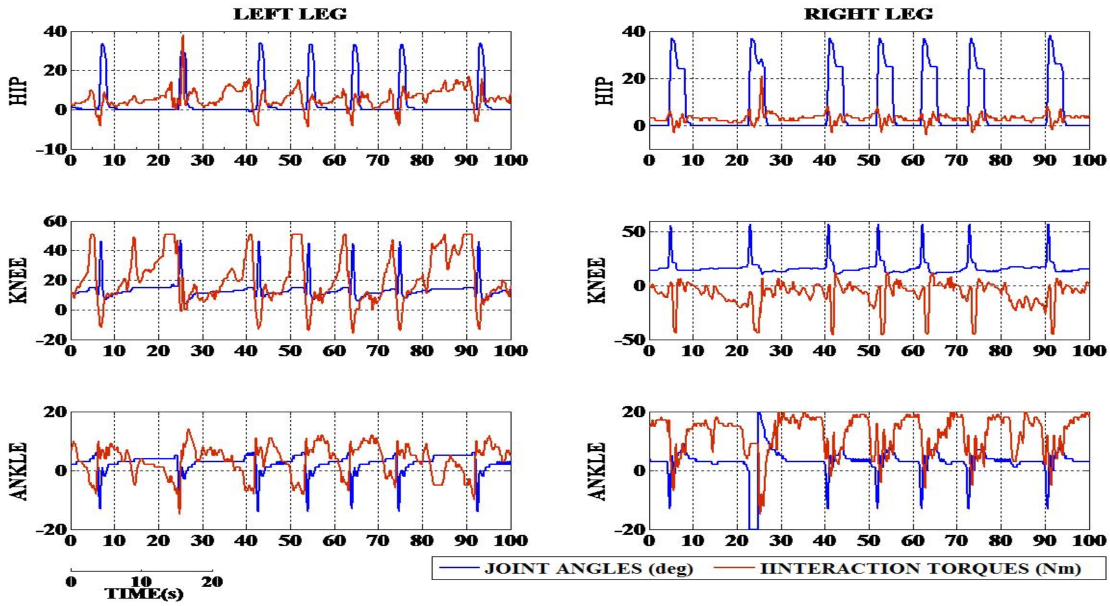

FIGURE B.5: Joint Trajectories of the patient P02 in presence of fixed stiffness (80N/m); The participants' Right leg was weaker than the right explaining the difference in movement. Right knee flexion movement is completely assisted by the exoskeleton indicated by the negative interaction torques

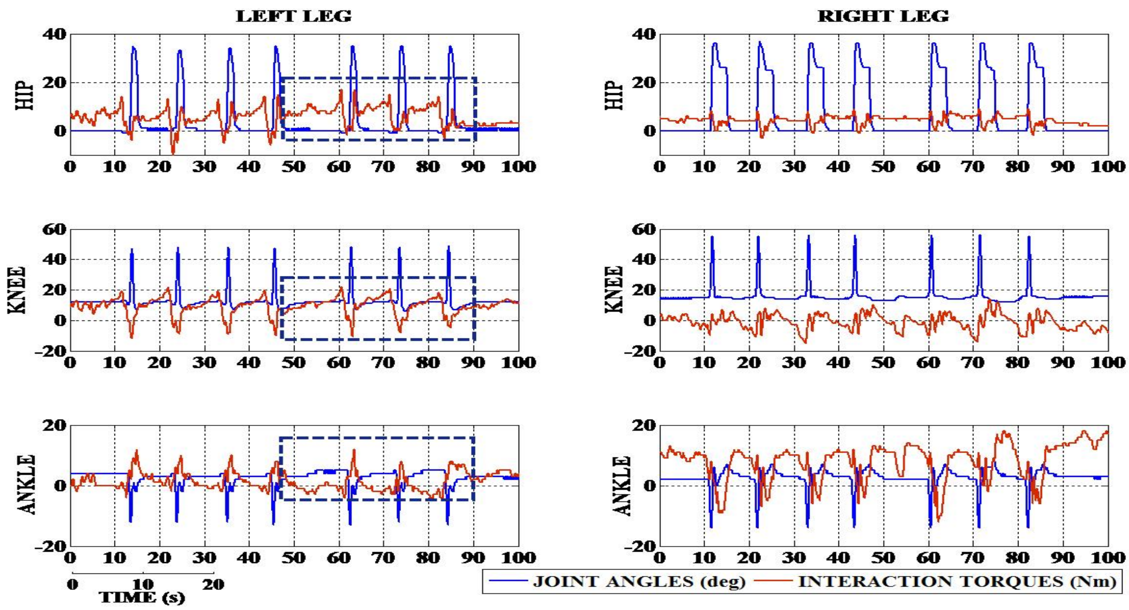

FIGURE B.6: Joint trajectories and interaction torques of the patient P02 with adaptive stiffness; Right leg of the patient was weaker but gradually there was a increase in the interaction torques (highlighted) due to the influence of the adaptive stiffness

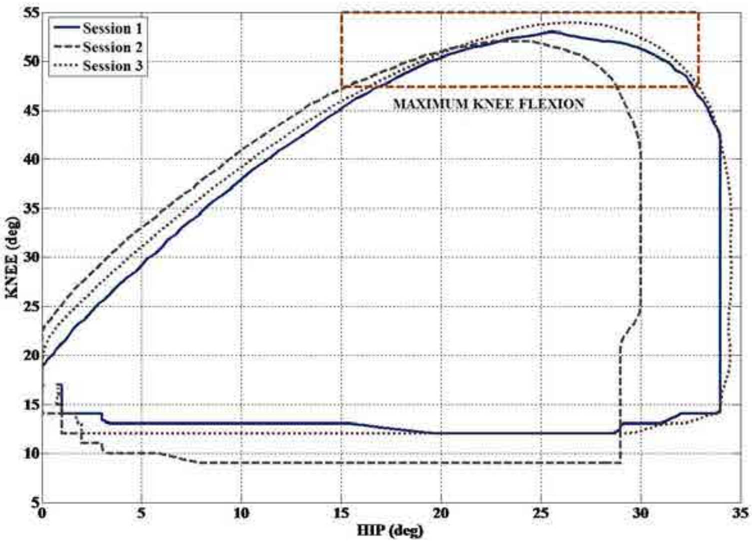

FIGURE B.7: Maximum flexion of Left knee of the patient P02, at the end of each session; the patient applied a compensatory movement in the hip joint which helped the user to achieve a maximum knee flexion in adaptive stiffness mode.

B.3 Patient P03

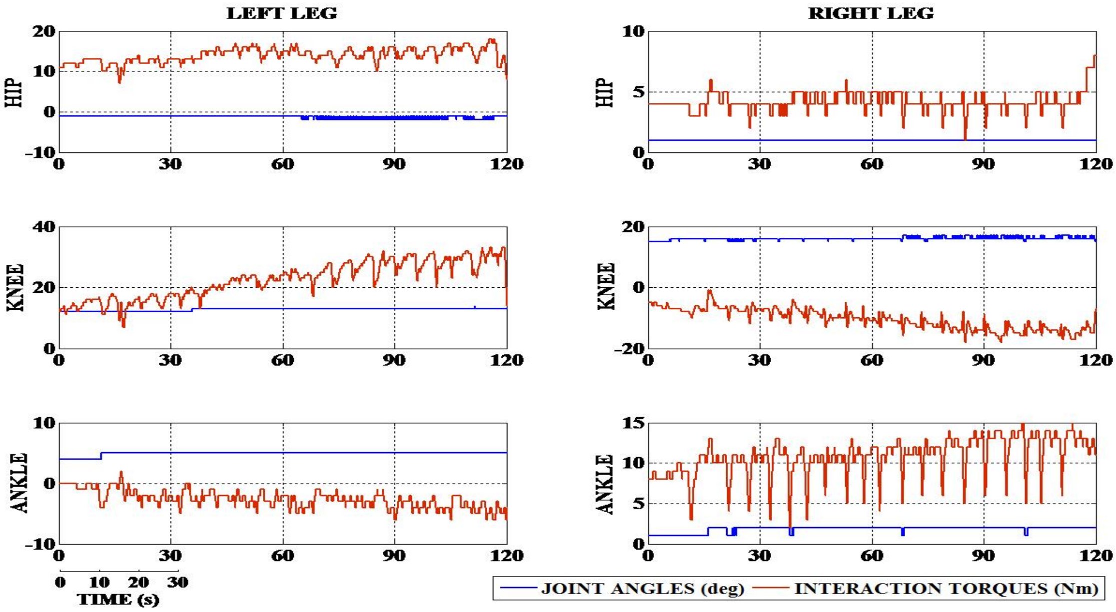

FIGURE B.8: Detection of movement intention by monitoring the joint angles and interaction torques of patient P03; Changes in the joint angles interaction torques and movements are observed while the motor activity is monitored using the neural signals through the BMI system. The joint links were maintained rigid with 90N/m stiffness, to maintain the joints in upright position.

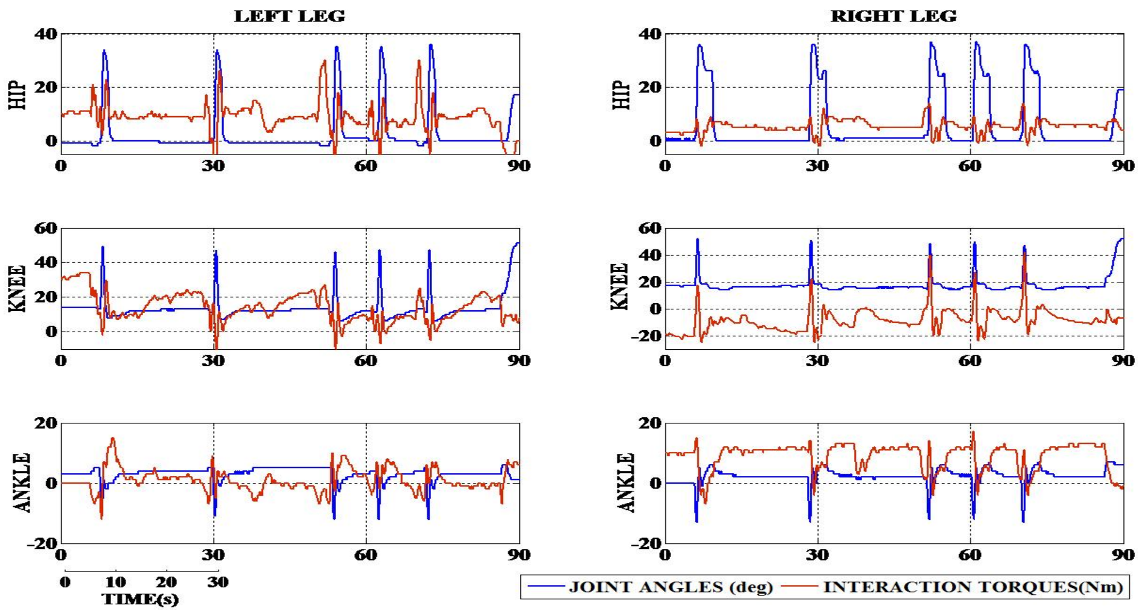

FIGURE B.9: Joint Trajectories of the patient P03 in presence of fixed stiffness (80N/m); Both the legs of the participant was weaker.

