Study and implementation of a gait rehabilitation system with capability for mobility and gait pattern generation

Lim, Hup Boon

2012

Lim, H. B. (2012). Study and implementation of a gait rehabilitation system with capability for mobility and gait pattern generation. Doctoral thesis, Nanyang Technological University, Singapore.

<https://hdl.handle.net/10356/49512>

<https://doi.org/10.32657/10356/49512>

STUDY AND IMPLEMENTATION OF
A GAIT REHABILITATION SYSTEM
WITH CAPABILITY FOR MOBILITY AND
GAIT PATTERN GENERATION

LIM HUP BOON

SCHOOL OF MECHANICAL & AEROSPACE ENGINEERING
NANYANG TECHNOLOGICAL UNIVERSITY

A thesis submitted to the Nanyang Technological University
in partial fulfillment of the requirement for the degree of
Doctor of Philosophy

2012

Abstract

Human motor performance appears to be remarkably flexible and easy. Yet, the underlying neuronal operations are only vaguely understood, even for well-studied movements. However, the walking capabilities that seem to be an easy task for healthy person often become a major problem for those with disability, especially those who have lost the capability of walking after spinal cord injury or stroke.

A brief introduction of spinal cord injury and stroke is provided for an insight of the potential gait disability after the lesion. Therapist assisted body weight supported gait rehabilitation is a modern gait rehabilitation method developed for gait rehabilitation mainly for spinal cord injury and stroke patients. However, this method is labor intensive and requires at least three therapists for each patient. In view of this issue, robotic systems have been introduced for body weight supported gait rehabilitation to reduce the intensive labor involved. Such systems, e.g., Lokomat, KineAssist, WalkTrainer, and HapticWalker are reviewed in depth. The pro and cons of these systems are also summarized.

The review on the state of the art in robotic assisted gait rehabilitation revealed that there is a need for development of a new gait rehabilitation methodology. The proposed gait rehabilitation methodology suggests that gait rehabilitation features such as natural pelvic motion assistance, lower limb gait locomotion assistance, and body weight support shall be provided together in the context of over-ground walking. The methodology also proposes a subject specific gait pattern planning.

Robotic gait rehabilitation system, *NaTure-gaits*, is developed for the implementation of the proposed features in the gait rehabilitation methodology. The proposed system provides lower limb gait locomotion assistance, pelvic motion assistance, body weight support, and actual over-ground walking experience. These features are implemented through three main modules: robotic orthosis, mobile platform, and pelvic assistance mechanism, of the robotic system. Pelvic assistance is designed and developed for the provision of natural pelvic motion together with body weight support, which has not been done before.

A gait pattern generation concept (*GaitGen*) is proposed for the generation of natural gait pattern tailored for each individual. Most research groups obtained their gait patterns by referring to Clinical Gait Analysis (CGA) data. The CGA data is often obtained from one subject, and it is unclear how well it suits other individuals. In addition, CGA data is limited to only one walking speed as per the published data. In first stage of *GaitGen*, neural network is used to predict the gait parameters (walking speed and stride length) with inputs: age, gender, body height, and weight. In second stage, generalized regression neural networks (GRNNs) were designed to predict the individual-specific Fourier coefficient vectors from given gait parameters and lower limb anthropometric data.

Two control strategies are proposed for robotic assisted gait rehabilitation. The first control strategy, passive control strategy, is proposed with position control approach for the provision of overall robotic assisted gait rehabilitation for human subject. The second control strategy proposes an active assistance for robotic assisted gait rehabilitation. Active pelvic motion assistance with adaptive body weight support is suggested for human subject during gait rehabilitation. The active assistance control strategy is implemented using Virtual Model Control.

Trials are arranged for the demonstration of the proposed robotic system, function performance with controllable joints, and passive control strategy. The preparation and procedures of the trials are presented. One patient and four healthy subjects have been recruited for the trials. Electromyography signals have been recorded during the trials for analysis of muscle activity to evaluate the effectiveness of the robotic assisted gait rehabilitation. Clinical trials have been carried out for three subjects, two male SCI subjects and one female stroke subject, in Tan Tock Seng Hospital of Singapore. The trial subjects felt comfortable using the *NaTure-gaits* for gait rehabilitation.

This thesis provides the complete information on the successful development of new robotic assisted gait rehabilitation methodology. The methodology is accompanied with the development of robotic system, human gait locomotion study, development of control strategy, and clinical trials.

