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Gait Locomotion Generation and Leg Muscle Evaluation for Overground Walking Rehabilitation Robots

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SCHOOL OF MECHANICAL AND AEROSPACE ENGINEERING

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Abstract

The number of people with difficulty in walking has increased with ageing issues worldwide. It is possible to regain walking ability after persistent locomotion training. In decades, robotic devices have been suggested to enhance motor recovery by replicating clinical manual-assisted training, which is labour-intensive and hard to persistent. This thesis presents a research work on a robotic-assisted training system based on overground walking (OGW), instead of the treadmill training. Proposed motion generation, control and evaluation of the lower limb rehabilitation are specified to the OGW training system for a more natural and effective gait training. In the present work, human motor control is first investigated to avoid a totally passive walking exercise on robotic system. In order to enhance effect of the active training, incorrect muscle activations on lower limbs for the pathological gait are identified. After investigating human motor control for one joint, motion generation algorithms of gait training are proposed for coordinating all joints on the pelvis and lower limbs. For different individual gaits and physical dimensions, an adjustable motion generation algorithm is proposed. Then, the intelligent control methods are also proposed for better tracking the generated motion under different walking speeds. All these proposed motion generation and control methods are implemented on an OGW gait system: *NaTUre-gaits (Natural and TUnable rehabilitation gait system)*. The synchronized control is implemented for the modular systems, i.e. pelvic arm, robotic orthosis and mobile platform. Comparisons of lower limb muscular activation walking with and without the robotic device are studied and discussed. Results of successful clinical-trials show that the performance of the device is able to provide comparable performance obtained by manual assistance by gait rehabilitation training. Further quantifying assessment of muscle dysfunction of human gait is investigated, which works as an effective tool and contributes to qualified evaluation and bio-feedback for gait rehabilitation. In conclusion, the present work has dealt successfully with a full-range of new robotic-assisted gait rehabilitation methodology, which includes human gait locomotion study, development of motion generation and control strategy, system implementations, clinical applications, and quantitative assessment for an OGW rehabilitation training robot.

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ABBREVIATIONS AND ACRONYMS

ADL	activities of daily live
AI	ambulation index
ATV	auto-tuning variation
BBT	Berg balance test
BF	bicep femoris
$B(s)$	biomechanical model
BWS	body weight support
BWSLT	body weight support locomotion training
BWSTT	body weight support treadmill training
CLME	complementary limb motion estimation
CNS	central nervous system
CP	Cerebral Palsy
DGI	dynamic gait index
DS	disease steps
EMG	electromyography
FFC	force-field controller
FWB	full weight bearing
GC	gait cycle
GM	gluteus maximus
$G(s)$	modeling human synovial joint
HC	heel contact
HMI	human machine interaction
IIR	infinite impulse response
OGW	overground walking
PNS	peripheral nervous system
PSD	power spectrum density

MVC	muscle voluntary contraction
MS	multiple sclerosis
MG	gastrocnemius medialis
OT	occupational therapist
OWG	overground walking
PT	physical therapist
RMS	root means square
RF	rectus femoris
SCI	spinal cord injury
SD	standard deviation
SO	soleus
ST	semitendinusus/semimembranous
TA	tibialis anterior
TUG	timed up and go
VMC	virtual model control
VL	vastus laterali
ZMP	zero moment point
A	actuator
C	controller
$C(q, \dot{q})$	matrix of the Coriolis and centrifugal forces
$g(q)$	gravity force
$H(x \cdot)$	dynamics equations
k_p	proportional parameter of PID controller
k_I	integral parameters of PID controller
k_D	derivative parameters of PID controller
$M(q)$	inertia matrix
q	joint position, actual trajectory
q_{dis}	disturbance for the joint motion trajectory
q_e	the error between the desired and actual trajectories

S	the set of the state
τ, Γ	torque
τ_H	muscular torque
τ_A	external assistive torque
u_H	brain commander
u_A	control signal
v	joint velocity
$\dot{v}$	joint acceleration
y_{Desired}, q_D	desired trajectory
y	actual trajectory

Chapter 1 INTRODUCTION

This thesis presents research work on motion generation, control and evaluation of overground gait robot for the lower limb rehabilitation. The current work aims at realization of robotic-assisted training of the pathological gait via control of rehabilitation robot including overground walking (OGW) motion generation, repetitive and adaptive synchronization control algorithm. The generated OGW motion and proposed control algorithms are tested and implemented into the modules of a gait robot, *NaTUre-gaits* (*Natural and TUnable rehabilitation gait system*), self-developed by our rehabilitation team in robotic research center (RRC), Nanyang Technological University (NTU). The performance and evaluations of the system are studied based on the muscle activation analysis upon healthy and subjects with walking problems. The research background, motivation, objective and scope of the thesis will be discussed in this chapter.

1.1 Background

With ageing and obesity rates rising worldwide, the number of people with difficulty in walking has increased. As we know, when getting old or fat, many diseases such as arthritis and diabetes may come along. Arthritis and diabetes are leading risk factors for mobility problem. At the same time, many other diseases and accidental falls increase with ageing and higher obesity rate [1]. Therefore, decreased locomotor ability is frequently encountered by many people with arthritis, diabetes, stroke, amyotrophia, Parkinson's disease, heart and lung diseases, trauma resulting in brain injuries or spinal cord injury (SCI), multiple sclerosis (MS), or other neurological diseases [2].

Walking impairments have an impact on patients and their families, primarily in terms of decreased self-independence, satisfactions and quality of life. For example, lower mobility affects the most trivial activities like going for shower, eating out with friends, going to

work or shopping and traveling abroad, etc. Hence, people with motor disabilities become a great burden on healthcare institutions and their families [3]. In order to regain normal lower-limb function, gait rehabilitation such as locomotion training is commonly employed as therapy, which covers many aspects including physical rehabilitation, gait assessment, as well as medical treatment and surgery.

The key to gait recovery from a clinical perspective is neuroplasticity. Neuroplasticity is the adaptive change in brain structure in response to a change in environment over time with a consequent change in function [4-6]. For example, walking disabilities are commonly encountered after spinal cord injury (SCI) [7-8], these locomotion disorders are attributed to neuronal circuitry being cut off within the spinal cord [9]. Owing to neuroplasticity [2, 10] and experiments carried out on cats [11-12], it is probable to regain the walking ability after persistent locomotion training. Various gait recovery approach and training strategy have been proposed in past years [10, 13-16].

The correlations between environment, adaptation and function are foundational capacity to recover gait function after injury. Major factors related to the gait rehabilitation are summarized in Table 1.1.

In order to highlight these factors, manual-assisted gait rehabilitation is described. Clinically, physical therapist (PT) and occupational therapist (OT) are two kinds of therapists who participate to physical rehabilitation. The PT focuses on the functional recovery of the patient, which is the independent locomotion in daily life. On the other hand, the OT aims to train the subject on how to return to normal daily activities, or even back to work. Although walking is remarkably flexible and easy for healthy people, conventional training procedures require up to about three therapists to lift, support and assist a patient (see Figure 1.1). These clinical exercises are helpful for gait recovery, but it is labour-intensive for therapists. As observed in the gait rehabilitation scenario, in addition to providing body-weight support for the patient to prevent them from falling,

therapists have to manually facilitate the movement for the lower limb to simulate a slow walking pattern with a proper posture. Since therapists have to concurrently support patient's body weight and facilitate propulsion of the legs, the training procedures become clumsy inconsistent and exhausting for all concerned. Even for these patients with trunk control, the therapists still need to constantly monitor them and assist them as needed, see Figure 1.1. In addition, the number of patients assigned to each therapist per day will often be kept to a level which does not exhaust therapists. Due to high labor cost and the shortage of skilled therapists, few patients can afford and benefit from such manual-assisted therapy. The number of clinical visits for gait rehabilitation training is, therefore, limited and insufficient.

Table 1.1 A Brief Summary of Factors Related to Gait Rehabilitation

Major Factors	Examples
<i>Common Gait Disorders</i>	Hip-hiking, circumduction (swing leg out to the side), trunk lean to the side or back, foot drag/slap (two catch), knee-lock, flexed, shuffling, vaulting (to clear the weak leg), veering to one side, etc
<i>Major Causes of Gait Disorders</i>	SCI (spinal cord injury), stroke, Parkinson's disease, etc
<i>Guideline for Effective Gait Training</i>	Repetitive, intensive, goal-orientated, functional, massed, active
<i>Balance and Gait Assessment</i>	Gait observation analysis, timed 10 meters walk, 2/6 minutes walk, ambulation index (AI), dynamic gait index (DGI), timed up and go (TUG), functional reach test, berg balance test (BBT), Tinetti gait and balance assessment

Figure 1.1 Clinical gait rehabilitation: (a) Patient walk with help of a walker [17]; (b) Physical therapists help patient to walk on a treadmill with a body weight support system for off-loading body weight burden [18]; (c) Therapists carefully help the patient with trunk control [19].

1.2 Motivation

To tackle the drawback of manual-assisted gait rehabilitation, it is a significant attempt to take best advantage of robotics integrating with the traditional manual-assisted rehabilitation. The analysis of upper limb function has been well-established in task-oriented rehabilitation for a long time [20]. Clinical rehabilitation for hands can be trained by actions such as pinching, griping, hooking, swinging, pulling and pushing, which focus on enhancing muscle force for better control of upper extremities [3]. However, such analysis is still lacking for lower extremities rehabilitation which aims at integrated walking process. The integrated walking process includes the body weight support (BWS), balance control, trunk control, pelvic control, lower limbs propulsion, correction of foot drop and activities of daily living (ADL)-related locomotion. In order to select suitable training programs, it is important to identify and assess the muscle dysfunction of SCI subjects. However, quantifying assessment of SCI gait is challenging as individual gait is usually changing, asymmetric, slow and instable based on variable nature of EMG signal [14].

As shown in Figure 1.2, robotic-assisted gait rehabilitation covers four major fields of study, i.e. the design of robotic system, the generation of gait pattern (the motion input for the developed robotic system), the control strategy, and the bio-feedback and clinical assessment for rehabilitation purposes.

Figure 1.2 Four major fields of study for robotic-assisted gait rehabilitation.

The enhancement of robotics and sensor technology has helped in the development of robotics rehabilitation gait systems, which can contribute to the fields of neuro-rehabilitation [21-27]. There are various alternative robotic therapies available to stimulate muscle activity pattern effectively with little supervision from therapists [28].

There are various types of gait robots developed for rehabilitation. Rehabilitation robots include treadmill and overground walking training. For example, Lokomat [27] is able to achieve active body weight support (BWS) and assistive locomotion for the patients to walk on a treadmill. LOPES [24, 29] contributed a more flexible actuator for the design of the exoskeleton. KineAssist [25] constructed mechanical arm to support the patient for the overground walking. The WalkTrainer [23] is also an impressive robotic system providing the overground walking by consideration of the pelvic motion. *NaTure-gaits (Natural and TUnable rehabilitation gait system)* is developed by our rehabilitation team in robotic research center (RRC), Nanyang Technological University (NTU) for overground walking (OGW) training [22]. The hybrid assistive limb (HAL) [21, 30] was developed in the

Cybernetics Laboratory at the University of Tsukuba in Japan, which aims at helping overground walking [31]. Schmidt [32] invented a pair of programmable footplates for floor walking and stair climbing. Powered ankle-foot-orthosis [21, 33] have been developed to provide assistance for the ankle joint and correction of drop foot, which allows the user to practice starting, turning, stopping, and avoiding obstacles during overground walking.

Table 1.2 Comparisons of Manual-assisted and Robotic-assisted Gait Rehabilitation

Principles of rehabilitation	Manual-assisted Rehabilitation	Robotic-assisted Rehabilitation
<i>Intensity of Practice</i>	Low, due to physical limitation of therapists	High, since motorized machine do not get tired
<i>Effect of Participation</i>	High, subject will participate actively	Low, if subject remain totally passive during training
<i>Independence of Participation</i>	Low, therapist limit participation of subject for the sake of safety	High, wide range of motion (ROM) with body weight supported by mechanical structure
<i>Specificity of Training (locomotion guidance)</i>	Less specifics since therapist not able to properly guide on the motion	High specific since machine can control the motion precisely
<i>Assessment and Evaluation</i>	Less frequently conducts when charge in/out by experienced therapists	High frequently involved advanced sensory for more prompt and accurate bio-feedback
<i>Training Durations</i>	Short training time due to limited manpower	High since machine can insist repetitive training
<i>Costs</i>	High, paying therapist manpower hours	High, machine setup cost and usage cost
<i>Risk of fall</i>	Low but over constrained manual assisted for body weight support	Low with mechanical support of body weight

The principles of rehabilitation are summarized and a comparison between manual-assisted and robotic-assisted gait rehabilitation are presented in Table 1.2, in which potential benefits for the robotics rehabilitation can be clearly observed.

Figure 1.3 Development of an overground rehabilitation robot for patient groups without trunk control.

NaTure-gaits is an overground rehabilitation robot and developed by our research team. Therefore, it is important to highlight some specifications of overground walking rehabilitation robots. (i) Although there are a wide range of patient groups, robotic device such as *NaTure-gaits* provide a gait training platform for patient groups without trunk control. That is, it is necessary to provide body weight supported and locomotion propulsion at the same time. (ii) The drawback of gait robots operated with partial body weight support on treadmill is that the majority of treadmills available commercially do

not provide proper speeds for patients [28]. Moreover, a stroke patient with hemiparalysis might have time-variance gait as well as unequal step durations for left and right legs, which are impossible to be covered in treadmill training. Furthermore, the muscular activation pattern between OGW and treadmill walking are slightly different [34]. Therefore, the development for OGW rehabilitation training for patient groups without trunk control [35] is required as shown in Figure 1.3.

Figure 1.4 Organization of the research scope and requirement for the overground walking rehabilitation robots.

Except overground walking, it is also found that only few studies [23] considered the control of pelvic motions [36]. However, doctors and therapists have emphasized the

importance of pelvic posture and its role in coordinating overall patient movement. Hence, *NaTure-gaits* is developed and includes pelvic motion control, active hip/knee/ankle assistance mechanism within one gait rehabilitation platform. After developing the prototype, it is essential to solve control issues such as bad coordination of sub-modules which would introduce foot-dragging and ground impact problem etc.

In conclusion, the major contributions are to initialize the overground training robot with pelvic control and active assistance to the lower limb. The research outline of overground walking training robots is demonstrated and expanded based on the four major fields, as shown in Figure 1.4. The gait motion generation, control system and evaluation methods of OGW rehabilitation robots are presented in this thesis for effective and less manual-intensive rehabilitation training as an alternative solution to conventional therapy methods.

1.3 Objectives and Scope

This thesis aims to contribute on gait locomotion generation and leg muscle evaluation for overground walking (OGW) rehabilitation robots. The gait locomotion generation of an adjustable OGW pattern for individuals is proposed together with selecting from assistance to resistance in rehabilitation, smoothed ground contact conditions and tracking performance for the repetitive motion. Then, evaluation of the walking quality with the overground walking rehabilitation robots is also studied for effective gait rehabilitation. In order to achieve the specified objectives, the following works have been accomplished:

1. A rehabilitation robot, *NaTure-gatis*, has been developed for the overground walking training for subjects with pathological gait. A systematic locomotion control and evaluation method are investigated with a more effective and less manual-intensive overground walking rehabilitation training as an alternative to conventional therapy methods. Furthermore, a method is proposed to model the human machine interface (HMI) for an active control.

2. It is challenging to obtain quantifying assessment of walking as individual gait is usually changing, asymmetric, slow and instable based on variable nature of EMG signal [14]. In order to design rehabilitation robot, it is necessary to identify and assess muscle dysfunction of human gait. According to proposed gait analysis method, the distribution of muscle activation power within one gait cycle is investigated for a quantitative assessment of pathological gait. It is an effective tool and likely contributes to qualified evaluation and bio-feedback of gait rehabilitation in future.
3. A method for the motion generation of gait is also proposed for different walking patterns on the gait system. In order to generate the OGW motion for individuals, the trajectory of ankle is then obtained via curve fitting based on specified points. Next, in order to keep the desired foot trajectory, the pelvic motion is compensated to modify hip and knee joint angles. Furthermore, the metatarsal trajectory is planned to satisfy conditions for a smooth OGW. The motion is generated based on both the inverse kinematics and OGW condition. Based on the OGW condition, an adjustable index is employed so as to generate motion that produce impact less and smooth footpad contact.
4. A repetitive controller, which is embedded with a periodic signal generator, is used in the control of this system so as to achieve steady tracking of cyclic reference signals and reject disturbance on the rehabilitation robot. It works as a stabilizer for the HMI model as mentioned. By selecting a suitable stabilizer, function of robotic-assisted system can switch between assistance and resistance for different rehabilitation requirements. Additionally, an autotuning controller is also developed to synchronize the motion of sub-systems.
5. These proposed control methods are implemented on a gait system: *NaTUre-gaits*. The coordinated control is used for the modular systems, i.e. pelvic arm (PA),

robotic orthosis (RO) and mobile platform (MP). Fourteen motors and three control sub-systems are set up for a smooth overground walking motion. Wireless commanders are sent by host computer to sub-systems. After coordination within these three control systems, synchronized motions are conducted. Healthy subjects and therapists have tried out the walking motion on system.

6. Successful clinical trials on the gait system are conducted to validate the proposed models. Initial results show that the performance of the device can provide impact-less overground walking, which is comparable to the performance obtained by manual assistance by gait rehabilitation training. Comparisons of lower limb muscular activation walking with and without the robotic device are studied and discussed. Furthermore, frequency evaluation method is also proposed to distinguish the gait impairment.

1.4 Organization of Thesis

In this thesis, modeling and compensation of the pathologic human motion control are presented and a stabilizer is introduced. First of all, a method is proposed to model the human machine interface (HMI) for healthy and pathological gait. Improved gait analysis is also proposed to assess pathological gait. Secondly, the synchronized motion generation for the exoskeleton modules is provided by virtue of the inverse kinematics analysis and OGW condition. The pelvic trajectory is predefined and key points are specified within one gait cycle to obtain the foot trajectory from the designated step length and height. The joint angles are calculated by the inverse kinematics. Based on the OGW condition, an adjustable index is employed so as to generate motion that produce impact less and smooth footpad contact. Thirdly, an internal model combined with repetitive control is proposed for better tracking performance. It works as a stabilizer according to HMI model. By selecting a suitable stabilizer, function of robotic-assisted system can switch between assistance and resistance for different rehabilitation requirements. The gait model

is investigated for the design of stabilizer used in control the human-machine interface (HMI). Furthermore, in order to enlarge the control zone, the performance of the exoskeleton is identified on-line by the relay feedback test. After modeling of controller and actuator, an autotuning controller is used for an optimized and adaptive control. Successful clinical trials have been conducted for the validation of the developed robotic system via muscle activation analysis.

The roadmap of this thesis is showing in Figure 1.5 and the organization of this thesis is listed as follows:

Chapter 1 has provided the background and introduction to the objective of this research. The motivation, objective and scopes of this research, research outline and organization of the thesis have been stated.

Chapter 2 covers literature review on design consideration related to clinical gait rehabilitation, state-of-the-art of gait rehabilitation devices, motion control for rehabilitation robot and evaluation of gait rehabilitation. This chapter ends with some research issues related to the proposed project.

Chapter 3 provides descriptions of subject, system and environment and the relationship between them. A method is proposed to model the human machine interface (HMI) for healthy and pathological gait. Improved gait analysis is also proposed to assess pathological gait. Four methods are used to achieve improved assessment of gait to contribute on qualified evaluation and bio-feedback for gait rehabilitation.

Chapter 4 provides synchronized motion generation for the gait system by virtue of the inverse kinematics analysis and OGW condition. The pelvic trajectory is predefined and key points are specified within one gait cycle to obtain the foot trajectory from the designated step length and height. Furthermore, to reduce, if not to

avoid, the ground impact and dragging of the footpad occurring during the course of overground walking with the help of the device, a gait motion generation is developed and specified to adjust the gait patterns for an impact-less and smooth footpad contact based on the walking model proposed in Chapter 3.

Chapter 5 provides control theories used in tracking the motion proposed in Chapter 4. In order to improve the tracking performance, an internal model combined with repetitive control. Repetitive controller works as a stabilizer which is designed according to HMI model. By selecting a suitable stabilizer, function of robotic-assisted system can switch between assistance and resistance for different rehabilitation requirements. An autotuning controller is developed by identification of the relay feedback test for a better tracking performance. Simulation examples and control implementation in Chapter 6 on the system are provided to demonstrate the effect of these intelligent control algorithms.

Chapter 6 introduces *NaTure-gaits*, which covers Robotic Orthosis (RO) synchronizing with a pair of Pelvic Arm (PA) mounting on Mobile Platform (MP) modules. The mechanical structure and control algorithm are provided. The implementation of the control theories proposed previously is tested on the system and the performance is presented in this Chapter.

Chapter 7 provides successful clinical trials arranged for the validation of the developed robotic system via electromyography (EMG) analysis. Comparisons of lower limb muscular activation between subjects walking with and without robotic device are also studied and discussed. Furthermore, frequency evaluation method is also proposed to distinguish the gait impairment.

Conclusions and recommendations are presented in **Chapter 8**.

Figure 1.5 Organization of this thesis.

Therefore, this thesis provides a full-range study for the motion generation, control and evaluations of OGW robotic-assisted gait rehabilitation methodology.

Chapter 2 LITERATURE REVIEW

In Chapter 1, the background and research gap on the robotic-assisted robot, which serves as a therapeutic device for gait rehabilitation by using of intelligence control, are identified. The research scope covers motion generation, control algorithm, and system evaluation. Review of these related works are briefly described in this chapter. The clinical aspects and summary of current gait robots are summarized in Section 2.1. The works on motion generation and control algorithm for rehabilitation robot, especially for the overground walking (OGW) device are discussed in Section 2.2. The quantitative assessment to diagnose pathologic gait and evaluate the progress of robotic-assisted therapy are investigated in Section 2.3. The summary of this chapter is provided in Section 2.4. The review is important as it can be used to assess the need to control a gait rehabilitation system and develop for assessment and evaluation of gait rehabilitation.

2.1 Clinical Aspects and State-of-the-Art in Gait Rehabilitation Robots

Rehabilitation robots can be designed to assist therapists, which is labour-intensive or even impossible to perform the gait training task manually. At the same time, the rehabilitation robots should target at training duration, training intensive, recovery function etc. In order to achieve these standards, literature survey and clinical observation on gait rehabilitation are conducted. Clinical gait rehabilitation and state-of-the-art in gait rehabilitation robots are provided. These studies are important for the guideline of gait rehabilitation robots.

2.1.1 Clinical gait rehabilitation

Promising results have been demonstrated with development of modern rehabilitation strategies. For example, Morris *et al.* [35, 37] have proposed theoretical basis for locomotor training (LT) as shown in Figure 2.1. The active training is encouraged and has

employed in *Constraint-Induced* (CI) therapy which involves continuous quantification and frequent feedback of improvement in lower extremity rehabilitation [38-41]. The clinical aspects are presented including (a) terminology in gait analysis, (b) common gait disorders, (c) gait training used in current clinical practice, (d) clinical-inspired functions, and (e) clinical gait assessment and evaluation.

Figure 2.1 Theoretical basis for the locomotor training (LT) [35].

First of all, in order to develop robot for gait training, it is essential to understand the biomechanics of human lower limb functions in walking. Furthermore, most of exoskeletons choose an almost anthropomorphic design, which means they are kinematical similar to human's leg but with reduced degree of freedom (DOF) to achieve more compact and power efficient design. Exoskeleton should not be collision between the robot and users when performing a task. At the same time, the structure cannot protrude out otherwise it will limit the mobility and maneuverability of the system. As a result, an anthropomorphic design which matches exactly to human leg is usually preferred compare to non-anthropomorphic design which usually cause collision between human and robot, then force human into unreachable configurations [42]. Therefore, this section investigates the perspective of kinematics and functionality of hip, knee, and ankle joints as follow.

(a) Terminology in gait analysis

Many terminologies are introduced in gait analysis such as stance phase, swing phase, right single support, right single support, double support etc. Gait can be inspected by the

length and width of step and stride, pressure distribution beneath foot and so on. There are so many parameters to be considered regarding to gait characteristics. Four groups of the major parameters in gait are listed in Table 2.1.

Table 2.1 Terminology in Human Gait [36]

<i>Temporal parameters</i>	Gait cycle time, stride time, step time, stance phase (% gait cycle), swing phase (% gait cycle), single support (% gait cycle), double support (% gait cycle), cadence, speed, average speed, mean normalized velocity and stride velocity
<i>Spatial parameters</i>	Stride length, step length, base width, toe in/toe out, distance traveled, step/extremity ratio, step width and stride width
<i>Kinetics parameters</i>	Pressure distribution, vertical force, shear force, joint movement, joint power and center of pressure
<i>Kinematics parameters</i>	Joint angles and foot clearance of ground

Gait cycle can be divided into temporal phases based on leg position and foot placement. For example, it is commonly defined the initial contact ground by the right foot heel as the beginning of one gait cycle [36]. The next right heel contact is the end of one gait cycle. Within one gait cycle, right toe off the group divide gait cycle into stance phase and swing phase. The duration of temporal phases is temporal parameters respectively.

According to the literature survey, four methods are commonly used to divide GC. Method 1: stance and swing phases [43] within GC are compared separately for the significant difference of walking between healthy subjects and subjects with walking problem. Method 2: completion percentage [44-45] of GC is commonly used. However, this standard GC cut off method is presented at the customary 83 m/min (1.36 m/s) [46] rate of walking cannot be used directly because walking speed is slower than 0.2 m/s when study the pathological gait such as SCI patient cannot walk as fast as normal group. Method 3: evenly divided GC [47] in each phase bin is used in the comparison of muscle

activity between healthy and pathological gait. Method 4: gait events [48-50] such as right heel contact (RHC), left toe off (LTO), left heel contact (LHC), and right toe off (RTO) can divide GC into right single support phase, double support phase, left single support phase and swing phase as shown in Figure 2.2.

Figure 2.2 Gait Cycle (GC) including right single support phase (% GC), double support phase (% GC), left single support phase (% GC) and swing phase (% GC).

(b) Common gait disorders and assessment of pathological gait

Joint angles for hip, knee, and ankle are the average joint motion in the Sagittal plane obtained from literature [51] as is shown in Figure 2.3. The movements of hip, knee, and ankle joint are related with major gait disorders. The details for major causes of gait disorders are explained in Appendix A. It appears, therefore, that systematic analysis of gait and walking function is necessary to develop for quantitative assessment, which may be utilized to determine the effectiveness interventions [52].

In order to assess walking to distinguish healthy and pathological gait, average muscle pattern in time domain analysis are commonly used [53-57]. For example, spinal cord injury (SCI) subjects produce synchronized pattern of muscle activity [58], which is significant different from that of healthy subjects [55, 59-61]. To our knowledge, specified muscles show most significant differences to distinguish subject with walking problem. These muscles could offer a good improvement to quantify EMG changes between healthy and pathological gaits. For example, SCI subjects have less extension of hip due to their weakness of rectus femoris (RF) [62]. Also, persistent plantar flexion

introduces compensation motions such as circumduction of affected limb to clear foot on the ground during the swing phase. Increased tibialis anterior (TA) EMG activity is presented in SCI subjects.

Figure 2.3 Time makers in one gait cycle and hip, knee and ankle joints within gait cycle.

Table 2.2 Guideline of Gait Training Used in Current Clinical Practice [37]

1	Promote the maximum capacity muscle strength, range, and endurance in muscles that can be voluntarily-activated above and below the lesion
2	Compensate for non-remediable deficits of paralysis, weakness, and balance by using parallel bars, standing frames, tilt tables, braces and assistive devices
3	Teach new behavioral strategies or skills for ambulation and mobility [i.e. w/c mobility, physics principles to compensate for weakness (levers, momentum, etc)]s.
4	Apply techniques to promote motor control (i.e. aquatic therapy).

(c) Gait training used in current clinical practice

The guideline of gait training used in current clinical practice is provided in Table 2.2. Furthermore, there are many rehabilitation methods, which can fill into compensation or recovery models. Assume that part of the nervous system is hard repair, and then patient is taught to compensate current motion or modify a task or environment [35, 63].

Compensation models and tools are commonly used. On the other hand, recovery techniques are applied to promote motor control because CNS is plastic and can learn to recover.

1) Compensation models

Parallel bars, standing frames, tilt tables, walking frame and other assistive devices are employed for non-remediable deficits of paralysis, weakness, and balance as shown in Figure 2.4.

Figure 2.4 Compensation models: (a) parallel bars; (b) standing frames [64]; (c) tilt tables [65]; (d) walking frame (walker) [44].

Figure 2.5 Compensation tools: (a) walking frame (walker); (b) walking crutch; (c) wheelchair.

New compensating behavioral strategies or skills are taught for ambulation and mobility such as how to walk with help of compensation tools such as walking brace, walking

crutch and wheelchair as shown in Figure 2.5. Compensation model improves the independence of patients at the initial stage of rehabilitation or those hard recovery cases. Furthermore, compensation model also works for consideration of safety to prevent second injury of patient.