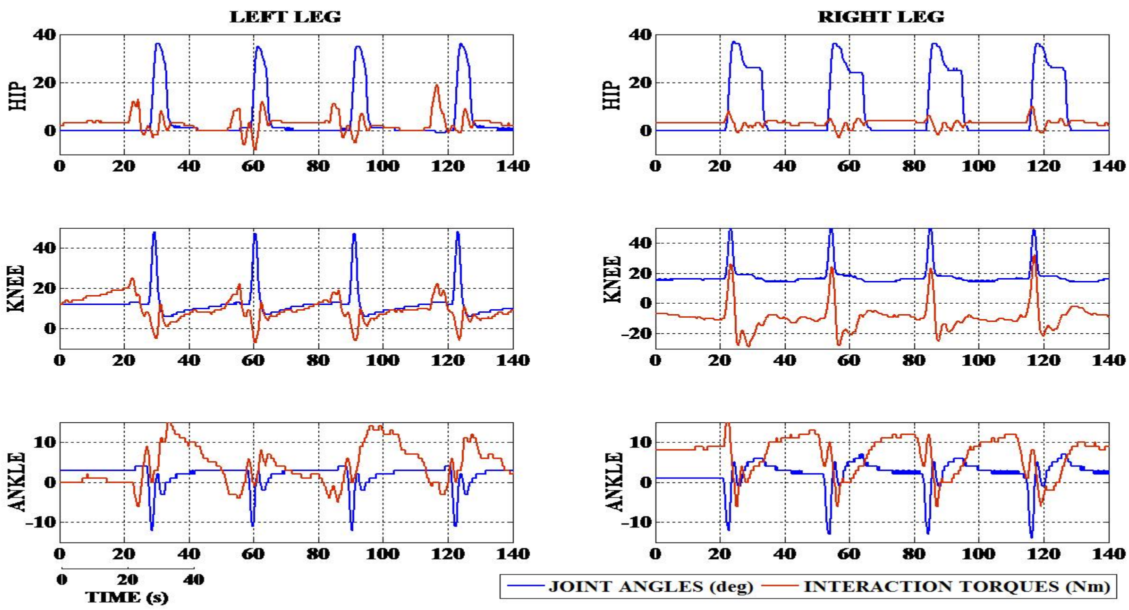

FIGURE B.10: Joint trajectories and interaction torques of the patient P03 with adaptive stiffness; Right leg of the patient was weaker but gradually there was a increase in the interaction torques due to the influence of the adaptive stiffness

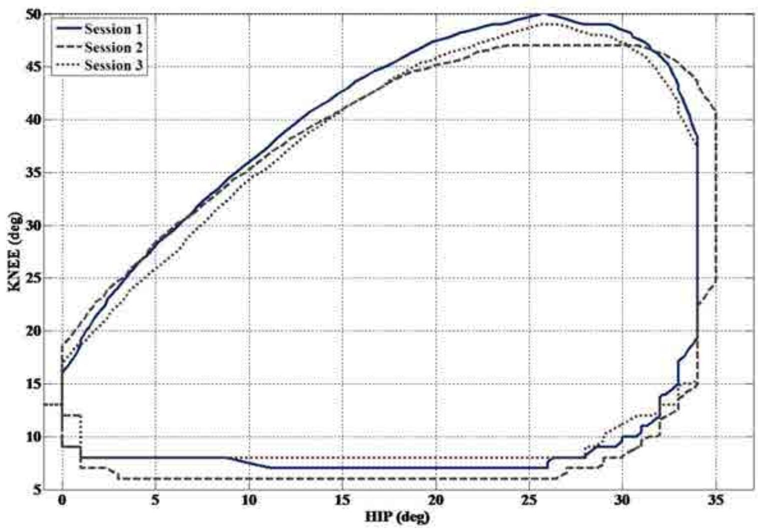

FIGURE B.11: Maximum flexion of Left knee of the patient P03, at the end of each session; the Patient applied a compensatory movement in the hip joint which helped the user to achieve a maximum knee flexion in adaptive stiffness mode.

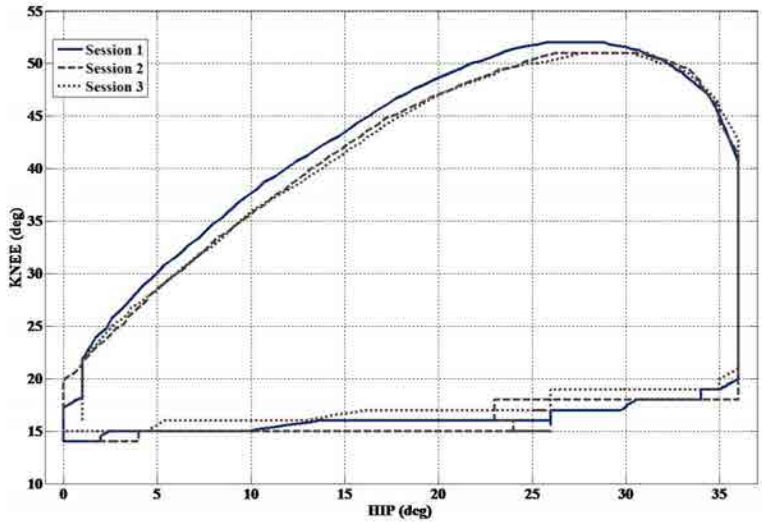

FIGURE B.12: Maximum flexion of right knee of the patient P03, at the end of each session; the Patient applied a compensatory movement in the hip joint which helped the user to achieve a maximum knee flexion in adaptive stiffness mode.

B.4 Patient P04

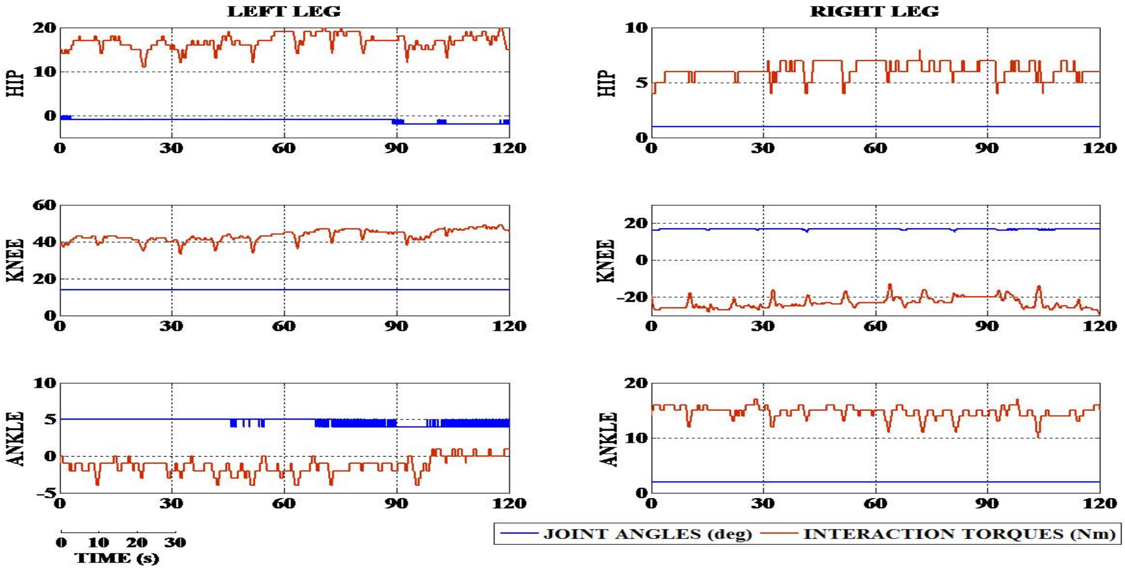

Figure B.13: Detection of movement intention by monitoring the joint angles and interaction torques of patient P04; Changes in the joint angles interaction torques and movements are observed while the motor activity is monitored using the neural signals through the BMI system. The joint links were maintained rigid with 90N/m stiffness, to maintain the knee and ankle joints in upright position.

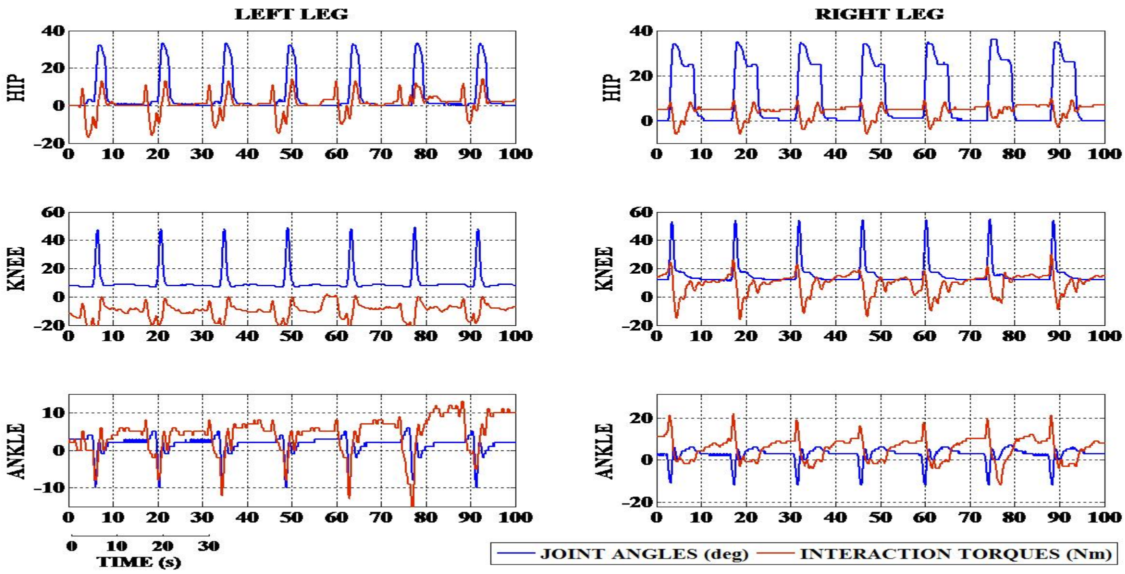

Figure B.14: Joint Trajectories of the patient P04 in presence of fixed stiffness (80N/m); Both the legs of the participant was weaker.