Acknowledgement

The author would like to express his gratitude to his supervisors, Professor Hoon Kay Hiang and Professor Low Kin Huat for their invaluable support, advice, and encouragement throughout the research. The progress in this research work was also made possible by their willingness to share their knowledge and give valuable advices.

The author would also like to give his heartfelt thanks to Mr. Lim Eng Cheng and Ms. Agnes Tan from the Robotics Research Center (RRC) for their unreserved support to the author, doing everything they could to help the author to collaborate with vendors/suppliers and settlement of all those tedious administrative works. The author is also very appreciative of the efforts by Mr. You Kim San and Ms. Toh Yen Mei for providing countless help and assistance in laboratory equipment and facilities.

The support and facilities from Motion Capture Laboratory and the Center for Human Factor and Ergonomics are acknowledged. The author would like to thank Professor Qu Xingda for his guidance in the completion of motion capture experiment. The help in arrangement for the use of motion capture laboratory by Mr. Edwin Lam is appreciated.

A special thank is given to Mr. Luu Trieu Phat, for his help in the completion of *GaitGen* and in the implementation of active control strategy. The support of Ms. Wang Ping in carrying out numerous experiments is greatly acknowledged. The stay in RRC is an enjoyable and good learning experience because of having good research mates like Ms. Huang Yunyun, Mr. Zhou Chun Lin, and Mr. Li Lei.

The author would like to thank Dr. Adela Tow, Mr. Wee Seng Kwee (physiotherapist), Ms. Chan Poh Leng (physiotherapist), Mr. Lim Pang Hung (physiotherapist) from the Department of Rehabilitation of Tan Tock Seng Hospital (Singapore) for their valuable feedback, suggestion, and opinion toward the completion of this research work. The valuable suggestions and comments of Professor Alison H McGregor from Imperial College of London are very much appreciated.

A special thanks to the funding agent, the National Medical Research Council (Research Grant, NMRC: 1051/2006) for the financial support provided for this project.

The author would like to thank the subjects, for their participation, time and consent in the human subject trials.

Author would like to thank all the teachers and professors who have taught the author before. They are the ones who have helped the author in gaining a strong foundation in Engineering.

Last but not least, I would like to thank my family, especially my wife, Zhou Li and my child Lim Yu Cheng. They have been with me every step of the way and I could not have done it without them

TABLE OF CONTENTS

Chapter 1	Introduction	1
1.1	Motivation of Research	4
1.2	Objectives and Scope	5
1.3	Outline of Thesis	6
Chapter 2	Review on Clinical and Engineering Aspects of Gait Rehabilitation....	9
2.1	Stroke, Spinal Cord Injured Patient and Associated Gait Deficits	11
2.1.1	Stroke	11
2.1.2	Spinal Cord Injury (SCI).....	12
2.1.3	Gait Deficit of Stroke and SCI Patients	14
2.2	Review on Clinical Aspect of Gait Rehabilitation.....	15
2.2.1	Three General Principles of Rehabilitation.....	17
2.2.2	Modern Model of Rehabilitation vs. Traditional Model of Rehabilitation	17
2.2.3	Therapist Assisted Body Weight Supported Treadmill Training (TABWS).....	18
2.3	State of the Art in Robotic Gait Rehabilitation.....	20
2.3.1	Lokomat	20
2.3.2	HapticWalker	23
2.3.3	WalkTrainer	25
2.3.4	KineAssist	26
2.4	Ability and Limitation of Existing Systems in Functional Walking Provision	28
2.4.1	Lower Limb Orthosis for Reciprocal Stepping.....	30
2.4.2	Review of Body Weight Support Approach	34
2.4.3	Pelvis Motion Provision with Pelvic Orthosis	36
2.4.4	Over-Ground or Treadmill Training?	41
2.5	Summary on Review of State of the Art	42