2) Recovery models

Morris [35, 37] presents critical guidelines for recovery model including task specific, intense practice and repetition, and sensory input associated with locomotion provided to neural axis and circuitry. Facilities such as rehabilitation cage [66], balance training machine [67], balance ball [40], arm swing assistance training [68], parallel bars, exercise steps, exercise footstool and treadmill are commonly used in clinical practice for recovery model, shown in Figure 2.6.

Figure 2.6 Recovery model: (a) rehabilitation cage [66]; (b) balance training machine [67]; (c) balance ball [40]; (d) arm swing training [68]; (e) parallel bars; (f) exercise steps; (g) exercise footstool; (h) treadmill.

1) Requirement A—Enhancing muscle force

Clinical rehabilitation, aimed at enhancing muscle force, encourages patients to train their own muscular power as much as possible. Circumduction, flexion and extension, and abduction and adduction of each joint are conducted for specified muscle training. For patients who cannot stand or walk due to weakness of their muscle, gait recovery is achievable via motor learning or brain plasticity after repeated exercises.

2) Requirement B—Maintaining balance control

After the training on enhancing the muscle force, therapists provide exercises for balance control. Balance control is fundamental to independent walking. Patients may fall down if they do not have good balance control. In such an exercise, patients are required to keep their upper body upright when in a sitting or standing position, through which they can relearn their trunk control. Equipments are used to train the sense of balance too. Center of Pressure (CoP) is an effective measurement method of evaluating the balance control ability.

3) Requirement C—Training walking locomotion

Apart from improvement for the muscular force and the sense of balance, it is also significant to practice rehabilitation exercises for walking locomotion. In order to achieve walking locomotion, the legs need to be trained specifically for body weight support (BWS) and providing propulsion. BWS is one of the essential functions of walking. In training program, therapists assist each patient by supporting his/her body weight when necessary. However, it is laborious for manual assistance and guidance. In addition, a way to compensate for the lack of BWS is to attach an arm-support onto the frame of the structure. Either with regained BWS ability or with external off-loading assistance, subjects carry out the next exercise which concerns leg propulsion. In many cases, the muscular strength of the patient limbs is weak. However, it may allow for slow walking

locomotion which is effective for rehabilitation. Therapists emphasize the muscle activation through leg movement that stimulates the nervous system to recall normal walking.

4) Requirement D—Providing pelvic control

Many patients walk in an abnormal way because of the loss of their pelvic control. Without proper pelvic control, incorrect posture, which adversely affects walking stability, may be observed. In this situation, oral reminder or necessary assistance may be provided. An example of assistance is through placing of both hands on the hips of a patient and moving with the patient while correcting the problems with the trajectory of pelvic movement as shown in Figure 2.8.

Figure 2.8 Pelvic control combines into the overground ambulation training with therapist assistance.

5) Requirement E—assisting mobility quality in activities of daily living (ADL)

ADL-related activities are provided when patients can walk slowly by themselves. Tasks such as stair climbing, slope walking, tuning and obstacle avoiding are common locomotion tasks in daily live. Those activities enable individuals to get ready for their day and get themselves to their place of employment, school, and/or recreation. In this

stage, the improvement of walking ability to perform ADL is targeted. Design, development and testing of key rehabilitation exercise related to specified ADL task is important.

In conclusion, the rehabilitation program starts from enhancing muscle force. Following which, after evaluating the subject's situation, the rehabilitation program may proceed to include other functions [69]. Therefore, the gait rehabilitation robot needs to meet principles of rehabilitation in individual rehabilitation stage.

(e) Clinical gait assessment and evaluation

Balance and gait dysfunction may be related to impairment such as spinal cord injury (SCI), stroke and other diseases. Many symptoms are presented such as loss of proprioception, weakness/muscle imbalance, numbness, tingling, spasticity, poor vision, vestibule-ocular dysfunction evident in slow, deliberate movement, hesitation to move, decreased step length, increased stance, stiff movement, and difficulty turning. In order to assessment these impairments, clinical evaluations are conducted such as gait observation analysis, timed 10 meter walk, 2-minute walk / 6-minute walk, ambulation index (AI), dynamic gait index (DGI), timed up and go (TUG), functional reach test, Berg balance test (BBT), Tinetti gait and balance assessment. Take three of them as examples to explain:

1) Disease steps (DS)

According to individual situation, walking problems are classified into groups including: 0 = normal, 1 = mild disability (no visible gait abnormality), 2 = moderate disability (abnormal gait but no aids), 3 = early cane (can walk 25 feet without cane), 4 = late cane (dependent on unilateral support), 5 = bilateral support (scooter for distance), 6 = confined to wheelchair (cannot walk 25 feet), U = unclassifiable (significant cognitive, visual, fatigue, bowel/bladder impairment).

2) Dynamic gait index (DGI)

This assessment index covers eight items graded 0 – 4 including (1) gait level surface, (2) change in gait speed, (3) gait with horizontal head turns, (4) gait with vertical head turns, (5) gait and pivot turn, (6) step over obstacle, (7) step around obstacles, and (8) steps on stairs.

3) Berg balance scale

This assessment score each graded 0 – 4 including (1) sitting to standing, (2) standing unsupported, (3) sitting unsupported, (4) standing to sitting, (5) transfers, (6) standing with eyes closed, (6) standing with feet together, (7) reaching forward with outstretched arm, (8) retrieving object from floor, (9) turning to look behind, (10) turning 360 degrees, (11) placing alternate foot on stool, (12) standing with one foot in front, and (13) standing on one foot.

2.1.2 State-of-the-art in gait robots

There are various types of gait robots which is listed as follow: (a) Rehabilitation robots: Lokomat, LOPES, ALEX [27] based on treadmill walking training and *NaTure-gaits*, Walktrainer, KineAssist based on overground walking training; (b) Prosthetic legs: C-leg, Tag Heuer prosthetic Leg; (c) Performance-augmenting exoskeletons: BLEEX [70], NTU-LEE [71], HAL [21, 30]; (d) Walking suits: Eleg, Rewalk, REX. And (e) Others products: Honda Leg, programmable footplates, powered ankle-foot-orthosis, etc.

Our research is focusing on the first group of gait robot (rehabilitation robots), which are different from the other gait robots. It is related to rehabilitation and is designed to provide therapy for the patient who suffers from neurological disorders such as spinal cord injury (SCI) stroke, Parkinson Disease, or Cerebral Palsy etc. To deliver automated gait therapy for the physically impaired patient, these devices are designed to replace

conventional therapy which requires intensive manual labour effort of two or three therapists for one patient. Rehabilitation robots and comparisons to other device are explained:

(a) Rehabilitation robots: treadmill and overground walking training

As previously mentioned, physical rehabilitation is necessary for gait recovery. Unfortunately, limits of rehabilitation sessions and shortage of PTs and OTs cannot satisfy rehabilitation consumers, which will result in compromising recovery. A potential solution is integration of robotics with more traditional manual rehabilitation. The rehabilitation robots can be divided into two sub-categories based on the training environment: treadmill-based rehabilitation robots and overground walking rehabilitation robots.

Figure 2.9 Treadmill-based rehabilitation robots: (a) Lokomat system with patients [27]; (b) LOPES [31]; (c) ALEX [72].

There are now various alternative robotic therapies available to stimulate muscle activity pattern effectively with little supervision from therapists via using treadmill-based rehabilitation robots on a corner of the rehabilitation department, as shown in Figure 2.9. Overground walking (OGW) rehabilitation robots have similar function as treadmill-based

rehabilitation robots while the OGW rehabilitation robots addressing the ability of overground locomotion gait training, as shown in Figure 2.10. Additionally, this thesis focuses on this type of rehabilitation robot.

Figure 2.10 Overground walking (OGW) rehabilitation robots: (a) a SCI subject walks with NaTure-gaits [22]; (b) WalkTrainer [23]; (c) KineAssist [25].

(b) Prosthetic legs

The prosthetic legs are different from the rehabilitation robot. They are used for low limb amputees. The prosthetic legs device is substituted for the low limb. It is different from the rehabilitation robot which is working on gait training. Two successful prosthetic legs are shown in Figure 2.11.

Figure 2.11 Prosthetic legs: (a) C-leg [73]; (b) Tag Heuer prosthetic leg[74].

(c) Performance-augmenting exoskeletons

Robotic exoskeletons are also developed to augment the performance of the able-bodied person are usually named as performance-augmenting exoskeletons [75]. It has similar function to rehabilitation robots and both of them are target to enhance the muscle power. The performan-augmenting exoskeletons are used for healthy subjects while rehabilitation robots are employed on subjects with walking problems. Some of the representative exoskeletons from this category include: BLEEX (Berkeley Lower Extremity Exoskeleton) [70], NTU-LEE (NTU Lower Extremity Exoskeleton) [71], and HAL-5 (Hybrid Assistive Limb) [30] as shown in Figure 2.12.

Figure 2.12 Performance-augmenting exoskeletons: (a) BLEEX [42]; (b) NTU-LEE [71]; (c) HAL-5 [30].

Figure 2.13 Mobile medical exoskeleton: (a) eLEGS [76]; (b) ReWalk [77]; (c) REX [78].

(d) Walking suits

A new kind of rehabilitation and assistive robots called mobile medical exoskeletons (Figure 2.13) emerge which are portable and can even be used at home. Walking suits can be used as either assistive walking device at home or rehabilitation device using for training. It is enhance the walking ability as same as the performance-augmenting exoskeletons and rehabilitation robots. Walking suits are different from the performance-augmenting exoskeletons which are used by healthy human. They are designed for person with gait disability. On the other hand, the rehabilitation robots usually provide a body weight support system to keep balance and help to support body weight, while the mobile medical exoskeletons requires the patient to balance themselves, which means the patient must have a healthy upper body, at least trunk control. Additionally, walking suits have a similar structure as performance-augmenting exoskeletons and rehabilitation robots. It should be emphasized the importance of the human robot interface. Since hip, knee and ankle are weight-bearing joints, it will collapse if there is not enough muscle force provided.

(e) Others products

There are other exoskeletons or assistive devices which do not belong to any of the categories previously mentioned such as Hand Leg, programmable footplates and powered ankle-foot-orthoses (Figure 2.14). Honda Leg [79] offers a novel mechanism of a body weight support. It proposes a solution for the subjects who cannot support their body weight to walk. Although this design is only aimed at healthy person, the design concept will benefit other rehabilitation robots design. Schmidt [32] invented a pair of programmable footplates for floor walking and stair climbing. This system provides complex gait cycle task-specific repetitive training. Except basic simulation for floor walking, the programmable footplates equipped with 6 DOF force sensors for the use of intelligent control. Powered ankle-foot-orthoses [80] have been developed to provide more power for the ankle joint and correction of drop foot, which allows practice starting,

turning, stopping, and avoiding obstacles during overground walking. It is feasible and practical owing to its lightweight than many other large robotic devices for assisting treadmill walking.

Figure 2.14 Other products: (a)Honda Leg [79]; (b) programmable footplates; (c) powered ankle orthosis.

The brief summary of the robotic devices as mentioned is listed in Table 2.3.

Table 2.3 Comparisons of Rehabilitation Robots, Performance-augmenting Exoskeletons and Walking Suits

		Purpose	Rehabilitation Robots					Performance-augmenting Exoskeletons			Walking Suits				
	Joints	Biological Limb	<i>Locomat</i>	<i>Nature-gaits</i>	<i>WalkTrainer</i>	<i>ALEX</i>	<i>LOPES</i>	<i>BLEEX</i>	<i>HAL-5</i>	<i>NTU-LEE</i>	<i>eLEGS</i>	<i>MIT Medical Exoskeleton</i>	<i>ReWalker</i>	<i>REX</i>	<i>Honda Leg</i>
DOF	Pelvis	6	N/A	5	N/A	3	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
	Hip	3	1	1	3	2	2	3	3	2	1	2	2	3	2
	Knee	2	1	1	1	1	1	1	1	1	1	1	1	1	1
	Ankle & Foot	4	N/A	3	3	1	N/A	4	1	1	1	1	N/A	2	3
ROM(deg)	Hip	140/15 ^a 40/30-35 ^b 15-30/60 ^c	N/A	50/15	N/A	40/40 ^a	60/30 15/15	121/10 ^a 16/16 ^b 35/35 ^c	115 ^a	140/15 ^a 40/30-35 ^b 15-30/60 ^c	N/A	N/A	N/A	N/A	N/A
	Knee	120-140/0-10 ^a	N/A	80/0	N/A	45-60/0	90/0	121/0 ^a	97 ^a		N/A	N/A	N/A	N/A	N/A
	Ankle & Foot	40-50/20 ^a 30-35/15-20	N/A	10/20	N/A	N/A	N/A	45/45 ^a 20/20	N/A	140/15 ^a	N/A	N/A	N/A	N/A	N/A
Torque (Nm)	Hip	140/120 ^a	N/A	31 ^a	N/A	50 ^a	65 30	150/120 ^a	N/A	118	N/A	N/A	N/A	N/A	N/A
	Knee	140/15 ^a	N/A	35.2 ^a	N/A	50 ^a	65	100/120 ^a	N/A	118	N/A	N/A	N/A	N/A	N/A
	Ankle & Foot	165 ^a	N/A	132 ^a	N/A	N/A	N/A	200/150 ^a	N/A	118	N/A	N/A	N/A	N/A	N/A

a. Flexion/Extension b. Adduction/Abduction c. Internal/External d. Inversion/Eversion

2.2 Motion Control of Rehabilitation Robot

The objective for control of rehabilitation robot is to encourage high voluntary muscle activation, which effectively benefits for gait recovery. With development of the control technology, motion control of rehabilitation robot is combined with intelligence control. Strategies on interaction control are proposed such as enlarged sensitivity of the human motion [81], loop shaping [82], “patient-cooperative” techniques via detecting voluntary efforts [83]. Compliant actuation is also developed by the need for active assistance training [84].

Inspired by the biped robot, the rehabilitation robot parallel attached to human and intended for ability to exhibit high impedance relative to the human subject [82], thus a high-force haptics to maximize transferred energy. A zero impedance joint control based on specifying the level of rehabilitation assistance [85]. Zero-force control [86], and force-field controller (FFC) [87] are proposed. Optimization of zero moment point (ZMP) is also applied to release load on the subject for less constraints for training [88]. To sum up various methods, human machine interaction, motion generation and control algorithm are investigated for the control of the rehabilitation robot.

Therefore, the control for rehabilitation robot is summarized as follow: (1) Human machine interaction (HMI) is important for the external force transferring from robot to subject walking in rehabilitation robots. (2) The motion generation for rehabilitation robot is another critical issue for the low limb walking training, which is similar to that of gait pattern generated for the humanoid robot. (3) The control algorithm for rehabilitation robot is to track the generated motion in order to improve the walking quality as comfort and natural as possible on the system. In current studies, the modeling of human machine interaction, motion generation and intelligent control algorithm used in rehabilitation robot are presented.

2.2.1 Modeling human machine interaction (HMI)

The most commonly used computed torque control based on dynamic analysis of the mechanical structure and impedance control for position/force control are first introduced. Then, the virtual force/ velocity field control used in LOPES and virtual model control to define the range of assist force are presented. Thirdly, improved of sensitivity and loop shaping for the improved HMI are provided based on the internal model, which is close related to proposed method in Chapter 3.

(a) Computed torque control and impedance control

It is an approach that makes direct use of complete dynamic model of the manipulator to compensate the effects of gravity, Coriolis and centrifugal force, friction and inertia tensor etc [89]. After cancellation of all the nonlinear dynamics, the linear dynamics are presented for control. However, it is not easy to estimate the exact identification of the robotic system. Complete cancellation of nonlinear effect is also prevented by non-completed model of system. The modeling error can be reduced by using high precision algorithm but expense of increased computational time. Modified algorithms [38] are developed based on this method combined with artificial neural network, sliding model control etc.

The impedance control technique focus on the interaction force of the manipulator-environment and generate “assisted-as-needed” force. The reference trajectory is calculated from stiffness of the environment [90]. However, it is hard to know the exact stiffness for optimal trajectory under the tracking force. Researchers have improved the impedance control combined hybrid position/force control, sliding model control etc. The active force control for the gait rehabilitation has been developed.

(b) Virtual force/ velocity field and virtual model control

Two algorithms were developed for assist-as-needed strategies of gait rehabilitation, which named as “Band” and “Window” [91]. The first strategy creates two fixed boundaries: an inner and outer boundary, forms a “band” surrounds the reference trajectory. Using ankle joint position as example, if the ankle stays within the band, it is guided by a small constant velocity field tangent to the desired trajectory. The velocity field does not enforce a specific timing for the leg movement along the desired trajectory. If the ankle leaves the band, the convergent velocity fields will move it back to the band region.

The research group of LOPES rehabilitation system has introduced a different gait rehabilitation methodology [92]. Instead of focusing on a joint level, they implement the controller to train the patient on a subtask level (i.e. foot clearance, balance control). In order to assist each of these tasks, they apply the Virtual Model Control (VMC) on LOPES. This type of controller basically does not contain any reference trajectory but provides the right amount of supportive force at the correct timing by using VMC. The basis of this control method is to define the physical interactions for the subject that would support him/her during the gait subtasks. These interactions are then translated into VMC, such as springs and dampers that can be turned on and off at appropriate timing in a gait cycle. The virtual forces that would be exerted by the VMC are then translated into joint torque commands for the joint actuators.

(c) Sensitivity control and fictions gain control

The basic requirement for the sensitivity control [93] used in BLEEX is to make the exoskeleton follow the wearer’s voluntary and involuntary movement immediately, without delay. With the high sensitivity, the quick response makes the exoskeleton move in concert with its wearer. However, disturbance and error are also enlarged leads to a loss of robustness in the system. Recently, Kong [81] has proposed fictitious gain control for the rehabilitation system. The fictitious gain is designed to increase/decrease the

sensitivity of the motion control. The concept of adjusting the sensitivity with robust control theory has facilitated in its application.

Robots intend for high-force interaction with human beings which requires low and tunable endpoint impedance as well as high force capacity [82]. A generally more robust approach to interaction control regulates dynamic behavior at the interaction port. In order to reduce the mechanical impedance of actuators, elastic or other compliant components are used [24, 94]. Kong [81] propose that the exoskeleton controller which is designed to realize the effect of a fictitious gain. The modeling of human interacting with the exoskeleton system is shown in Figure 2.15. $y_{Desired}$ is the desired motion and y is the actual motion for the human joint H . By wearing gait robot, the human joint, H is actuated by torque τ , the sum of muscular torques τ_H and external assistive torques τ_A . The muscular torques are exerted by muscle under control of brain commander u_H . The external assistive torques are exerted by actuator A under control of controller C on the gait robot. In this thesis, improved modeling of human machine interaction is developed for adjustable assistance level of the active control in Chapter 3.

Figure 2.15 Modeling of Human Machine Interface [81].

2.2.2 Walking pattern generation

Both the human model (such as Hodgins's human model [95]) and walking motion are important for design of rehabilitation robot, especially, the joint motion of human in the Sagittal plane can be applied to the parallel mechanism directly as shown in Figure 2.16. Motion capture experiment [96-97] is commonly used to obtain the walking motion for

gait robot. With development of technology, the video-based motion capture experiment reaches a high accuracy for capture the walking motion. However, personal walking patterns and simplified robot structure compared to that of human being restrict the direct application of the human motion to robot motion.

Figure 2.16 A SCI subject walks with NaTure-gaits, in which the hip/pelvic, knee and ankle trajectories are illustrated in one gait cycle [22].

A gait profile generator is needed to customize the gait pattern for individuals. Numerous motion planning methods have been proposed for the exoskeleton and robots used in gait rehabilitation. One analytical way is pre-planned the ankle, hip trajectory first then computed for joint angle profile via inverse kinematic analysis [98-100]. The other method via optimal control is also commonly used for searching the suitable gait pattern [26, 88, 95, 101-103]. Researchers have also combined the two methods and proposed improved methods to solve this problem [104]. Furthermore, online trajectory generation is proposed for the active joint control [99, 103]. Three examples as follows are presented for each category as mentioned. In this thesis, an analytical lower limb walking motion generation is proposed combined with OGW condition optimization in Chapter 4.

(a) Inverse kinematic based on analytical computation

This is a straightforward method with complicated analytical computation caused by the inverse kinematic and the calculation is highly sensitive to the position of end effectors. Considering a three-joint robotic manipulator for the lower limb, it is coupled three uncertainties to final trajectory on the end point. Other than the trajectory of end point, all those joint trajectories are needed to plan as all of them are attached to the human. Same trajectory for end point of chain linkage can obtain more than one solution. Therefore, the motion plan for the rehabilitation robot includes the trajectories of the end points and all the joint angles.

An analytical inverse kinematic motion plan is developed for a four-links rehabilitation robot walking on the treadmill. Constraints of swing limb tip and hip motion are analytical formulated by cubic polynomial interpolation, by varying those conditions, different motion trajectory can be obtained. Then, joint angles on the lower limbs are computed according to those trajectories [98]. It works well on treadmill walking training with a fixed supporting frame. But it cannot directly extend to the biped robot because of the imbalance may caused by the predefined trajectory. Therefore, the result is suitable for the treadmill walking training but cannot extend to the OGW. More complicated algorithm is needed based on the kinematic analysis of system with conditions such as double stance, single stance and swing phase etc.

(b) Optimal motion generation for an OGW

Walking with intermittent ground contact leads to discontinuities in state which is hard to optimize due to high dimensionality, nonlinearity, the impact between the foot and the ground, foot clearance requirement and foot-ground conditions. The optimal control for a walking robot covers three steps. First of all, the modeling of the robotic system is built up. Based on the system model, control law is deduced. Secondly, the optimal equations and cost function are provided. Especially, the cost function is different from the biped

robot focusing on joint torque to improve the efficiency. Rehabilitation robot focuses on interaction force between human and robot. After optimal objective is built, the optimal objective is approached by combination of control model and optimal equations. The procedure is explained as follows:

1) Walking model

The overall characteristics are determined by an ordinary differential equations with discontinuous [101] in form of

$$\dot{x} = f(x, y) \quad x, y \in S \quad (2.1)$$

$$x^+ = H(x^-) \quad x^- \in S \quad (2.2)$$

where $x^- = \lim_{t \rightarrow t_0^-} x(t)$ is the state at the moment before the ground contact. $x^+ = \lim_{t \rightarrow t_0^+} x(t)$ is the state at the moment after the ground contact. $y^- = \lim_{t \rightarrow t_1^-} y(t)$ is the state at the moment before the other lower limb contact ground. $y^+ = \lim_{t \rightarrow t_1^+} y(t)$ is the state at the moment after the other limb contact ground. $H(x \cdot)$ is the dynamics equations and S is the set of the state.

2) Control law

In the single support phase, human walking is modeled as a rotational joint open-chain manipulator with linkage. The major joints cover metatarsal, ankle, knee, hip joint on both sides. The trunk and upper body are considered as load. Let the joint angle $q := (q_1, \dots, q_n)^T$ be the generalized coordination describing the configuration of the human body. The dynamics equations can be derived using the Lagrange equations [105]. The result is a second order system

$$\tau = M(q)\ddot{q} + C(q, \dot{q})\dot{q} + g(q) \quad (2.3)$$

where τ is the torque, q is the joint position, $M(q)$ is the inertia matrix, $C(q, \dot{q})$ is the matrix of the Coriolis and centrifugal forces, and $g(q)$ is the gravity force. The state equations of the robotic manipulator are of following form:

$$\begin{cases} \dot{q} = v \\ \dot{v} = M^{-1}(q)(\tau - C(q, \dot{q}) - g(q)) \end{cases} \quad (2.4)$$

where v is the joint velocity, and $\dot{v}$ is the joint acceleration, assume

$$u = \ddot{q} \quad (2.5)$$

The dynamic equations of the robotic manipulator are highly nonlinear and coupled. In order to design a controller, the following linearization of the dynamic:

$$\begin{cases} \dot{q} = v \\ \dot{v} = \hat{M}^{-1}(q)(\tau - \hat{C}(q, \dot{q}) - \hat{g}(q)) \end{cases} \quad (2.6)$$

where $\hat{M}$, $\hat{C}$ and $\hat{g}$ are estimation of the manipulator dynamics. In the absence of disturbance and when the dynamic model is exactly known, $\hat{M} = M$, $\hat{C} = C$ and $\hat{g} = g$, then, the robotic manipulator is simplified as a second order linear differential dynamics. The dynamics of robotic manipulator can be decoupled, that is, u_i is only variable q_i independently of the motion of the other joints.

3) Trajectory optimization

Consider the walking motion or part of walking motion, the infinite horizon cost functional is

$$J = \int_0^\infty e^{-\beta t} c(x(t), u(t)) dt \quad (2.7)$$

Or

$$J = \frac{1}{2} \int_0^\infty \omega_e (\tau_1^2 + \tau_2^2 + \dots + \tau_6^2) dt \quad (2.8)$$

Those cost function is defined based on different objectives, for example, Equation (2.8) is a cost function for a treadmill walking training rehabilitation robot.

(c) Online trajectory generation

Researchers propose online trajectory generation to achieve the “correct” gait adapt both to the individual subject and to the situation. For example, CLME (complementary limb motion estimation) are proposed for the online reference trajectory generation [106]. Desired states for disabled leg are generated online based on the movements of the sound leg applied for hemiparetic stroke subjects. The tests on the healthy subjects show that CLME can achieve more natural walking. Although online trajectory generation has some distinct advantages, the limitations of the response time and safety consideration obstruct its application on rehabilitation robots.

2.2.3 Control algorithm for gait rehabilitation

Once proper trajectories for the end pint and all joints are planned, problems remain on issuing the commands to the joint actuators. Actuators under control will push the manipulator to track or follow the desired trajectory and produce suitable torque. Numerous techniques [99, 103, 107-111] have been proposed for the robotic control. For example, fuzzy logic combined with PID control into the force/position hybrid control [108], modified impedance control [82, 90, 112] and neural network technology [38, 111, 113-115] adaptively change of subject’s condition. In this section, the classic PID control is introduced first, as shown in Figure 2.17. k_P , k_I , and k_D are proportional, integral and derivative parameters of PID controller. q_d is the desired trajectory. q is the actual trajectory. q_{dis} is the disturbance for the joint motion trajectory of $G(s)$. $q_e = q_d - q$ is the error between the desired and actual trajectories.

Figure 2.17 Control structure of a classic PID control for a single-axis robotic joint

Other than classic PID control, intelligent control algorithms are also employed. The suitable control algorithms are proposed to use in rehabilitation application. The high-level control such as making decisions on the task that might be needed for the different therapies. The discrete control scheme are also developed [111]. The sequence of major events is arranged for the coordination of the sub-systems. For example, motions of the body weight shifting from one side to the other side should be completed before moving that lower limb in the walking training.

Figure 2.18 Diagram of the repetitive control system, modified from reference [116].

To our knowledge, repetitive control algorithm is suitable for the “repetitive” motion input for the lower limb. So the intelligent control algorithms for the periodic signals named repetitive control are used. Furthermore, in order to control the walking velocity on

time for the OGW system, an adaptive control is proposed based on on-line model identification algorithm. An autotuning controller is developed to synchronized motions for the sub-system. Development of repetitive control and autotuning control are presented in Chapter 5. The basis theory and related works are presented in this section.

(a) Repetitive control

The diagram of the repetitive control is shown in Figure 2.18 and the brief development of repetitive control is presented. The state-space description of the repetitive controller is as follows:

$$\begin{cases} \dot{x}_r(t) = -\omega_c x_r(t) + \omega_c x_r(t-L) + \omega_c e(t) \\ v(t) = x_r(t-L) + e(t) \end{cases} \quad (2.9)$$

By using the augmented state vector $x = [x_p^T, x_r^T]^T$, combine Equation (2.9) and state feedback stabilized system with $r(t) \equiv 0$ and $\omega(t) \equiv 0$ to yield the closed-loop system:

$$\dot{x}(t) = Ax(t) + A_1x(t-h) + A_2x(t-L) + A_3x(t-h-L) \quad (2.10)$$

where

$$A = \begin{bmatrix} A_p + B_p F - B_p C_p & 0 \\ -\omega_c C_p & -\omega_c \end{bmatrix}, A_1 = \begin{bmatrix} B_h F - B_h C_p & 0 \\ 0 & 0 \end{bmatrix}, A_2 = \begin{bmatrix} 0 & B_p \\ 0 & \omega_c \end{bmatrix}, A_3 = \begin{bmatrix} 0 & B_h \\ 0 & 0 \end{bmatrix} \quad (2.11)$$

To clarify, define the scalars as $d_1 := h$, $d_2 := L$, $d_3 := h + L$, and $0 < d_j \leq \bar{d}_j$, for $j=1, 2, 3$. The stability based on LMI and Lyapunov are computed and the design problem of $Q(s)$ will be discussed, which is equivalent to finding out the maximum cutoff frequency ω_c satisfying the LMI constraints given in stable condition. Represent ω_c as the sum of $\hat{\omega}_c$ and $\delta\omega_c$:

$$\omega_c = \hat{\omega}_c + \delta\omega_c \quad (2.12)$$

where $\hat{\omega}_c$ is an estimated value and $\delta\omega_c$ is an unknown value to be decided. Then, the matrices A and A_2 can be represented as the following form:

$$A = \hat{A} + \tilde{A} \times \delta\omega_c \quad (2.13)$$

$$A_2 = \hat{A}_2 + \tilde{A}_2 \times \delta\omega_c \quad (2.14)$$

where

$$\hat{A} = \begin{bmatrix} A_p + B_p F - B_p C_p & 0 \\ -\hat{\omega}_c C_p & -\hat{\omega}_c \end{bmatrix}, \tilde{A} = \begin{bmatrix} 0 & 0 \\ -C_p & -I \end{bmatrix},$$

$$\hat{A}_2 = \begin{bmatrix} 0 & B_p \\ 0 & \hat{\omega}_c \end{bmatrix}, \tilde{A}_2 = \begin{bmatrix} 0 & 0 \\ 0 & I \end{bmatrix}. \quad (2.15)$$

By introducing a new variable $\lambda := \delta\omega_c^{-1}$, it is obtained that the maximum $\delta\omega_c$ by solving the following LIM-constrained optimization problem:

$$\text{Minimize } \lambda \text{ over } P, Q_{j0}, Q_{j1}, X_j, Y_j, Z_j$$

$$\text{Subject to } \tilde{\Pi} < \lambda \times (-\hat{\Pi}), Q_{j1} < \lambda \times Q_{j0}$$

The constraints in optimization problem have the standard forms of generalized eigenvalue minimization problem. Thus can be solved numerically using the corresponding solve.