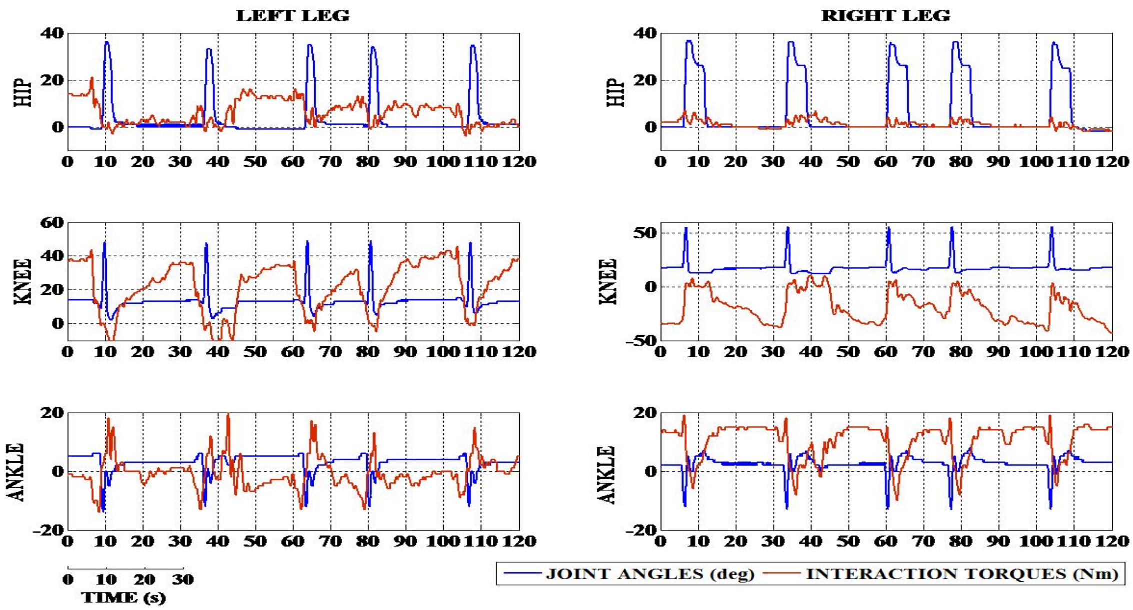

FIGURE B.15: Joint trajectories and interaction torques of the patient P04 with adaptive stiffness; Right leg of the patient was weaker but gradually there was a increase in the interaction torques due to the influence of the adaptive stiffness

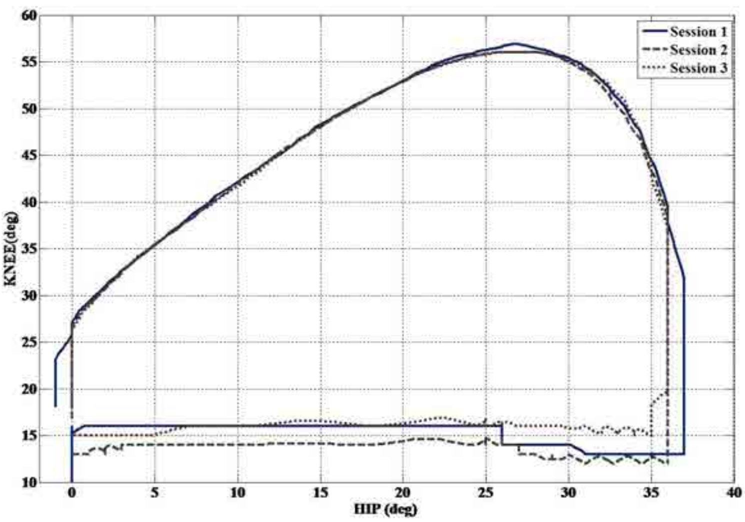

FIGURE B.16: Maximum flexion of right knee of the patient P04, at the end of each session; the Patient applied a compensatory movement in the hip joint which helped the user to achieve a maximum knee flexion in adaptive stiffness mode.

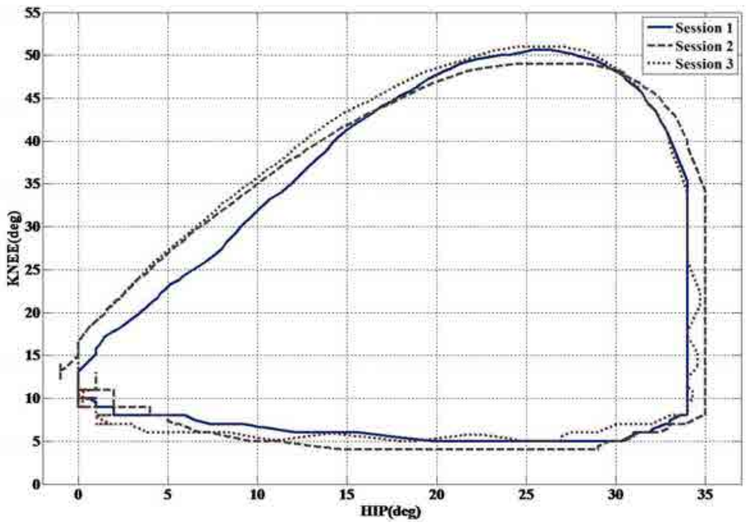

FIGURE B.17: Maximum flexion of Left knee of the patient P04, at the end of each session; the Patient applied a compensatory movement in the hip joint which helped the user to achieve a maximum knee flexion in adaptive stiffness mode.

Appendix C

BMI calibration

The BMI calibration of both the healthy subjects and incomplete SCI patients are presented in this appendix. The complete analysis study involved in the calibration and detection of the movement using the volitional orders has been explained in [240].

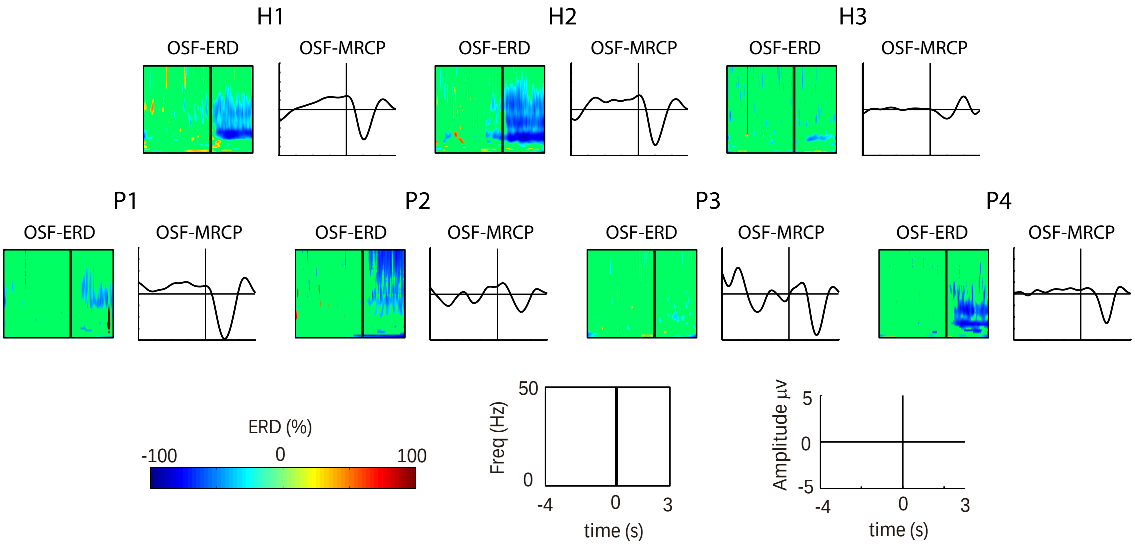

FIGURE C.1: Significant ERD and MRCP for each subject in the channels obtained by applying optimized spatial filtering. For each of the 7 subjects (the 3 healthy subjects on top and the 4 patients at the bottom), the left plot shows the ERD, and the right plot shows the MRCP.

Bibliography

[1] J L Pons, J C Moreno, F J Brunetti, and E Rocon. Lower-limb wearable exoskeleton. In S S Kommu, editor, *Rehabilitation Robotics*. I-Tech Education and Publishing, 2007.

[2] C Azevedo, P Poignet, and B Espiau. Artificial locomotion control: from humans to robots. *Robotics and Autonomous Systems*, 47(4):203–223, 2004.

[3] H Hauser, G Neumann, A J Ijspreet, and W Maass. Biologically inspired kinematic synergies provide a new paradigm for balance control of humanoid robots. In *Proceedings of 7th IEEE-RAS International Conference on Humanoid Robots*, 2007.

[4] J L Pons, editor. *Wearable Robots: Biomechatronic Exoskeletons*. John Wiley and Sons Ltd, USA, 2008.

[5] *Neurological Disorders: public health challenges*. 2006.

[6] N M Crewe and J S Krause. Spinal cord injury. In M G Brodwin, F W Siu, J Howard, and E R Brodwin, editors, *Medical, Psychological and Vocational Aspects of Disability*, pages 289–303. Elliott and Fitzpatrick Inc, 2009.

[7] National Spinal Cord Injury statistical center. Annual report for the model spinal cod injury care systems. *Birmingham, AL*, 2005. URL http://www.spinalcord.uab.edu.

[8] I Diaz, J J Gil, and E Sanchez. Lower-limb robotic rehabilitation: Literature review and challenges. *Journal of Robotics*, 2011.

[9] K.H. Low. Robot-assisted gait rehabilitation: From exoskeletons to gait systems. In *Defense Science Research Conference and Expo (DSR), 2011*, pages 1–10, 2011.

[10] W Young. Spinal cord injury levels and classification. 2006. URL http://www.sci-info-pages.com/levels.html.

[11] J W McDonald and C Sadowsky. Spinal-cord injury. *The LANCET*, 359:417–425, 2002.

[12] B Winter and H Pattani. Spinal-cord injury. *Anesthesia and Intensive care Medicine*, 12(9):403–405, 2011.

[13] D A Winter. Human balance and posture control during standing and walking. *Gait & Posture*, 3:193–214, 1993.

[14] W H Gage, D A Winter, and J S Frank. Kinematic and kinetic validity of the inverted pendulum model in quiet standing. *Gait & Posture*, 19(5):124–132, 2004.

[15] A D Kuo and F E Zajac. Human standing posture: multi-joint movement strategies based on biomechanical constraints. *Progress in brain research*, 97:349–358, 1993.

[16] D A Winter, A Patla, and F Prince. Stiffness control of balance during quiet standing. *Gait & Posture*, 5:154–155, 1997.

[17] D A Winter, A Patla, M Ishac, and W H Gage. Motor mechanism of balance during quiet standing. *Journal of Electromyogrpahy & Kinesiology*, 13(3):49–56, 2003.

[18] L Marchal-Crespo and D J Reinkensmeyer. Review of control strategies for robotic movement training after neurologic injury. *Journal of NeuroEngineering and Rehabilitation*, 6(20), 2009.

[19] S Oh, E Baek, S-ki Song, S Mohammed, D Jeon, and K Kong. A generalized control framework of assistive controllers and its application to lower limb exoskeletons. *Robotics & Autonomous Systems*, online, 2014.

[20] T G Hornby, D D Campbell, J H Kahn, T Demott, J L Moore, and Roth H R. Enhanced gait-related improvements after therapist-versus robotic.assisted locomotor training in subjects with chronic stroke: a randomized controlled study. *Stroke*, 39: 1786–1792, 2008.

[21] D Erol and N Sarkar. Intelligent control for robotic rehabilitation after stroke. *Journal on Intelligent Robot Systems*, 50:341–360, 2007.

[22] S Wehner and M Bennewitz. Humanoid gait optimization based on human data. *Automatika*, 52(3):223–232, 2011.

[23] H Krebs, J J Palazzolo, L Dipietro, M Ferraro, J Krol, K Rannekleiv, and N Hogan. Rehabilitation robotics: Performance-based progressive robot assisted therapy. *Autonomous Robots*, 15:7–20, 2003.

[24] S K Banala, S K Agrawal, and J P Scholz. Active leg exoskeleton (ALEX) for gait rehabilitation of motor-impaired patients. In *IEEE 10th International Conference on Rehabilitation Robotics*, pages 401–407, 2007.

[25] R Riener, L Lünenburger, S Jezernik, M Anderschitz, G Colombo, and V Dietz. Patient-cooperative strategies for robot aided treadmill training: First experimental results. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 13 (3):380–394, 2005.