2.6	State of the Art from Clinical Perspective.....	45
2.6.1	Three Principles of Gait Rehabilitation and Robotic Gait Rehabilitation Systems.....	46
2.6.2	Realization of Eight Sensory Cues in Existing Robotic Gait Rehabilitation Systems.....	48
2.7	Reference Gait Pattern of the Developed Robotic Gait Rehabilitation System.....	49
2.8	Review of Control Strategies for Robotic Gait Rehabilitation	50
2.8.1	Principles of Patient Cooperative Control Strategy	51
2.8.2	State of the Art of Patient Cooperative Control Strategy.....	52
2.8.3	Summary of Control Strategies for Robotic Gait Rehabilitation	55
2.9	Gaps in the Current Research.....	56
2.9.1	Gait Pattern Generation for Robotic Assisted Gait Rehabilitation	56
2.9.2	Pelvic Motion Assistance and Body Weight Shifting.....	57
2.9.3	Over-Ground Walking for More Realistic Walking Experience.....	57
2.9.4	Control Strategy	57
2.10	Concluding Remarks and Overview of Research Approach.....	58
2.10.1	Research Approach	59
Chapter 3	Individual Specific Gait Pattern Planning for Robotic Assisted Gait Rehabilitation.....	63
3.1	Gait Pattern Generator – <i>GaitGen</i>	64
3.2	Prediction of Individual-Specific Basic Gait Parameters for Over-Ground Walking	65
3.2.1	Preliminary Study for Verification of Normalized Gait Parameters Equation	66
3.2.2	Methods.....	69
3.2.3	Results	73
3.2.4	Discussion	74
3.3	Individual-Specific Gait Pattern Prediction Model via Generalized Regression Neural Network	76
3.3.1	Methods.....	76

3.3.2	Results	85
3.3.3	Discussion	88
3.4	Concluding Remarks	89
Chapter 4	Accomplishment of Proposed Robotic Assisted Gait Rehabilitation Features	93
4.1	Proposed Essential Features for Robotic Assisted Gait Rehabilitation	93
4.1.1	Overview of Proposed Features and Accomplishment of the Features	94
4.2	Realization of Unrestricted Pelvic Motion Assistance and Body Weight Support	98
4.2.1	Pelvic Assistance Mechanism (PA) – Mechanism Design and Kinematic Model.....	99
4.2.2	Provision of Pelvic Motion	103
4.2.3	Body Weight Shifting and Body Weight Support	104
4.3	Realization of Over-Ground Mobility for Robotic Assisted Gait Rehabilitation	104
4.3.1	Kinematic Model of the Mobile Platform.....	105
4.4	Provision of Reciprocal Lower Limb Stepping with Robotic Orthosis (RO)	110
4.4.1	Robotic Orthosis (RO)	110
4.5	<i>NaTure-gaits</i> – Completed Prototype	114
4.5.1	Control Modules and Power Source	116
4.6	Concluding Remarks.....	117
4.6.1	Features of <i>NaTure-gaits</i>	117
Chapter 5	Control Strategy for Robotic Assisted Gait Rehabilitation.....	121
5.1	Patients Oriented Control Strategies	121
5.1.1	Categorization of Patients	122
5.1.2	Rehabilitation Strategies for Patients	124
5.2	Passive Control Strategy	127
5.2.1	Human Gait Locomotion Acquisition via Motion Capture Experiment	127

5.2.2	Lower Limb Locomotion Play Back with Robotic Orthosis.....	133
5.2.3	Pelvic Motion Play Back with Pelvic Assistance Mechanism and Mobile Platform	134
5.2.4	Improvement of Foot Contact during Walking via Pelvic and Lower Limb Motions Coordination.....	138
5.2.5	Implementation of Passive Control Strategy on <i>NaTure-gaits</i>	150
5.3	Active Pelvic Motion Assistance with Adaptive Body Weight Support for Robotic Gait Rehabilitation.....	152
5.3.1	Function of Body Weight Support	152
5.3.2	Concept of the Proposed Control Strategy	154
5.3.3	Design of Controller for Adaptive Body Weight Support with Natural Pelvic Motion Assistance	155
5.3.4	Experiments and Results	161
5.4	Concluding Remarks	167
5.4.1	Passive Control Strategy Implementation on <i>NaTure-gaits</i>	167
5.4.2	Active Pelvic Motion Assistance with Adaptive Body Weight Support	168
Chapter 6	Gait Locomotion Assistance with <i>NaTure-gaits</i> on Human Subject	169
6.1	Protocol of Trials.....	169
6.1.1	EMG Signal Acquisition	172
6.2	Healthy Subjects' Trials	173
6.2.1	Observation and Subjects' Survey	174
6.2.2	Muscle Activity during Robotic Assisted Gait Locomotion.....	174
6.3	Patient Trial	180
6.3.1	Pre-Trial	180
6.3.2	Observation and Discussion	182
6.4	Clinical Trials	187
6.5	Practical Implementation of <i>GaitGen</i> and Motions Coordination on <i>NaTure-gaits</i>	190
6.5.1	Subject Trials.....	192