Consequently, according to the discussions mentioned above, following procedures are designed the optimal cutoff frequency of the repetitive controller.

Step 1: select $\hat{\omega}_c$ as a small positive value;

Step 2: check the feasibility;

Step 3: if feasible, go to the next step. Otherwise, replace $\hat{\omega}_c$ with $\hat{\omega}_c + \varepsilon$ and go to step 2;

Step 4: solve the optimization problem.

The design technique proposed in this study is also applicable by just modifying the dimensions of some matrices. The tracking performance for the cyclic input signal is improved based on the repetitive control, especially when rapid changes presents and the classical PID controller cannot track it.

(b) Autotuning control

With consideration of dynamic, nonlinear and coupled of the robotic manipulation, on-line adaptive control is used. Compared with the traditional computed torque control, a better tracking performance behavior has been shown [38, 111, 113-114]. The applications of these intelligent control have widely used in the process control. However, the huge on-line computation limits their extension to robotic control cause the loss of the quick response. This situation has been improved with development of hardware and modified algorithm. For example, Astrom [117] proposed an automatic tuning controller and Luyben [118] summarized the applications and proposed auto-tuning variation (ATV) method. This autotuning method soon became a superior alternative to the conventional trial-and-error method. A relay identification component is added to the PID control loop in parallel as shown in Figure 2.19. r is the reference input. e is the tracking error. $u(t)$ is the input for the process under control. $y(t)$ is the output of the process. If it is necessary to tune the PID parameters, relay control is chosen by the switch between relay and PID control. The tuning process is around the working point is one of the advantages of the system. The improved performance can be satisfied most industrial applications. In the robotic control, the manual tuning procedures are time consuming, in particular, for those

high-dimensional joints synchronizing move. A modified relay-based autotuning controller is developed for quick response and track performance in this thesis.

Figure 2.19 The autotuning control based on relay feedback system.

In this thesis, the repetitive control are investigated and introduced to control application in the gait rehabilitation robot. The autotuning control algorithm is developed to allow walking velocity updates on various situations in Chapter 5. All the methods proposed from Chapter 3 to Chapter 5 have implemented into the motion generation and control of *NaTure-gaits* as presented in Chapter 6.

2.3 Evaluation of Gait System

Gait robots reduce the use of hands on manpower and enhance the duration of gait re-training programs with adjustable walking speeds and body weight off-loading capabilities. Studies are required for quantifying the muscle contribution based on the normalized walking speeds and body weight off-loading. Furthermore, although the effect of rehabilitation walking on neuromuscular co-ordination and motor control has not been fully understood, clinical studies show that prolonged repetitive walking training with an automated device can contribute to the recovery of functional gait [15]. Currently available training approaches include body weight support treadmill training (BWSTT) [27, 119] and overground walking (OGW) locomotion training [23, 54, 120-121]. However, it is still controversial that robotic-assisted gait training lead to effective

neuromotor rehabilitation. Therefore, this study forces on comparisons of manual-assisted and robotic-assisted gait training in order to show the effect of the robotic-assisted gait training.

The ability to identify the underlying regulation of muscle activation pattern via electromyography (EMG) analysis, which is a substantial component in the assessment of muscle activity and variability in the gait periodic events, could lead to a better understanding of progressive recovery, and provide a baseline for robotic rehabilitation devices. However, it is challenging to develop a uniform and standard evaluation methodology as the walking pattern varies by individuals. The various gait patterns pose significant restrictions on the evaluation's effectiveness and generality.

The adjustable rehabilitation gait system *NaTure-gaits* is targeted at normal overground walking (OGW), named as body weight support locomotion training (BWSLT). At same time, it has been developed with modulation of walking speeds and weight off-loading levels. In order to determine suitable ranges of training speeds and off-loading levels, it is necessary to study the muscle contribution changes with speed and weight off-loading level during OGW. Furthermore, as a preliminary evaluation, system trials are conducted on healthy and pathologic gait subjects, respectively. The effects between subjects walking with and without the robotic device are compared. Moreover, the systematic evaluation of walking quality is summarized.

2.3.1 Assessment of human gait via walking speed and off-loading level

It is useful to evaluate the effects of lower limb function during walking qualitatively and quantitatively under different conditions for setting the control parameters of a robotic training/ rehabilitation device. Before evaluation of walking speed and off-loading level with walking device, the effects of them are evaluated on healthy subject walking. Muscle activity is required to initiate movement and govern walking speed; however, at the same

time the muscle must also work to sustain body weight and equilibrium against gravity. Therefore, the magnitude and duration of muscle activation changes accordingly. Previous work has demonstrated variability in muscle activity related to different walking speeds [122-124], and it has been reported that there is a modification in the patterns of muscle activity in normal subjects between normal walking and very slow walking [34, 60]. As well as EMG quantification via different walking speeds, muscle activation patterns have been quantified to different body-weight support levels [125-126]. For example, van Hedel [127] suggested that patients using automated bodyweight-supported treadmill training (BWSTT) should walk at velocities higher than 2.5 km/h (0.69 m/s) and less than 50% of body weight unloading. In addition, muscles are further classified into functional groups, in terms of propulsion and support during walking at different speeds [128] or be classified in terms of timing and energy production [123].

2.3.2 Comparison between manual-assisted and robotic-assisted rehabilitation

Researchers can translate results on human motor control to facilitate modulation parameters for robotic rehabilitation applications. However, there is difference between manual-assisted and robotic-assisted rehabilitation. In order to obtain the natural walking training, robotic-assisted rehabilitation is evaluated. For example, treadmill gait training study has also concluded that the interaction of body weight support and stride frequency should be taken into consideration to optimize motor output during locomotor training [129]. At the same time, researchers have pointed out that the muscle activation pattern for the treadmill walking is different from the overground walking [34]. Thus clearly further exploration for “overground” robotic gait devices is needed.

2.3.3 Systematic evaluation of human gait training system

The average muscle activation pattern within gait cycle is commonly used in evaluation of human gait. Furthermore, it is useful to include the auto-correlation as a factor to be

considered as the ambulatory variation limited the average pattern investigations. Some researchers [45, 127-128] presented standard deviation, correlation coefficients, and analysis of variance, etc. It has been highlighted that repetition and symmetry estimation will be measured by portable sensors [54]. None application of the autocorrelation analysis is used to evaluate robotic-assisted overground walking training. In addition, the frequency spectrum widely used in muscle fatigue analysis [122] is rarely extended to this area. It would be distinguish some post-stroke patients developed a syndrome called spasticity, which changes the motion of the affected joint. A pre-programmed gait trajectory without considering spasticity might cause injury to the patient. Therefore, a gait evaluation in these aspects is required for the robotic-assisted overground gait training system if compared to manual-assisted gait training.

2.3.4 Frequency analysis method of EMG for a quantitative assessment

Advocated gait assessment has conducted. However, observational gait analysis is still an essential domain in order to qualitatively assess walking function in subject with walking problem. According to the recent studies [130-131], frequency analysis focusing on computation of power spectrum density (PSD) analysis are successfully used to detect the reduction in muscle activation associated with frequency band. It is important result to achieve the quantifying assessment for pathological gait. Furthermore, in statistical significance testing, previous works [132] mentions that the confidence interval value ($P \leq 0.01$) chosen may be too restrictive for such research, a P -value of 0.05 is usually chosen when study gait.

In conclusion, effects on lower limb muscular activation during walking of different walking speeds and off-loading levels are investigated. Furthermore, the successfully clinical trials are introduced for patients walking on *NaTure-gaits*. Its system description and control implementation are presented in Chapter 6. Their muscle activations have been monitored for demonstration of walking effectiveness. Machine-related effect is

highlighted in comparison of the average muscle pattern. After these studies, a systematic evaluation is proposed for assessment of walking quality in Chapter 7. Furthermore, quantitative assessment of pathological gait is provided based on frequency analysis.

2.4 Summary of literature survey

Figure 2.20 Clinical-related overview for the development of a rehabilitation robot.

First of all, this thesis emphasizes that the importance of the clinical consideration in the designing of rehabilitation robots. This is actually the driving force in the changes of the rehabilitation robots. The clinical-related overview for the development of a rehabilitation robot is shown in Figure 2.20.

In conclusion, the rehabilitation robots should be designed to assist therapies to perform the task, which is hard or impossible for the therapist to cope with and provide the gait rehabilitation exercise to patient. Meanwhile, it should be easy to use for the therapists.

According to this objective, the contribution and research scope are presented in Figure 2.21.

2.4.1 “Assist-as-needed” joint control

Enabling active participation with control strategy is the relatively new research field in robotic assisted gait rehabilitation. In comparison, therapists can “feel” the patient and encourage the patient to participate in the process of rehabilitation. This is the major advantage of therapist manual-assisted gait rehabilitation. Although active participation is being researched intensively in those robotic system, but the therapist input for this segment can be hardly replaced [133]. Doctors often emphasize rehabilitation training should be provided in the range of the patient’s strength. Too much external assistance will negate the rehabilitation effect. In a word, totally passives cannot supply energy to the lower limbs [87, 134].

In order to improve the active participation level, the current work has studied the control of external robotic-assisted force, either assistance or resistance force. The function of the stabilizer is investigated on the human model to re-establish the stable motion control for walking training. The resultant augmented process with the stabilizer provides assistance or resistance in the rehabilitation training program.

2.4.2 Individual motion generation for walking

Walking is a crucial and complex activity. The two-legged locomotion allows the humans to view the world in an upright manner and liberates upper extremities for complex tasks [135]. Although most healthy individuals take walking for granted, this activity represents an essential threshold for an independent lifestyle in daily living. Furthermore, walking displays certain personal peculiarities superimposed upon the basic pattern of bipedal locomotion. That is why individuals can be recognized by friends in a great distance.

Therefore, In view of the difference in gait profiles of each individual subject, it is crucial to include the their own gait profile as a factor to be considered for planning gait rehabilitation for patients [102].

Most developed systems have pre-programmed gait trajectory independent of individuals [99, 136]. It has been highlighted that some patients developed a compensation walking pattern which will change the motion for the affected joint, a non-adjustable walking motion will injure the joint or muscle. Therefore, a customized gait trajectory is generated for a specific training task in clinical application. Also, a ground impact problem might surface during overground walking for the exoskeleton. Poor coordination to walking speed would introduce foot-dragging problems. Planning for a suitable gait motion is not easy as in the development of the biped robot [101].

In order to generate the OGW motion for individuals, the trajectory of ankle is then obtained via curve fitting based on specified points. Next, in order to keep the desired foot trajectory, the pelvic motion is compensated to evaluate hip and knee joint angles. Furthermore, the metatarsal trajectory is planned to satisfy the condition of a smooth OGW. Therefore, the motion generation of an adjustable OGW pattern for individuals is proposed by virtue of the inverse kinematics analysis together with smoothed ground contact conditions to optimize the generated joint angles.

2.4.3 Repetitive and adaptive control

In order to track the OGW motion, rehabilitation robot motion control has entered the intelligent control stage. For example, fuzzy logic combined with PID control into the force/position hybrid control [108], modified impedance control [82, 90, 112] and neural network technology [38, 111, 113-115] to adapt to the change of subject's condition. Furthermore, virtual force/velocity field control [87], path control [137] and virtual model control [92] are proposed for specified devices. In this thesis, the current work focuses on

the “repetitive” motion input for the lower limb, the intelligent control algorithm for the periodic signals are used. The improved repetitive control algorithm based on the characteristics of a cyclical reference signal and the internal model control principle has developed by implanted the periodic signal generator into the closed-loop system in order to achieve steady tracking of cyclic reference signals and rejecting disturbance on the rehabilitation robot. Not only the tracking performance is improved, the assist-as-needed force can also be under control. Furthermore, in order to control the walking velocity on time for the OGW system, an adaptive control is proposed based on on-line model identification algorithm. An autotuning controller is developed to adjust motions for the sub-systems.

2.4.4 Gait assessment and evaluation

As for system evaluation with subjects, it is useful to provide the comparison between robotic-assisted and manual-assisted rehabilitation training methods. However, there is a lack of quantitative evaluations on the effectiveness of the exoskeleton or orthotic devices in performing specific tasks [75]. It is challenging to develop a uniform and standard evaluation methodology as the walking pattern varies by individuals. Recent works [45, 127] investigated machine-related effects with unnatural muscle activations and modulation of the machine through the study of the average muscle pattern of subjects training with the respective gait system. By extending the average muscle profile analysis to the overground gait system, walking speed on the ground is a crucial parameter to control if compared with that of a treadmill system.

Therefore, a systematic motion control method is proposed and three major contributions are listed and will be presented in Chapters 3-5. The implementation of the motion control on an OGW device is provided in Chapter 6. Clinical trials and system evaluations are presented in Chapter 7.

Figure 2.21 Brief summary of this thesis on control of OGW rehabilitation robot.

Chapter 3 MODELING OF HUMAN-MACHINE INTERACTION (HMI)

Based on the literature review and discussion in Chapter 2, the gap in current research has been provided between clinical application and robotics control technology, whilst technical solutions are being achieved in the field of robotics and knowledge is developing in clinical rehabilitation engineering. In order to provide engineering solutions to clinical problems confronted by individuals with disabilities, the work target is first to build a bridge linking between the clinical and engineering aspects in rehabilitation research. Therefore, subject, system and environment and their relationship are provided in Figure 3.1 for design specification and control guideline in the robotic system.

Figure 3.1 Subject, system and environment.

In this Chapter, modeling of human-machine interaction (HMI) is introduced. The neuromusculoskeletal motion is modeled as an internal model control and the perfect controller acted by brain are presented. After the general description of healthy human

motion, the pathologic joint is modeled as an unstable model. The distribution of muscle activation power within one gait cycle is also presented for a quantitative assessment of pathological gait. Different gait divisions are used for quantitative assessment.

3.1 Modeling of Human-motion Control

As nervous system allows communication between the brain and joint motion including sensory function, integrative function (processing of information, decision making) and motor function, human motor control needs nervous system to detect, stimulate and coordinate. Nervous system made up of brain, spinal cord and nerves can be anatomically divided into the central nervous system (CNS) and peripheral nervous system (PNS). Top-down analysis for human motor control starts from a nerve impulse in the central nervous system and ends with the generation of movements and external forces or stimulus [138]. The motion control system of a synovial joint is regards as a feedback control loop shown in Figure 3.2.

Figure 3.2 Human motion control for a synovial joint.

Brain refers to primary function is to control the actions as the “high-level controller”, which extracts relevant sensory information and manages complex task after decision making, motion planning, etc. q_d is the desired motion and q is the actual motion for the human synovial joint. The desired motion q_d is generated by brain. In order to follow desired motion, biomechanical model acts as the “low-level controller”, including pair of antagonist muscle, one of them contract τ_+ while the other relax τ_- under control of spinal cord control output u_+ and u_- . Other coupling muscles introduce disturbance caused by

torque τ_{dis} and position q_{dis} . The sensory organ feedback the actual motion q close the control loop.

The control structure of human motor control is shown in Figure 3.3. $G(s)$ is modeling of human synovial joint. $B(s)$ is biomechanical model for the spinal and a pair of antagonist muscle. τ is the sum of muscles. q_d is the desired motion and q is the actual motion for the human synovial joint.

Figure 3.3 Control structure of human motion control.

3.1.1 Internal model

$G(s)$ is the synovial joint model and $B(s)$ is the brain. The above unity feedback loop can be equivalent to an internal model control (IMC) structure through

$$B(s) = \frac{Q(s)}{1 - G^*(s)Q(s)} \quad (3.1)$$

where $Q(s)$ is an IMC controller [139]; $G(s)$ is the joint muscular skeleton model; and $G^*(s)$ is the joint motion-torque estimator and the estimated musculoskeletal model of joint motion, as shown in Figure 3.4.

When the internal model $G^*(s)$ is equal to $G(s)$ as in healthy subjects' motion control, $Q(s)$ acts as a perfect controller. Because the healthy human control behaves identically to structure of the internal model, the motion output is same to the desire value. The perfect controller can be expressed according *theorem 2* of IMC [140].

$$Q(s) = G^{*-1}(s) \quad (3.2)$$

Then, the unity feedback loop can be equivalent to an internal model control (IMC) structure and the brain model $B(s)$ can be obtained.

Figure 3.4 Assumption of internal model control (IMC) of human motion control: (a) IMC; (b) alternative form of IMC.

Under the nominal condition, the sensitivity $S(s)$ of closed-loop system can be expressed as

$$S(s) = \frac{1}{1-G(s)B(s)} = 1 - G(s)Q(s) \quad (3.3)$$

and the complementary sensitivity $T(s)$

$$T(s) = \frac{G(s)B(s)}{1-G(s)B(s)} = G(s)Q(s) \quad (3.4)$$

Denote with $H(s)$ the transfer function matrix from desire motion q_d and disturbance torque τ_{dis} to actual motion $q(s)$ and torque $\tau(s)$

$$\begin{bmatrix} q(s) \\ \tau(s) \end{bmatrix} = H(s) \begin{bmatrix} q_d(s) \\ \tau_{dis}(s) \end{bmatrix} \quad (3.5)$$

where

$$H(s) = \begin{bmatrix} G(s)Q(s) & G(s)(1 - G(s)Q(s)) \\ Q(s) & -G(s)Q(s) \end{bmatrix} = \begin{bmatrix} T(s) & G(s)S(s) \\ Q(s) & -T(s) \end{bmatrix} \quad (3.6)$$

The closed-loop system is supposed to be internal stable if all transfer function in $H(s)$ are stable. The sufficient and necessary condition of internal stability is

(1) $Q(s)$ is stable;

(2) $G(s)(1 - G(s)Q(s))$ is stable.

The above conditions are equivalent to that $Q(s)$ is stable and satisfies the following constraints:

$$\lim_{s \rightarrow \infty} (1 - T(s)) = 0 \quad (3.7)$$

$$\lim_{s \rightarrow \infty} \frac{d}{ds} (1 - T(s)) = 0 \quad (3.8)$$

In fact, these two constraints above have to be satisfied in the desired closed-loop transfer function $T(s)$ in order to maintain the internal stability of the closed-loop control system, including loading disturbance rejection and dynamic tracking.

3.1.2 Perfect control of the brain

Assume that the $G(s)$ is stable process under control of brain, the model is simply presented as a linear first-order system given by

$$G(s) = \frac{k}{\tau s + 1} e^{-\theta s} \quad (3.9)$$

where k is the control gain, τ is the time constant, and θ is the time delay. Then, the perfect controller is under exact modeling are

$$Q(s) = \frac{\tau s + 1}{k e^{-\theta s}} \quad (3.10)$$

It is non-proper. A filter $F(s) = \frac{1}{(\lambda s + 1)^n}$ is added to model the brain control function, yielding

$$q(s) = \frac{B(s)G(s)}{1-B(s)F(s)[G(s)-G^*(s)]} q_d(s) + \frac{1-B(s)F(s)G^*(s)}{1-B(s)F(s)[G(s)-G^*(s)]} q_{dis}(s) \quad (3.11)$$

worked as a H_∞ optimal controller for further understand the function, an executable $T(s)$ in terms of optimal performance objective is proposed as:

$$T(s) = Q(s)G(s)F(s) = \frac{e^{-\theta s}}{(\lambda s + 1)^n} \quad (3.12)$$

where n is chosen so that the control is proper both for $Q(s)$ and $\frac{1}{Q(s)}$. Here λ is a positive real constant, by adjusting λ , the nominal performance and robustness are tuning for different situations. For example, the nominal performance can arbitrarily approach by decreasing λ . If $\lambda = 2$, The corresponding brain control $B(s)$ of the unity feedback system is

$$B(s) = \frac{1}{k} \frac{\tau s + 1}{(\lambda s + 1)^2 - e^{-\theta s}} \quad (3.13)$$

If $B(s)$ is a standard PID controller,

$$B(s) = k_p + \frac{1}{k_i} + k_d s \quad (3.14)$$

The brain control should be realized by approximated by a rational transfer function in Equation (3.13). So the condition of asymptotical performance results in

$$\lim_{s \rightarrow 0} ((\lambda s + 1)^2 - e^{-\theta s}) = 0 \quad (3.15)$$

Here $B(s)$ is expanded in Maclaurin series as

$$B(s) = \frac{f(s)}{s} = \frac{1}{s} (f(0) + \dot{f}(0)s + \frac{\ddot{f}(0)}{2!} s^2 + \dots) \quad (3.16)$$

From the first three terms of the above expansion can be interpreted as standard PID controller. $k_p = \dot{f}(0)$, $k_i = \frac{\ddot{f}(0)}{2!}$ and $k_d = f(0)$. Therefore, the brain control can be predicted by the model of human motion control. It is similar to perfect control for the internal model control. So the control theory is extent to the human motion control. The brain acts as the perfect in our study. However, the pathologic joint acts differently that out of control for brain. In order to design controller to compensate the brain, the model of the pathological joint motion control is also studied.

3.1.3 Modeling of motion control for the pathologic joint

In the normal human motion control, the neuro-musculoskeletal system acts as a perfect controller so that various parts can be coordinated including the body segment, muscles, sensors which are all over the body under control of nervous system. A robust and adaptive tracking behavior present when transfer from the actual point to the set point under disturbance forces such as wind or friction may have effect on the body. However, if the muscles are physically impaired or load exceeds the maximum muscular face, the resulting position deviates from the intended position and intrigue the instability motion. Therefore, the mismatch between the internal model and the actual model causes wrong torque output for the pathological gait. Assume the transfer function from the actual τ to q is human motion control behavior:

$$G(s) = \frac{q(s)}{\tau(s)} \quad (3.17)$$

For example, an unstable process with time delay could be used as a simple model for the relationship between torque and position with unstable performance of pathologic joint. The right-half plane (RHP) poles makes system unstable. Suppose that a more detailed model has a time delay, yielding the transfer function:

$$\tilde{G}(s) = \frac{k}{\tau s - 1} e^{-Ts} \quad (3.18)$$

Some pathologic gait like Parkinsonian gait suffers from involuntary motion, which is similar to the sensitive system out of control. In this case, reduce the control gain or enlarge the error tolerance may help to return to the homeostatic mechanisms of human body. When muscle spasms are present, the resistive force provides a heavy load for the muscle to stabilize. All those assistive forces can be provided from an exoskeleton, either to assist or resist extremities is effective for people with impaired motion control capability. Although the brain is intelligence in the normal motion control as an optimal controller, it cannot work properly caused it is an unstable model for the pathologic motion control.

According to the Maximum modulus theorem

$$\|W(s)S(s)\|_{\infty} = \|W(s)(1 - G(s)Q(s)\|_{\infty} \geq |W(s_0)S(s_0)| \quad (3.19)$$

The optimal controller is obtained as follows

$$Q_{opt}(s) = \frac{W(s) - W(s_0)S(s_0)}{W(s)G(s)} \quad (3.20)$$

Using 0/1 order Pade approximation, the process is

$$G(s) = \frac{k}{(\tau s - 1)(1 + Ts)} \quad (3.21)$$

As $G(s)$ has zeros in the right half plane, there is a positive lower bound for the maximum value of $|W(s)S(s)|$. Let s_0 be the zero of $G(s)$ in the right half plane and should be greater than the bound, then $|W(s_0)S(s_0)| = |W(s_0)| \geq \frac{1}{s_0}$. When $G(s)$ has no zeros in the right half plane, assume

$$\|W(s)(1 - G(s)Q(s))\|_{\infty} \geq 0 \quad (3.22)$$

In this case, the optimal controller is

$$Q_{opt}(s) = \frac{(\tau s - 1)(1 + Ts)}{k} \quad (3.23)$$

As $Q_{opt}(s)$ is improper. A filter with the pole-zero excess in the form of are introduced

$$F(s) = \frac{as + 1}{(\lambda s + 1)^3} \quad (3.24)$$

$$a = \frac{\lambda^3}{\tau^2} + \frac{3\lambda^2}{\tau} + 3\lambda \quad (3.25)$$

$$Q(s) = Q_{opt}(s)F(s) = \frac{(\tau s - 1)(1 + Ts)(as + 1)}{k(\lambda s + 1)^3} \quad (3.26)$$

It is derived that the stability condition as the ratio of the time delay to the time constant being less than 1 for controller [141].

3.1.4 Comparison of motion control between healthy and pathologic joint

Assume the healthy gait model have $\tau = 2$, $k = 1$, $T = 1$ while the pathologic gait model also have $\tau = 2$, $k = 1$, $T = 1$, the optimal controller is calculated as $B(s) = 2.198 \cdot (1 + \frac{1}{2.418s} + 0.332s)$, the performance of the step response for the motion control is compared in Figure 3.5. As shown, under control of optimal control, the motion control of healthy joint present stable characteristics and quick response follow the command. However, the pathological joint turns out to be unstable process out of control.

Figure 3.5 Comparison the performance between stable (healthy joint) and unstable (pathologic joint) process with optimal controller $B(s)$.

3.2 Assessment of Pathological Gait

Alterations in muscle activation activity are investigated for pathological gait compared to that of healthy subjects walking. To distinguish the pathological gait, the distribution of muscle activation power within one gait cycle is studied. The coordination and adaptation of muscles for pathological gait are presented in frequency analysis for a quantitative assessment, which is an effective tool and likely contribute to qualified evaluation and bio-feedback of gait rehabilitation in future.

3.2.1 Recruitment of participants

Ten healthy subjects and seven SCI subject with walking problem but as ASIA D can regain walking ability participated in this study (Table 3.1 and Table 3.2). Informed written consent was obtained and subjects were informed about the study procedure, but not about the purpose of the investigations so as to avoid any influence on the parameters recorded. Safety precaution certification and approval from the local Ethics Committee were obtained before the trial.

Table 3.1: Basic Information of Healthy Subjects

Subject	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	Mean	S. D.
<i>Age</i>	23	23	26	24	23	25	22	25	29	23	24.3	1.56
<i>Height (m)</i>	1.75	1.75	1.71	1.71	1.79	1.70	1.73	1.73	1.73	1.74	1.73	0.02
<i>Weight (kg)</i>	60	68	67	62	72	60	80	74	65	57	66.5	5.7

Table 3.2: Detailed Information of SCI Subject Group

Subject Number	P1	P2	P3	P4	P5	P6	P7
<i>Age</i>	47	54	37	57	63	64	60
<i>Years of injury</i>	3	3	9	6	3	8	9
<i>Impairment</i>	SCI	SCI	SCI	SCI	SCI	SCI	SCI
<i>Skeletal Level Lesion</i>	C6-7	C3-5	C6	C3-4	C5	C5	C4/5

3.2.2 EMG data collection

The EMG data were collected at 1500 Hz with Telemetry 2400T G2 (Noraxon, USA Inc.). Noraxon self-adhesive silver-silver-chloride dual snap electrodes for surface were attached to muscles of the lower limbs. The EMG signals were then sent the signal via Bluetooth transmission to the host personal computer. Electrodes using silver/silver chloride (Ag/AgCl) surface electrodes (Ambu Blue Sensors diameter 8 mm) were placed on the muscle surface are able to remotely detect EMG signals of different muscles during walking, as shown in Fig. 1 on vastus lateralis (VL), rectus femoris (RF), semitendinosus /semimembranosus (ST), bicep femoris (BF), tibialis anterior (TA), soleus (SO) and gastrocnemius medialis (MG).

Figure 3.6 Superficial muscles of the right leg in anterior view and posterior view: soleus (SO), tibialis anterior (TA), gastrocnemius medialis (MG), vastus lateralis (VL), rectus femoris (RF), semitendinosus / semimembranosus (ST) and gluteus maximus (GM) and patients walking trials with assistance.