[26] L I Cai, A J Fong, C K Otoshi, Y Liang, J W Burdick, R R Roy, and Edgerton V R. Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning. *Journal of Neuroscience*, 26(41): 10564–10568, 2006.

[27] Y Stauffer, Y Allemand, M Bouri, J Fournier, R Clavel, R Metrailler, R Brodard, and F Reynard. The walktrainer - a new generation of walking reeducation device combining orthoses and muscle stimulation. *IEEE Neural Systems and Rehabilitation Engineering*, 17:38–45, 2009.

[28] H Vallery, E H F vas Asseldonk, M Buss, and H van der Kooij. Reference trajectory generation for rehabilitation robots: Complementary limb motion estimation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 17(1):23–30, 2009.

[29] L Lünenburger, G Colombo, R Riener, and V Dietz. Biofeedback in gait training with the robotic orthosis Lokomat. In *Proceedings of the 26th Annual International Conference of the IEEE EMBS*, 2004.

[30] L Lünenburger, G Colombo, and R Riener. Biofeedback for robotic gait rehabilitation. *Journal of NeuroEngineering and Rehabilitation*, 4(1), 2007.

[31] D Aoyagi, WE Ichinose, SJ Harkema, DJ Reinkensmeyer, and JE Bobrow. A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 15:387–400, 2007.

[32] L E Kahn, P S Lum, W Z Rymer, and D J Reinkensmeyer. Robot-assisted movement training for the stroke-impaired arm: does it matter what the robot does? *Journal of Rehabilitation Research*, 43(5):619–629, 2006.

[33] J F Veneman, R Ekkelenkamp, R Kruidhof, F C T van der Helm, and H van der Kooij. A series elastic and bowden -cable-based actuation system for use as torque actuator in exoskeleton-type robots. *International Journal on Robotic Research*, 25: 261–281, 2006.

[34] L Marchal-Crespo and D J Reinkensmeyer. Haptic guidance can enhance motor learning of a steering task. *Journal of Motor Behaviour*, 40:545–557, 2008.

[35] S Hussain, S Q Xie, and G Liu. Robot assisted treadmill training: Mechanisms and training strategies. *Medical Engineering and Physics*, 33:527–533, 2011.

[36] J L Emken and D J Reinkensmeyer. Robot-enhanced motor learning: Accelerating internal model formation during locomotion by transient dynamic amplification. *IEEE Transaction on Neural systems and Rehabilitation engineering*, 13(1):33–39, 2005.

[37] J L Emken, R Benitez, and D J Reinkensmeyer. Human-robot cooperative movement training: learning a novel sensory motor transformation during walking with robotic assistance-as-needed. *Journal of NeuroEngineering and Rehabilitation*, 4, 2007.

[38] J L Patton, M E Stoykov, M Kovic, and F A Mussa-Ivaldi. Evaluation of robotic training forces that either enhance or reduce error in chronic hemiparetic stroke survivors. *Experimental Brain Research*, 168(3):368–383, 2006.

[39] A Kaelin-Lane, L Sawaki, and L G Cohen. Role of voluntary drive in encoding an elementary motor memory. *Journal of Neurophysiology*, 93(2):1099–1103, 2005.

[40] E T Wolbrecht, V Chan, and D J Reinkensmeyer. Optimizing compliant, model-based robotic assistance to promote neurorehabilitation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 16(3):286–297, 2008.

[41] L Villani and J D Schutter. Force control. In B Siciliano and O Khatib, editors, *Handbook of Robotics*, pages 161–185. Springer, 2008.

[42] M Bernhardt, M Frey, G Colombo, and R Riener. Hybrid force-position control yields cooperative behaviour of the rehabilitation robot LOKOMAT. In *IEEE 9th International Conference on Rehabilitation Robotics*, pages 536–539, 2005.

[43] R Shadmehr and S Wise. *The computational neurobiology of reaching and pointing : a foundation for motor learning.* MIT Press, 2005.

[44] J De Schutter and H van Brussel. Compliant robot motion ii : A control approach based on external control loops. *International Journal on Robot Research*, 7(4):18–33, 1988.

[45] H Kazerooni, T B Sheridan, and P K Houpt. Robust compliant motion for manipulators. part 1: the fundamental concepts of compliant motion. *IEEE Journal of Robotics and Automation*, 2:83–92, 1986.

[46] S Chiaverini and L Sciavicco. The parallel approach to force/position control of robotic manipulators. *IEEE Transactions on Robotics and Automation*, 9:361–373, 1993.

[47] J C Perez Ibarra, W M dos Santos, H I Krebs, and A A G Siqueira. Adaptive impedance control for robot aided rehabilitation of ankle movements. In *Fifth IEEE RAS/EMBS International Conference on Biomedical Robotics & Biomechatronics (BioRob)*, 2014.

[48] J K Salisbury. Active stiffness control of a manipulator in cartesian coordinates. In *19th IEEE Conference on Decision Control*, pages 95–100, 1980.

[49] M Kordasz, K Kuczkowski, and P Sauer. Study on possible control algorithms for lower limb rehabilitation system. In *Proceedings of IEEE RAS/EMBS International Conference on Rehabilitation Robotics (ICORR)*, pages 1–6, 2011.

[50] N Hogan and S P Buerger. Impedance and interaction control. In T S Kurfess, editor, *Robotics and Automation Handbook*, pages 19:1–19:24. CRC Press, 2004.

[51] B Vanderborght, A Albu-Schaeffer, A Bicchi, E Burdet, D G Caldwell, R Carloni, M Catalano, O Eiberger, W Friedl, G Ganesh, M Garabini, M Grebenstein, G Grioli, S Haddadin, H Hoppner, A Jafari, M Laffranchi, D Lefeber, F Petit, S Stramigioli, N Tsagarakis, M Van Damme, R Van Ham, L C Visser, and S Wolf. Variable impedance actuators: A review. *Robotics and Autonomous Systems*, 61:1601–1614, 2013.

[52] O Unluhisarcikli, M Pietrusinski, B Weinberg, P Bonato, and C Mavroidis. Design and control of a robotic lower extremity exoskeleton for gait rehabilitation. In

IEEE/RSJ International Conference on Intelligent robots and systems, pages 4893–4898, 2011.

[53] G Aguirre-Ollinger, J E Colgate, M A Peshkin, and A Goswami. Active-impedance control of a lower-limb assistive exoskeleton. In *IEEE 10th International Conference on Rehabilitation Robotics*, pages 188–195, 2007.

[54] B T Volpe, H I Krebs, N Hogan, O I Edelstein, C Diels, and M Aisen. A novel approach to stroke rehabilitation: robot aided sensorimotor stimulation. *Neurology*, 54:1938–1944, 2000.

[55] M Cempini, S M Maria De Rossi, T Lenzi, N Vitiello, and M C Carrozza. Self-alignment mechanisms for assistive wearable robots: A kinetostatic compatibility method. *IEEE Transactions on Robotics*, 29(1):236–250, 2013.

[56] J.C. Arevalo and E. Garcia. Impedance control for legged robots: An insight into the concepts involved. *IEEE Transactions on Systems, Man, and Cybernetics, Part C: Applications and Reviews*, 42(6):1400–1411, 2012.

[57] D.A. Lawrence. Impedance control stability properties in common implementations. In *Proceedings of IEEE International Conference on Robotics and Automation*, volume 2, pages 1185–1190, 1988.

[58] E J Rouse, L J Hargrove, E J Perreault, and T A Kuiken. Estimation of human ankle impedance during the stance phase of walking. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 22(4):870–878, 2014.

[59] N Hogan. Impedance control: an approach to manipulation. *ASME Journal on Dynamic System Measurement Control*, 107:1–24, 1985.

[60] N L Tagliamonte, F Sergi, D Accoto, G Carpino, and E Guglielmelli. Double actuation architectures for rendering variable impedance in compliant robots: a review. *Mechatronics*, 22:1187–1203, 2012.

[61] J C Perez Ibarra and A A G Siqueira. Impedance control for rehabilitation robots for lower limbs. review. In *Joint Conference on Robotics: SBR-LARS Robotics Symposium and Robocontrol*, pages 235–240, 2014.

[62] I D Landau, R Lozano, M M'Saad, and A Karimi. Introduction to adaptive control. In *Adaptive control: Algorithms, Analysis and Applications.* Springer, 2011.

[63] T Anwar and A Al Jumaily. Patient cooperative adaptive controller for lower limb robotic rehabilitation device. In *IEEE International Advance Computing Conference (IACC)*, pages 1469–1474. IEEE, 2014.

[64] T Anwar and A Al Jumaily. Adaptive trajectory control based robotic rehabilitation device. In *Middle East Conference on Biomedical Engineering (MECBME)*, pages 269–272. IEEE, 2014.

[65] S Mefoued. A robust adaptive neural control scheme to drive an actuated orthosis for assistance of knee movements. *Neurocomputing*, 140:27–40, 2014.

[66] A Erhan and M A Adli. The design and control of a therapeutic exercise robot for lower limb rehabilitation:physiotherabot. *Mechatronics*, 21:509–522, 2011.

[67] W Hasani, S Mohammed, H Rifaï, and Y Amirat. Powered orthosis for lower limb movements assistance and rehabilitation. *Control Engineering Practice*, 26:245–253, 2014.

[68] S Jezernik, G Colombo, and M Morari. Automatic gait-pattern adaptation algorithms for rehabilitation with a 4-dof robotic orthosis. *IEEE Transactions on Robotics and Automation*, 20(3):574–582, 2004.

[69] K Kiguchi and Y Yokomine. Walking assist for a stroke surivor with a power assist exoskeleton. In *IEEE 11th International Conference on Systems, Man & Cybernetics*, pages 1888–1892, 2014.

[70] J-H Kim, J W Han, D Y Kim, and Y S Baek. Design of a walking assistance lower limb exoskeleton for paraplegic patients and hardware validation using cop. *International Journal of Advanced Robotic Systems*, 10(113), 2012.

[71] T Yan, M Cempini, C M Oddo, and N Vitiello. Review of assistive strategies in powered lower-limb orthoses and exoskeletons. *Robotics and Autonomous Systems*, 64:120–136, 2015.

[72] G Aguirre-Ollinger, J E Colgate, M A Peshkin, and A Goswami. Design of an active one-degree-of-freedom lower limb exoskeleton with inertia compensation. *International Journal of Robotics Research*, 30(4):486–499, 2011.

[73] G Aguirre-Ollinger, J E Colgate, M A Peshkin, and A Goswami. Inertia compensation of a one-degree of- freedom for lower-limb assistance: Initial experiments. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 20(1):68–77, 2012.

[74] K H Low, X Liu, C H Goh, and H Yu. Locomotive control of a wearable lower exoskeleton for walking enhancement. *Journal of Vibration and control*, 12(12):1311–1336, 2006.

[75] K H Low, X Liu, and H Yu. Design of an assistive gait device for strength endurance and rehabilitation. In M Hackel, editor, *Humanoid robots: Human-like machines*. I-Tech, 2007.

[76] R J Wong and J A Smith. Regenerative effects in the sit-to-stand and stand-to-sit movement. *Robotica*, 33:107–126, 2015.

[77] H Takemura, T Onodera, D Ming, and H Mizoguchi. Design and control of a wearbale stewart platform-type ankle-foot assistive device. *International Journal of Advanced Robotics Systems*, 9, 2012.