6.6	Concluding Remarks	194
Chapter 7	Conclusion and Future Work	197
7.1	Summary of Contributions	197
7.1.1	Individual-Specific Gait Pattern Generation	197
7.1.2	Realization of Robotic System – <i>NaTure-gaits</i>	198
7.1.3	Control Strategy	199
7.1.4	Human Subject Trials	200
7.2	Suggestion for Future Works	200
Appendix - A	Information on Stroke and Spinal Cord Injury	203
Appendix - B	Basic Information of Human Walking	215
Appendix - C	Detailed Mathematics for Computation of Lower Limb Joint Angles	218
Appendix - D	Clinical Trials – Protocols and Procedures	224
Appendix - E	Clinical Trial Survey Form	232
Appendix - F	Experimental Data of Gait Parameters for the Fifty Subjects	238
Appendix - G	Predicted Basic Gait Parameters using Trained GRNNs	246
Appendix - H	Adjusted Lower Limb Joints Angle	257
Appendix - I	List of Relevant Publications Published	263

LIST OF FIGURES

Figure 1-1: Four major fields of study of robotic assisted gait rehabilitation	3
Figure 2-1: Review of clinical and engineering aspects for gait rehabilitation in this chapter	10
Figure 2-2: Gait disability of SCI patient	13
Figure 2-3: Stroke related gait deficits.....	14
Figure 2-4: Gait related issues after spinal cord injury	15
Figure 2-5: Patient having over-ground ambulation training with therapist assistance...	16
Figure 2-6: Therapist assisted body weight supported gait rehabilitation [39]	19
Figure 2-7: (a) Lokomat [51] (b) Schematic of Lokolift Mechanical Setup [50]	22
Figure 2-8: (a) HapticWalker [64] (b) Foot binding on the footplate [65]	24
Figure 2-9: (a) Concept of WalkTrainer [68] (b) WalkTrainer tested on human [69]	25
Figure 2-10: KineAssist in operation [12]	26
Figure 2-11: DoFs in (a) Trunk control mechanism; (b) Pelvis control mechanism [71]	28
Figure 2-12: Three elements required for gait training.....	29
Figure 2-13: Overview of gait rehabilitation features realization with module of the developed robotic gait rehabilitation	30
Figure 2-14: Robotic orthosis of treadmill based systems, from left to right, ALEX [73], Lokomat [74], LOPES [75], and POGO [61]	31
Figure 2-15: From left to right, robotic orthosis of the WalkTrainer (Over-ground based system) [10], and footplate of HapticWalker (Footplate based system).....	31
Figure 2-16: Body weight support module, from left to right, Lokolift [82], POGO (produced by Robomedica Inc., POGO is not shown in the photo) [63], KineAssist [12], and WalkTrainer [69]	34
Figure 2-17: (a) Movement definition of the pelvis (b) Coordination assignment for left side (balloon L) and right side (balloon R) of the pelvis	38
Figure 2-18: (a) PAM prototype (b) Schematic illustration of PAM [89].....	38
Figure 2-19: (a) Close up of pelvic orthosis on WalkTrainer [68] (b) CAD model of the pelvic orthosis [83] (the arrows indicate motors placement and actuated movements).....	39