Raw EMG data were saved for subsequent offline processing, this was done in conjunction with the synchronized video recordings of the walking. In order to divide gait cycle into phases, GAITRite walkway is implemented. GAITRite has recorded footfall provide gait events such as heel contact and toe off in the right and left sides. After offset the starting point by the synchronized video recording, gait events are automatically linked to EMG data. Similar methods to detect the gait events are presented by pressure sensor [142-143] for the online monitoring. Furthermore, time markers were placed at right heel strike as determined from the synchronized video records for the EMG recording. At the same time, the starting point is also synchronized by the footfall recorded down by GAITRite as shown in Figure 3.6.

3.3 Distribution of Muscle Activation

To classify the muscle groups, the gait cycle (GC) was divided into several specific phases. The peak value instead of MVC at the initial (0 – 30% of gait cycle), middle (30 – 60% of gait cycle) and final phase (60 – 100% of gait cycle) in GC under normal speed were used, see Table 3.3.

Figure 3.7 summarizes the mean processed and normalized EMG with standard deviation (SD) for these three phases on eight of the lower limb muscles respectively. Group 1: VO and GM are grouped together as they predominately functioned during the initial phase of the loading response and mid-stance stages (0 – 30% of gait cycle); Group 2: SO and MG were grouped together since they predominately functioned during the middle phase of the terminal stance and pre-swing (30 – 60%) stages; Group 3: TA, RF, BF and ST were classified into multi-phases active muscle group in that they activate more than one phase of the GC involved both the stance and swing stages, which may contribute to balance control throughout GC.

Major movements of lower limb joints in walking are studied. Muscles involved in ankle plantar flexion are SO and MG in Group 2, which act mainly before toe off (around 60% of GC) in GC to provided propulsion for walking. GM in Group 1 is the major extensor muscle and plays an important role in hip extension, which occurs in the initial phase of GC before bearing weight transferred from the other low extremity. As a result, different muscle patterns are shown under different functions named propulsion-related (P-R), support-related (S-R) and multiple phases active muscle group as grouped to balance-related (B-R). Muscle groups to achieve the same function in the gait cycle shared similar patterns of muscle activation. Although muscles work multi-functionally, significant factors for each function were tested. Further analyses are provided using this functional classification.

Table 3.3 EMG Functions in the Initial, Middle and Final Phases of Gait Cycle at Free Speed^a

<i>Phase of gait cycle</i>	<i>Muscle</i>	<i>Sbj 1</i>	<i>Sbj 2</i>	<i>Sbj 3</i>	<i>Sbj 4</i>	<i>Sbj 5</i>	<i>Sbj 6</i>	<i>Sbj 7</i>	<i>Mean</i>	<i>Standard deviation</i>
<i>Initial phase: Loading response and mid-stance (0-30%) :</i>	SO	0.51	0.49	0.27	0.07	0.33	0.31	0.20	0.31	0.16
	TA	0.88	0.15	0.48	1.00	0.28	0.75	0.50	0.58	0.31
	MG	0.20	0.68	0.28	0.23	0.23	0.51	0.33	0.35	0.18
	VO	1.00	0.21	0.65	0.44	0.60	0.15	0.13	0.45	0.32
	RF	0.41	0.09	0.17	0.08	0.17	0.20	0.13	0.18	0.11
	BF	0.25	0.62	0.19	0.21	0.43	0.44	0.17	0.33	0.17
	ST	0.73	0.35	0.53	0.24	0.56	0.57	0.16	0.45	0.20
	GM	0.19	0.25	0.10	0.03	0.34	NA ^b	NA ^b	0.18	0.12
<i>Middle phase: Terminal stance and pre-swing (30-60%) :</i>	SO	0.81	0.75	0.69	0.20	1.00	0.12	0.84	0.63	0.33
	TA	0.52	0.08	0.18	0.37	0.27	0.54	0.22	0.31	0.17
	MG	0.51	0.89	1.00	0.16	0.73	0.12	1.00	0.63	0.38
	VO	0.29	0.13	0.44	0.15	0.34	0.75	0.09	0.31	0.23
	RF	0.09	0.05	0.13	0.04	0.10	0.59	0.06	0.15	0.20
	BF	0.21	1.00	0.11	0.23	0.37	0.49	0.04	0.35	0.32
	ST	0.36	0.34	0.22	0.22	1.00	0.74	0.05	0.42	0.33
	GM	0.09	0.08	0.05	0.03	0.20	NA ^b	NA ^b	0.09	0.07
<i>Final phase: Initial swing and terminal swing (60-100%) :</i>	SO	0.11	0.16	0.58	0.03	0.23	0.50	0.33	0.28	0.20
	TA	0.51	0.17	0.31	0.45	0.46	0.86	0.36	0.45	0.21
	MG	0.05	0.08	0.29	0.04	0.10	1.00	0.08	0.23	0.35
	VO	0.35	0.07	0.20	0.16	0.24	0.43	0.05	0.22	0.14
	RF	0.14	0.05	0.09	0.05	0.51	0.38	0.06	0.18	0.19
	BF	0.26	0.22	0.19	0.13	0.17	0.28	0.08	0.19	0.07
	ST	0.68	0.11	0.25	0.09	0.27	0.52	0.13	0.29	0.23
	GM	0.10	0.04	0.05	0.03	0.19	NA ^b	NA ^b	0.08	0.07

^a free speed is normalized speed 3 in the experiment; it is also preferred walking speed of the subject; ^b not available in recording

Figure 3.7 EMG distribution in phases of gait cycle (GC) and stance time changes: mean value of EMG within initial, middle and final phases of GC.

3.4 Distribution of Muscle Activation via Frequency Analysis

With help of GAITRite, footfall is recorded so that gait events such as heel contact and toe off in both sides is synchronized to EMG recording. Then, the average gait cycle (GC) are divided four phases: right single stance phase, double stance phase, left single stance phase and swing phase. Other commonly used methods to divide GC are compared.

The root mean square value (RMS) of each FFT for muscle groups was normalized against their corresponding peak RMS value to represent the muscle energy. After the EMG signal in frequency domain of each interval was then calculated by the formula:

$$fRMS = \frac{1}{N} \sqrt{\sum x^2(f)} \quad (3.27)$$

where N is the total number of samples within the window considered for processing.

As reported, weakness of muscles produce synchronized pattern of all for a quick swing. It is assumed that inappropriate coactivations emerged, which will distinguish

pathological gait from control groups. The ill-timing for the muscle activation changes the distribution of muscle activation.

Table 3.4: Mean (SD) normalized EMG activation in frequency domain for each group during select phases of the gait cycle. Each power spectrum of muscle' activity level was normalized to the peak rectified activity

Subjects	Muscle	Right Single Support Phase (% GC)	Double Support Phase (% GC)	Left Single Support Phase (% GC)	Swing Phase (% GC)
Healthy Subjects	<i>VL</i>	0.343(0.204)	0.393(0.234)	0.113(0.067)	0.028(0.017)
	<i>RF</i>	0.056(0.033)	0.034(0.020)	0.008(0.005)	0.079(0.047)
	<i>ST</i>	0.815(0.484)	1.122(0.666)	0.338(0.201)	0.034(0.020)
	<i>BF</i>	0.023(0.014)	0.031(0.019)	0.031(0.019)	0.007(0.004)
	<i>TA</i>	0.231(0.137)	0.346(0.205)	0.097(0.057)	0.300(0.178)
	<i>SO</i>	0.038(0.023)	0.145(0.086)	0.115(0.068)	0.005(0.003)
	<i>MG</i>	0.019(0.011)	0.087(0.052)	0.040(0.024)	0.001(0.001)
SCI Subjects	<i>VL</i>	0.040(0.049)	0.065(0.078)	0.150(0.182)	0.015(0.018)
	<i>RF</i>	0.011(0.013)	0.008(0.010)	0.029(0.035)	0.006(0.008)
	<i>ST</i>	0.191(0.232)	0.134(0.163)	0.052(0.064)	0.074(0.089)
	<i>BF</i>	0.272(0.329)	0.216(0.261)	0.114(0.138)	0.092(0.112)
	<i>TA</i>	0.143(0.173)	0.005(0.006)	0.033(0.040)	0.119(0.144)
	<i>SO</i>	0.104(0.126)	0.103(0.125)	0.176(0.213)	0.040(0.049)
	<i>MG</i>	0.078(0.094)	0.078(0.094)	0.040(0.048)	0.029(0.035)

Therefore, divided phases within gait cycle are studied for the muscle activation distribution as shown in Table 3.4. Mean (SD) normalized EMG activation in frequency domain for each muscles on every divided phases of the gait cycle. In statistical significance testing, the confidence interval P -value ($P \leq 0.01$) is chosen. Although

previous works [132] mentions that the P -value ($P \leq 0.01$) chosen may be too restrictive for such research, a P -value of 0.05 is usually chosen when study gait. The confidence interval is set as $P \leq 0.01$. The smaller P -value makes our diagnosis more accurately for detection of pathological gait.

Figure 3.8 Summary of the mean integrated muscle activation: (a) vastus lateralis (VL), (b) rectus femoris (RF), (c) semitendinosus / semimembranosus (ST), (d) tibialis anterior (TA), (e) soleus (SO) and (f) gastrocnemius medialis (MG).

As shown in Figure 3.8, small confidence interval P -value ($P \leq 0.01$) occur on thigh muscles other than calf muscles. The muscle activation distribution of VL and RF are similar, especially in first three phases (i.e. right single support, double support and left single support, totally stance phase). At first two phases activation of healthy subject is bigger than that of SCI subject. Then, muscle activity of SCI subject enlarges and turns out to be bigger than that of healthy subject in the third phase. Hence, increasing muscle activation tendency of SCI subjects is presented in stance phase. However, muscle activity

of SCI subjects returns to less than that of healthy in the fourth phase (i.e. swing phase). Especially, a great increasing of normal muscle RF is shown in the fourth phase, while the SCI clearly lost this increasing activation. Muscle ST shows much less muscle activity in first three phases on SCI subject. In general, less muscle activation are shown in VL, RF and ST of SCI subjects than control group. Another thigh muscle BF is quite different from VL, RF and ST because the muscle activity of SCI subject is larger than the healthy subject throughout entire gait cycle.

Although there is not as much significant different in calf muscles between SCI and healthy subjects as P -value ($P \leq 0.01$), the most significant difference occurs on TA in the double stance phase. Less muscle activation of TA is shown for four phases. SO and MG do not show obvious difference between SCI and healthy subjects. It should also be pointed out that the SCI muscle activity of SO and MG in the swing phase are larger than that of healthy subjects.

According to the literature survey, four methods are commonly used to divide GC. Method 1: stance and swing phases [27] within GC are compared separately for the significant difference of walking between healthy subjects and subjects with walking problem. Method 2: completion percentage [31-32] of GC is commonly used. However, this standard GC cut off method is presented at the customary 83 m/min (1.36 m/s) [33] rate of walking cannot be used directly because walking speed is slower than 0.2 m/s in this study as the SCI patient cannot walk as fast as normal group. Method 3: evenly divided GC [34] in each phase bin is used in the comparison of muscle activity between healthy and SCI subject. Method 4: gait events [26, 35-36] such as right heel contact (RHC), left toe off (LTO), left heel contact (LHC), and right toe off (RTO) can divide GC into right single support phase, double support phase, left single support phase and swing phase is selected for our study. The P -value of four methods is presented in Table 3.5. The most significant differences are highlighted.

Table 3.5: Comparisons of P-value of four methods to divide gait cycle

P-value	GC	VL	RF	ST	BF	TA	SO	MG
Method 1 [24]	Stance	0.0121	0.0713	<u>0.0029</u>	0.0536	<u>0.0085</u>	0.7671	0.7026
	Swing	0.0941	<u>0.0012</u>	0.2539	0.0368	0.0320	0.0488	0.0291
Method 2 [25-26]	0-10%	<u>0.0007</u>	<u>0.0007</u>	<u>0.0040</u>	<u>0.0299</u>	<u>0.0055</u>	0.0421	0.0235
	10-30%	<u>0.0008</u>	<u>0.0007</u>	<u>0.0016</u>	0.0383	<u>0.0003</u>	0.6927	0.7219
	30-60%	<u>0.0024</u>	0.0346	<u>0.0005</u>	0.2239	<u>0.0006</u>	0.1603	0.3120
	60-73%	0.0043	0.9442	0.7538	0.1105	0.0566	0.1033	0.4627
	73-87%	0.0499	<u>0.0004</u>	0.0897	0.0336	0.1473	0.0326	0.0236
	87-100%	<u>0.0013</u>	<u>0.0004</u>	0.4151	0.0244	0.0139	0.0439	0.0262
Method 3 [28]	0-25%	<u>0.0045</u>	<u>0.0025</u>	<u>0.0073</u>	0.0306	0.0034	0.1951	0.1058
	25-50%	<u>0.0049</u>	0.0115	<u>0.0020</u>	0.1285	0.0012	0.3740	0.5254
	50-75%	<u>0.0079</u>	0.0881	<u>0.0049</u>	0.1509	0.0922	0.0203	0.2244
	75-100%	<u>0.0068</u>	0.0019	0.2181	0.0247	0.0456	0.0356	0.0213
Method 4 [29-31]	RHC-LTO	<u>0.0017</u>	<u>0.0038</u>	<u>0.0067</u>	0.0287	0.2562	0.1230	0.0641
	LTO-LHC	<u>0.0029</u>	<u>0.0073</u>	<u>0.0017</u>	0.0394	<u>0.0006</u>	0.4227	0.7874
	LHC-RTO	0.5581	0.0779	<u>0.0026</u>	0.0765	0.0238	0.4059	0.9988
	RTO-RHC	0.1296	<u>0.0011</u>	0.1841	0.0270	0.0420	0.0354	0.0214

In conclusion, the healthy and pathological gait is modeled for HMI. Then, according to the detected heel contact (HC) and toe off (TO), the average gait cycle can be divided into

four phases. It is noted that the distribution of muscle activation in gait cycle of the SCI subjects was significantly changed in comparison with that of the healthy subjects. Analysis of the muscular activation in seven major muscles, it is find that tibialis anterior (TA) in left single support phase and rectus femoris (RF) in swing phase have comparatively significant differences between the healthy and SCI subjects. Meanwhile, the confidence interval is set as $P \leq 0.01$, which is usually chosen $P \leq 0.05$ for gait research. The smaller P value makes our diagnosis more accurately for the detection of pathological gaits.

Chapter 4 ADJUSTABLE OVERGROUND WALKING PATTERNS GENERATION VIA GAIT ANALYSIS

In Chapter 3, the human biomechanical model is built up to model the healthy or pathological human motor control. In order to improve the active participation level, the function of external robotic-assisted system is simulated by HMI model. The HMI model describes the human motion control and function of the assistive force provided by external device for one joint angle. At same time, the kinematic model is developed for the trajectory plan to coordinate all joints. In this study, the kinematic model is used on generation of walking motion for the lower limb, which is investigated in Chapter 4.

Since the exoskeleton is attached to lower limbs in parallel, generation of walking motion for the lower limbs can be extended to that of rehabilitation device. Joint motion of lower limbs can be obtained by the motion capture experiment for individuals. The experiments will be presented in Section 4.1. However, it is impossible to conduct motion capture experiment for all subjects. In view of the difference in gait profiles of individual subject, it is crucial to include their own gait profile as a factor to be considered for planning gait rehabilitation for subjects [102]. Most developed systems have pre-programmed gait trajectory independent of the subject [99, 136]. Furthermore, it has been highlighted that some patients has a compensation walking pattern which changes the motion for the affected joint, a non-adjustable walking motion would injure the joint or muscle. Therefore, a customized gait trajectory is generated for a specific training task in clinical application, see Section 4.2.

It is necessary to consider the ground conditions because possible problems that may surface during overground walking pattern generation include: 1. ground impact problem. 2. poor coordination to walking speed would cause foot dragging. To reduce, if not to

avoid, the ground impact and dragging of the footpad occurring during the course of overground walking with help of the device, a gait motion generation is specified to adjust index for an impact-less and smooth footpad contact, see Section 4.3.

4.1 Motion Capture Experiment

In order to obtain the precise lower limb motion of human walking, motion capture experiment is carried out for individual subjects. The data obtained during motion capture experiment is then analyzed and processed to obtain the required motions of lower limb and pelvic. The procedures of the motion capture experiment will be briefly explained.

Table 4.1 Numbering and Position of the Motion Capture Markers

Marker Number	Denotation	Anatomical Position
M1	P_1	Right metatarsal head II
M2	P_2	Right heel
M3	P_3	Right lateral malleolus
M4	P_4	Right mid-shank
M5	P_5	Right femoral epicondyle
M6	P_6	Right mid-thigh
M7	P_7	Right ASIS [#]
M8	P_8	Left metatarsal head II
M9	P_9	Left heel
M10	P_{10}	Left lateral malleolus
M11	P_{11}	Left mid-shank
M12	P_{12}	Left femoral epicondyle
M13	P_{13}	Left mid-thigh
M14	P_{14}	Left ASIS [#]
M15	P_{15}	Sacrum

Figure 4.1 Markers placement on test subject during motion capture experiment (Labels are inserted for visible markers only, sacrum and most of the left side markers are not visible): (a) motion capture experiment setup for the joint angle data collection; (b) motions of all the markers in the frontal view while subject walk for several gait cycles.

The recoding of gait motion for this study is carried out using Eagle digital motion capture system distributed by Motion Analysis Corporation [144]. The motion capture system is a passive marker based on motion capture system. Eight cameras are used to capture a work envelope for approximately three complete strides. Adequate walkway space (the walkway is ten meters long, and capture volume is positioned at the middle of the walkway) is provided to ensure the subject enters the capture volume at rhythmic walking stage. Data are recorded walking of a subject at self-paced walking speed. The data are used to compute lower limb locomotion and pelvic motion. This work adopts the markers set introduced by Christopher et al. [145]. A total of fifteen markers are placed on subject's lower limb as per the anatomical positions labeled in Figure 4.1. The trajectory for these markers in frontal view is also shown with the anatomical positions depicted.

Raw data of 40 markers putting on whole body are tracked by different colors for three gait cycles. It is shown that the lateral shifting of the upper body during walking. In our application, the pelvic and lower limb motions are useful for our application.

4.1.1 Pelvic motion

Three markers placed at right ASIS, left ASIS, and sacrum are used to compute the trajectory of hip joint center (HJC) and then the computed HJC trajectories are needed to transfer to the side of the pelvis, as shown in Figure 4.2. In general, the computed trajectories (as shown in Figure 4.3) of the left and right sides are similar with a phase lag of 50% gait cycle. This phenomenon is more observable on trajectory of pelvis in the z -axis. For movement of pelvis on the y -axis, the trajectories of left and right sides are phase lag of 50% and mirrored with respected to the anteroposterior axis.

Figure 4.2 Motion assistance provided by PA mechanism and the relationship between motion assistance provided by PA mechanism and computed HJC trajectories are highlighted.

To achieve a reference motion for the repetitive locomotion, the average motion of one cycle will be smoothed for a desired repetition. The pelvic motion of the right side is used for further processing to obtain consecutive gait cycles. The final constructed pelvic motions of the right side, which will be used as the reference trajectories for pelvic in three dimensions, see figure 4.3.

Figure 4.3 Post processed pelvic motion constructed for the right side (three gait cycles), original data obtained via motion capture experiment: (a) in x-axis; (b) in y-axis; (c) in z-axis.

4.1.2 Lower limb locomotion

By carrying out motion capture experiment and using the coordinate of the respective markers, hip, knee and ankle joints trajectories are recoded. By using these coordinate data, the angle of the lower limb joints can be computed of trajectories for hip, knee, and ankle joints. To obtain a repetitive locomotion in robotic gait rehabilitation, the motion of one cycle can be duplicated for several repetitions as desired. Walking locomotion during rhythmic stage is remarkably consistent, thus motion of one gait cycle obtained during rhythmic stage can be re-produced to several consecutive gait cycles. However, it does not follow one single motion cycle exactly during the rhythmic phase, as shown in Figure 4.4. The slight change in landscape, stride length, cadence, and other factors caused the small variation in the walking motion.

The concept and procedure on the construction of the lower limb joint angles during walking are explained with one set of walking data. Since the gait locomotion for the left and right lower limb are symmetry with a phase lag of 50%, the following explanation will be provided for right lower limb only. The same process is applicable for the left lower limb. One gait cycle is trimmed from the data obtain via motion capture experiment. It is noticeable that the same gait cycle does not connect smoothly inter cycle, if it is duplicated to form several consecutive gait cycles. It is most visible for hip joint angles, the portion where two cycles are connected is not a smooth curve. The angle of the lower limb joints will be processed with software filter to obtain a smooth curve at the transition portion between two cycles.

Once the trajectories are filtered, the middle cycle (100% – 200%) can be trimmed and repeat to any number of repetitive cycles with smooth transition from one cycle to another. It is noteworthy that only the middle cycle gives the smoothed curve when connected. Thus, the first cycle and third cycle cannot be used for the construction of consecutive gait cycles. By using the middle cycle, three cycles are constructed. The joint

angles obtained from the motion capture experiment work fine, if the orthosis does not touch ground. However, those data might cause collision problem for the overground walking, which will be explained in next section.

Figure 4.4 Lower limb joint angles during walking (at slow speed), data obtained via motion capture experiment, four walks of same subject: (a) hip joint; (b) knee joint; (c) ankle joint.

4.2 Key-point Generation

In order to generate foot trajectories, key points on the trajectory are designated at first. In view of human walking, five events are marked for one side: heel contact (HC), foot flat (FF), heel off (HO), toe off (TO), and toe high (TH). TH occurs when the stride height reaches maximum at the swing phase. There another event marked events for the other side. Therefore, ten key points are picked up for one gait cycle. The occurrence of these events is shown in Figure 4.5. The trajectory for knee, ankle and metatarsal joints are also illustrated. The stance phase of both sides is labeled so that the single stance duration (SSD), double stance duration (DSD) and swing duration (SD) are deduced.

Figure 4.5 Event markers for one gait cycle, the red color shadow indicate the stance phase of the right side which the green for the left. So the single stance duration (SSD), double stance duration (DSD) and swing duration (SD) are labeled. The trajectory for the knee, ankle, and metatarsal joints are also demonstrated in green, red and blue line.

For the overground walking, these key points from the initial point at right heel contact to toe off are located at stance phase. The ankle trajectory is fixed on the ground from heel contact (HC) to heel off (HO). The metatarsal trajectory is fixed on the ground from foot flat (FF) to toe off (TO). The distance between these two fixed points is the distance between heel and metatarsal joint. Stance phase covers either ankle or metatarsal joint contact the ground, and both of them contact ground. In the swing phase, the stride height reaches maximum at toe high (TH) as specified by the stick diagram of walking pattern, see Figure 4.6. The step length, stride length, cadence, and ratio between stance phase and swing phase can also be designated. Even the maximum flexion and extension joints angle for hip, knee, and ankle can be constrained in the proposed method. The dimension of the

lower limbs can be pre-defined for the motion generation, too. Therefore, various gait patterns can be generated.

Figure 4.6 Key points definition as seen from fixed external observer: stick diagram of the walking pattern, with zooming in the stance phase.

To simplify analysis, only the right side is presented. Since the walking gait cycle is assumed to be bilaterally symmetric, the left side is assumed to be 50% phase delay to the right side. As consider our mobile design on a platform, the global coordination is converted to local origin on a point on hip. The stick diagrams are shown in Figure 4.7. The ten key points and one hundred points are illustrated for the walking motion. As can be seen in the illustration, the foot trajectory decreases in the x direction and then returns to the initial point to complete the cyclic trajectory in the local coordination.

pelvic are planned as mentioned, the hip and knee joint angles can be obtained combined the planned ankle trajectory in x -direction and y -direction. The ankle angle can be obtained from the ankle and metatarsal trajectory in y -direction. Those three trajectories planned according to pre-defined stride length and height is presented.

Figure 4.8 Three rocker bio-mechanisms of foot contact during the stance phase: (a) rocker at an angle on heel above the ground level with fixed contact point on the ground (b) the foot is kept on contacting the ground in flat on a fixed place while ankle joint rotating. (c) rocker at an angle on metatarsal joint above the ground level with fixed contact area on the ground.

If the maximum of hip flexion angle and extension angle are assumed to be same for simplicity, they can be calculated from the step length. So l_3 is the length of the thigh; l_4 is the length of the calf; l_5 is the length between ankle joint to metatarsal. Therefore, the maximum and minimum angle for flexion and extension are expressed as

$$\theta_{hip-max} = \sin^{-1} \frac{S_L}{2(l_3+l_4)} \quad (4.1)$$

$$\theta_{hip-min} = -\sin^{-1} \frac{S_L}{2(l_3+l_4)} \quad (4.2)$$

$$v_{mp} = \frac{S_L}{S_T} \quad (4.3)$$

where $\theta_{hip-max}$ and $\theta_{hip-min}$ are the maximum and minimum hip joint angle; v_{mp} is the speed of mobile platform; S_L is step length; S_H is step height; and S_T is step period. Actually, the proposed trajectory generating on the assumptions of $|\theta_{hip-max}| = |\theta_{hip-min}|$, the knee joint is not bend at the initial heel contact, and the heel is half of the step length away from the origin in the horizontal plane for the sake of simplification. However, these assumptions ignore the complicated details in normal human walking, as shown in Figure 4.9. For example, although with large angle of hip joints, the center of gravity (CG) drop to the lowest position, the walking process moves the center of gravity (CG) forward compared to stance feet. So without measuring the precise value of the angle, approximation is made for the analytical motion generation.

Then, in order to generate the trajectory for ankle on the key points as mention, specified points are expressed in x -direction and y -direction of the Sagittal plane. Pre-described key points at ankle and hip are indicated, taken ten points as an example:

$$X_a = x_a(0 \sim 10); Y_a = y_a(0 \sim 10) \quad (4.4)$$

$$X_h = x_h(0 \sim 10); Y_h = y_h(0 \sim 10) \quad (4.5)$$

Before generation of trajectory for ankle joint, specified points in x -direction x_a are calculated. First of all, the $x_a(0)$ and $x_a(6)$ indicated critical point from stance to swing or from swing to stance, are presented:

$$x_a(0) = (l_3+l_4) \sin \theta_{hip-max} \quad (4.6)$$

$$x_a(6) = (l_3+l_4) \sin \theta_{hip-min} \quad (4.7)$$

$$x'_a(0) = -v_{mp} \quad (4.8)$$

$$x'_a(6) = -v_{mp} \quad (4.9)$$

where $x_a(0) \sim x_a(6)$ indicate the stance phase while foot stay on a fixed point from the global coordinates. It is converted into local coordinate on the mobile platform at v_{mp} . $x_a(0)$ is the starting point of one gait cycle, i.e. heel contact (HC); $x_a(6)$ is the turning point from stance to swing, i.e. toe off (TO). The intermittent impact between the foot and the ground are discontinuities in state. Therefore, according to the difference between stance and swing model, two curve fitting are conducted for obtaining the other point using interpolation algorithm. If assume a constant speed for the walking, it turns out to be a linear interpolation in this application.

Figure 4.9 The relationship between maximum and minimum hip joint angle.

The estimation of y_a in the y-direction is more complicated than that in the x-direction. Except the turning point between the stance and swing phase, the movement and direction of feet is also participated the analytical expressions of the angle position. In the stance phase, heel of is a turning point $y_a(3)$ of lifting up the ankle but keeping part of foot

contacting ground. The angle motion draws a circle-curve based on the origin of the metatarsal joint fixed on the ground. Before that the ankle joint is fixed on the ground, so

$$y_a(0\sim3) = -l_3 - l_4 \quad (4.10)$$

$$y_a(6) = l_5 \sin \theta_{hip-max} - l_3 - l_4 \quad (4.11)$$

In the swing phase, another turning point $y_a(7)$ occurs when the step height reaches the maximum value. These critical points in y-direction y_a are calculated:

$$y_a(7) = S_H - l_3 - l_4 \quad (4.12)$$

where

$$\alpha_M = \frac{D \cdot S_H}{l_5} \quad (4.13)$$

D is scaling index. Therefore, the specified ten points are generated. Assume $l_3 = 40cm$, $l_4 = 40cm$, $l_5 = 10cm$, $S_L = 100cm$, $S_H = 10cm$, and $D = 0.9$ then specified ten points at ankle joint are shown in Figure. 4.10.

Figure 4.10 Ankle trajectory at specified points as seen from an external observer moving with pelvic velocity in x-direction.

Then, the inverse kinematics are provided as

$$\sigma_a = \frac{X^2 + Y^2 + l_3^2 - l_4^2}{2l_3 \sqrt{X^2 + Y^2}} \quad (4.14)$$

$$\sigma_b = \frac{X^2 + Y^2 - l_3^2 + l_4^2}{2l_3l_4} - 1 \quad (4.15)$$

$$\theta_{hip} = \sin^{-1} \sigma_a - \tan^{-1} \frac{X}{Y} \quad (4.16)$$

$$\theta_{knee} = \theta_{hip} + \cos^{-1} \sigma_b \quad (4.17)$$

where $X = X_a - X_h$, and $Y = Y_a - Y_h$. If $\sigma_a > 1$ and $\sigma_b > 1$, assume them to be 1, then, the ankle trajectory is modified. Therefore, the pelvic motion is used to compensate the modification.

Figure 4.11 The key-point generation (a) the generated joint angles for hip, knee, and ankle; (b) lower limb motion as seen from an external observer moving with pelvic velocity in x direction.