[78] W Meng, Y Zhu, Z Zhou, K Chen, and Q Ai. Active interaction control of a rehabilitation robot based on motion recognition and adaptive impedance control. In *IEEE International Conference on Fuzzy Systems*, pages 1436–1441, 2014.

[79] C. Fleischer, C. Reinicke, and G. Hommel. Predicting the intended motion with emg signals for an exoskeleton orthosis controller. In *IEEE/RSJ International Conference on Intelligent Robots & Systems (IROS)*, pages 2029–2034, 2005.

[80] R Jimenez-Fabian and O Verlinden. Review of control algorithms for robotic ankle systems in lowerlimb orthoses, prostheses and exoskeletons. *Medical Engineering and Physics*, 34:397–408, 2012.

[81] M R Tucker, J Olivier, A Pagel, H Bleuler, M Bouri, O Lambercy, J del R Millan, R Riener, H Vallery, and R Gassert. Control strategies for active lower extremity prosthetics and orthotics: a review. *Journal of NeuroEngineering and Rehabilitation*, 12(1), 2015.

[82] R Lu, Z Li, C-Yi Su, and A Xue. Development and learning control of a human limb with a rehabilitation exoskeleton. *IEEE Transactions on Industrial Electronics*, 61 (7):3776–3785, 2014.

[83] M Goršič, R Kamnik, L Ambrožič, N Vitiello, D Lefeber, G Pasquini, and M Munih. Online phase detection using wearable sensors for walking with a robotic prosthesis. *Sensors*, 14:2776–2794, 2014.

[84] D Novak, P Reberšek, S M Maria de Rossi, M Donati, J Podobnik, T Beravs, T Lenzi, N Vitiello, M C Carozza, and M Munih. Automated detection of gait initiation and termination using wearable sensors. *Medical Engineering and Physics*, 35:1713–1720, 2013.

[85] E Aertbeliën and J De Schutter. Learning a predictive model of human gait for the control of a lower-limb exoskeleton. In *5th IEEE RAS/EMBS International Conference on Biomedical Robotics & Biomechatronics (BioRob)*, pages 520–525, 2014.

[86] R Williamson and B J Andrews. Gait event detection for FES using accelerometers and supervised learning. *IEEE Transactions on Rehabilitation Engineering*, 8(3): 312–319, 2000.

[87] P Rebersek, D Novak, J Podobnik, and M Munih. Intention detection during gait initiation using supervised learning. In *IEEE 11th International Conference on humanoid robots*, pages 34–39, 2011.

[88] R Begg and J Kamruzzaman. A machine learning approach for automated recognition of movement patterns using basic, kinetic and kinematic gait data. *Journal of Biomechanics*, 38:401–408, 2005.

[89] J Bae and M Tomizuka. Gait phase analysis based on a hidden markov model. *Mechatronics*, 21:961–970, 2011.

[90] J Bae and M Tomizuka. A gait rehabilitation strategy inspired by an iterative learning algorithm. *Mechatronics*, 22:213–221, 2012.

[91] M A Gomes, G L M Silveira, and A A G Siqueira. On-line trajectory adaptation for active lower limbs orthoses based on neural networks. In *Proceedings of International Congress of Mechanical Engineering*, 2009.

[92] Hsiu-Chin Lin, M Howard, and S Vijayakumar. A novel approach for generalising walking gaits across embodiments and behaviours. In *IEEE 5th RAS/EMBS International conference on Biomedical robotics & Biomechatronics (BioRob)*, pages 1009–1015, 2014.

[93] K Anam and A A Al-Jumaily. Active exoskeleton control systems: State of the art. *Procedia Engineering*, 41:988–994, 2012.

[94] J Cao, S Q Xie, R Das, and G L Zhu. Control strategies for effective robot assisted gait rehabilitation: the state of the art and future prospects. *Medical Engineering and Physics*, 2014.

[95] A Koenig, X Omlin, J Bergmann, L Zimmerli, M Bolliger, F Müller, and R Riener. Controlling patient participation during robot-assisted gait training. *Journal of NeuroEngineering and Rehabilitation*, 8(14), 2011.

[96] M J Johnson, K J Wisneski, J Anderson, D Nathan, E Strachota, J Kosasih, J Johnston, and R O Smith. Task-oriented and purposeful robot-assisted therapy. In S S Kommu, editor, *Rehabilitation Robotics*, pages 221–242. I-Tech Education and Publishing, 2007.

[97] S Kousidou, N G Tsagarakis, C Smith, and D G Caldwell. Task oriented biofeedback systems for the rehabilitation of the upper limb. In *IEEE 10th International Conference on Rehabilitation Robotics*, pages 376–384, 2007.

[98] H Sveistrup. Motor rehabilitation using virtual reality. *Journal of NeuroEngineering and Rehabilitation*, 1(10), 2004.

[99] A W Dromerick, P S Lum, and J Hidler. Activity based therapies. *NeuroRx: The Journal of the American Society for experimental Neurotherapeutics*, 3(4):428–438, 2006.

[100] V S Huang and J W Krakauer. Robotic neurorehabilitation: a computational motor learning perspective. *Journal of NeuroEngineering and Rehabilitation*, 6(5), 2009.

[101] K V Kammen, A Boonstra, H Reinders-Messelink, and R den Otter. The combined effects of body weight support and gait speed on gait related muscle activity: A comparison between walking in the lokomat exoskeleton and regular treadmill walking. *PLOS one*, 9(9), 2014.

[102] G Colombo, M Joerg, R Schreier, and V Dietz. Treadmill training of paraplegic patients using a robotic orthosis. *Journal of Rehabilitation Research and Development*, 37:693–700, 2000.

[103] Wang P, KH Low, A Tow, and PH Lim. Initial system evaluation of anoverground rehabilitation gait training robot (nature-gaits). *Advanced Robot*, 25:1927–1948, 2011.

[104] S Hussain, S Q Xie, and P K Jamwal. Control of a robotic orthosis for gait rehabilitation. *Robotics and Autonomous Systems*, 61:911–919, 2013.

[105] J F Veneman, R Kruidhof, E E G Hekman, R Ekkelenkamp, E H F Van Asselsdonk, and H van der Kooij. Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 15(3):379–386, 2007.

[106] H Barbeau and M Visintin. Optimal outcomes obtained with body-weight support combined with treadmill training in stroke subjects. *Archives of Physical Medicine and Rehabilitation*, 84(10):1458–1465, 2003.

[107] K J McCain, F E Pollo, B S Baum, S C Coleman, S Baker, and Smith P S. Locomotor treadmill training with partial body weight support before overground gait in adults with acute stroke: a pilot study. *Archives of Physical Medicine and Rehabilitation*, 89(4):684–691, 2008.

[108] X Xiao, D Huang, and B O'Young. Gait improvement after treadmill training in ischemic stroke survivors: A critical review of functional mri studies. *Neural Regenration Research*, 7(31):2457–2464, 2012.

[109] S Jezernik, R G Wassink, and T Keller. Sliding mode closed loop control of fes:controlling the shank movement. *IEEE Transactions on Biomedical Engineering*, pages 263–272, 2004.

[110] A Domingo and T Lam. Reliability and validity of using the lokomat to assess lower limb joint position sense in people with incomplete spinal cord injury. *Journal of NeuroEngineering and Rehabilitation*, 11(167), 2014.

[111] A Esclarin-Ruz, M Alcobendas-Maestro, R Casado-Lopez, G Perez-Mateos, M A Florido-Sanchez, E Gonzalez-Valdizan, and J L Martin. A comparison of robotic walking therapy and conventional walking therapy in individuals with upper versus lower motor neuron lesions: a randomized controlled trial. *Archives of Physical Medicine and Rehabilitation*, 95(6):1023–1031, 2014.

[112] J Hidler, W Wisman, and N Neckel. Kinematic trajectories while walking withinthe lokomat robotic gait-orthosis. *Clinical Biomechanics*, 23:1251–1259, 2008.

[113] H van der Kooij, J Veneman, and R Ekkelenkamp. Design of a compliantly actuated exo-skeleton for an impedance controlled gait trainer robot. In *28th Annual International Conference of Engineering in Medicine and Biology Society (EMBC)*, pages 189–193, 2006.

[114] H van der Kooij, J F Veneman, and R Ekkelenkamp. Compliant actuation of exoskeletons. In A Lazinica, editor, *Mobile Robots, Towards new applications*, pages 129–148. Sage Publications, 2006.

[115] H van der Kooij, B Koopman, and E H F van Asseldonk. Body weight support by virtual model control of an impedance controlled exoskeleton LOPES for gait training. In *30th Annual International Conference of Engineering in Medicine and Biology Society (EMBC)*, pages 1969–1972, 2008.

[116] R Ekkelenkamp, P Veltink, S Stramigioli, and H van der Kooij. Evaluation of a virtual model control for the selective support of gait functions using an exoskeleton. In *Proceedings of IEEE 10th International Conference of on Rehabilitation Robotics*, pages 1–7, 2007.

[117] B Koopman, E H F van Assesldonk, and H van der Kooij. Selective control of gait subtasks in robotic gait training: foot clearance support in stroke survivors with a powered exoskeleton. *Journal of NeuroEngineering and Rehabilitation*, 10(3), 2013.

[118] H Vallery, J F Veneman, E H F Van Asselsdonk, R Ekkelenkamp, Buss M, and H van der Kooij. Compliant actuation of rehabilitation robots. *IEEE Robotics and Automation Magazine*, pages 60–69, 2008.

[119] K N Winfree, P Stegall, and S K Agrawal. Design of a minimally constraining, passively supported gait training exoskeleton: ALEX II. In *IEEE 12th International Conference on Rehabilitation Robotics*, pages 1–6, 2011.

[120] T Lenzi, M C Carrozza, and S K Agrawal. Powered hip exoskeletons can reduce the user's hip and ankle muscle activations during walking. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 21(6):938–948, 2013.

[121] S K Banala, S K Agrawal, S H Kim, and J P Scholz. Novel gait adaptation and neuromotor training results using an active leg exopskeleton. *IEEE/ASME Transactions on Mechatronics*, 15(2):216–225, 2010.

[122] S Hussein and J Krüger. Patient adaptive control of end-effecter based gait rehabilitation devices using a haptic control framework. In *Proceedings of IEEE International Conference on Rehabilitation Robotics*, 2011.

[123] X Liu and K Xi. Feedback- assisted iterative learning model predictive control with nonlinear fuzzy control. *Mathematical problems in engineering*, 2014.

[124] M Guidali, P Schlink, A Duschau-Wicke, and R Riener. Online learning and adaptation of patient support during ADL training. In *Proceedings of the IEEE International Conference on Rehabilitation Robotics*, pages 1–6, 2011.

[125] Y Wang, J Tuo, Z Zhao, and F Gao. Optimal structure of learning-type set-point in various set-point-related indirect ILC algorithms. *Industrial and Engineering Chemistry Research*, 50:13427–13434, 2011.

[126] J L Emken, J H Wynne, S J Harkema, and D J Reinkensmeyer. A robotic device for manipulating human stepping. *IEEE Transactions on Robotics*, 22(1):185–189, 2006.