Figure 2-20: (a) Pelvic movement provided by LOPES [60] (b) Parallelogram of Lokomat [90] (c) Pelvic mechanism of KineAssist [12]	40
Figure 2-21: Overview of features found on treadmill based robotic gait rehabilitation systems.....	43
Figure 2-22: Overview of features found on footplate based robotic gait rehabilitation systems.....	43
Figure 2-23: Overview of features found on robotic gait rehabilitation systems provides gait rehabilitation in the context of over-ground walking	44
Figure 2-24: Comparisons of therapist and robotic assisted body weight supported gait rehabilitation for three principles of gait rehabilitation	47
Figure 2-25: Overview of research approach	59
Figure 2-26: Realization of the research approach of this research work	62
Figure 3-1: Overview of proposed model, <i>GaitGen</i> , for gait pattern planning in robotic assisted gait rehabilitation.....	65
Figure 3-2: Gait parameters equation verification for 15 male subjects	68
Figure 3-3: Gait parameters equation verification for 15 female subjects	69
Figure 3-4: Overall concept of gait pattern generation in robotic assisted gait rehabilitation	70
Figure 3-5: Effect of smoothing factor represented by MSE of predicted outcome for normal walking speed	73
Figure 3-6: Individual-specific gait pattern prediction model for robotic assisted gait rehabilitation	77
Figure 3-7: 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)	78
Figure 3-8: Lower limb joint angles during walking (at slow speed), data obtained via motion capture experiment, four walks (indicates by W1, W2, W3, and W4) of same subject.....	81
Figure 3-9: Trajectories of pelvic motion during normal gait locomotion, obtained via motion captured experiment. Motion of pelvis in the: (a) <i>x-axis</i> (Anteroposterior	

direction); (b) <i>y-axis</i> (Mediolateral direction); (c) <i>z-axis</i> (Superior-inferior direction). (Solid line – right side, dashed line – left side)	82
Figure 3-10: Integral of power of lower limb joints angle waveforms	83
Figure 3-11: Actual lower limb joints angle waveforms and reconstructed waveforms with the Fourier coefficient vectors associated with 700 walks	86
Figure 3-12: Actual lower limb joints angle waveforms and predicted waveforms from gait pattern prediction model associated with 100 walks	87
Figure 4-1: Conceptual idea of the gait rehabilitation methodology and its realization (* pelvic motion and body weight shifting)	95
Figure 4-2: Main modules on <i>NaTure-gaits</i> and corresponding fourteen actuated motions.....	96
Figure 4-3: Planes assignment on human body [131].....	97
Figure 4-4: Local coordinates for each module of <i>NaTure-gaits</i>	98
Figure 4-5: PA module (a) Robotic arm providing pelvic motion in the sagittal plane (b) Lateral shift mechanism of the right side (Arrows indicates the actuated motion)	100
Figure 4-6: Configuration of robotic arms of pelvic assistance mechanism	100
Figure 4-7: Kinematic representation of PA sub-mechanism.....	101
Figure 4-8: Detailed design of mechanism for lateral pelvic movement actuated by actuator 3, located at the right side of pelvis (Arrow indicates the actuated motion)	102
Figure 4-9: Effective and designed workspace for sub-mechanism of PA mechanism, with the end effector set as reference coordinates ($x = 0$, $z = 0$ at end effector)....	102
Figure 4-10: Illustration of pelvic motion provision by the pelvic assistance mechanism	103
Figure 4-11: CAD model of mobile platform	105
Figure 4-12: Denotations for kinematic modeling of mobile platform.....	106
Figure 4-13: Time function description for key variable of mobile platform (The coordinate system is assigned as per Figure 4-3).....	107
Figure 4-14: CAD model of robotic orthosis (a) Hip module, (b) Knee module, (c) Ankle module.....	112
Figure 4-15: Denotation of parameters for ankle module kinematics	113

Figure 4-16: Conceptual design and features of robotic orthosis for <i>NaTUre-gaits</i>	114
Figure 4-17: Prototype of <i>NaTUre-gaits</i>	115
Figure 4-18: Attachment of <i>NaTUre-gaits</i> to subject.....	116
Figure 4-19: Control system of pelvic assistance mechanism, mobile platform, and robotic orthosis	117
Figure 5-1: Patient categorization algorithm	124
Figure 5-2: Proposed possible rehabilitation strategies for the patients grouped according to this research work (Solid line indicates passive assistance and dashed line indicates active assistance)	126
Figure 5-3: Several gait cycles constructed by duplication of same gait cycle ('A' marked the interconnection between two cycles, which does not connect smoothly)	129
Figure 5-4: Post processed lower limb joint angles constructed for the right side (three gait cycles), original data obtained via motion capture experiment	130
Figure 5-5: Motion assistance provided by PA mechanism. The relationship between motion assistance provided by PA mechanism and computed HJC trajectories are highlighted	131
Figure 5-6: Post processed pelvic motion constructed for the right side (three gait cycles), original data obtained via motion capture experiment	132
Figure 5-7: Reference trajectories constructed for both lower limbs (Solid – right lower limb, dashed – left lower limb).....	133
Figure 5-8: Overview of conversion of constructed reference trajectories to input for the controller of robotic orthosis	134
Figure 5-9: Conversion from pelvic motion to motion for mobile platform and pelvic assistance mechanism	136
Figure 5-10: Final constructed motion for PA mechanism (a) Motion in the direction of x -axis; (b) Motion in the direction of y -axis; (c) Motion in the direction of z -axis (Solid line – right side, dashed line – left side)	137
Figure 5-11: Trajectories of pelvic motion input to PA sub-mechanism (motion in the x and z axes, which is the sagittal plane)	137