For those out of range point, the pelvis motion trajectory is modified. The update of pelvic trajectory compensates those out of range points. Furthermore, line space interpolation are used to set the movement in horizon direction is coordinate to the velocity of the mobile to avoid foot dragging on the ground during the stance phase. The single and double stance can also be specified by dispersing motion trajectory. Therefore, the specified ten points are generated. The calculated joints angle and trajectory are shown in Figure 4.11.

Procedures of the ten point generation are listed as:

- 1 In order to generate suitable joint angle for overground walking, several procedures are listed.
- 2 The joint angles for the hip and knee can be obtained with known X_h, Y_h and X_a, Y_a . $X_h = A_x \sin(2t + \theta)$, $Y_h = A_y \cos(2t + \theta)$ at every specified point.
- 3 The ankle trajectory can be calculated by the direct kinematics. Compare the difference between the calculated trajectories to the pre-described one in y-direction. Then, subtract the difference at pelvic motion in y-direction. Then the ankle trajectory is satisfied pre-described one.
- 4 From the new hip trajectory, calculate the ankle trajectory. Assume ankle joint do not change, the metatarsal trajectory are obtained. Then, update the trajectory when metatarsal supports to be on the ground. Within the key points

$$y_m(3\sim7) = -l_3 - l_4$$

- 5 The ankle joint angle can be calculated from the ankle and metatarsal trajectory as

$$\alpha_{ankle} = \sin^{-1} \frac{y_a - y_m}{l_5}$$

- 6 Curve fitting are used based on the specified points, the planned trajectories are obtained. Use the line space interpolation during the stance phase. The coordinated v_m is calculated.
- 7 Then, another inverse kinematics are used based on the new trajectory of pelvic and ankle. Finally, smoothing algorithm is used for the motor input.

Figure 4.12 Kinematic model used in workspace analysis: relationship between 10 Dof rotations/2 Dof translations and stride length/height is shown in the Sagittal plane;

4.3 Adjustable Locomotion for Individuals

In order to generate trajectories, key points on the trajectory are designated at first. Curve fitting all those key point can obtain trajectory that meet overground walking requirement. Then joint angles can be calculated by inverse kinematics of the system. However, there are angles cannot be obtained based on the pre-defined trajectories because they are out of physical range of motion. For example, the following condition cannot be satisfied, the joint angle cannot be calculated.

$$\sqrt{(x_h - x_a)^2 + (y_h - y_a)^2} < l_3 + l_4 \quad (4.18)$$

where $(x_h, y_h) \in \text{workspace of hip}$ and $(x_a, y_a) \in \text{workspace of ankle}$.

The kinematic model of the robotic orthosis (RO) is shown in Figure 4.12, from the initial point O to the foot point at metatarsal joint M is

$$\begin{cases} x_k = x_h + l_1 \sin(\theta_{hip}) \\ y_k = y_h + l_1 \cos(\theta_{hip}) \end{cases} \quad (4.19)$$

$$\begin{cases} x_a = x_k + l_2 \sin(\theta_{knee} - \theta_{hip}) \\ y_a = y_k + l_2 \cos(\theta_{knee} - \theta_{hip}) \end{cases} \quad (4.20)$$

$$\begin{cases} x_m = x_a + l_3 \cos(\theta_{ankle} - \theta_{knee} + \theta_{hip}) \\ y_m = y_a + l_3 \sin(\theta_{ankle} - \theta_{knee} + \theta_{hip}) \end{cases} \quad (4.21)$$

where l_1 is the length of thigh, l_2 is the length of calf, and l_3 is the length from heel to metatarsal on the foot. The kinematic model can be simplified into vector expression as shown in Figure 4.13.

$$\mathbf{P}_{\text{ankle}} = \mathbf{P}_{\text{OH}} + \mathbf{P}_{\text{HK}} + \mathbf{P}_{\text{KA}} \quad (4.22)$$

$$\mathbf{P}_{\text{metatarsal}} = \mathbf{P}_{\text{OH}} + \mathbf{P}_{\text{HK}} + \mathbf{P}_{\text{KA}} + \mathbf{P}_{\text{AM}} \quad (4.23)$$

The condition for collision-less during the overground walking is set as

$$\mathbf{P}_{\text{ankle-y}} \approx l_1 + l_2, \text{ when } t_{\text{heel_contact}} < t < t_{\text{heel_off}} \quad (4.24)$$

$$\mathbf{P}_{\text{metatarsal-y}} \approx l_1 + l_2, \text{ when } t_{\text{foot_flat}} < t < t_{\text{toe_off}} \quad (4.25)$$

where $t_{\text{heel_contact}}$ is the time when heel contact occurs within a gait cycle; $t_{\text{heel_off}}$ is the time when heel off within a gait cycle; $t_{\text{foot_flat}}$ is the time when foot flat within a gait cycle; and $t_{\text{toe_off}}$ is the time when toe off within a gait cycle.

The stride length is then given by $L = |\mathbf{P}_{\text{metatarsal-x_right}} - \mathbf{P}_{\text{metatarsal-x_left}}|$, when $t = t_{\text{heel_contact}}$. The walking speed can be obtained by $v = L/t_c$, where t_c is the step time.

Figure 4.13 Kinematic model using vector expression.

The ten point generation data is used as the initial points for the gait motion. At this connection, $\mathbf{P}_{\text{ankle-y}}$ and $\mathbf{P}_{\text{metatarsal-y}}$ are calculated based on the forward kinematic for the whole set of joint angle data within a gait cycle and compare the result to equations. The points leading to the ground impact need to re-generate. One simple way is to enlarge the hip and knee joint angles, to shorten the vertical length at the moment. Another way is study the inverse kinematic and gait planning for the lower limb and pelvic.

Instead of using the inverse kinematic and dynamic analysis as mentioned in Chapter 2, an adjustable index γ is proposed to scale hip and knee joint angle given by

$$\begin{bmatrix} \theta_{hip} \\ \theta_{knee} \end{bmatrix}_{n+1} = (\gamma + \Delta\gamma) \begin{bmatrix} \theta_{hip} \\ \theta_{knee} \end{bmatrix}_n \quad (4.26)$$

in which $\Delta\gamma$ is the step to enlarge γ before θ_{hip} and θ_{knee} reaching their mechanical constraints. The optimization and curve fitting toolbox in Matlab (The MathWorks Inc., Natick, MA) are used to provide smooth trajectories. It compares the walking of the raw data and processed data of joint angle. The processed walking approaching a normal walking shows the effective of the proposed methods.

Figure 4.14 The joint angles in the Sagittal plane for hip, knee and ankle and comparison between the raw data get from the motion capture experiment in solid line and the processed data for impact-free in the stand phase in dash line: (a) hip joint; (b) knee joint; (c) metatarsal joint.

(a)

(b)

(c)

Figure 4.15 The trajectory in Sagittal plane of the knee, ankle and metatarsal joints based on the original data get from the motion capture experiment and the obtained data: (a) knee joint at K; (b) ankle joint at A; (c) metatarsal joint at M.

The trajectory for knee, ankle, and metatarsal joints in Sagittal plane are shown in Figure 4.14. In the knee joint trajectory, the original trajectory with raw angle data from motion capture experiment is compared to the modified knee trajectory. The trajectory after processing indicates less hip extension and more flexion. In the ankle joint trajectory, the original trajectory with raw angle data from motion capture experiment is compared to the modified ankle trajectory. The trajectory after processing indicates more knee flexion. In the metatarsal joint trajectory, the original trajectory with raw angle data from motion capture experiment is compared to the modified metatarsal joint. The trajectory after processing as shown in Figure 4.15, indicates more dorsiflexion and less plantarflexion.

In conclusion, in order to generate individual walking motion for different body size, stride length, foot clearance height etc. Key points are specified within one gait cycle to obtain the foot trajectory from the designated step length and height. The trajectory is obtained via curve fitting based on these specified points. On the other hand, in order to keep the desired foot trajectory, the pelvic motion is compensated to accommodate hip and knee joint angles. Next, the metatarsal trajectory is planned to satisfy the condition of a smooth OGW. Furthermore, in order to solve possible problems that may surface during overground walking include: 1. Ground impact problem. 2. Poor coordination to mobile platform (MP) which would cause foot dragging. A gait motion generation is developed and specified to adjust the gait patterns for an impact-less and smooth footpad contact based on the walking model proposed. To reduce, if not to avoid, the ground impact and dragging of the footpad occurring during the course of overground walking with help of the device, an adjusting index is proposed to guarantee the impact-less overground walking pattern based on optimizations.

Chapter 5 CONTROL FOR PERIODIC GAIT TRAINING AND EFFECTIVE REHABILITATION

The proposed key points overground motion generation and modeling of human-machine interface proposed in Chapter 3 will be implemented in control strategies presented in this Chapter. Based on the generation of walking motion introduced in Chapter 4, the reference commands are obtained for the rehabilitation robot to follow as shown in Figure 5.1. The tracking performance of the adaptive motion and repetitive control will be discussed in this Chapter. The performance of the motion plan in Chapter 4 and control in Chapter 5 will be provided in Chapter 6.

Figure 5.1 Concept of conversion of constructed reference trajectories to controller input.

This Chapter focuses on the “repetitive” motion input for the lower limb, the intelligent control algorithm for the periodic signals are used. The repetitive control algorithm based on the characteristics of a cyclical reference signal and the internal model control principle, implanted the periodic signal generator has developed into the closed-loop system in order to achieve steady tracking of cyclic reference signals and rejecting disturbance on the rehabilitation robot. Not only the tracking performance is improved, the assist-as-needed force can also be under control. This iterative learning algorithm for the repetitive gait task control will be explained in Section 5.1. Furthermore, in order to control the walking velocity on time for the OGW system, an adaptive control is proposed based on on-line model identification algorithm. The repetitive controller works as a stabilizer is proposed to enlarge the control zone. With the help of stabilizer functioned by the exoskeleton and other external force, the augmented stable joint will be presented. By

selecting a proper stabilizer, the function of robotic-assisted system can switch between assistance and resistance for different rehabilitation requirements. The robustness and tracking performance are investigated based on increasing gain of the stabilizer.

The autotuning modeling and control using PID controller with feed-forward feedback will be introduced in Section 5.2. Additionally, in this Chapter, all those methods are studied and tested on a simulation platform. They will be implemented on the sub-systems of a self-developed gait trainer in Chapter 6.

5.1 Repetitive Control

In application to gait system, the tracking systems have to deal with periodic reference signals. Repetitive control based on the internal model principle has been proved to be a useful control strategy for this class of systems. Other than the repetitive controller with time delay as mentioned in Chapter 2, the diagram of the repetitive control used in robotic system is shown in Figure 5.2. $M(q)$ is mass matrix of the robotic joint. $H(q, \dot{q})$ is summation of the matrix of coriolis torques and the gravity torques. A repetitive controller is designed.

Figure 5.2 Diagram of the repetitive control system used in the robotic system.

The block of repetitive control system is shown in Figure 5.3.

Figure 5.3 Diagram of the repetitive control system.

The open-loop transfer function is

$$\mathbf{D(s)} = \frac{\mathbf{Q(s)e^{-Ls}}}{\mathbf{1-Q(s)e^{-Ls}}} \quad (5.1)$$

The closed-loop transfer function is

$$\mathbf{T(s)} = \frac{\mathbf{G(s)}}{\mathbf{1-Q(s)e^{-Ls}+G(s)}} \quad (5.2)$$

The periodic joint angles can be represented by the Fourier series as

$$\mathbf{q(t)} = \sum_{k=1}^{\infty} \mathbf{a_k \cos(k\omega t)} + \mathbf{b_k \sin(k\omega t)}, \quad \mathbf{k = 1, 2 \dots} \quad (5.3)$$

Figure 5.4 Comparison between the original signals (solid line) and signals that cut off high harmonic Fourier (dotted line).

For the conventional controller such as PID controller commonly used in robotic control, while cut off high frequency disturbance they also lost high harmonic Fourier of period signal. Therefore, the cut off signal are slightly different from the original signal as shown in Figure 5.4. The tracking performance is influenced by this reason.

Compared to ordinary controller such as PID controller, the periodic enlarge high frequency domain is suitable for the periodic input of joint angle. More high harmonic series can pass through the controller. The frequency response of the repetitive controller is shown in Figure 5.5. When the internal model oscillation is the same to the input signal, then it enhance the capacity of periodic signal tracking and suppress the periodic disturbance.

Figure 5.5 Frequency response of the repetitive control system.

As for the repetitive controller, the fundamental frequency is

$$\omega_k = \frac{2\pi}{L} \quad (5.4)$$

The instability's condition covers $G(s)$ asymptotic stability and $T(j\omega) = 1$. That is, $|1 + G(j\omega)| > 1$ refers to Equation (5.2).

A robotic manipulator with a revolute joint is used for simulation study. The conventional PD controller and repetitive controller with simulation of a 3 Dof robotic system is shown in Figure 5.6. Three joints follow the cyclic commands from hip, knee and ankle joints. The machine for model conduct the overground walking pattern as generated as shown in Figure 5.7.

Figure 5.7 Walking machine for the simulation model.

Then, the proposed repetitive control is used in this simulation platform, the track performance for hip, knee and ankle joints are shown in Figure 5.8. The track performance improved with time with the help of the repetitive controller. The overall frequency response compared between the conventional PD controller and repetitive

controller is shown in Figure 5.9. The augment process gain is increased based on the repetitive controller which indicates less assist force is exerted from the robotic device according to the modeling of HMI in Chapter 3.

Figure 5.8 Tracking performance of hip, knee and ankle joint with repetitive controller:
(a) hip joint; (b) knee joint; (c) ankle joint.

Figure 5.9 Frequency response compared between PD controller and repetitive controller.

Then, the proposed repetitive control is used in this simulation platform, the tracking error is compared to the conventional PD controller as shown in Figure 5.10. The performance is compared in Figure 5.11.

Figure 5.10 Tracking error of hip, knee and ankle joints compared between PD controller and repetitive controller.

(a)

(b)

Figure 5.11 Tracking performance of hip, knee and ankle joints: (a) PD controller; (b) repetitive controller.

In conclusion, repetitive controls in the rehabilitation robot control suppress the tracking error and improve the active level of the test subject by increasing the augment process gain.

5.2 Stabilizer

A stabilizer $K(s)$ is introduced as used in the typical control structure with an inner stabilizing loop shown in Figure 5.12. Since the impaired human control act as an unstable system, the exoskeleton serve as a stabilizer to stabilize the unstable process with an inner loop and the resultant augmented process is stable. The augmented stable process can be written as

$$G_a(s) = \frac{G(s)}{1+G(s)K(s)} \quad (5.5)$$

where

$$K(s) = A(s)C(s) \quad (5.6)$$

Figure 5.12 Control structure with an inner stabilizing loop: (a) stabilizer; (b) external assistance.

The function of stabilizer is provided by the external assistive device like exoskeleton as a compensation for the underestimated or overestimated joint torque through the actuator $A(s)$ and the robotic controller $C(s)$. The assistive force will transfer from motorized

power to human via the human machine interface. Augmented stable processes instead the unstable process under the brain to control.

The closed-loop transfer function is

$$\begin{aligned}
 H_m(s) &= \frac{G_a(s)B(s)}{1+G_a(s)B(s)} \\
 &= \frac{G(s)B(s)}{1+G(s)K(s)+G(s)B(s)} \\
 &= \frac{G(s)B(s)}{1+G(s)A(s)C(s)+G(s)B(s)} \quad (5.7)
 \end{aligned}$$

Even for simplicity $K(s)$ is taken to be a proportional gain, then the $B(s)$ can be designed for the stabilized process with larger control zone and stable performance.

If there is 20% error in estimating the time delay, it is unstable and the closed-loop responses of process without stabilizer and the proposed method. The uncertainty profile is

$$\Delta(\omega) = \begin{cases} |e^{-j0.1\omega} - 1| & \omega < 10\pi \\ 2 & \omega \geq 10\pi \end{cases} \quad (5.8)$$

The closed-loop system is robust stability if and only if

$$\left\| \Delta(\omega) \frac{G(j\omega)C_a(j\omega)}{1+G(j\omega)C_a(j\omega)} \right\|_{\infty} < 1, \forall \omega \quad (5.9)$$

An unstable process is applied as:

$$G(s) = \frac{1}{s-1} e^{-0.5s} \quad (5.10)$$

The stabilizer is assumed to be a second-order process, its frequency response is shown in Figure 5.13. Then, stabilizing gain is $|G(s)| = 1.5927$, the augmented stable process is

$$G_a(s) = \frac{1.6872e^{-0.34s}}{0.3913s^2 + 0.493s + 1} \quad (5.11)$$

The step response of the unstable process and unstable process with the stabilizer are compared in Figure 5.13. With the help a stabilizer of gain, the unstable process turned into a stable process under control. Then, the control for the stable process is same as mentioned. The optimal controller can be design for a perfect controller. As shown the unstable process of the pathologic gait performs out of control, even with the optimal controller designed. The controller works fine on the stable process of the healthy gait model. The response time and overshoot are small. The bode diagram shown the stabilizer makes the unstable process to stable process when the gain of stabilizer satisfy the Equation (5.9).

Figure 5.13 Comparisons of unstable system and unstable system with stabilizer: (a) step response; (b) bode diagram.

5.3 Assist or Resist

Although the ideal of perfect control must be abandoned in the process control procedure with right-half plane (RHP) zeros, time delay, modeling error and constraints on the

manipulated variables, human motion control presents the perfect control performance and the existence of the internal model has been proofed by animal experiments and literature reference. The brain is capable of moving the extremities in an environment with many unpredictable perturbations [147]. These perturbations will disturb the limb and deviate from the desired position. So a feedback system of human motion control is intelligent enough to establish this internal model to stabilize the body robustly and adaptively. The closed-loop system is affected by the model-plant mismatch which can be exploited according to robustness criterion, support the norm bounded uncertainty is defined by

$$\left| \frac{G^*(s) - G(s)}{G(s)} \right| \leq \Delta_m(s) \quad (5.12)$$

Perturbations and mismatch of the internal model is assumed as $\Delta_m(s)$, then the sufficient and necessary condition that guarantees the robust stability of closed-loop system is

$$\|T(s)\Delta_m(s)\|_\infty < 1 \quad (5.13)$$

for the whole frequency domain, where $\Delta_m(s)$ is a random linear stable transfer function which represent the mismatch of the internal model in the human body dynamics. The magnitude of Δ is assumed to be bounded by $|\Delta_f|$, where Δ_f is fixed boundary of the mismatch. According to different modeling function, the mismatch occurs at gain uncertainty, time delay uncertainty, or time constant uncertainty. Take the gain uncertainty as an example

$$\Delta_m(s) = \frac{\Delta K}{K}, \forall \omega > 0 \quad (5.14)$$

where ΔK is mismatch of the gain; K is the system gain. For $|\Delta_m| = 1$, it is the perfect controller and the internal model provide precise torque command for exact motion. However, if the muscle cannot exert force following brain command, the healthy motor

control requires assists. For $|\Delta_m| > 1$, the internal model of human motion is enlarged the input torque command and the joint motion needs resist to normal. For $0 < |\Delta_m| < 1$, the internal model underestimated the input torque command and joint output motion needs assisted to normal. The resist and assist needs to be provided by the external force. By adjusting the gain of the stabilizer, the augmented process providing assist or resist force in the rehabilitation training program.

Take the model of Equation (5.14) for an example, with increasing k , the $|\Delta_m|$ increase with the augmented process gain as shown in Figure 5.14. Therefore, the human machine interaction (HMI) acts from assist force to resist force in the control range. When $|\Delta_m| = 1$, there is the zero force control. These criteria will be used for proposed control algorithm in Chapter 5.

Figure 5.14 Effect of increasing stabilizer gain k switching between assistance and resistance.

In conclusion, descriptions of subject, system and environment and the relationship between them are provided in this chapter first. Then, in order to improve the active

participation level, control of external robotic-assisted force is studied, either assist or resist force. An internal model is introduced for the human motion control, especially, for the pathologic gait, a stabilizer is proposed for a stable motion with help of the exoskeleton. The function of the stabilizer is investigated on the human model to re-establish the stable motion control for walking training. The resultant augmented process with the stabilizer with providing assist or resist force in the rehabilitation training program. Therefore, a new method is proposed for modeling of human-machine interface (HMI) and a stabilizer is designed accordingly. By selecting a proper stabilizer, the function of robotic-assisted system can switch between assistance and resistance rehabilitation. The performance and robustness of the overall external assistance are studied for future control algorithm used in robotic device.

5.4 Synchronized Control

In order to control the walking velocity on time for the OGW system, an adaptive control is proposed based on on-line model identification algorithm. An autotuning controller is developed to synchronized motions for the sub-system. The optimal control is commonly used in the nonlinear and high dynamic process for better tracking performance. Major functions are presented including internal modeling, reference trajectory, performance index and control variable. In conclusion, after modeling of controller and actuator, an autotuning control is proposed for better tracking performance. A new concept is proposed, called phase derivation, is introduced to compensate for the loss of those truncated high-order harmonics. The optimal controller is proposed based on the model.

5.4.1 Model identification

With help of the nonlinear component relay, the limit cycle is presented. In the describing function analysis, the negative reciprocal curve is represented by $-1/N(a)$. The sustained oscillation occurs at the critical point when $-1/N(a)$ of relay cross the Nyquist plot of

the dynamic system, see Figure 5.15. The cross point is located at the negative real axis of the Nyquist plot when it is the idea relay, detailed explanations refers to Appendix B.

Figure 5.15 The sustained oscillation occurs at the critical point when $-1/N(a)$ of relay cross the Nyquist plot of the dynamic system.

A general second order process with dead time (SOPDT) are employed as

$$G(s) = \frac{1}{as^2+bs+c} e^{-\theta s} \quad (5.15)$$

Parameters are identified by the analytical expression [148], k_n are calculated by Equation (5.15).

$$a = \frac{m_2}{8\omega_{osc}^2} \quad (5.16)$$

$$b = \frac{\sqrt{m^2+2pq}}{8\omega_{osc}} \quad (5.17)$$

$$c = \frac{m_1}{8} \quad (5.18)$$

where

$$p = \sqrt{\frac{150k_1^2 - 25k_3^2 + 3k_5^2}{2}} \quad (5.19)$$

$$p = \sqrt{\frac{2k_1^2 - 3k_3^2 + k_5^2}{6}} \quad (5.20)$$

$$m = \sqrt{\frac{-34k_1^2 + 39k_3^2 - 5k_5^2}{3}} \quad (5.21)$$

5.4.2 Optimal control

The PD controller is represented as

$$G_c(s) = k_p + k_d s \quad (5.22)$$

$$G(s)G_c(s) = \frac{k_p + k_d s}{as^2 + bs + c} e^{-\theta s} \quad (5.23)$$

If it is chosen that the gain margin is A_m and phase margin is φ_m

$$1/A_m = |G(j\omega_m)G_c(j\omega_m)| \quad (5.24)$$

$$\varphi_m = \arg(G(j\omega_m)G_c(j\omega_m)) \quad (5.25)$$

$$G(s)G_c(s) = \frac{k_p + k_d s}{as^2 + bs + c} e^{-\theta s} \quad (5.26)$$

When $c \approx 0$ in the motor control case

$$G(s)G_c(s) = \frac{e^{-\theta s}}{A_m s} \quad (5.27)$$

$$k_p = b/A_m \quad (5.28)$$

$$k_d = a/A_m \quad (5.29)$$

For the digital motion control, ZOH and DAC are added. The controller can be written in the discrete equivalent form.

Figure 5.16 PD controller application.

5.4.3 Simulation example

The transfer function of the plant is $G_p(s) = \frac{1}{2s+1} e^{-2s}$, and the relay feedback identification model transfer function is $\widetilde{G}_p(s) = \frac{1}{1.658s+1} e^{-2s}$. The parameter in other controllers list in the Table 5.1. Assume the holder to be zero holder, then the discrete plant is obtained. The PD controller is derived as shown in Figure 5.16. Meanwhile, the other typical PID controller using the same relay feedback control will be compared with the proposed method. These are Cohen-Coon tuning approach (i.e. CC), Ziegler-Nichols tuning method (i.e. ZN) and Revise Ziegler-Nichols tuning method (i.e. RZN). The simulation performance is illustrated in Figure 5.17.

Table 5.1 Simulation Results

CC	ZN	RZN ($\beta=0.7381$)	Relay-PD
$k_p=1.621$	$k_p=1.356$	$k_p=1.356$	$\frac{1 - 0.547z^{-1}}{0.453}$
$k_i=3.75$	$k_i=3.102$	$k_i=3.102$	
$k_d=0.6167$	$k_d=0.775$	$k_d=0.775$

Figure 5.17 Simulation results: (a) The step signal imposes on the input port; (b) step input and disturbance in $t=20$; (c) step input and +20% model fluctuation; (d) step input, disturbance and +20% fluctuation.

In conclusion, the author focuses on the “repetitive” motion input for the lower limb, the intelligent control algorithm for the periodic signals are used. The improved repetitive control algorithm based on the characteristics of a cyclical reference signal and the internal model control principle with the periodic signal generator has developed into the closed-loop system in order to achieve steady tracking of cyclic reference signals and rejecting disturbance on the rehabilitation robot. Additionally, the reference signals are generated by the proposed method in Chapter 4. Not only the tracking performance is improved, the assist-as-needed force can also be under control with the modeling of human-machine interaction (HMI) mentioned in Chapter 3. Furthermore, in order to control the walking velocity on time for the OGW system, an adaptive control is proposed based on on-line model identification algorithm. With the help of stabilizer functioned by

the exoskeleton and other external force, the augmented stable joint will be presented. By selecting a proper stabilizer, the function of robotic-assisted system can switch between assistance and resistance for different rehabilitation requirements. The robustness and tracking performance are investigated based on increasing gain of the stabilizer. An autotuning controller is developed to synchronize motions for the sub-system. The simulation examples have shown their effectiveness. The motion generation and control for the overground walking will be tested on our system, which will be introduced in Chapter 6.

Chapter 6 CONTROL IMPLEMENTATION ON A GAIT REHABILITATION SYSTEM: *NaTure-Gaits*

The robotic system is developed and named as *NaTure-gaits*, as a *Natural and TUnable rehabilitation gait system*, see Figure 6.1. As a major contributor for the development of *NaTure-gaits*, especially the motion generation and control of whole system, the overall objective of this research is to develop and initiate an overground training incorporated with pelvic control and active assistance to lower limb so as to improve walking capabilities of a subject in rehabilitation mobility progressing stages.

Figure 6.1 Overview of NaTure-gaits: Pelvic Arm (PA) to provide pelvis motion and body weight support (BWS); Robotic Orthosis (RO) for active assistance to hip/knee/ankle joints in the Sagittal plane; and Mobile Platform (MP) to allow the locomotion training.

It is important to highlight some conceptual specifications of the gait system developed. First of all, the research is aimed to design a device that can be applied to rehabilitation training progress of therapy for reducing manpower and increasing efficiency. The gait

system is not to replace the conventional manual training with therapists, but serves as an assistive platform to reduce the physical and work burden of a therapist and thus enhance the quality of the rehabilitation training extended to the patient. Secondly, although the robotic device provides a gait training platform for a wide range of patient groups, the design consideration in its first version caters for those patients, who have lost their trunk control. Finally, the proposed robotic system can be served as a useful platform for clinical-engineering research, progress comparisons, and quantitative outcome evaluation of rehabilitation progress.

Figure 6.2 Conceptual idea of the gait rehabilitation methodology and its realization (pelvic motion and body weight shifting).*

In this chapter, several clinical-relevant factors serve as the fundamental guideline for the system development, control, and implementation as shown in Figure 6.2. The main features of developed overground gait system are described. The training scheme with modular design is highlighted. The mechanical structure of three sub-systems and the architecture of control framework are described respectively. The application of control theories mentioned before are tested on the system. The clinical trials on healthy and SCI/stroke subjects have conducted successfully. The performance and evaluation of the system will be presented in next chapter.

6.1 Clinical-relevant Issues Considered for System Development

The development of *NaTure-gaits* started since the beginning of candidature. The first prototype of *NaTure-gaits* was delivered in year June, 2008. On the same year it is presented to doctor and therapist from Tan Tock Seng hospital. One of the therapists has tested the system personally. Remarks and opinions were collected after the test, and the prototype is further improved and second prototype of *NaTure-gaits* is introduced in September, 2009. Health subjects have been recruited for testing the system. Walking motion and control algorithms have been updated. After rounds of improvement to obtain the approval by therapists and ethic committee, clinical trials are conducted on Tan Tock Seng hospital in May, 2011. The final version has be conducted the successful clinical trials on the SCI/stroke subjects in hospital with improvement on attachment, harness, arm rest *etc.* The roadmap of the development of *NaTure-gaits* is presented in Figure 6.3.

Figure 6.3 Development roadmap of *NaTure-gaits*.