[127] D Reinkensmeyer, D Aoyagi, J Emken, J Galvez, W Ichinose, G Kerdanyan, J Nessler, S Maneekobkunwong, B Timoszyk, K Vallance, J Weber, R Wynne, R de Leon, J Bobrow, S Harkema, and V Edgerton. Robotic gait training: Toward more natural movements and optimal training algorithms. In *Proceedings of 26th Annual International Conference of Engineering in Medicine and Biology Society (EMBC)*, pages 4818–4821, 2004.

[128] D J Reinkensmeyer, J L Emken, and S C Cramer. Robotics, motor learning and neurologic recovery. *Annual Review Biomedical Engineering*, 6:497–525, 2004.

[129] S Hesse and D A Uhlenbrock. A mechanized gait trainer for restoration of gait. *Journal of Rehabilitation Research & Development*, 37(701), 2000.

[130] H Schmidt, C Werner, R Bernhardt, S Hesse, and J Kruger. Gait rehabilitation machines based on programmable footplates. *Journal of NeuroEngineering and Rehabilitation*, 4:2–7, 2007.

[131] H Schmidt, S Hesse, R Bernhardt, and J Kruger. Hapticwalker - a novel haptic foot device. *ACM Transactions on Applications and Perception*, 2:166–180, 2005.

[132] S Hesse, A Waldner, and C Tomelleri. Innovative gait robot for the repetitive practice of floor walking and stair climbing up and down in stroke patients. *Journal of NeuroEngineering and Rehabilitation*, 7:30–39, 2010.

[133] P Metrailler, R Brodard, Y Stauffer, R Clavel, and R Frischknecht. Cyberthrosis : Rehabilitation robotics with controlled electrical muscle stimulation. In S S Kommu, editor, *Rehabilitation Robotics*. I-Tech Education and Publishing, 2007.

[134] M Vukobratovic, D Hristic, and Z Stojiljkovic. Development of active anthropomorphic exoskeletons. *Medical & Biological Engineering*, pages 66–80, 1974.

[135] M Vukobratovic. Active exoskeletal systems and beginning of the development of humanoid robotics. *Advances in Nonlinear Sciences II- Sciences and Applications*, 2: 329–348, 2008.

[136] R Bogue. Robotic exoskeletons: a review of recent progress. *Industrial Robot: An International Journal*, 42(1):5–10, 2015.

[137] A M Dollar and H Herr. Lower extremity exoskeletons and active orthoses: Challenges and state of the art. *IEEE Transactions on Robotics*, 24(1):144–158, 2008.

[138] A Waldner, C Werner, and S Hesse. Robot assisted therapy in neurorehabilitation. *EUR MED PHYS*, 44(1-3), 2008.

[139] S Mohammed and Y Amirat. Towards intelligent lower limb wearble robots: Challenges and perspectives - state of the art. In *Proceedings of IEEE International Conference on Robotics & Biomimetics*, pages 312–317, 2008.

[140] D J Reinkensmeyer and M L Boninger. Technologies and combination therapies for enhancing movement training for people with a disability. *Journal of NeuroEngineering and Rehabilitation*, 9(17), 2012.

[141] Z Taha, A P P Abdul Majeed, M Y W Paul Tze, and A G A Rahman. Preliminary investigation on the development of a lower extremity exoskeleton for gait rehabilitation: A clinical consideration. *Journal of Medical and Bioengineering*, 4(1):1–5, 2015.

[142] R Riener, L Lünenburger, and G Colombo. Human centered robotics applied to gait training and assessment. *Journal of Rehabilitation Research and Development*, 43(5): 679–694, 2006.

[143] A Duschau-Wicke, A Caprez, and R Riener. Patient-cooperative control increase active participation of individuals with sci during robot-aided gait training. *Journal of NeuroEngineering and Rehabilitation*, 7(43), 2010.

[144] H Kazerooni, J L Racine, L Huang, and R Steger. On the control of the berkeley lower extremity exoskeleton BLEEX. In *IEEE International Conference on Robotics and Automation, ICRA*, 2005.

[145] H Kazerooni, R Steger, and L Huang. Hybrid control of the berkley lower extremity exoskeleton BLEEX. *International Journal of Robotics Research*, 25(5-6):561–573, 2006.

[146] H A Quintero, R J Farris, C Hartigan, I Clesson, and M Goldfarb. A powered lower limb orthosis for providing legged mobility in paraplegic individuals. *Topics in Spinal Cord Injury Rehabilitation*, 17(1):25–33, 2011.

[147] S A Murray, K H Ha, C Hartigan, and M Goldfarb. An assistive control approach for a lower limb exoskeleton to facilitate recovery of walking following stroke. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 23(3):441–449, 2015.

[148] R J Farris, H A Quintero, and M Goldfarb. Preliminary evaluation of a powered lower limb orthosis to aid walking in paraplegic individuals. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 19:652–659, 2011.

[149] T Hayashi, H Kawamoto, and Sankai Y. Control method of robot suit hal working as operator's muscle using biological nd dynamical information. In *Proceedings of IEEE/RSJ International Conference on Intelligent Robots and Systems*, pages 3063–3068, 2005.

[150] H Kawamoto, S Taal, H Niniss, T Hayashi, K Kamibayashi, K Eguchi, and Y Sankai. Voluntary motion support control of robot suit hal triggered by bioelectrical signal for hemiplegia. In *32nd International Conference of the IEEE EMBS*, 2010.

[151] H Kawamoto, H Kandone, T Sakurai, R Ariyasu, K Eguchi, and Y Sankai. Development of an assist controller with robot suit hal for hemiplegic patients using motion data on the unaffected side. In *36th Annual International Conference of Engineering in Medicine and Biology Society (EMBC)*, pages 3077–3080, 2014.

[152] H Watanabe, N Tanaka, T Inuta, H Saitou, and H Yanagi. Locomotion improvement using a hybrid assistive limb in recovery phase stroke patients: A randomized controlled pilot study. *Archives of Physical Medicine and Rehabilitation*, 95:2006–2012, 2014.

[153] A Tsukahara, Y Hasegawa, K Eguchi, and Y Sankai. Restoration of gait for spinal cord injury patients using hal with intention estimator for preferable swing speed. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 23(2):308–318, 2015.

[154] K A Strausser and H Kazerooni. The development and testing of a human machine interface for a mobile medical exoskeleton. In *IEEE/RSJ International Conference on Intelligent Robots and Systems*, pages 4911–4916, 2011.

[155] K A Strausser, T A Swift, A B Zoss, B C Bennett, and H Kazerooni. Mobile exoskeleton for spinal cord injury: Development and testing. In *4th Annual Dynamic Systems and Controls Conference*, pages 419–425, 2011.

[156] A Esquenazi, M Talaty, A Packel, and M Saulino. The rewalk powered exoskeleton to restore ambulatory function to individuals with thoracic-level motor-complete spinal cord injury. *American Journal of Physical Medicine and Rehabilitation*, 91:911–921, 2012.

[157] M Talaty, , and J E Briceño. Differentiating ability in users of the rewalk powered exoskeleton. In *IEEE International Conference on Rehabilitation Robotics*, 2013.

[158] J A Blaya and H Herr. Adaptive control of a variable-impedance ankle foot orthosis to assist drop-foot gait. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 12(1), 2004.

[159] S M Mizanoor Rahman and R Ikeura. A novel variable impedance compact compliant ankle robot for overground gait rehabilitation and assistance. *Procedia Engineering*, 41:522–531, 2012.

[160] N L Tagliamonte, F Sergi, G Carpino, D Accoto, and E Guglielmelli. Human-robot interaction tests on a novel robot for gait assistance. In *IEEE 13th International Conference on Rehabilitation Robotics*, 2013.

[161] S H Collins, M B Wiggin, and G S Sawicki. Reducing the energy cost of human walking using an unpowered exoskeleton. *Nature*, 522:212–215, 2015.

[162] J E Pratt, B T Krupp, C J Morse, and S H Collins. The roboknee: An exoskeleton for enhancing strength and endurance during walking. In *Proceedings of the IEEE International Conference on Robotics and Automation*, pages 2430–2435, 2004.

[163] S K Agrawal, S K Banala, A Fattah, V Sangwan, V Krishnamoorthy, J P Scholz, and Wei-Li Hsu. Assessment of motion of a swing leg and gait rehabilitation with a gravity balancing exoskeleton. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 15(3):410–420, 2007.

[164] S Wang, L Wang, C Meijneke, E van Asseldonk, T Hoellinger, G Cheron, Y Ivanenko, V La Scaleia, F Sylos-Labini, M Molinari, F Tamburella, I Pisotta, F Thorsteinsson, M Ilzkovitz, J Gancet, Y Nevatia, R Hauffe, F Zanow, and H van der Kooij. Design and control of the mindwalker exoskeleton. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 23(2):277–286, 2015.

[165] J Oh, S Lee, M Lim, and J Choi. A mechanically adjustable stiffness actuator MASA of a robot for knee rehabilitation. In *IEEE International Conference on Robotics & Automation (ICRA)*, pages 3384–3389, 2014.

[166] E Baek, S-ki Song, S Oh, S Mohammed, D Jeon, and K Kong. A motion phase-based hybrid assistive controller for lower limb exoskeletons. In *Advanced Motion Control*, pages 197–202, 2014.

[167] N Costa and D G Caldwell. Control of a biomimetic soft-actuated lower body 10dof exoskeleton. In *Proceedings of the 8th International IFAC Symposium Robot Control SYROCO*, pages 6–8, 2006.

[168] M Wehner, B Quinlivan, P M Aubin, E Martinez-Villalpando, M Baumann, L Stirling, K Holt, R Wood, and C J Walsh. A lightweight soft exosuit for gait assistance. In *IEEE International Conference on Robotics and Automation*, pages 3347–3354, 2013.

[169] D Ferris, J Czerniecki, B Hannaford, and VA Puget Sound Healthcare System. An ankle-foot orthosis powered by artificial pneumatic muscles. *Journal of Applied Biomechanics*, 21(2):189–197, 2005.

[170] K E Gordon and D P Ferris. Learning to walk with a robotic ankle exoskeleton. *Journal of Biomechanics*, 40:2636–2644, 2007.

[171] K Knaepen, P Beyl, S Duerinck, F Hagman, D Lefeber, and R Meeusen. Human-robot interaction: Kinematics and muscle activity inside a powered compliant knee exoskeleton. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 22(6):1128–1137, 2014.

[172] K Kora, A McDaid, and S Xie. Non-linear model based control for the human-inspired robotic exoskeleton (HuREx) gait trainer. In *Proceedings of 19th World congress, International Federation of Automatic Control*, pages 11830–11835, 2014.

[173] H Aguilar-Sierra, R Lopez, W Yu, S Salazar, and R Lozano. A lower limb exoskeleton with hybrid actuation. In *5th IEEE RAS/EMBS International conference on Biomedical Robotics and Biomechatronics (BioRob)*, 2014.

[174] D Maeda, K Tominaga, T Oku, H T T Pham, S Saeki, M Uemura, H Hirai, and F Miyazaki. Muscle synergy analysis of human adaptation to a variable stiffness exoskeleton: Human walk with a knee exoskeleton with pneumatic artificial muscles. In *12th IEEE RAS International Conference on Humanoid Robots*, pages 638–644, 2012.

[175] D Novak and R Riener. A survey of sensor fusion methods in wearable robotics. *Robotics & Autonomous systems*, 73:155–170, 2015.

[176] D Lloyd and T Bessier. An emg driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *Journal of Biomechanics*, 36(6):765–776, 2003.