Figure 5-12: Overview of conversion of final constructed pelvic motion to input for the controller of robotic orthosis.....	138
Figure 5-13: Ideal ankle reference angle during human walking	140
Figure 5-14: Heel trajectories in the direction of <i>Z-axis</i> , computed from lower limb and pelvic motion obtained by motion capture experiment.....	141
Figure 5-15: Ankle reference angle computed from data of motion capture experiment	142
Figure 5-16: Identification of foot flat and heel off using ankle reference angle: Balloon “1” marked the foot flat event (The angle changing from +ve to 0); Balloon “2” marked the heel off event (The angle changing from 0 to –ve).....	143
Figure 5-17: Denotations for variables of PA and RO	145
Figure 5-18: Ankle reference angle: before and after adjustment	147
Figure 5-19: Heel position in the <i>Z-axis</i> direction: before and after adjustment	147
Figure 5-20: Lower limb joints angle: Before and after adjustment.....	148
Figure 5-21: Adjustment of walking motions based on the data of motion capture experiment: before and after (Red triangle representing the foot, graph unit is mm)	149
Figure 5-22: Scenario of <i>NaTUre-gaits</i> gait locomotion play back, (a) – (c): Heel strike to foot flat of the right limb, (d): Heel strike of contra lateral limb and heel off of right limb, (e) Foot flat of control lateral limb, (f) Heel strike of the right limb and completion of one gait cycle	151
Figure 5-23: Illustration of dynamic equilibrium of walking (a) Without external force component, (b) With external force components, BWS (Forces location indicate in this figure does not necessary denotes the actual location of force acting)	152
Figure 5-24: Dynamic equilibrium during walking (a) without external support (b) with body weight support.....	153
Figure 5-25: Denotations use for the design of controller (dotted line is the reference trajectory for the pelvic motion; dash line is the maximum allowable deviation for pelvic motion; figure is for illustration purposes only and do not necessary denote the actual pelvic motion).....	156
Figure 5-26: Control algorithm of the proposed control strategy	157

Figure 5-27: Bipolar sigmoid function	159
Figure 5-28: Stiffness value of virtual spring as a function of slope of the sigmoid function and allowable deviation for pelvic position.....	160
Figure 5-29: Control algorithm of the proposed control strategy.....	161
Figure 5-30: Linear moving platform built for the controller verification	162
Figure 5-31: Hardware and software configuration of the controller.....	162
Figure 5-32: Control block layout in Simulink for the controller design	163
Figure 5-33: Response of the controller for scenario (I-a) (solid line-actual trajectory, dotted line-desired trajectory, and dashed line-maximum allowed deviation).....	165
Figure 5-34: Response of the controller for scenario (I-b) (solid line – actual trajectory, dotted line – desired trajectory, dashed line – maximum allowed deviation)	165
Figure 5-35: Slope δ and the stiffness k of the virtual spring in scenario (I-a) (solid line – stiffness (unit, N/mm), dotted line – slope (no unit))	166
Figure 5-36: Slope δ and the stiffness k of the virtual spring in scenario (I-b) (solid line – stiffness (unit, N/mm), dotted line – slope (no unit))	167
Figure 5-37: Response of the controller for scenario (II) (solid line – actual trajectory, dotted line – desired trajectory, dashed line – interaction force (unit, mN)).....	167
Figure 6-1: Snap shot of the set up for trials	170
Figure 6-2: Overview of prototype trial protocol	171
Figure 6-3: Anatomical position of the selected muscle and electrode placement sites	172
Figure 6-4: Average EMG signals of Vastus Medialis (VM) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over- ground).....	175
Figure 6-5: Average EMG signals of Tibialis Anterior (TA) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over- ground).....	176
Figure 6-6: Average EMG signals of Gluteus Maximus (GM) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground	