Patients with spinal cord injury (SCI) walking with our developed system for the overground environment. The hip/pelvic, knee and ankle trajectories are given in the figure for illustration. A subject is a 66-year old Asian male, C5 spinal cord injury, with resulting in loss of function below shoulder. Progressive physical exercises have improved his sense of control. However, routines to strengthen his muscles are costly, with training involving at least three assistants. He required assistance for his lower limbs, pelvis, trunk and upper body in order to be able to transfer, support, balance, and ambulate per session. Through the rehabilitation, he hopes to improve ambulation abilities as well as recovery of motor function. Safety precaution certification and approval from the local ethics committee were obtained and documented before the trial. It is worth mentioning that the subject was excited throughout the walking testing with *NaTure-gaits*. He felt physiologically to be able to walk again, without the assistance by any caregiver.

Table 6.1 Differences Outlined for NaTure-gaits and WalkTrainer

	<i>NaTure-gaits</i>	WalkTrainer
<i>Pelvic Motion Assistance</i>	By a pair of robotic arms	Parallel structure
<i>Body Weight Support</i>	Active BWS, provide by robotic arms	Cable BWS, provided by overhang system
<i>Lower Limb Motion Assistance</i>	By robotic orthosis	FES and robotic orthosis

Gait rehabilitation helps the patients to regain or enhance muscle strength, balance, coordination, endurance, locomotion, correct walking posture and functional mobility activities of daily living (ADL). Furthermore, the therapeutic approaches cover compensation and recovery models. Traditional therapy after, for example, incomplete SCI includes transfers, W/C mobility, ambulation with assistive device or braces training. Model therapy is driven more by principles of recovery such as an intensive and task-specific walking training. The research emphasizes functional walking in the aspects of overground ambulation and pelvic movement. According to the publications, most

research does not combine pelvic motion control, active hip, knee, ankle assistance and overground gait guiding mechanism within one gait rehabilitation platform. Other than our system, only one project called WalkTainer has these features. Hereby, the comparisons between *NaTure-gaits* and WalkTrainer are provided in Table 6.1. The research of *NaTure-gaits* fills a gap in rehabilitation field.

6.2 System Descriptions

The structure of *NaTure-gaits* is described at first, both the manual adjustable design and motorized joint motion are introduced. The control architecture of the hardware setup is further presented as well as the software programming in the following sections.

Figure 6.4 Main modules of *NaTure-gaits* and its corresponding 14 joint motions.

Figure 6.4 depicts an overall design of *NaTure-gaits*. A pair of pelvic arms (PA) in parallel is to generate the pelvic motion for the left and right sides. The end of the PA is connected to the harness on the subject for body weight support (BWS), body weight lateral shifting, and pelvic control (PC). A pair of robotic orthosis (RO) is connected to the PA to provide gait and pelvic motion for both sides in the Sagittal plane. The RO provides the assistive locomotion for lower limbs. Furthermore, a passive cable-BWS is added for the safety consideration. The PAs and ROs are mounted on a mobile platform (MP), providing an overground ambulation. The controller for gait control, wireless communication, and batteries are also included to implement the natural and tunable gait rehabilitation.

To accommodate a broad range of subject groups, *NaTure-gaits* has incorporated manually adjustable structure to cover most body sizes. All these adjustments are conducted during the setup procedures before walking on the system. Therefore, most of the modules of *NaTure-gaits* can modulate to different body dimensions.

6.2.1 Modulation of harness and strapping

In order to support the subject's trunk body, a customized body harness made of nylon is designed. Since dimensions are specific to the subject, it fits subjects comfortable and ensures that external support properly transferred. The body weight is supported by the robotic arm contacting around the pelvic area. A bracket at the end point of the Pelvic Arm (PA) is connected to harness to increase rigidity of control. Straps can be strategically placed to secure the harness to the subject more efficiently for weight bearing, see Figure 6.5. Consider the different body sizes, two set of customized harnesses are prepared. The overhanging straps prevent falling down in case of emergency.

Figure 6.5 Function of the harness and bracket connection.

In addition of the harness transfer the external force to subjects, brace straps and footpads are designed properly. Velcro strapping and padding are used for the adjustment and comfort, see Figure 6.6.

Figure 6.6 Strapping and padding connections.

6.2.2 Modulation of Pelvic Arm (PA)

The mounting base is designed with modulate height and weight catering different body sizes of subjects. The estimated payload is $260kg$ inclusive weight of MP ($25kg$), onboard batteries ($25kg$), additional payload such as Pelvic Arm ($40kg$), Robotic Othosis ($40kg$), subject's weight ($100kg$) and many the peripheral. The mounting base for the Pelvic Arm (PA) through hardware connection is located on top of Mobile Platform (MP) and will have operating velocity at maximum of 0.9 m/s . The stable and slow adjustment of the mounting base is designed for manual operation with screw.

6.2.3 Modulation of other components

The highly adjustable strapping is connected to the hard support of thigh and shank. All contact areas are sponged and three fingers space clearance from joint/bony points, as shown in Figure 6.7. Furthermore, the length of arm rest is adjustable for better ergonomic dimensions. The back support is also adjustable for different subjects. A fall prevention overhanging structure is also adjustable, as shown in Figure 6.8.

Figure 6.7 Thigh and shank strapping.

Figure 6.8 Adjustment of arm rest, back support and fall prevention.

6.2.4 Motorized joints on PA, RO and MP

Other than the manual adjustable movement of system, motorized joints for powered assistance are developed on *NaTure-gaits*. Six DC brushless motors are in charge of Pelvic Arm (PA)'s movement in three directions. Another six DC brushless motors is for Robotic Othosis (RO) and another two motors is for Mobile Platform (MP). The mechanical structure and actual prototype of PA on one side and the other side is laterally symmetric. Pelvic motion in the Sigittal plane and lateral shifting are shown in Figure 6.9.

Figure 6.9 Actual prototype of Pelvic Arm (PA) with motors.

The lower limb robotic orthosis with a total of six active DoFs is developed as one of the modules of *NaTUre-gaits*. It provides active assistance to rotation of hip, knee, and ankle of both lower limbs at the Sagittal plane. Every joint is powered by a DC brushless motor sized to the appropriate power to actuate human joint with the robotic orthosis as shown in Figure 6.10.

Figure 6.10 Actual prototype of Robotic Orthosis (RO) with motors.

The mobile platform is developed to have a large foot print ($1200\text{ mm} \times 1000\text{ mm}$) for two reasons. The large foot print increases the stability of the entire system. The large foot print also allows a more spacious space to be provided for the patient to walk within the support polygon of the system. The mobile platform is actuated by two 300 watts DC brushed motors coupled with a 50: 1 gear head as shown in Figure 6.11.

Figure 6.11 Actual build of the mobile platform.

6.3 Control Architecture and Implementation

The power source/battery pack for the *NaTure-gaits* is placed at the rear of the mobile platform to increase the stability of the entire system as shown in Figure 6.12. Control system for sub-system PA, RO and MP are built up separately and communicate to each other via wireless connections to host computer. The power supply modules are also put on the mobile system.

The control framework is shown in Figure 6.13. The architecture of control system is comprised of three sub-control systems, such as pelvic arm (PA), robotic orthosis (RO), and mobile platform (MP), respectively, depicted in Figure 6.14. The host computer, which is the central process unit to connect the three sub-systems, receives high level commands by wireless communication. Subsequently, it interprets commands and sends individual commands to the sub-controller via TCP/IP communication. By this, three sub-systems act as slave stations while the host controller runs as a master station. These

independent motion control modules are integrated by sets of PIC and Galil controllers, developed specifically according to the system requirements. GUI software is also developed for the initializing and operating of the integrated gait system. In addition, the three modules can be synchronized to provide a “natural” gait of a subject.

Figure 6.12 Overall control system for NaTure-gaits II.

Figure 6.13 Architecture of control system for NaTure-gaits.

Figure 6.14 Control framework of NaTure-gaits.

6.3.1 Control of Pelvic Arms (PA) for pelvic control and body weight support

The PA provides two functions, body weight support (BWS) and pelvic control. It provides the suitable support to subject's trunk during gait rehabilitation. The provision of BWS using PA effectively eliminates the constraints caused by overhanging BWS system. The BWS support is realized by two independent sets of pelvic arm as shown in Figure 6.15. Apart from BWS, PA are designed to provide pelvic control as well. The waist can be modeled as a body with six degrees of freedom (DoFs), translational and rotational movements in three axes. It is found that during gait locomotion, the pelvic rotates at transverse axis, tilts at frontal plane, and shifts laterally. A pair of PA provides unrestricted pelvis motion using three translational movements to provide workspace for rotational movement of pelvis during walking.

$$x_h = L + \frac{Ll_2}{\sqrt{l_1^2 + l_2^2}} \quad (6.1)$$

$$y_h = \frac{l_1 l_2}{\sqrt{l_1^2 + l_2^2}} \quad (6.2)$$

where $L = 20cm$. The range of motion of the actuator is $-5cm < l_1 < 5cm$, $5cm < l_2 < 10cm$, and $-4cm < s_3 < 4cm$. Where $\theta_x = 0$; $\theta_y = 0$; $\theta_z = 0$, with the end effectors set as reference coordinates $x = 0$, $y = 0$, $z = 0$ at end effector.

Figure 6.15 Sub-mechanism and the kinematic model of PA, with the end effectors set as reference coordinates ($x = 0$, $y = 0$, $z = 0$ at end effector).

The trajectory of pelvis is provided based on the motion capture experiment as mentioned. The location of a marker is located on the pelvic area. The 3D motion of this point is also demonstrated. Within one gait cycle, the trajectory in the Saggital plane and later direction are investigated. Those trajectories can be converted to the motor input for three motors for movement of PA.

Other than motion capture experiment, the motor input reference trajectory can be calculated as

$$f(x, y, z) \quad (6.3)$$

where $x = A_x \sin(2t + \theta_x)$, $y = A_y \cos(2t + \theta_y)$, and $z = A_z \sin(t + \theta_z)$ [96, 149].

In order to follow those desired trajectories, control system is presented for those motors. The movement of Pelvic Arms (PAs) in the Saggital plane are driven by four brushless servo motors. The command signals are sent to the servo motors from four microcontroller chips, dsPIC33FJ128GP202, via four servo amplifiers, HUM-130508A. To achieve synchronized pelvic movement, another microcontroller chip is employed to send out commands to the four microcontroller chips at a specific timing. A brief diagram of the interfacing of the components is shown in Figure 6.16.

Figure 6.16 Overall control system architecture of pelvic arms.

Microcontroller chip is used as an overall master control of the PA system. The communication method used between the four microcontrollers is Universal Asynchronous Receiver/Transmitter (UART). After receiving commands from the master controller, the four slave controllers execute the commands according to the requirements. Similar to the communication method between master and slaves, the slave controllers use

UART to converse with the servo amplifiers. Printed Circuit Boards (PCB) is designed by the team. The same PCB design is used by the four slave controller and the master controller, and each is powered by four AA batteries.

The movement of PAs in the lateral direction are driven by another two brushless servo motors. The control system is combined to the Robotic Orthosis (RO) and will be introduced in next section.

6.3.2 Control Robotic Orthosis (RO) for assisting the lower limb

The key considering issue of the lower limb robotic orthosis is to develop a mechanism as shown in Figure 6.17, which can better reflect the real movement of human's lower limb during walking. However, to include all the DoFs of a human's lower limb for a mechanism is almost impossible and not cost effective. Furthermore, it has been pointed out by most of the therapists that the movement in Sagittal plane control during initial gait rehabilitation is sufficient for the patient having too much DoFs for the subject to take care of would create a burden to the walking subject [150]. This is the reason the therapists 'lock' certain joint movement of the patient during gait rehabilitation without over-constraining the natural walking dynamic.

Figure 6.17 Sub-mechanism and kinematic model of RO, with the end effectors set as reference coordinates ($x = 0$, $y = 0$ at end effector).

In the present work, a lower limb robotic orthosis provides active assistance to rotation of hip, knee, and ankle of both lower limbs at the Sagittal plane. Every joint is powered by a DC brushless motor sized to the appropriate power to actuate human joint with the robotic orthosis. Therefore, the robotic orthosis mechanism consists of two sets of linkage resembled human lower limb, which provide assistance for walking during gait rehabilitation. At each of the joint, a Maxon brushless motor is meshed with spur gears to drive the movement of the linkage. The controlled joints on each limb of the robotic orthosis are (a) hip joint control, (b) knee joint control, and (c) ankle joint control.

Figure 6.18 Brief illustration of components interfacing.

In order for the six motors, i.e. all the joints on both sides of the linkage, to achieve synchronized orthosis movement, a Galil motion control card, DMC-2183

Ethernet/RS232 Econo Motion controller, is employed to control the six motors. As the voltage and power requirement of the Maxon motors are higher than that of the control card, two units of Galil opto-isolator modules, ICM-20105, and six brushless PWM servo amplifiers, ZBDC12A8B, are used to interface between the control card and the motors. A brief diagram of the interfacing of the components is shown in Figure 6.18.

Figure 6.19 Overall control system architecture of robotic orthosis (RO).

The DMC-2183 is an 8-axis programmable motion control card. Other than those six motors, two motors for the lateral shifting is also under control of DMC-2183. It can interfaces with desktop via Ethernet cable or RS232 cable. Two units of Opto-Isolator modules are mounted onto the control card. The overall control system architecture of robotic orthosis is depicted in Figure 6.19. To program the DMC-2183, Window Servo Design Kit (WSDK), a program provided by Galil motion control, allows users to use basic assembly language to control the motion profile of the motors. Users can program the motors to run simultaneously using different modes, such as Electronic CAM mode, contour mode, and absolute positioning mode. At current stage, the project team makes use of the WSDK to program the gait pattern for the robotic orthosis.

Galil motion control card is deployed to control the eight motors (8 axes). Two units of Galil opto-isolator modules (each module provides connection for up to four amplifiers) and six brushless PWM servo amplifiers are used to interface between the control card and the motors. The control box of 8 axes motion control is shown in Figure 6.20.

Figure 6.20 Cable connection design in Robotic Orthosis (RO) system.

6.3.3 Control of Mobile Platform (MP)

The idea of allowing the patient to have a gait rehabilitation that is not confined to the treadmill inspired the development of a mobile platform (Figure 6.21) to provide body weight support and overground walking. The mobile platform is designed to withstand the body weight of the patient without structural failure and risk of overturn during operation. The design considerations for the main frame of the mobile platform are the safety issue of the mechanical structure and the patient supporting method. The design of a large base provides a relatively large support polygon at the base. The mobile platform is driven by two individually controlled motors, which provided omni-direction driving capability. In

In addition to the mobility capability, the mobile platform is designed to house parallelogram arms, robotic orthosis, controllers, and other sub-components. It will minimize the weight of these items from acting on the patient during gait rehabilitation.

Figure 6.21 CAD model of mobile platform.

6.4 Control Algorithm

After setting up the hardware of control and power supply for the system, control algorithm are tested for better performance. The initial tests present unstable performance of mobile platform because of the large load and dynamic disturbance. The vibration of the whole system affect the walking process on the system. Thus, the PID filters needs to tuning and the autotuning control method as mentioned is applied on control of MP. Furthermore, the tracking performance of RO and PA of cyclic reference is studied. Repetitive controller is used for improvement.

6.4.1 Repetitive control of Robotic Othosis (RO) with periodic input

In order to enable the periodic synchronization of eight motors, all they axes can be slaved to one master axis. The master axis is fictitious axis which allows a table-based relationship to synchronize all the controller axes. In the system, the DMC-2182

controller has a fictitious axis as one master and eight slaves for all the motors on RO and body weight shifting. The synchronized six axes motion for the lower limbs is generated by the proposed ten point optimal motion plan and two axes motion for the body weight shifting is generated by the motion plan for pelvic in the lateral direction. The procedures for the synchronized motion are listed:

Step 1: Define a fictitious axis as the master axis.

Step 2: Specify the master cycle and the change in the slave axes.

Step 3: Specify the master interval and starting point.

Step 4: Specify the slave positions.

Step 5: Enable the ECAM synchronized motions.

Step 6: Engage the slave motion.

Step 7: Disengage the slave motion.

The performance of RO is present in Figure 6.22 and Figure 6.23. The position error and motor torque are compared.

The same problem is faced in *NaTure-gaits* in the usability design to minimize the set-up time. Effort is exerted in developing quick adjusting mechanisms in attachments and harness to allow the patients to quickly put on the harness and attachments. Even during the day of the clinical trial, together with the therapist a fast way is discovered to get the patients from the wheel chair to stand onto *NaTure-gaits*. Continuously efforts are being put on to make the machine easy to use. If the patients can be quickly set up on the machine, time can be optimized to let the patient to do more training.

Figure 6.22 Position error of hip, knee and ankle joints.

Figure 6.23 Torque of hip, knee and ankle joints.

The future trend of the rehabilitation robots is determined by the requirement of clinical. *Nature-gaits* is a good example to illustrate this. Compared to the treadmill-based rehabilitation robots, the wheeled mobile platform in *Nature-gaits* was designed to allow the patient to have a real overground walking feeling. The new trends of rehabilitation robots now focus on the home-based rehabilitation. According to therapist point of view, the neurological injury episode should not be just confined within the several months after injury but a long time disease need to get treatment and exercise. The transportation cost and the limited facilities in the hospital make it impossible for the patients to get trained every day. The mobility-based rehabilitation robots come out to meet the demand of the therapist.

6.4.2 Autotuning control of Mobile Platform (MP) with large load

The encoder provides the performance feedback of the MP. The comparisons of actual encoder reading and reference trajectory before and after tuning are shown in Figure 6.24. Then, the control method is tested on our system. The relay test is shown in Figure 6.25. The encoder reading are recorded down for comparison of different control method, see Figure 6.26. First of all, A PD controller is applied. The P parameter keeps increasing until the system vibrated. Consider the obvious fluctuation of the output, D parameter is set at a small value. Compare the PD controller with the default controller, the more stable control output are shown. The MP can run much smoothly than before. Then, consider the heavy load on the MP, a forward feedback is added on the PD controller. The comparison of control output with/without forward feedback is shown. Less vibrating of the control output is shown. Furthermore, the setting error of the system is tuning to smaller value. The MP moves much smoother at the end. The encoder error is less than 0.1% of the reference input. The proposed control method is tested in a simulation model which has been built up in Matlab (control toolbox), PD parameters are obtained as $KP: 30, KD: 44$.

Figure 6.24 Comparisons between different control methods testing on NaTure-gaits: (a) PD control with default parameter (b) PD control with proposed parameters.

An autotuning method will be developed in the future to allow online adjustable walking patterns with different MP velocity updates in the NaTure_gaits system. Different MP velocities can be

obtained different PD parameter online, so the performance of MP with heavy load and dynamic disturbance can operate to satisfy a stable and robust specification.

Figure 6.25 Position error comparisons between different control methods testing on NaTure-gaits (a) PD control with different k_p , (b) forward feedback + PD controller with further improvement of small error toleration (c) PD control with default, trial and error, and proposed parameter.

Figure 6.26 Relay test with application on the gait robotic trainer: NaTure-gaits.

In conclusion, the robotic mechanism together with the modulation of the structure of *NaTure-gaits* are provided. The modeling of human-machine interaction (HMI), individual OGW pattern generation and intelligent control algorithm have implemented on *NaTure-gaits* successfully. The repetitive control and synchronized control is used for the modular system in terms of pelvic arm (PA) sub-system, robotic orthosis (RO) sub-system and mobile platform (MP). Therefore, it provides lower limb motion assistance in the Sagittal plane, natural pelvic movement assistance, body weight support with no restriction on the pelvic movement, and over ground walking experience. A unique robotic mechanism is designed for the provision of pelvic movement, body weight shifting and body weight support together.

Chapter 7 MUSCULAR ACTIVATION EVALUATION OF OVERGROUND WALKING (OGW)

Motion generation and control algorithm have been proposed and implemented in *NaTure-gaits*. Then, subject trials are carried out on healthy and SCI/stroke subjects to verify the basic function and safety of the developed robotic system. However, currently clinical assessments of gait motor function are routinely based on qualitative assessments obtained from therapists' observations. It is subjective and depends on the skills and experience of the therapist. As such there is a clear need for an objective and reliable assessment of gait function to monitor and quantify patients' progress during therapy and to validate the outcome of the robotic-assisted rehabilitation treatment. Therefore, quantitative assessment of walking is needed in order to evaluate the gait system. Furthermore, as for robots hold advantages over traditional therapy, evaluation of the performance of the robot is important to guarantee long time training, in a consistent and precise manner, whilst accounting and adjusting for training effect.

The performance of the overground gait training with the gait system is evaluated by comparing the EMG signals from the muscles on the lower limbs of subject. With several disturbance factors such as walking speed, step length, and kinematics data influencing the evaluation, it poses a restriction on the validity of evaluation results by virtue of the average pattern analysis commonly used in the treadmill walking.

Assessment and evaluation of walking quality are investigated in this Chapter. Effects on lower limb muscular activation during walking of different speeds and body-weight off-loading levels are investigated. Secondly, the successful clinical trials are introduced on SCI/stroke subjects walking on *NaTure-gaits*. Their muscle activation was monitored for demonstration of walking effectiveness. Machine-related effect is highlighted in

comparison of the average muscle pattern. Thirdly, in view of various walking patterns put a limitation on the quantitative evaluation result. A systematic methodology is proposed on walking quality. Comparison between regularity, repetition, symmetry and smoothness are carried out between the healthy and pathologic gait. Symptoms of spinal cord injury subjects recruited from clinical trials are highlighted in the frequency domain analysis.

7.1 Effects of Walking Speeds and Body-weight Off-loading Levels

With the evolution of robotic systems to facilitate overground walking rehabilitation, it is important to identify the effect of walking speed and robotic-aided body-weight supported loading on lower limb muscle activity. Modulations of walking speed and body-weight supported are commonly used on robotic devices to optimize neuromotor recovery. To optimize the setting of rehabilitation devices, current study collects and studies electromyography (EMG) data from key muscles in the lower extremity from seven healthy young volunteers walking over a wide range of speeds and a range of body-weight off-loading conditions as provided by a bespoke gait robot.

7.1.1 Participation and protocol of trials

Seven healthy subjects including five males and two females, aged between 24 to 29 years old, mean 24.5 ± 1.99 (S.D.), weight ranging from 48 to 72kg, mean 62.3 ± 7.74 (S.D.), and height from 157 to 179cm, mean 171.7 ± 7.04 (S.D.) were recruited from the local university population. None of the subjects presented with any neurological injury or orthopaedic disorder at the time of testing. Informed written consent was obtained and subjects were informed about the study procedure, but not about the purpose of the investigations so as to avoid any influence on the parameters recorded.

Subjects were asked to walk barefoot with a normal arm swing across a GaitRite [151] platform. This is a portable floor-mat walkway embedded with pressure-activated sensors

that provides measure of spatial and temporal gait characteristics [120, 152]. During the recording, walking stride length and other gait parameters were self-determined based on their interpretation of the different speed categories.

In the loading-related test, the subject walked barefoot with the body weight support locomotion training (BWSLT) system, as shown in Figure 7.1. The subject wore a harness attached to the lever of the BWS system. By increasing the height of the lever, the weight offloaded from the subject was increased. The rate of off-loading was obtained in the initial stage of stance before walking, e.g. 0% (full weight bearing, FWB), 5%, 10%, 20%, 40%, and 70% weight offload were normalized by individual body weight.

Figure 7.1 Experimental set-up: (a) Prototype of the bespoke body weight support locomotion training (BWSLT) gait robot; (b) Photo of the BWSLT gait robot and setup with weighing scale; (c) Experiment setup on healthy subject for walking with different walking speeds; (d) Experiment setup on healthy subject for walking with different body weight off-loading levels.

7.1.2 EMG data acquisition and procedures

Electromyography (EMG) was used to measure the muscle activity of the subject. The EMG data were collected at 1000 Hz with Telemyo 2400T G2 (Noraxon, USA Inc.). Noraxon self-adhesive silver-silver-chloride dual snap electrodes for surface were

attached to muscles of the lower limbs. The EMG signals were then obtained by using the TeleMyoTM 2400T G2 Transmitter, which sent the signal via Bluetooth transmission to the host PC. Therefore, electrodes placed on the muscle surface are able to remotely detect EMG signals of different muscles. Raw EMG raw data were sent to the host computer by the TeleMyoTM 2400 G2 Telemetry System for subsequent offline processing, this was done in conjunction with the video recordings of the walking.

Then, signals were full-wave rectified. This processing will retain the signal's energy, then Root Means Square (RMS) was used for smoothing and the size of the window was 50 milliseconds. A 6Hz low-pass Infinite Impulse Response (IIR) filter was used with Butterworth approximation [54]. The absolute magnitude of EMG is less consistent for individuals and research therefore focuses on the shape of EMG profile, times of peak or onset/cessation of myoelectric activity [153]. Detailed descriptions are provided in Appendix C.

Each walking trial was video recorded and the corresponding EMG signals were saved simultaneously. The overground walking speed was measured by Gaitrite, while the off-loading was determined by the BWS walking device through weighing scales. The percentage of off-loading was obtained by these weighing scales. As the absolute magnitude of raw EMG signal are not consistent between individuals, some researchers focus on the shape of EMG profile, times of peak or onset/cessation of myoelectric activity [153]. In this study, current work choose to record the overall recruitment EMG profile, i.e. envelopes of EMG. Time markers were placed at each right heel strike as determined from the synchronized video records. From these records, different stages of the gait cycle were identified using established marker criteria [36]. Envelops of EMG raw data were obtained for each muscle group in every subject for all gait cycles recorded (time normalized from 0 to 100% completion of a gait cycle). As opposed to using MVC (maximum voluntary contraction) to normalize the EMG activity, it is opted to use the

maximum value of processed EMG signal data at the normal walking speed instead MVC to normalize the EMG data.

Walking speeds were normalized to individual body length by the non-dimensional walking speeds as Liu, Anderson et al [128]

$$v_{ni}^* = v_n / \sqrt{gL_{leg-i}} \quad (7.1)$$

where v_{ni}^* is the normalized speed of subject i for walking speeds. v_n is actual measured walking speed for walking speed catalog n , g is acceleration of gravity ($= 9.8 \text{ ms}^{-2}$), and L_{leg-i} is leg height of subject i .

Table 7.1 Normalized Walking Speeds of Subjects

Subject (Gender)	Leg length (m)	Speed (m/s)/Normalized speed ^a					
		1 ^b very slow (<0.08)	2 slow (0.08- 0.20)	3 free (0.20- 0.30)	4 fast (0.30- 0.50)	5 very fast (0.50- 0.70)	6 fastest (>0.70)
Sbj 1 (Male)	0.86	0.13/0.05	0.22/0.08	0.74/0.25	0.96/0.33	1.53/0.52	2.10/0.72
Sbj 2 (Male)	0.84	0.18/0.06	0.32/1.11	0.57/0.20	1.13/0.39	1.43/0.50	NA ^c
Sbj 3 (Male)	0.79	0.14/0.05	0.23/0.08	0.56/0.20	0.84/0.30	1.41/0.51	2.45/0.88
Sbj 4 (Male)	0.80	0.19/0.07	0.23/0.08	0.57/0.20	0.99/0.35	1.56/0.56	2.49/0.89
Sbj 5 (Male)	0.90	0.15/0.05	0.25/0.08	0.55/0.21	0.95/0.32	1.64/0.55	2.43/0.82
Sbj 6 (Female)	0.71	0.10/0.04	0.31/1.12	0.61/0.23	1.02/0.39	1.76/0.67	2.01/0.76
Sbj 7 (Female)	0.85	0.18/0.06	0.33/1.11	0.62/0.22	1.06/0.37	1.78/0.62	2.23/0.77
Mean	0.82	0.05	0.52	0.22	0.35	0.56	0.81
Standard deviation	0.06	0.01	0.55	0.01	0.04	0.06	0.01

^a Normalized speed is the actual speed normalized by $v_{ni}^* = v_n / \sqrt{gL_{leg-i}}$; ^b Speeds from very slow, slow, free, fast, very fast and fastest; ^c not available in recording

The normalized walking speeds for the seven subjects are summarized in Table 7.1. Normalized speeds are cataloged into six groups. Catalog 1 to Catalog 6 represent the walking at very slow, slow, free (self-preferred, median or normal), fast, very fast and

fastest possible speed, respectively. After processing and normalization of EMG signals, the activation pattern of the lower limb muscles reveals almost symmetry to the contralateral leg with a 50% phase lag. For clarity, only one side is presented in this study.

The time of stance phase for different walking speeds are illustrated in Figure 7.2. The stance time decreases when walking speed increases. As shown, the stance phase decreased with increasing walking speed. At the same time, the change of stance phase has less relationship to the change of off-loading level. The muscle activation duration is related to the change of stance time so this is depicted by the shaded region in Figure 7.3 and Figure 7.4.

Figure 7.2 Comparison of the stance time in different walking speeds: (a) normalized walking speed; (b) normalized off-loading level.

7.1.3 Speed-related changes during overground walking

In the speed experiment, the magnitudes of muscles increase with speed increase while the activation durations decrease in the general observation. As mentioned in Chapter 3, different muscle patterns are shown under different functions: Group 1, Group 2 and Group 3. Muscle groups to achieve the same function in the gait cycle shared similar patterns of muscle activation.

Group 1: VO and GM were active during the initial phase of the loading response stage after heel strike in the gait cycle, see Figure 7.3 (c) and (d). The magnitude of the EMG profile increases with walking speed for propulsion function. Meanwhile, activation duration decreased and then increased.