[177] S K Au, P Bonato, and H Herr. An emg-position controlled system for an active ankle-foot prosthesis: An initial experimental study. In *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics*, pages 375–379, 2005.

[178] J Rosen, M Brand, M Fuchs, and M Arcan. A myosignal-based powered exoskeleton system. *Part A:Systems and Humans, IEEE Transactions on Systems, Man and Cybernetics*, 31(3):210–222, 2001.

[179] F Sylos-Labini, V La Scaleia, A d'Avella, I Pisotta, F Tamburella, G Scivoletto, M Molinari, S Wang, L Wang, E van Asseldonk, H van der Kooij, T Hoellinger, G Cheron, F Thorsteinsson, M Ilzkovitz, J Gancet, R Hauffe, F Zanov, F Lacquaniti, and Y P Ivanenko. Emg patterns during assisted walking in the exoskeleton. *Frontiers in Human Neuroscience*, 8(423), 2014.

[180] A Nilsson, K S Vreede, V Häglund, H Kawamoto, Y Sankai, and J Borg. Gait training early after stroke with a new exoskeleton - the hybrid assistive limb : a study of safety and feasibility. *Journal of NeuroEngineering and Rehabilitation*, 11(92), 2014.

[181] B M Elbagoury, T Schrader, M M Altaf, S A Banawan, and M Roushdy. Design of neural networks for rehabilitation robotics. In *Proceedings of the International Conference on Circuits, Systems and control*, pages 89–95, 2014.

[182] M Aach, O Cruciger, M Sczesny-Kaiser, O Höffken, R Ch Meindl, M Tegenthoff, P Schwenkreis, Y Sankai, and T A Schildhauer. Voluntary driven exoskeleton as a new tool for rehabilitation in chronic spinal cord injury: a pilot study. *The Spine Journal*, 14:2847–2853, 2014.

[183] W van Dijk, B Koopman, R Ronsse, and H van der Kooij. Feed-forward support for human walking. In *Proceedings of the IEEE RAS/EMBS 4th International Conference on Biomedical Robotics & Biomechatronics*, pages 1955–1960, 2012.

[184] Tzu-Hao Huang, Ching-An Cheng, and Han-Pang Huang. Self learning assistive exoskeleton with sliding mode admittance control. In *Proceedings of IEEE/RSJ International Conference on Intelligent Robots & Systems (IROS)*, pages 698–703, 2013.

[185] T Noda, N Sugimoto, J Furukawa, Masa-aki Sato, Sang-Ho Hyon, and J Morimoto. Brain-controlled exoskeleton robot for BMI rehabilitation. In *12th IEEE-RAS International Conference on Humanoid Robots*, pages 21–27, 2012.

[186] J L Contreras-Vidal, A Presacco, H Agashe, and A Paek. Restoration of whole body movement- toward a noninvasive brain-machine interface system. *IEEE Pulse*, pages 34–37, 2012.

[187] J C Moreno, A J del Ama, A de los Reyes-Guzman, A Gil-Agudo, R Ceres, and J L Pons. Neurorobotic and hybrid management of lower limb motor disorders: a review. *Medical and Biological Engineering and Computing*, 49:1119–1130, 2011.

[188] K H Ha, H A Varol, and M Goldfarb. Volitional control of a prosthetic knee using surface electromyography. *IEEE Transactions on Biomedical Engineering*, 58(1):144–151, 2011.

[189] J Gancet, M Ilzkovitz, E Motard, Y Nevatia, S Wang, F Zanow, R Hauffe, and F Thorsteinsson. MINDWALKER: Going one step further with assistive lower limbs

exoskeleton for sci condition subjects. In *Fourth IEEE RAS/EMBS International Conference on Biomedical Robotics & Biomechatronics*, 2012.

[190] M Hasan, H Kadone, K Suzuki, and Y Sankai. Wearable gait measurement system with an instrumented cane for exoskeleton control. *Sensors*, 14:1705–1722, 2014.

[191] D G Liebermann, A S Buchman, and I M Franks. Enhancement of motor rehabilitation through the use of information technologies. *Clinical Biomechanics*, 21:8–20, 2006.

[192] S-C Yeh, M-C Huang, Pa-C Wang, Te-Y Fang, Mu-C Su, Po-Yi Tsai, and A Rizzo. Machine learning based assessment tool for imbalance and vestibular dysfunction with virtual reality rehabilitation system. *Computer methods and programs in Biomedicine*, 116:311–318, 2014.

[193] A Koenig, X Omlin, J Bergmann, L Zimmerli, M Sapa, C Krewer, M Bolliger, F Muller, and R Riener. Psychological state estimation from physiological recordings during robot-assisted gait rehabilitation. *Journal of Rehabilitation Research & Development*, 48(4):367–386, 2011.

[194] A Koenig, D Novak, X Omlin, M Pulfer, E Perreault, L Zimmerli, M Mihelj, and R Riener. Real-time closed-loop control of cognitive load in neurological patients during robot-assisted gait training. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 19(4):453–464, 2011.

[195] R Labruyere and H J A van Hedel. Strength training versus robot-assisted training after incompleet spinal cord injury: a randomized pilot study in patients depending on walking assistance. *Journal of NeuroEngineering and Rehabilitation*, 11(4), 2014.

[196] M M Mirbagheri, M Ladouceur, H Barbeau, and R E Kearney. The effects of long-term fes assisted walking on intrinsic and reflex dynamic stiffness in spastic spinal-cord-injured subjects. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 10(4):280–289, 2002.

[197] C Senanayake and S Senanayake. Emerging robotics devices for therapeutic rehabilitation of the lower extremity. In *IEEE/ASME International Conference on Advanced Intelligent Mechatronics*, pages 1142–1147, 2009.

[198] C T Freeman, E Rogers, A-M Hughes, J H Burridge, and K L Meadmore. Iterative learning control in health care. *IEEE Control systems Magazine*, pages 18–43, 2012.

[199] C T Freeman. Newton-method based iterative learning control for robot-assisted rehabilitation using fes. *Mechatronics*, 2014.

[200] I Cikajilo and T Bajd. Fes gait re-education: The swing phase estimation. *Neuromodulation: Technology at the Neural Interface*, pages 122–127, 2003.

[201] A J Del-Ama, A D Koutsou, J C Moreno, A de los.Reyes, A Gil-Agudo, and J L Pons. Review of hybrid exoskeletons to restore gait following spinal cord injury. *Journal of Rehabilitation Research and Development*, 49(4):497–514, 2012.

[202] A J Del-Ama, J C Moreno, A Gil-Agudo, and J L Pons. Hybrid fes-robot cooperative control of ambulatory gait rehabilitation exoskeleton for spinal cord injured users. In J L Pons, D Torricelli, and M Pajaro, editors, *Converging Clinical and Engineering Research in Neurorehabilitation*, pages 155–159. Springer, 2013.

[203] M Bortole, A J del Ama, E Rocon, J C Moreno, F Brunetti, and J L Pons. A robotic exoskeleton for overground gait rehabilitation. In *2013 IEEE/RSJ International Conference on Robotics and Automation*, 2013.

[204] M Bortole and J L Pons. Development of an exoskeleton for lowerlimb rehabilitation. In J L Pons, D Torricelli, and M Pajaro, editors, *Converging Clinical and Engineering Research in Neurorehabilitation*, pages 85–90. Springer, 2013.

[205] M Bortole. *Robotics Exoskeleton with an Assist-As-Needed Control Strategy for Gait Rehabilitation after Stroke.* PhD thesis, Universidad Carlos III de Madrid, 2014.

[206] A J Del-Ama, E Bravo-Esteban, J C Moreno, J Gomez-Soriano, S Piazza, A D Koutsou, J L Pons, and A Gil-Agudo. Knee muscle fatigue estimation during isometric artificially elicited contractions in incomplete spinal cord injury subjects. In J L Pons, D Torricelli, and M Pajaro, editors, *Converging Clinical and Engineering Research in Neurorehabilitation*, pages 327–332. Springer, 2013.

[207] T Nef, M Mihelj, and R Riener. Armin: a robot for patient-cooperative arm therapy. *Medical and Biological Engineering and Computing*, 45:887–900, 2007.

[208] D Erol, V Mallapragada, N Sarkar, G Uswatte, and E Taub. A new control approach to robot assisted rehabilitation. In *IEEE 9th International Conference on Rehabilitation Robotics*, pages 323–328, 2005.

[209] C Thomas and B Ross. Distinct patterns of motor unit behavior during muscle spasms in spinal cord injured subjects. *Journal of Neurophysiology*, 77(5):2847–2850, 1997.

[210] A Kralj, R Acimovic, and U Stanic. Enhancement of hemiplegic patient rehabilitation by means of functional electrical stimulation. *Prosthetics and Orthotics International*, 17(2):107–114, 1993.

[211] M H Rahman, M Saad, C Ochoa-Luna, J P Kenne, and P S Archambault. Cartesian trajectory tracking of an upper limb exoskeleton robot. In *IECON 2012-38th Annual Conference on IEEE Industrial Electronics Society*, pages 2668–2673, 2012.

[212] H Yali and W Xingsong. Kinematics analysis of lower extremity exoskeleton. In *Control and Decision Conference*, pages 2837–2842, 2008.

[213] D Chugo, H Kaetsu, N Miyake, K Kawabata, H Asama, and K Kosuge. Force assistance system for standing-up motion. In *Proceedings of the 2006 IEEE International Conference on Mechatronics and Automation*, pages 1103–1108, 2006.

[214] Y Hirata, J Higuchi, T Hatsukari, and K Kosuge. Sit-to-stand assist system by using handrail and electric bed moving up and down. In *2nd IEEE RAS EMBS International Conference on Biomedical Robotics and Biomechatronics*, pages 187–192, 2008.

[215] S M Taslim Reza, N Ahmad, I A Choudhury, and R A Raja Ghazilla. A fuzzy controller for lower limb exoskeletons during sit-to-stand and stand-to-sit movement using wearable sensors. *Sensors*, 14:4342–4363, 2014.

[216] K Kim, W Cho, G Yuk, H Yang, B-Rim Jo, and Byung-Hoon Min. Kinematic analysis of sit-to-stand assistive device for the elderly and disabled. In *2011 IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 1–5, 2011.

[217] P. Médèric, V. Pasqui, F. Plumet, and P. Bidaud. Sit to stand transfer assisting by an intelligent walking-aid. In *Climbing and Walking Robots*, pages 1127–1135. Springer Berlin Heidelberg, 2005.

[218] K Junius, B Brackx, V Grosu, H Cuypers, J Geeroms, M Moltedo, B Vanderborght, and D Lefeber. Mechatronic design of a sit-to-stance exoskeleton. In *IEEE RAS/EMBS 5th International Conference on Biomedical Robotics & Biomechatronics (BioRob)*, pages 945–950, 2014.

[219] F B Horak, S M Henry, and A Shumway-Cook. Postural perturbations: New insights for treatment of balance disorders. *Physical Therapy*, 77(5):517–533, 1997.

[220] J Patton, D A Brown, M Peshkin, J J Santos-Munne, A Makhlin, E Lewis, J E Colgate, and D Schwandt. KineAssist: Design and development of a robotic overground gait and balance therapy device. *Topics on Stroke Rehabilitation*, pages 131–139, 2008.