contact (off ground), and gait locomotion assistance with ground contact (over-ground)	177
Figure 6-7: Average EMG signals of Biceps Femoris (BF) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)	178
Figure 6-8: Average EMG signals of Gastrocnemius Lateral (GL) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (of ground), and gait locomotion assistance with ground contact (over-ground)	179
Figure 6-9: Over-ground walking training of the patient at pre-trial (a) The patient walk with the help of two trainers (b) The right leg of the patient is moved forward with the help of one trainer (the trainer is working at an ergonomically unfavorable posture)	181
Figure 6-10: Suspending the patient into <i>NaTUre-gaits</i>	182
Figure 6-11: Gait locomotion assistance for the patient	183
Figure 6-12: Average EMG signals of Vastus Medialis (VM) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)	185
Figure 6-13: Average EMG signals of Tibialis Anterior (TA) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)	185
Figure 6-14: Average EMG signals of Gluteus Maximus (GM) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)	186
Figure 6-15: Average EMG signals of Biceps Femoris (BF) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)	186
Figure 6-16: Average EMG signals of Gastrocnemius Lateral (GL) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)	187

Figure 6-17: Pre-setting for system trials: (a) Harness setup for one of the SCI subjects; (b) and (c) <i>NaTure-gaits</i> strapping for the same SCI subject	189
Figure 6-18: Snapshot of clinical trials of: (a) SCI subject – I, (b) SCI subject – II, and (c) Stroke subject	190
Figure 6-19: Rating of two patterns by the subjects based on their trial experience	193
Figure 6-20: Snapshots of video taken during the trial.....	193

List of Tables

Table 2-1: Features of Lower Limb Orthosis in Robotic Gait Rehabilitation Systems...	33
Table 2-2: Features of Body Weight Support Module in Robotic Gait Rehabilitation Systems	36
Table 2-3: Pelvic Movement Provision by Different Robotic Gait Rehabilitation Systems	40
Table 3-1: Information of Subjects and Numbers of Walks	71
Table 3-2: Smoothing Factor Selected for Corresponding GRNNs	73
Table 3-3: Mean Square Error of Predicted Outcome	73
Table 3-4: Predicted Outcomes with Highest MSE, Least MSE, and Average MSE for the Trained GRNNs	74
Table 3-5: Accuracy of Trained GRNNs in Predicting the Basic Gait Parameters	75
Table 3-6: Numbering and Position of the Motion Capture Markers	79
Table 3-7: Denotations of Anthropometric Data Recorded	80
Table 3-8: Division of Experimental Data for GRNN.....	85
Table 3-9: Mean (STD) of Assessment Parameters from 700 Walks.....	86
Table 3-10: Mean (STD) of Assessment Parameters from 100 Walks in Test Set.....	86
Table 4-1: Design Requirement based on Anthropometric Data of Targeted User Group	95
Table 4-2: The Designed Range of Motion (ROM) for Robotic Orthosis (Presented with Permissible ROM of Human and ROM of Human during Walking)	111
Table 4-3: Pelvic Movement Assistance of <i>NaTure-gaits</i> Compared to Pelvic Movement Assistance Available on other Robotic Gait Rehabilitation Systems	118
Table 4-4: Body Weight Support Feature of <i>NaTure-gaits</i> Compared to State of the Art Gait Rehabilitation System	119
Table 4-5: Features of Robotic Orthosis of <i>NaTure-gaits</i> Compared to Lower Limb Orthosis of State of the Art Gait Rehabilitation System	119
Table 6-1: Information of Subjects Participated in the Trials.....	173
Table 6-2: Detailed Information of SCI/Stroke Subjects.....	188
Table 6-3: Basic Information and Anthropometry Parameters of the Trial Subjects	190

Table 6-4: Selected Basic Gait Parameters for the Trial Subjects Based on the Predicted Values	191
---	-----