Group 2: activation bursts of SO and MG were located during the middle phase of supporting leg in the gait cycle (see Figure 7.3 (a) and (b)). Both of them have a “hill” shape in different walking speeds. It is noted that the magnitude of the profile indicates that the muscle force increased at faster walking speeds. The activation duration decreases from slow to median walking speed and then increases from median to fast. The very fast speeds intrigue large activation magnitude and duration; meanwhile a large variance is demonstrated from cycle to cycle.

Group 3: TA, RF, BF and ST were active during more than one phase of the GC (see Figure 7.3 (e) - (h)). It was noted from the figure that bursts of activity decreased and even disappeared from the slow speed to fast walking speed during middle phase of the gait cycle. This could help to provide greater power to extend the limb during this longer stance phase. Other bursts of this group, the magnitude increase and the duration decrease then increase with increasing speed.

Figure 7.3 Muscle activation pattern changes with normalized walking speed (a) SO in propulsion-related muscle group; (b) MG in propulsion-related muscle group; (c) VO in support-related muscle group; (d) GM in support-related muscle group; (e) TA in multiple phases active muscle group; (f) RF in multiple phases active muscle group; (g) BF in multiple phases active muscle group; (h) ST in multiple phases active muscle group.

7.1.4 Body-weight support-related changes during overground walking

The stance-phase duration of walking under different off-loading rates is illustrated in Figure 7.2. The stance phase does not reveal a linear relationship with off-loading changes. The shaded region in Figure 7.4 highlights the average stance phase. In the body weight supported experiment, the magnitudes of muscles decrease and then increase with increasing off-loading while the activation durations decrease in general observation.

Group 1: the power generated by VO and GM decreased initially then increased with the increasing body-weight off-loading (see Figure 7.4 (c) and (d)). It is effective to increase the off-loading weight to decrease the muscle power of body weight support. However, the large off-loading of body weight stimulated muscle activity to high level by pulling up the trunk elicits extra dorsiflexion/plantarflexion to compensate walking motion that altered the muscle pattern greatly.

Group 2: muscle activity patterns of SO and MG were sensitive to the off-loading changes; see Figure 7.4 (a) and (b). They varied with respect to shape, duration and volume. In several cases, they remained inactive which would have no benefit to rehabilitation. The reason for this is that the device provides all of the forward propulsion for the subject instead of the muscles. However, the large off-loading achieves high levels of assistance as same as Group 1. This may be due to the height of the BWS top lever being so high that subjects were unable to make contact with the ground during for walking which in turn introduced foot dorsi and plantar flexion.

Group 3: muscles in multiple phase active muscle group simplified their activity pattern compared to walking without the help of the BWS device (see Figure 7.4 (e) – (h)). Independent experiments are carried out of major movement of lower limb including hip hyperextension/extension/flexion, knee flexion/extension and ankle dorsoflexion/platerflexion in stance, sitting and lying down position. Muscles can be activated to different levels for the same motion of same subject. Gravity varied in different testing postures is one of possible explanations. Muscles take advantage of gravity to move the skeleton during the dynamic walking process, while it is do not happen in sitting and lying position. Therefore, the adaptation to gravity changes walking with and without constrains from BWS device might result in the simplification of muscle patterns.

Figure 7.4 Muscle activation pattern changes with normalized off-loading rate (a) SO in propulsion-related muscle group; (b) MG in propulsion-related muscle group; (c) VO in support-related muscle group; (d) GM in support-related muscle group; (e) TA in multiple phases active muscle group; (f) RF in multiple phases active muscle group; (g) BF in multiple phases active muscle group; (h) ST in multiple phases active muscle group.

Muscle activation changes with speed, in general, the magnitude increase and the duration decrease with speed increased. The study is extended to integral EMG [154] of the muscle activation activity shown in Figure 7.5. The integral area of muscles activation for each subjects are calculated and normalized to the maximum value for comparison. The integral area index is obtained and demonstrated. The integral area index did not change in normalized walking speeds (< 0.08 , $0.08 - 0.20$, $0.20 - 0.30$) and then increased with faster speeds. It is suggested that muscle physiological energy remains unchanged with limited walking speed. In the loading study, the integral index varied from person to person may caused by the machine constrains.

Figure 7.5 Integrated area index of muscle activity in different normalized walking speeds.

Study of different walking speeds indicates that contribution bursts of muscle activation from all muscle groups generally increased with faster walking speeds. Few of them decreased and even disappeared of muscles in Group 3. It is also noted that activation duration decreased before it increased with faster walking speed. It is believed that these decreased/increased may be attributed to two reasons. Firstly, evidence shows that

skeleton alignment contribute to the gravity resistance when walking slowly. Muscles activate to hold skeleton alignment in the upright position. Therefore, there is greater demand placed upon the duration of muscle activation when walking slowly. Secondly, this holding demand is substituted by sustaining stability during fast walking because contra-lateral limb swing with larger propulsion affect the balance. Those two facts acting together alter muscles activation. The minimum value of muscle activation probably occurred in normal walking speed which is also proposed by least energy cost of walking [155].

In the support-related test, it is found that muscle activations were very sensitive to the change of off-loading and that muscles simplified their activity pattern. It may caused by constraints from the machine which provide body weight support in the hanging manner which influence walking. Furthermore, a faster than 0.7 normalized walking speed and a larger than 50% off-loading of over-hanging BWS lead to extra motion such as foot dorsiflexion/plantarflexion, it is not suitable for effective rehabilitation overground walking training program assisted by robotic gait devices.

7.2 Clinical Trials and Evaluation of *NaTure-gaits*

After subject test on the BWSLT device, healthy subject are tested *NaTure-gaits* with pelvic and lower limb assistance. Pre-trial is conducted before the walking trials to allow overground walking experience. Feet of the subject will come into contact with the ground during the trial. With the ground contact as generated motion in Chapter 4, clinical trials are carried out successfully. The body weight will be partially supported by the subject, and thus the study focus on muscle activation pattern which could be different between the healthy and pathologic walking. Furthermore, the EMG results are presented associated to three subjects: (1) complete SCI, (2) incomplete SCI, and (3) stroke. Three representative subjects are selected to this study. Their muscle activation is presented.

Additionally, the effect of the robotic-assisted overground walking is highlighted by comparing to the totally robotic-assisted walking above ground as sky walking alike.

Table 7.2 Basic Information of All Subjects

Subject	No.	Age	Weight(kg)	Height(m)
Healthy	1	23	1.68	65
Healthy	2	23	1.79	68
Healthy	3	26	1.72	70
Healthy	4	24	1.60	75
Group mean		24	1.69	69.5
Standard Deviation (SD)		1	0.058	3
SCI	1	32	1.78	50
SCI	2	66	1.64	65
Stroke	1	73	1.58	70
Group mean		57	1.67	61.67
Standard Deviation (SD)		16.67	0.076	7.78

7.2.1 Participation and protocol of trials

After the in-house testing has been completed, SCI/stroke subjects are recruited from local hospital for the clinical trials. In order to study the functionality of the developed gait rehabilitation with SCI/stroke subjects, three routine SCI/stroke subjects participated in this study. The descriptive information of all subjects involved in this study is presented in Table 7.2.

Although comparison is currently tailored to three specific individual suffering from spinal cord injury or stroke, it is believed that the subject's condition is still a good representation of the demographics of spinal cord. Description of three SCI/stroke subject is induced and listed in Table 7.3.

- P1 is a 32 year old Asian male, with a complete C6 spinal cord injury. He there has lost his motor control and sensory function of his lower limb. Therefore, maximum

assistance is needed for his daily activities and he is reliant on a wheelchair for mobility.

- P2 is a 64 year old Asian male. He subscribes to a regular rehabilitation program. After years of rehabilitation, the subject has since regained some control and sensation in the lower limbs from an initial state of paralyzed from the trunk downwards. However, the subject is still largely unable to walk without assistance.
- P3 is a 73 year old Asian female. She can walk with aid of walking frame after years of rehabilitation after a stroke three years ago. The left side is weaker than the right side of her body. She is still largely unable to walk without assistance, too.

Table 7.3 Detailed Information of SCI/Stroke Subjects

Subject Number		P1		P2		P3	
Gender		m		m		F	
Age		32		67		73	
Years of injury		5		10		3	
Impairment		TBI		SCI		Stroke (Left)	
Skeletal Level Lesion		C6		C5		-	
		complete		incomplete		-	
ASIA impairment scale		A		D		-	
Ambulating		N		Y		Y	
Level of assistance		max assistance		moderate		Moderate	
		right	left	right	left	right	Left
Range of motion	Hip	-		Full		Full	
	Knee	-		Full		Full	
	Ankle	-		20-40	30-40	NA*	0-20
Spasticity of limbs	Hip	-		-	-	-	-
	Knee	-		2	1+	-	1
	Ankle	-		-	-	-	-

*not available with block of AFO

7.2.2 Walking with/without robotic-assisted gait locomotion

The first study is carried out as the control group, while the comparison between other two groups is to show the effectiveness of the gait system by virtue of the study on muscle activation via EMG analysis. Three experiments are shown in Figure 7.6.

- First experiment is for the healthy subjects walking. They were asked to perform repeated walking trials across a six meters floor-mat at self-preferred speed. Force sensors within the floor-mat were recorded walking speed of every step. Subjects then performed a total of ten repetitive trials. Before that, they were given 10 to 15 minutes to practice the constant speed around 0.15 m/s (slow walking). They started their walking three meters away from the floor-mat and stopped three meters after stepping off. This is to exclude interfere acceleration and deceleration factors that may affect them to gain and remain constant velocity. Gait cycle are accepted, if the measured speed was within 5% of the target speed [44]. If not, subjects were asked to repeat the trial.
- Second experiment is for the SCI subject walking with help of three persons. He was assisted with two care-givers and one helper for supporting and shifting the body-weight. Due to the inability of the subject to lift his right foot for moving forward, one helper pull the foot up and bring it forward to simulate the scenario that the subject was walking on his own. He was asked to perform repeated walking trials across a 6 m floor-mat at 0.1 m/s (maximum speed for safety) for five times. The wheelchair was next to him for taking rest in between the experiments.
- Third experiment is for the same SCI subject walking with the proposed gait. Body weight support, pelvic control, and active assistance to all joints on the lower limb and functional over ground walking without the subject was tested. The motion generation for the overground walking is next introduced. With the gait pattern implemented, the subject wearing EMG units then performed repeated walking trials with the gait system along 10 m with 0.1 m/s constant walking speed for five times. It takes around

15 mins for setting up and attaching subject to the system. A caregiver is next to the gait system for the emergency shutdown.

Figure 7.6 Three experimental setups: (a) healthy subject walking; (b) SCI subject walking with the help of three persons assisting; (c) the same SCI subject walking with NaTure-gaits.

Comparison between SCI walking with and without assistance of the device is shown in Figure 7. 7. From the right side, the muscle groups remain much smaller activation volume compared to the walking with assistance of three therapists. Therefore, the effect of the *NaTure-gaits* has less effectiveness of muscle activation. Although the activation volume of some muscle groups are lower than the clinical rehabilitation with help of three therapists. Muscle pattern shows correct activation which is helpful for recovery using no external help from therapists. Therefore, saving professional manpower from labor-intensive physical assistance is one of contributions of this device. Therapists can pay their attentions to supervise and advice. However, the constraining of the device limits the trunk and pelvic motion which contributes to the changes in muscle pattern.

Figure 7.7 Comparison between the SCI walking with and without BWSLT device (SCI walking without device used the dark color; with the device used the light color) (a) right side; (b) left side.

7.2.3 SCI/stroke subjects walking with rehabilitation robot: *NaTure-gaits*

This study aimed to test this device on healthy subjects and two SCI and one stroke subjects with a view to evaluating the ability of the system to train gait and to permit future system developments. The overall objective of this evaluation is to initialize overground gait training in sense of the robotic assistance coupled with standard physiotherapy with the ultimate aim of improving the walking ability of the patient.

Figure 7.8 Clinical trials: (a) complete SCI subject; (b) incomplete SCI subject; (c) stroke.

It has been suggested that 0.1m/s walking using a gait rehabilitation system is ideal for clinical gait training as. All of trials were set at this walking speed. The coordination of the system has been optimized to prevent the foot from being dragged across the ground. Following initial trials without EMG measurements, the test collecting EMG signals is repeated to investigate the level of activity in the lower limb muscles during the use of the system as shown in Figure 7.8.

Four healthy subjects used the system. They expressed that the harness strapping was comfortable and tolerable and that the system “felt” like walking. During the data collection six major muscles on both legs were selected for measurement. In Figure 7.9, cyclic muscle activation for all the tested muscles is shown.

Figure 7.9 EMG for a healthy subject on left and right lower limbs including major muscles VLO, TA, and GM.

Figure 7.10 EMG for a complete SCI subject on left and right lower limbs including major muscles VLO, TA, and GM.

(a) Case 1: complete SCI subject

Although the EMG signal from the complete SCI subject is small, it is still obtained these measures for continuity. After zoom out the activation level, certain cyclic muscle activation on his right side is presented compared to that activation before moving legs.

However, the activation level is much smaller than the healthy activation, even may be introduced by the background noise, see Figure 7.10.

(b) Case 2: incomplete SCI subject

For this subject, his right side is weaker than the left side. As in the right side, he cannot motion his right leg, some of muscle groups on his right leg show correct muscle activation such as GM. TA, RF muscle groups affects his propulsion and balance control do not shown the cyclic muscle activation. The right leg can support his weight from this analysis and this concurs with his clinical observations by his therapists. Therefore, inactivation of the TA and RF muscle is one of the main problems with respect to independent locomotion. RF affects hip flexion. Exercise for TA and RF strength needs to be enhanced. The others muscle group active levels accordingly with minor difference compared to the healthy walking, see Figure 7.11.

Figure 7.11 EMG for an incomplete SCI subject on left and right lower limbs including major muscles VLO, RF, TA, and GM.

For the left side comparison, most of this muscle groups shows a “normal” muscle activation pattern. His propulsion, balance control and body weight support have been

tested in the clinical setting and noted to move correctly. However, his therapists feel that many issues arise from a lack of sensory feedback on the left side. That is, he can move his left side but cannot control or modify the movement due to poor sensory feedback in turn limiting his function. Therefore, the training program needs to address this sensory deficit.

(c) Case 3: stroke subject

This participant's muscle activation patterns are greatly disturbed when compared to the normal group even with the use of the system. The overall activation level is lower than that with Case 2, see Figure 7.12.

Figure 7.12 EMG for a stroke subject on left and right lower limbs including major muscles VLO, RF, TA, and GM.

As in the right side, although she can motion her right leg to certain limit, most of muscle groups on her right leg present muscle activation and deactivation. The results present TA and RF do not show obvious cyclic activation. TA contributes to walking balance and flexion of knee and RF mainly in charge of flexion of hip. The future rehabilitation for her would enhance the muscle power on them.

For the left side comparison, the majority of muscle activation patterns appear abnormal.

Her propulsion, balance control and body weight support have been tested in clinical which cannot move properly. She wears an ankle foot orthosis (AFO) to stabilize her ankle joint. Spastic muscle patterns disturb her movement and can also be detected by the EMG measurements. For example, the GM can activate but has problem to de-activate on time. This is also a problem for rehabilitation. Therefore, the motion control for the left side is still not in good condition. But with help of system, her walking stimulates muscles on the lower limbs which are benefit for her brain to recall the correct walking pattern and recover.

7.2.4 OGW effect during robotic-assisted gait locomotion

The comparison between robotic-assisted overground walking and totally robotic-assisted walking above the ground as sky walking alike are shown in Figure 7.13. The muscle groups in sky walking remain much smaller activation volume compared to the overground walking. Therefore, the effect of the robotic-assisted overground walking has more effective of muscle activation than the sky walking.

Figure 7.13 Comparison between the overground walking and sky walking.

The gait trainer, *NaTure-gaits* (natural and tuneable rehabilitation gait system), has been developed to assist the gait rehabilitation. Comparison on the healthy subject shows the effects of the device via muscle pattern analysis. In the experiment, the muscle activation of overground walking is compared between the manual-assisted and robotic-assisted methods. The initial results show that the performance of the device can provide impact-less overground walking and it is comparable to that by manual assistance in gait rehabilitation training.

Furthermore, the EMG results are presented associated to three subjects: (1) complete SCI, (2) incomplete SCI, and (3) stroke. Three representative subjects are selected to this study. Complete SCI subject present smallest muscle activation volume near zero. The incomplete SCI shows his left side is in better condition including activation level and duration for activation and deactivation. The left side stroke subject has stronger muscle activation on the right leg. The excessive or uncontrolled activity patterns provide large disturbance for her recording.

Additionally, the proposed generation method has been tested on the gait system for the overground walking. Experiments are conducted to compare the overground walking to off-ground walking based on the joint angles obtained from the motion capture experiment. It is found that the effect of the robotic-assisted overground walking has more muscle activation. The results have also demonstrated the effectiveness and smoothness of the overground walking on the gait system.

7.3 Systematic EMG Evaluations for Walking

The performance of the overground gait training with the gait system is evaluated by comparing the EMG signals from the muscles on the lower limbs of subject. With several disturbance factors such as walking speed, step length, and kinematics data influencing the evaluation, it poses a restriction on the validity of evaluation results by virtue of the

average pattern analysis commonly used in the treadmill walking. In addition, the auto-correlation and power spectrum are applied in the present evaluation. In order to evaluate the performance of the developed overground walking pattern, four factors including regularity, repetition, symmetry and smoothness are proposed to evaluate the quality of the overground walking with *NaTure-gaits*.

- **Regularity:** it is a key character appearing in every walking cycle with a standard profile. For example, lateral gastrocnemius, located at back of calf, remains non-activation except for the duration that it is sustaining the alignment of the lower limb in the stance phase. Patients with walking impairment may lack activation or have excessively prolonged activation beyond the stance phase, thus losing the standard regularity.
- **Repetition:** it describes an identical or similar characteristic presenting in every gait cycle of an individual. Healthy person shows high repetitive level without external disturbance, while this might be the case for patients. The trend can serve as an evaluation for steady walking capability.
- **Symmetry:** it describes an identical or similar characteristic that alternates left and right sides. However, symmetry evaluation is useless for individuals with unbalanced compromising walking style.
- **Smoothness:** it describes the quality of motor control. The gait movement shows smoothness and accuracy in healthy walking. However, patients might have quivering and spasticity interference to movement.

7.3.1 Regularity

The regularity, which appears in the average activation profile, represents the significant performance of the overground walking. Although walking speed and other gait

parameters alter the dynamics [156], a standard regularity for healthy walking are presented. That is, major activation bursts are mounted for a general observation. As a subject walking with the device, decreased activation volumes are shown. The comparison of muscle (lateral gastrocnemius) for three healthy subjects, randomly picked, is shown in Figure 7.14. As can be seen, the muscle activation volume decreased with the assistive device. One of the possible reasons is that the locomotion and support is provided by using the device with motorized power instead of muscle. Another observation demonstrates a similar muscle pattern with and without device. The difference occurs when equipped robotic device restricts the motion of trunk and lower limbs. In fact, abnormal sequence and timing of muscle activation is to be expected during the dynamic cycle of walking.

The second comparison extended to SCI subject walking with/without the gait device. Three healthy subjects walking without the device are kept as the control group. Another muscle (gluteus maximus) on thigh is compared as representative. Although muscle activation increased in the manual-assisted walking, the muscle activation pattern is different from standard pattern.

In contrast, the muscle activation pattern is closer to that of a healthy walking. As it is observed, the magnitude of the strength decreases if compared to the control group. This is the same to the healthy subject walking with the gait device, as the device compensates the muscle force during walking process and generates normal muscle activation. In addition, the activation volume can be increased with faster walking, which does not increase for safety consideration at this stage.

The EMG activation pattern of muscles on the lower limb was recorded for each respective subject, including healthy volunteers and an individual with incomplete SCI. Three experiments carried out involve (1) healthy volunteers walking at self-preferred walking speed, (2) a SCI subject walking with the help of caregivers, and (3) the same

SCI subject walking with the help of the gait device, as shown in Figure 7.14. The initial results show that the performance of the gait device is comparable to that by manual assistance.

Figure 7.14 Comparisons of the average muscle activation pattern: (a) lateral gastrocnemius muscle comparison between three healthy subjects walking with/without the device; (b) gluteus maximus muscle comparison between three healthy subjects and a SCI subject walking with/without the device.

7.3.2 Repetition

Repetition is important for an intensive training program, which hypothetically serves as a positive stimulation for neuroplasticity to rebuild motor control ability. Evidences have obtained from the successful animal experiments. In order to evaluate developed gait device, the above-mentioned autocorrelation are compared between SCI subject walking with/without the device. The healthy subject's walking profile is included for assessment.

The repetition index for muscle (tibialis anterior) is shown in Figure 7.15 (a) and repetition zoomed in the neighboring cycle is highlighted in Figure 7.15 (b). As it illustrated, every local peak is approach to 1.0 for the healthy subject which indicates high similarity per cycle. For the SCI subject walking with device the periodic characteristic is shown with a decreasing repetition index. For the SCI subject walking with three helpers, the periodical characteristic is presented with lowest repetition index. Therefore, the

repetition of the SCI subject walking with the gait device is better than that of his walking with the helpers.

Figure 7.15 Repetition of gait cycle: (a) comparisons between healthy walking and SCI subject walking with/without the device; (b) comparisons within two neighboring gait cycles.

7.3.3 Symmetry

By studying the quality of walking and by applying the autocorrelation to both sides of walking, the symmetry of muscle (tibialis anterior) pattern is compared in Figure 7.16. For the healthy subjects, the result of left side is almost the same to the right as presented before. For the SCI subject walking with the device, the difference between right and left leg is shown. The right side has poorer symmetry than the left side, which is due to the weakness of right side. For the SCI subject walking with three helpers, the difference between right and left legs is also shown. A closer look at the two neighboring cycles demonstrates that the effectiveness of the device with less manpower is the same if not better than that of manual assistive rehabilitation training.

Figure 7.16 Symmetry of gait cycle: (a) comparisons between healthy walking and SCI subject walking with/without the device; (b) comparisons within two neighboring gait cycles.

7.3.4 Smoothness

As mentioned in Introduction, the muscle activation with spasticity might form an unstable motion. The evaluation of the smoothness is therefore important. The spectral power of EMG signal for the SCI subject walking with/without the gait device is illustrated in Figure 7.17. In the comparison, the power spectrum associated to the SCI subject walking with three helpers requires a higher energy compared to the SCI subject walking with the device. It indicates a less power exerted for the SCI subject walking with the gait device.

Figure 7.17 Spectrum power of SCI subject walking with/without the gait device.

As the robotic overground gait device can provide a longer duration of therapy if compared to that of conventional therapy. In order to evaluate the effective of the gait system systematically, four factors of the gait muscle strength via EMG signals are observed: regularity, repetition, symmetry, and smoothness. The systematic evaluation is helpful to the future development of rehabilitation gait systems.

7.4 Detection of Spinal Cord Injury via EMG Frequency Analysis

The spastic symptom is commonly presented in SCI subject and it can be detected from power spectrum study in Section 7.3. Therefore, an extensive frequency analysis is conducted. EMG data were recorded from seven major muscles in the lower extremity upon ten healthy and seven SCI subjects. The frequency analysis, computed by the power spectral density (PSD), exhibit substantially difference between healthy and SCI subject. The results indicate that the peak values of EMG spectrum in SCI subjects shift towards lower frequency band. The approach presented for assessment of gait problems in this thesis is an effective tool and likely contribute to qualified evaluation and bio-feedback of gait rehabilitation in future.

7.4.1 EMG power spectral density

As the absolute magnitude of raw EMG signal are not consistent between individuals, some researchers focus on the shape of EMG profile, times of peak or onset/cessation of myoelectric activity [153]. In this study, the power spectral density (PSD) is chosen to analyze the EMG in frequency domain. Analysis of the EMG signal of each muscle was performed with a MATLAB-based program (Math works, Inc., Natick, MA). PSD of EMG raw data were obtained for each muscle group in every subject for all gait cycles recorded.

Figure 7.18 Mean power spectrum of the full-wave rectified EMG of major muscles on the lower limbs for healthy and SCI subjects: (a) vastus lateralis (VL), (b) rectus femoris (RF), (c) semitendinosus / semimembranosus (ST), (d) tibialis anterior (TA), (e) soleus (SO) and (f) gastrocnemius medialis (MG).

Power spectrum density (PSD) analysis is widely used to detect changes of muscle activation and is shown in Figure 7.18. In general observation, all muscles decrease the muscle activation level compared to that of healthy subjects. This demonstrates the weakness of all leg muscle for SCI subject. Furthermore, the location of the peak PSD has a clear shift in EMG frequency band for SCI subjects. That is, SCI subjects exhibit less activation power during the higher frequency domain, which may be influenced by spinal

reflexes. It is assumed that the muscles lost their ability for high vibration in high frequency band. Therefore, examination on frequency performance on the lower frequency band can distinguish EMG signal obtained from SCI subject from healthy subject, which provided alternation for the clinical application of EMG.

7.4.2 Clinical applications

Comparison of power spectrum density (PSD) is provided for muscle activity between healthy and SCI subjects. Our results show SCI patients adopt a neuromuscular power spectrum that has less high frequency energy because the peak values of EMG spectral in SCI subjects shift towards lower frequency band. To diagnose and further qualify the motor disability, assessment is frequently applied for Parkinsonian patients [157], hemiparetic stroke [158], cerebral palsy [43, 159-161] vestibular impairment [162] and so on. To our knowledge, assessment of muscle activity for SCI patient is not commonly provided [163-164]. At same time, it has been proposed control mechanism exist beyond spinal circuits underlying locomotion such as central neuronal program or central pattern generator (CPG) [165-166]. Therefore, investigations on SCI patient muscle activities have a potential value to in future. In this study, it is assumed that the EMG frequency property is close linked to the response to a probably stimulus generation in the spinal circuits control loop for motion. Low sensitivity of motor control on SCI patients may lead to the lack of high frequency vibration in all muscles' activity. For example, two of SCI patients have slight motor weakness and can walk independently without problem. It is hard to observe their previous spinal cord injury, the power spectrum is also approach to normal. Therefore, it is possible to develop a quantitative assessment of SCI gait in the frequency analysis.

Furthermore, a successful application is reported that the location of the peak PSD has a clear shift in EMG frequency band under isometric fatigue conditions. It is interesting that the muscle pattern of SCI subjects has the same peak shift to that of fatigue subjects. The

peak value of SCI subject is smaller than the fatigue subjects. The relationship between SCI subject and fatigue subjects can be investigated. Additionally, age [44], fatigue [167], walking speed [60-61] have been studied for their contribution to the muscle pattern change. The time-frequency analysis [168] and principle component analysis (PCA) [159] are also powerfully provided in this area to diagnose the patients from the normal. Therefore, it would be great if EMG data analysis in the frequency domain can also be extend widely applied to assessment of locomotor tasks for SCI subject [122].

With development of machine-assisted gait training including treadmill walking [169], body weight supported treadmill training (BWSTT) [170], body weight supported locomotion training (BWSLT) [152] and functional electrical stimulation (FES) [164], the compensation and evaluation of those method are critical for SCI rehabilitation. Our initial investigation is also contributes to quantification of EMG signals in frequency analysis, which will be useful for the bio-feedback on the intelligent compensation provided by machine.

EMG data are collected on lower-extremity from ten healthy subjects and seven SCI patients with walking problems at a slow walking speed. Processed EMG signals were normalized to peak spectrum over a gait cycle for individuals. Compared to the healthy subjects, SCI subjects exhibit less activation power during the higher frequency domain, suggesting gait problem can be diagnosed by whether activation can reach a normal frequency band, which provide a diagnostic tool for spinal cord injury subjects. In conclusion, the assessment of gait problems in frequency domain, is an effective tool and likely contribute to quantified compensation and evaluation of pathological gait.

7.5 Summary

In conclusion, the robotic trainer, *NaTure-gaits* (*Natural and TUneable rehabilitation gait system*), has been developed for overground walking rehabilitation exercise. Before,

evaluation of walking on *NaTure-gaits*, walking speed and off-loading level effects on the lower limb muscular activation are provided. In general, experimental results from different walking speeds demonstrated bursts of muscle activation changes in terms of magnitude and duration from very slow to very fast walking. Furthermore, fast walking or large off-loading introduce disturbance of muscle activation pattern, led to greater motion of the foot in terms of dorsiflexion/plantarflexion and as such were found to be unsuitable for the bespoke robotic gait rehabilitation device. It was found that muscle activation patterns are sensitive to the walking speed and level of off-loading. Therefore, constraints on those major factors are important to keep the validation of the evaluation.

Then in order to evaluate the effectiveness of the device, four healthy volunteers and two spinal cord injury (SCI) or one stroke subjects were invited to participation of the walking experiments with the gait system. The EMG signals are obtained simultaneously during the walking with system. Major muscles are recorded and show active participation while walking on the system. The EMG results are presented associated to three subjects: (1) complete SCI, (2) incomplete SCI, and (3) stroke. Three representative subjects are selected to this study. Their muscle activation is presented. Complete SCI subject present smallest muscle activation volume near zero. The incomplete SCI shows his left side is in better condition including activation level and duration for activation and deactivation. The left side stroke subject has stronger muscle activation on the right leg. The comparison between healthy and SCI/stroke subjects walking with the device demonstrates the effect of device based on muscle pattern analysis. Therefore, the initial results show that the performance of overground walking with the gait system with minimum manual assistance provides effective training effect.