[221] D A Winter, F Prince, and A Patla. Validity of the inverted pendulum model of balance in quiet standing. *Gait & Posture*, 5:153–154, 1997.

[222] H Vallery, A Bögel, C O'Brien, D Li, and R Riener. Robotic assistance for human balance. In *Proceedings of the Dynamic Walking Conference*, 2012.

[223] R A Clark, A L Bryant, Y Pua, P McCrory, K Bennell, and M Hunt. Validity and reliability of the nintendo wii balance board for assessment of standing balance. *Gait & Posture*, 31(3):307–310, 2010.

[224] J J Eng and P F Tang. Gait training strategies to optimize walking ability in people with stroke: A synthesis of the evidence. *Expert review of Neurotherapeutics*, 7(10): 1417–1436, 2007.

[225] R Kobetic, C S To, J R Schnellenberger, M L Audu, T C Bulea, R Gaudio, G Pinault, S Tashman, and R J Triolo. Development of hybrid orthosis for standing, walking and stair climbing after spinal cord injury. *Journal of Rehabilitation Research & Development*, 46(3):447–462, 2009.

[226] A L Behrman and S J Harkema. Locomotor training after human spinal cord injury: A series of case studies. *Physical Therapy*, 80:688–700, 2000.

[227] B M Fleerkotte, B Koopman, J H Buurke, E H F Van Asseldonk, H van der Kooij, and J S Rietman. The effect of impedance controlled robotic gait traning on walking ability and quality in individuals with chronic incomplete spinal cord injury: an explorative study. *Journal of Rehabilitation Research & Development*, 11(26), 2014.

[228] A J Del-Ama, A Gil-Agudo, J L Pons, and J C Moreno. Hybrid fes-robot cooperative control of ambulatory gait rehabilitation exoskeleton. *Journal of Neuroengineering and Rehabilitation*, 11(27), 2014.

[229] J M Belda-Lois, S M Horno, I Bermejo-Bosch, J C Moreno, J L Pons, D Farina, M Iosa, M Molinari, F Tamburella, A Ramos, A Caria, T Solis-Escalante, C Brunner, and M Rea. Rehabilitation of gait after stroke: a review towards a top-down approach. *Journal of NeuroEngineering and Rehabilitation*, 8(66), 2011.

[230] Y Kim. The effects of EMG-triggered functional electrical stimulation on upper extremity function in stroke patients. *Physical therapy Rehabilitation Science*, 2(1): 1–6, 2013.

[231] F Wang, S Wen, C Wu, Y Zhang, and J Li. Motion intention analysis based coordinated control for amputee-prosthesis interaction. *Journal of Robotics*, 2010.

[232] J-Y Jung, Min-Gi Chae, In H Jang, and H Park. A hybrid control method of an exoskeleton robot for intention-driven walking rehabilitation of stroke patients. In *International Conference on Ubiquitous Robots and Ambient Intelligence (URAI)*, pages 58–60, 2012.

[233] E Rocon, A F Ruiz, J L Pons, J M Belda-Lois, and J J Sanchez-Lacuesta. Rehabilitation robotics: a wearable exoskeleton for tremor assessment and suppression. In *Proceedings of IEEE International Conference Robotics & Automation*, pages 2271–2276, 2005.

[234] G Pfurtscheller and F H Lopes de Silva. Event-related EEG/MEG synchronization and desynchronization: basic principles. *Clinical Neurophysiology*, 110:1842–1857, 1999.

[235] H Shibasaki and M Hallett. What is the bereitschaftspotential. *Clinical Neurophysiology*, 117:2341–2356, 2006.

[236] E Lopez-Larraz, J M Antelis, L Montesano, A Gil-Agudo, and J Minguez. Continuous decoding of motor attempt and motor imagery from eeg activity in spinal cord injury patients. In *Engineering in Medicine and Biology Society (EMBC), 2012 34th Annual International Conference of the IEEE*, 2012.

[237] E Lopez-Larraz, L Montesano, A Gil-Agudo, and J Minguez. Continuous decoding of movement intention of upperlimb self initiated analytic movements from premovement eeg correlates. *Journal of NeuroEngineering and Rehabilitation*, 11:153–167, 2014.

[238] E Lopez-Larraz, F Trincado-Alonso, V Rajasekaran, S Pérez-Nombela, A J del Ama, J Aranda, J Minguez, A Gil-Agudo, and L Montesano. Control of an ambulatory exoskeleton with a brain-machine interface for spinal cord injury gait rehabilitation. *Frontiers in Neuroscience*, under review.

[239] F Trincado-Alonso, E Lopez-Larraz, F Resquin, A Ardanza, S Perez-Nombela, J L Pons, L Montesano, and A Gil-Agudo. A feasibility study of hand rehabilitation for spinal cord injury patients using brain triggered feedback based on electrical stimulation and virtual reality. *Medical Engineering and Physics*, under review, 2015.

[240] E Lopez-Larraz. *Brain-machine interfaces for motor rehabilitation and restoration after spinal cord injury.* PhD thesis, University of Zaragoza, Spain, 2015.

[241] H Vallery, A Bogel, C O'Brien, and R Riener. Cooperative control design for robot-assisted balance during gait. *Automatisierungstechnik*, 60(11):715–720, 2012.

[242] R Prashanth Kumar and Jungwon Yoon. Improved stability in lower extremity exoskeletons using foot extensions. In *Proceedings of ICROS-SICE International joint conference*, pages 1212–1217, 2009.

[243] G Capellini, Y P Ivanenko, N Dominici, R E Poppele, and F Lacquaniti. Motor patterns during walking on a slippery walkway. *Journal of Neurophysiology*, 103(2): 746–760, 2010.

[244] A T Asbeck, R J Dyer, A F Larusson, and C J Walsh. Biologically-inspired soft exosuit. In *IEEE International Conference on Rehabilitation Robotics*, pages 1–8, 2013.

[245] R W Bohannon and M B Smith. Interrater reliability of a modified ashworth scale of muscle spasticity. *Physical Therapy*, 67(2):206–207, 1987.

[246] R W Bohannon. Manual muscle test scores and dynamometer test scores of knee extension strength. *Archives of Physical Medicine and Rehabilitation*, 67(2):390–392, 1986.

[247] J R Mendell and J Florence. Manual muscle testing. *Muscle & Nerve*, 13:S16–S20, 1990.

[248] C Fekete, I Eriks-Hoogland, M Baumberger, A Catz, M Itzkovich, H Lüthi, M W Post, E von Elm, A Wyss, and M W Brinkhof. Development and validation of a self-report version of the spinal cord independence measure (SCIM III). *Spinal Cord*, 51 (1):40–47, 2013.

[249] M Itzkovich, I Gelernter, F Biering-Sorensen, C Weeks, MT Laramee, BC Craven, M Tonack, SL Hitzig, E Glaser, G Zeilig, S Aito, G Scivoletto, M Mecci, RJ Chadwick, A El Masry, WSand Osman, CA Glass, P Silva, BM Soni, BP Gardner, G Savic, EM Bergström, V Bluvshtein, J Ronen, and A Catz. The spinal cord independence measure (SCIM) version iii: reliability and validity in a multi-center international study. *Disability and Rehabilitation*, 29(24):1926–1933, 2007.

[250] P L Ditunno and J F Ditunno Jr. Walking index for spinal cord injury (WISCI II): scale revision. *Spinal cord*, 39:654–656, 2001.

[251] A B Jackson, C T Carnel, J F Ditunno, M S Read, M L Boninger, M R Schmeler, S R Williams, and Donovan. W H. Gait and ambulation subcommittee: Outcome measures for gait and ambulation in the spinal cord injury population. *Journal of Spinal Cord Medicine*, 31(5):487–499, 2008.

[252] J F Ditunno, H Barbeau, B H Dobkin, R Elashoff, S Harkema, R J Marino, W W Hauck, D Apple, D M Basso, A Behrman, D Deforge, L Fugate, M Saulino, M Scott, and J Chung. Validity of the walking scale for spinal cord injury and other domains of function in a multi-center clinical trial. *Neurorehabilitation & Neural Repair*, 21 (6):539–550, 2007.

[253] M M Priebe, A M Sherwood, J I Thornby, N F Kharas, and J Markowski. Clinical assessment of spasticity in spinal cord injury: a multidimensional problem. *Archives of Physical Medicine and Rehabilitation*, 77(7):713–716, 1996.

[254] M M Adams, K A Ginis, and A L Hicks. The spinal cord injury spasticity evaluation tool: development and evaluation. *Archives of Physical Medicine and Rehabilitation*, 88(9):1185–1192, 2007.

[255] H J van Hedel, M Wirz, and V Dietz. Assessing walking ability in subjects with spinal cord injury: validity and reliability of 3 walking tests. *Archives of Physical Medicine and Rehabilitation*, 86(2):190–196, 2005.

[256] T Lam, V K Noonam, J J Eng, and S R Team. A systematic review of functional ambulation outcome measures in spinal cord injury. *Archives of Physical Medicine and Rehabilitation*, 88(9):1185–1192, 2007.

[257] G A Borg. Psychophysical bases of perceived exertion. *Medicine and Science in Sports and Exercise*, 14(5):377–381, 1982.

[258] G A Borg. Perceived exertion as an indicator of somatic stress. *Scandinavian journal of rehabilitation medicine*, 2:92–98, 1970.

[259] L Demers, M Monetter, Y Lapierre, D L Arnold, and C Wolfson. Reliability, validity and applicability of the quebec user evaluation of satisfaction with assistive technology (QUEST 2.0) for adults with multiple sclerosis. *Disability Rehabilitation*, 15(24(1-3)):21–30, 2002.

[260] M Wirz, D H Zemon, R Rupp, A Scheel, G Colombo, V Dietz, and T G Hornby. Effectiveness of automated locomtor training in patients with chronic incomplete spinal cord injury: a multicenter trial. *Archives of Physical Medicine and Rehabilitation*, 86 (4):672–680, 2005.

[261] M Lotze, Braun C, N Birbaumer, S Anders, and L G Cohen. Motor learning elicited by voluntary drive. *Brain*, 126(4):866–872, 2003.

[262] C F J Nooijen, N T Hoeve, and E C Field-Fote. Gait quality is improved by locomotor training in individuals with SCI regardless of training approach. *Journal of NeuroEngineering and Rehabilitation*, 6(36), 2009.

[263] M Bortole, A Venkatakrishnan, F Zhu, J C Moreno, G E Francisco, J L Pons, and J L Contreras-Vidal. The h2 robotic exoskeleton for gait rehabilitation after stroke: early findings from a clinical study. *Journal of Neuroengineering and Rehabilitation*, 12(54), 2015.

[264] R J Farris, H A Quintero, S A Murray, K H Ha, C Hartigan, and M Goldfarb. A preliminary assessment of legged mobility provided by a lowerlimb exoskeleton for persons with paraplegia. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 22(3):482–490, 2014.

[265] E J Goldberg and R R Neptune. Compensatory strategies during normal walking in response to muscle weakness and increased hip joint stiffness. *Gait & Posture*, 25: 360–367, 2007.

[266] L Wang, S Wang, E H F van Asseldonk, and H van der Kooij. Actively controlled lateral gait assistance in a lower-limb exoskeleton. In *IEEE/RSJ Interantional Conference on Intelligent Robots & Systems IROS*, pages 965–970, 2013.