In order to evaluate the effective of the gait system systematically, four factors of the gait muscle strength via EMG signals are observed: regularity, repetition, symmetry, and smoothness. The initial evaluation of walking performance presented in this Chapter

provides a better insight, which is helpful to the future development of rehabilitation gait systems. The approach presented for assessment of gait problems in this Chapter is an effective tool and likely contribute to qualified evaluation and bio-feedback of gait rehabilitation.

Chapter 8 CONCLUSIONS AND RECOMMENDATIONS

This chapter summarizes the works presented in this thesis. Results and conclusions in the previous chapters will be highlighted and discussed. An outlook of recommendations will be provided.

This thesis presents a research work on the robotic-assisted overground walking (OGW) training system for lower limb rehabilitation. Background and research gap on robotic assisted devices are summarized in Chapter 1.

In Chapter 2, the literature survey covers clinical aspects on gait rehabilitation, summary of current gait robot, current research work on motion generation and control algorithms for the overground walking (OGW) rehabilitation robot, and evaluations of the robotic therapy.

Based on those studies, human motor control is first investigated to avoid a totally passive exercise on robotic system in Chapter 3. In order to enhance effect of the active training, incorrect muscle activations on lower limbs for the pathological gait are identified. It is an effective tool and contributes to qualified evaluation and bio-feedback for gait rehabilitation.

After investigating human motor control for one joint, motion generation algorithms of gait training are proposed for coordinating all joints on the pelvis and lower limbs in Chapter 4. For different individual gaits and physical dimensions, an adjustable motion generation algorithm is proposed. The pelvic trajectory is predefined and key points are specified within one gait cycle to obtain the foot trajectory from the designated step length and height. This result can work as the reference trajectory for the parallel exoskeletons

attached on the lower limbs. The walking trajectory is compared to the result obtained from the motion capture experiments. The application of the method shows the adjustability and feasibility of the motion generation. Furthermore, to reduce, if not to avoid, the ground impact and dragging of the footpad occurring during the course of overground walking with the help of the device, a gait motion generation is developed and specified to generate the gait patterns in consideration of ground conditions for an impact-less and smooth footpad contact.

In order to track the desired trajectory generated in Chapter 4, control algorithms have been investigated in Chapter 5. A repetitive control algorithm based on the characteristics of a cyclical reference signal and the internal model control principle, implanted the periodic signal generator has developed into the closed-loop system in order to achieve steady tracking of cyclic reference signals and rejecting disturbance on the rehabilitation robot. Not only the tracking performance is improved, the assist-as-needed force can also be programmed under control. The repetitive controller works as a stabilizer. By selecting a proper stabilizer, the function of robotic-assisted system can be extended from assistance to resistance in the rehabilitation. In order to synchronize the walking velocity, the exoskeleton is identified on-line by the relay feedback test. After modeling of controller and actuator, Galil controller are used for an optimized and adaptive control with an autotuning control. The simulations model of the system show the effectiveness of those methods.

Chapter 6 provides the implementation of these proposed algorithms to the modular system including: pelvic arm (PA), robotic orthosis (RO) and mobile platform (MP). Each control scheme for individual sub-system and the synchronized control of them are introduced. The implementations of the motion generation and control methods proposed in Chapter 3-5 are also presented in this chapter.

The evaluation of the whole system is presented in Chapter 7. In order to evaluate the effectiveness of the device, four healthy volunteers and two spinal cord injury (SCI) or one stroke subjects were invited to participate of the walking experiments with the gait system. Three representative subjects are selected for this study. Their muscle activation is presented. Complete SCI subject presents smallest muscle activation volume near zero. The incomplete SCI shows his left side is in better condition including activation level and duration for activation and deactivation. The left side stroke subject has stronger muscle activation on the right leg. All these measuring outcomes agree with the observation results. Furthermore, experimental results from different walking speeds and off-loadings demonstrated bursts of muscle activation changes in terms of magnitude and duration to select suitable walking speed and off-loadings.

The comparison between healthy and SCI/stroke subjects walking with the device demonstrates the effect of device based on muscle pattern analysis. In order to evaluate the effectiveness of the gait system systematically, four factors of the gait muscle strength via EMG signals are observed: regularity, repetition, symmetry, and smoothness. Furthermore, frequency evaluation method is also proposed to distinguish the gait impairment.

8.1 Summary of Current Works

The major contributions and originality of this work are summarized as follows:

Modeling of human-machine interaction (HMI):

In order to improve the subjects' active participation level, this thesis has studied the control of robotic-assisted force, either assist or resist. An internal model is introduced for the human motion control, especially, for the pathologic gait, a stabilizer is proposed for a stable motion with help of the exoskeleton. The function of the stabilizer is investigated on the human model to re-establish the stable motion control for walking training. By

selecting a proper stabilizer, the function of robotic-assisted system can be extended from assistance to resistance in the rehabilitation. The performance and robustness of the overall external assistance are studied for future control algorithm used in robotic device.

Human gait motion study:

In view of the difference in gait profiles of each individual subject, it is crucial to include their own gait profile as a factor to be considered for planning gait rehabilitation for patients. The proposed motion generation provides walking pattern for different body size, stride length, foot clearance height *etc.* Key points are specified within one gait cycle to obtain the foot trajectory from the designated step length and height. The trajectory is obtained via curve fitting based on these specified points. On the other hand, in order to keep the desired foot trajectory, the pelvic motion is compensated to accommodate hip and knee joint angles. Next, the metatarsal trajectory is planned to satisfy the ground condition to conduct a smooth OGW. Furthermore, to reduce, if not to avoid, the ground impact and dragging of the footpad occurring during the course of overground walking with help of the device, a gait motion generation is developed and specified to adjust the gait patterns for an impact-less and smooth footpad contact based on the walking model proposed. Therefore, an adjusting index is proposed to guarantee the impact-less overground walking pattern based on optimizations.

Control algorithm:

The repetitive control algorithm based on the characteristics of a cyclical reference signal and the internal model control principle, implanted the periodic signal generator has developed into the closed-loop system in order to achieve steady tracking of cyclic reference signals and rejecting disturbance on the rehabilitation robot. Not only the tracking performance is improved, the assist-as-needed force can also be under control. Furthermore, in order to control the walking velocity on time for the OGW system, control strategy is proposed based on on-line model identification algorithm. An

autotuning controller is developed to synchronize motions for the sub-systems. The application of relay identification and optimal control provide a better tracking performance combined with repetitive controller improve the performance cycle by cycle as an iterative learning method.

Robotic system and successful trials conducted in clinical application:

Control of a robotic gait rehabilitation system, *NaTure-gaits* is one of major contributions. The system is completed with fourteen degrees of freedom, and control system for flexible motion programming for the entire system. *NaTure-gaits* provides lower limb motion assistance in the Sagittal plane, natural pelvic movement assistance, body weight support with no restriction on the pelvic movement, and overground walking (OGW) experience. The modular and reprogrammable architecture of the developed system allows control strategy to be implemented and validated on the system.

After the in-house testing has been completed, the current work recruited SCI/stroke subjects from our local hospital for the clinical trials. Three routine SCI/stroke subjects participated in this study. The coordination of the system has been optimized to prevent the foot from being dragged across the ground. The initial results show that the performance of overground walking with the gait system with minimum manual assistance provides effective training effect.

System evaluation via muscle activation analysis:

Comparisons of lower limb muscular activation between subjects walking with and without robotic device are shown that the performance of the device can provide impact-less overground walking, which is comparable to the performance obtained by manual assistance in gait rehabilitation training. Experimental results from different walking speeds demonstrated bursts of muscle activation changes in terms of magnitude and duration from very slow to very fast walking. When examining the impact of body-weight

off-loading, it was found that muscle activation patterns were sensitive to the level of off-loading. In general, fast walking or large off-loading introduce disturbance of muscle activation pattern, led to greater motion of the foot in terms of dorsiflexion/plantarflexion and as such were found to be unsuitable for the bespoke robotic gait rehabilitation device.

Quantitative assessment for the pathological gait:

In order to evaluate the effective of the gait system systematically, four factors of the gait muscle strength via EMG signals are observed: regularity, repetition, symmetry, and smoothness. Furthermore, in order to select suitable training programs, it is important to identify and assess the muscle dysfunction of SCI subjects. However, quantifying assessment of SCI gait is challenging as individual gait is usually changing, asymmetric, slow and instable based on variable nature of EMG signal. In order to demonstrate the phase transition and pathological synchronization of muscle activation elicited by the spastic gait/muscle dystonia of pathological gait, the results were compared via the distribution of muscular activation in frequency domain within the average gait cycle (GC). Focusing on those specified phases of certain muscles would make a better detection for the pathological gait. It could be concluded that the frequency approach presented in this thesis for assessment of gait problems is an effective tool and likely contribute to qualified evaluation and bio-feedback of SCI gait rehabilitation in future.

8.2 Recommendations

There are several possible directions for future work in the areas of research presented in this thesis. Some suggestions are listed as follows:

Effective rehabilitation to motivate subject for an active training

After fabrication of prototype, it is important to transfer the assistive force effectively from machine to human. However, overwhelming external force works as constrain for the subjects to active their own muscles. Effective rehabilitation assistance is to motivate

the subjects and stimulate their muscles to active. It is also helpful to develop an effective visual feedback system to indicate the human muscle system for patient and therapists.

Online Generation of walking motion in 3dimensions

Current motion generation is used for the exoskeleton movement in the Sagittal plane. Kinematic and dynamic studies can be carried out for the 3D assistant motions. The offline optimization can be substituted by online trajectory generation for more flexible walking motion with faster computation speed.

Enhanced Control Strategy

In clinical practice, the therapists “feel” the muscle strength of patients and encourage them to move from time to time during the rehabilitation training. Current system has realized repetitive control for tracking the generated motion. The next stage for the more flexible optimal assistance is a crucial control strategy from clinical aspects for effective robotic rehabilitation.

Then, from the qualification of EMG related to the muscles strength, the compensation force can be obtained for the interface force in HMI in the proposed optimal assistance model. The muscle patterns for healthy walking can be qualified from different walking speeds. The designed gait system can also provide different walking speeds by virtue of the velocity control of the mobile platform (MP). The trajectory for the displacement of pelvic has to be synchronized with the gait of the patient’s walking. Resistance control model will be realized based on the optimal control.

Bio-feedback

Complete bio-feedback architecture can be created with the incorporation of several bio-signals feedback. EMG information can be used to identify the muscle activation pattern, and encourage the patient to activate correct muscle during gait locomotion training.

Ground reaction force feedback provides the patient and therapist about the information about the body weight shifting, which is essential prior to swing phase of the non-supporting leg. Other useful bio-signals can be further explore, combines these information could provide both patient and therapist a complete information on the rehabilitation progress, therapeutic outcome, and as a reference for the planning of future rehabilitation intervention.

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APPENDIX A BASIC ANATOMICAL TERMINOLOGY

Basic anatomical terminology allows us to describe the locomotion clearly and precisely. Every direction terminology is described based on the anatomical position as the subject stands upright facing the observer, with the head level and the eyes facing forward. The feet are flat on the floor and directed forward, and the arms are at the sides with palms turned forward. Briefly, there are three reference planes and eight fundamental directions as shown in Figure A.1.

In the reference planes, the sagittal plane means a plane which divides subject into right and left portions; the coronal plane (also named frontal plane) divides subject into front and back portions; the transverse plane divides subject into upper and lower portions. In the eight fundamental directions, medial and lateral means nearer to or farther from the midline; anterior and posterior means nearer to the front or back of the body; superior and inferior means toward or away from the head or the upper part of the subject; proximal and distal means nearer to or farther from the attachment of a limb to the trunk.

A.1 Terminology of human body

The human body is divided into several major regions as head, neck, trunk, upper limbs and lower limbs. The head consists of skull and face; the neck attaches head to the trunk; the trunk consists of the chest, abdomen and pelvis; the upper limb consists of shoulder, arm, forearm, wrist and hand; the lower limb consists of hip, thigh, knee, leg, ankle and foot. The specific body regions are shown in Figure A.2.

(a) Skeletal system

Walking almost involves every bone and muscle in the body, especially two legs provide both support and propulsion alternately. Each lower limb (lower extremity) has 30 bones

in four location – (1) the femur in the thigh; (2) the patella (kneecap); (3) the tibia and fibula in the leg; (4) and the 7 tarsals in the tarsus (ankle), the 5 metatarsals in metatarsus, and the 14 phalanges (bones of the digits) in the foot (Figure A.3). Pelvis connects the bones of the lower limbs to the axial skeleton. It also provides a strong and stable support for the trunk and upper body while walking.

(b) Muscular system

Muscles involved in ankle plantar flexion are gastrocnemius and soleus. Gastrocnemius is a strong plantarflexor when the ankle is in motion, while soleus is more active when standing at ease. For ankle dorsi flexion, tibialis anterior plays a major role. Extensor digitorum longus and extensor hallucis longus are also involved in dorsi flexion, but only generate little force as they are weak muscles. Knee flexion involves the activation of hamstrings i.e. bicep femoris, semitendinosus and semimembranosus. For knee extension, it uses the bicep femoris and the quadriceps femoris i.e. vastus lateralis, vastus medialis and rectus femoris. Gastrocnemius and soleus are also believed to contribute to knee extension. Hip flexion involves the muscles like iliopsoas, rectus femoris, tensor fasciae latae, sartorius and pectineus. Tensor fasciae latae plays an important role here in stabilizing the hip and knee joints when walking. Lastly, muscles responsible for hip extension include gluteus maximus, bicep femoris and erector spinae. Gluteus maximus is the major extensor muscle, hence plays an important role in hip extension. The bicep femoris contributes to walking by knee flexion, while erector spinae stabilizes the lumbar spine and pelvis during hip extension.

Figure A.1 Anatomical position with three reference planes and six fundamental directions.

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<http://www.cyberorthotics.com/new123/members/ebook/Chapter%201.html>]

Figure A.2 Specific body regions name related to walking.

A.2 Terminology of human gait

Gait cycle is the time interval between two successive occurrences of one of the repetitive events of walking. A friend can be recognized from their personal walking peculiarities from a long distance because those peculiarities show by every gait cycle. Although slight differences occurred in the magnitude of movements, basic gait patterns are similar in all

age and height groups cycle by cycle. There is a slight tendency for the shorter subjects show the greater flexion than the tall subjects [36, 51]. Gait cycle joint motion and muscle activity are provided in Figure A.3.

Figure A.3 Gait cycle joint motion and muscle activity [171].

A.2 Pathologic gait

(a) Stroke

Stroke of ‘brain attack’ is a type of cardiovascular disease. It occurs when a blood clot blocks an artery or a blood vessel bursts and interrupts the blood flow to an area of the brain. When either one occurs, part of the brain which cannot get the blood (or oxygen) it needs, the brain cells start to die and the brain becomes damaged.

The effects of stroke depend on which area of the brain is damaged. The abilities that are controlled by that area will be weakened or lost. The severity of the disability depends on how much the brain is damaged. For example, a minor stroke might weaken a person’s muscle; a major stroke might result in paralysis on one side of the body. Majority of the stroke survivors will have some form of disability.

Considering an isolated blood vessel, the blood flow to the brain tissue can be hampered in two ways:

Ischemic Stroke – clogs within the blood vessel

Hemorrhagic Stroke – the vessel ruptures, causing blood to leak into the brain

Ischemic Stroke

About 83% of stroke cases are Ischemic stroke. It is a result of an obstruction within a blood vessel that supplies blood (or oxygen) to the brain. The obstruction is usually caused by development of fat/cholesterol deposits on the vessel walls, which is called atherosclerosis. Ischemic stroke can occur in two ways.

For **embolic stroke**, a blood clot is initially formed elsewhere in the body before traveling through the bloodstream to the brain. The clot will eventually travel to a blood vessel in

the brain that is too small for the passage of the clot, in effect, blocking the passage. The clot will lodge there and block the blood flow, causing a stroke. The medical term for this type of blood clot is embolus.

Thrombotic stroke occurs when there is blockage of one or more of the arteries supplying blood to the brain, impairing the blood flow. The process whereby a clot forms on a blood vessel deposit and results in a blockage, is known as thrombosis, and the medical term for the clot is thrombus. The clot can also happen as a result of a buildup of fat deposits and cholesterol in unhealthy blood vessels. The body regards these as multiple, tiny and repeated injuries to the blood vessel wall. The body reacts to it just as it would if the body was bleeding from a wound by forming clots.

Hemorrhagic Stroke

Hemorrhagic stroke results from a weakened vessel that ruptures and bleeds into the surrounding brain, causing the blood to accumulate and compress the surrounding brain tissue. There are a number of disorders that affect the blood vessels, including long-standing history of high blood pressure and cerebral aneurysms that cause hemorrhages. An aneurysm is a weak or thin spot on a blood vessel wall, which is usually present at birth. It will develop over a number of years and usually does not cause any detectable problems until it breaks. There are two types of hemorrhagic stroke.

Intracerebral hemorrhage refers to bleeding that occurs from vessels within the brain itself. The primary cause is high blood pressure. **Subarachnoid hemorrhage** (SAH) occurs when an aneurism bursts in a large artery on, or near the thin, delicate membrane surrounding the brain. The burst causes blood to spill into the area around the brain, which is filled with a protective fluid. This causes the brain to be surrounded by blood contaminated fluid.

Transient Ischemic Attacks (TIAs)

Transient ischemic attacks are minor or warning strokes. It indicates an ischemic stroke is present and the typical stroke warning signs develop. This type of stroke is usually caused by blockage of blood vessel in the brain for a short period of time, which results in a slowdown or stoppage of the blood flow to that area of the brain. However, the blockage occurs only for a short period of time and tends to resolve by itself. Although TIAs generally does not damage the brain permanently, but it is a serious warning sign of a major stroke and should not be ignored. Among those who have experienced a TIA, one-third of them will have an actual stroke.

(b) Spinal Cord Injury (SCI)

Most spinal injuries are caused by direct or indirect trauma to the vertebral column, resulting in either temporary or permanent change in normal motor, sensory or autonomic function. There are a few terms used to classify spinal cord injuries. Tetraplegia (replacing the term quadriplegia) refers to the condition of complete or incomplete paralysis that involves all four extremities. The paralysis of lower extremities is called paraplegia. The definitions of complete and incomplete SCI are based on American Spinal Injury Association (ASIA) definition with sacral-sparing. Complete SCI refers to absence of sensory and motor functions in the lowest sacral segments. The condition of preservation of sensory or motor function below the level of injury, including the lowest sacral segments, is referred to as an incomplete SCI. ASIA Impairment Scale further defines the extent of injury using the following categories [172]:

A (Complete): No sensory or motor function is preserved in sacral segments S4 – S5.

B (Incomplete): Sensory, but not motor, function is preserved below the neurologic level and extends through sacral segments S4 – S5.

C (Incomplete): Motor function is preserved below the neurologic level, and most key muscles below the neurologic level have muscle grade less than 3.

D (Incomplete): Motor function is preserved below the neurologic level, and most key muscles below the neurologic level have muscle grade greater than or equal to 3.

E (Normal): Sensory and motor functions are normal.

APPENDIX B PROOF OF RELAY FEEDBACK THEORY

Based on the describing function analysis, when a sinusoidal input

$$e(t) = asin\omega t \quad (B.1)$$

is applied, the relay output, $u(t)$ will be a square waveform. Expand it into Fourier's series, can be rewritten as

$$u(t) = \sum_{n=1,3,5}^{\infty} F_n \sin n\omega t, \quad (n = 1,3,5, \dots) \quad (B.2)$$

where $F_n = \frac{4h}{\pi n}$, $(n=1,3,5, \dots)$

Since

$$F_n = \frac{2}{T} \left| \int^T u(t) e^{-jn\omega t} \right|, \quad (n = 1,3,5 \dots) \quad (B.3)$$

and

$$\omega = 2\pi/T \quad (B.4)$$

Then

$$\left| \int^T u(t) e^{-jn\omega t} \right| = \frac{4h}{wn}, \quad (n = 1,3,5 \dots) \quad (B.5)$$

Substituting (B.5) into (B.2) yields

$$u(t) = \frac{4h}{\pi} \sum_{n=1,3,5}^{\infty} \frac{\sin n\omega t}{n}, \quad (n = 1,3,5 \dots) \quad (B.6)$$

with regard to the relay feedback system, the output is

$$y(t) = \frac{4h}{\pi} A^* \sum_{n=1,3,\dots}^{\infty} \frac{\sin(n\omega t + \varphi^*)}{n}, \quad (n = 1, 3, 5 \dots) \quad (\text{B.7})$$

where

$$A^* = |G(jn\omega)| \quad (\text{B.8})$$

$$\varphi^* = \arg(G(jn\omega)) \quad (\text{B.9})$$

(B.8) and (B.9) can be obtained by

$$G(jn\omega) = \frac{\int_0^T y(t) e^{-jn\omega t} dt}{\int_0^T u(t) e^{-jn\omega t} dt}, \quad (n = 1, 3, 5 \dots) \quad (\text{B.10})$$

For the reason of simplicity, it might as well let

$$k_n = \left| \frac{1}{G(jn\omega)} \right|, \quad (n = 1, 3, 5 \dots) \quad (\text{B.11})$$

Substituting (B.7) and (B.10) into (B.11) yields

$$k_n = \left| \frac{4h}{n\omega \int_0^T y(t) e^{-jn\omega t} dt} \right|, \quad (n = 1, 3, 5 \dots) \quad (\text{B.12})$$

k_n are key parameters in the proposed identification method.

Theorem: Assume the relay feedback system establish the limit cycle, ω_{osc} is the oscillation frequency, T_{osc} is the oscillation period, $y(t)$ is the output of process, let

$$F_{sn} = \frac{2}{T_{osc}} \int_0^{T_{osc}} y(t) \sin(n\omega_{osc} t) dt \quad (\text{B.13})$$

$$F_{cn} = \frac{2}{T_{osc}} \int_0^{T_{osc}} y(t) \cos(n\omega_{osc} t) dt \quad (\text{B.14})$$

Then, a phase derivation is defined as

$$\delta_{pn} = \begin{cases} \tan^{-1} \left(\frac{F_{cn}}{F_{sn}} \right), \frac{F_{cn}}{F_{sn}} \geq 0 \\ \tan^{-1} \left(\frac{F_{cn}}{F_{sn}} \right) + \pi, \frac{F_{cn}}{F_{sn}} < 0 \end{cases}, n = 1, 2, 3 \dots \quad (\text{B.15})$$

Proof The output of the relay $u(t)$ is expanded into Fourier series, given by odd harmonics as

$$\begin{aligned} u(t) &= u_1(t) + u_3(t) + u_5(t) + \dots = \sum_{n=1,3,5,\dots} u_n(t) \\ &= \sum_{n=1,3,5,\dots} \frac{4h}{n\pi} \sin(nw_{osc}t) \end{aligned}$$

where $u_n(t)$ represent n -order component, then the fundamental component can be denoted as

$$u_1(t) = \frac{4h}{\pi} \sin(w_{osc}t)$$

When the output of the relay $u(t)$ work as the input of the process in the system, the corresponding output of process $y(t)$ is expanded into the Fourier series in the form of

$$\begin{aligned} y(t) &= y_1(t) + y_3(t) + y_5(t) + \dots = \sum_{n=1,3,5,\dots} y_n(t) \\ &= \sum_{n=1,3,5,\dots} \frac{4h}{n\pi} |G(jnw_{osc})| \sin[nw_{osc}t + \delta_{pn}] \end{aligned}$$

The n -order component of $y(t)$ is expressed as

$$\begin{aligned} y_n(t) &= \frac{4h}{n\pi} |G(jnw_{osc})| \sin[nw_{osc}t + \delta_{pn}] \\ &= \frac{4h}{n\pi} |G(jnw_{osc})| [\sin(nw_{osc}t) \cos \delta_{pn} + \cos(nw_{osc}t) \sin \delta_{pn}] \end{aligned}$$

In order to obtain the δ_{pn} , implement the complete orthogonal vector in complex plane as

$$\mathbf{v} = [\sin(w_{osc}t), \cos(w_{osc}t), \sin(2w_{osc}t), \cos(2w_{osc}t), \\ \sin(3w_{osc}t), \cos(3w_{osc}t), \dots]^T$$

Then, $\mathbf{v}$ is multiplied to two parts of $y_n(t)$, respectively, the coefficient F_{sn} of the part of $\sin \delta_{pn}$ and F_{cn} of the part of $\cos \delta_{pn}$ is obtained as

$$F_{sn} = \frac{2}{T_{osc}} \int_0^{T_{osc}} y(t) \sin(nw_{osc}t) dt \\ = \frac{4h}{n\pi} |G(jnw_{osc})| \cos \delta_{pn} \\ F_{cn} = \frac{2}{T_{osc}} \int_0^{T_{osc}} y(t) \cos(nw_{osc}t) dt \\ = \frac{4h}{n\pi} |G(jnw_{osc})| \sin \delta_{pn}$$

In the relay feedback experiment, the sample values of $y(t)$ is obtained on sampling point. With this information available, both of the above equations are transferred into numeric integrations. In our utilization, the trapezoid shape area division is applied to get results in high accuracy.

Finally, δ_{pn} is obtained by division of the equations

$$\delta_{pn} = \tan^{-1}\left(\frac{F_{cn}}{F_{sn}}\right) + k\pi, \quad k=0,1$$

i.e.,

$$\delta_{pn} = \begin{cases} \tan^{-1}(F_{cn}/F_{sn}), & F_{cn}/F_{sn} \geq 0 \\ \tan^{-1}(F_{cn}/F_{sn}) + \pi, & F_{cn}/F_{sn} < 0 \end{cases}, \quad n=1,2,3\dots$$

APPENDIX C MUSCLE ACTIVATION

C.1 Average muscle pattern

The main purpose of averaged muscle activation profiles is to detect prototypical activation patterns the cyclic walking motion. The averaging process requires a continuous, standardized repetition sequence [173]. My studies demonstrate that the processed data proportionally reflects the magnitude/ duration of muscle activation. Subsequently, the processed data are divided cycle by cycle at every repetitive events of the right heel contact [124]. Cycles are obtained with corresponding $\pm$ SD are presented. As a common knowledge, the absolute magnitude of EMG has less consistent for individuals. Therefore, researches focus on the shape of EMG profile, times of peak or onset/cessation of myoelectric activity [34]. The peak EMG values of the average muscle pattern were obtained for normalization [174], the processing procedures are demonstrated in Figure C.1.

Figure C.1: Average muscle pattern of healthy subject walking begins with right heel contact. Those events are picked up by the synchronized video recording.

C.2 Autocorrelation

Autocorrelation [175]: is commonly used in signal processing to test the repeating pattern of a signal, especially a periodic signal with large noise. Autocorrelation coefficient (A) is the sum of the products of time series x_i ($i=1,2,\dots,N$) multiplied by a delay replication of the time series (x_{i+m}), where the lag parameter m is the phase shift in number of samples:

$$A = \sum_{i=1}^{N-|m|} x_i x_{i+m} \quad (C.1)$$

The unbiased alternative $A_{unbiased}$ is produced by dividing the raw autocorrelation coefficient by the number of samples representing the overlapping part of the time series and the delay replication:

$$A_{unbiased} = \frac{1}{N-|m|} \sum_{i=1}^{N-|m|} x_i x_{i+m} \quad (C.2)$$

Figure C2: Central part of repetition index sequence from EMG signal during normal walking in gait cycle: (a) repetition index; (b) symmetry index.

Matlab Signal Processing Toolbox 6.0 (The MathWorks Inc., Natick, MA) was used in the computation of the autocorrelation coefficients. The algorithms also provide the normalization magnitude from zero to one as in Figure C.2. The time domain

normalization is studied of the gait cycle. The gait cycle has normalized to see the similarity of points from certain distance. Every dominant period of gait cycle, the similarity reach the local peak. The similarity of forward neighboring gait cycle is shown in the first local peak which is approach to 1.0. Both the neighboring and longer distance repetition is demonstrated. At the same time, the similarity of two body sides can also compared by this method to get the symmetry index.

C.3 Power spectrum analysis

A FFT (fast Fourier transform) is an efficient algorithm to computes DFT (discrete Fourier transform). The raw data of EMG in time domain are converted by DFT. Let $x_0, \dots, x_{N-1}$ be complex numbers. The DFT is defined by the formula

$$X_k = \sum_{n=0}^{N-1} x_n e^{-i2\pi k \frac{n}{N}} \quad k = 0, \dots, N-1 \quad (C.3)$$

The power spectrum comparison of healthy gait and pathological gait with spastic problem is shown in Figure C.3. The raw EMG data of the two is also provided. In the spastic muscle, more spikes than the healthy muscle and the power spectrum on high frequency are higher. This analysis will be used to evaluate the smoothness of muscle activation.

Figure C.3: EMG signal for health gait and pathologic walking: (a) healthy walking and pathologic walking with spastic muscle; (b) power spectrum comparison.