List of Abbreviations

ASIS	Anterior Superior Iliac Spine
BF	Biceps Femoris
BWS	Body Weight Supported/Body Weight Support
cBWS	Cable-harness Body Weight Support
CNS	Central Nervous System
COM	Center of Mass
CPS	Central Pain Syndrome
DFT	Discrete Fourier Transform
DoF	Degree of Freedom
EMG	Electromyography
FFT	Fast Fourier Transform
FT	Fourier Transform
GC	Gait Cycle
GL	Gastrocnemius Lateral
GM	Gluteus Maximus
GRF	Ground Reaction Force
GRNN	Generalized Regression Neural Network
GUI	Graphical User Interface
HJC	Hip Joint Center
MP	Mobile Platform
PA	Pelvic Assistance (Mechanism)
RMS	Root Mean Square
RO	Robotic Orthosis
ROM	Range of Motion
sBWS	Structural Body Weight Support
SCI	Spinal Cord Injury
TA	Tibialis Anterior
TABWS	Therapist Assisted Body Weight Supported Treadmill Training
VM	Vastus Medialis
VMC	Virtual Model Control

Chapter 1

INTRODUCTION

Walking, an important task for most human, achieved through the repetition of a well-defined movement. To achieve a bipedal walking, it requires the integration of the central nervous system with many peripheral sensory systems to control the many muscles acting on a skeletal system with many degrees of freedom.

Spinal cord injury (SCI) and stroke are the leading cause of permanent disability around the world. In United States, there are approximately 795,000 new stroke cases [1], and an estimated 12,000 new spinal cord injury cases [2] occurring each year. Loss of walking ability is a debilitating outcome in post-stroke and spinal cord injury, with more than 50% of the post-stroke patients demonstrating persistent walking deficits, and more than 90% of the SCI patients lose their sensory and motor control of the lower limbs.

Even after a conventional rehabilitation, some patients are still unable to walk well, or walk safely. Three required components of locomotion, maintaining equilibrium during propulsive movement, body weight shifting and reciprocal stepping (foot movement in the direction of progression) [3, 4] are not able to be trained by conventional rehabilitation simultaneously. The inability of the patients to support body weight while moving forward, disallows proper weight shifting and stepping during rehabilitation session, even with the help of therapist.

Body weight supported (BWS) gait rehabilitation training strategy was introduced two decades ago [5]. It is a strategy proposed based on the recovery model of rehabilitation. The strategy proposes offloading of patient using a BWS apparatus, whereby a certain percentage of body weight is offloaded during rehabilitation training. Barbeau et al. show that with partial body weight supported, the patients are more capable of carry out stepping motion [5]. Being suspended in a harness provides a safe and effective environment for gait rehabilitation. The harness substitutes the equilibrium reflexes deficit, and the body weight support compensated for the paresis and muscle weakness

[6]. Although a similar intervention might be carried out by therapist, the amount of support force can be provided are limited. In particular, those patients with much severe condition often required greater supporting force that exceeds the physical ability of the therapist.

Conventionally, therapist assisted BWS gait rehabilitation is carried out by suspending patient over a treadmill using cable-harness apparatus. The cable-harness system offloads a certain amount of the patient's body weight (hence the training method named BWS). The offloading of the patient's body weight facilitates walking training. With the help of several therapists, the patient's pelvic and lower limbs are guided to complete the gait locomotion. Over the last 10 years, studies have shown that SCI and stroke patients who receive body weight supported treadmill training demonstrated improved EMG activation patterns, more natural walking characteristics, increment in weight bearing on their legs, and functional improvement in walking ability. The review of the evidences can be found in [7, 8].

However, for the therapist assisted BWS gait rehabilitation approach, the labor costs are high as up to three therapists can be required to assist each patient. Thus the popularity of this approach has dwindled due to initial high cost outlay for the equipment, following by high labor not to mention staff training in the use of the system. Clearly there is a need to improve the approach, both with respect to the design and use interfaces, particularly as with technology such system should reduce therapist input as opposed to increasing it.

Some examples, such as Lokomat [9], WalkTrainer [10], HapticWalker [11], and KineAssist [12] are developed to automate the BWS gait rehabilitation training process. These robotic gait rehabilitation systems provide features such as lower limb assistance, pelvic movement assistance, and body weight support. However, the features provided by these robotic systems are not complete. Majority of these systems are treadmill based as opposed to actual over-ground walking. An essential feature, pelvic movement assistance, is lacking in majority of these systems, or constrained by the body weight support apparatus.

Robotic assisted gait rehabilitation covers four major fields of study: (1) design of robotic system, (2) generation of gait pattern (the motion input for the developed robotic

system), (3) control strategy, and (4) bio-feedback and clinical assessment for rehabilitation purposes, as shown in Figure 1-1. The most common field is the design of robotic system to achieve the provision of desired gait locomotion assistance, which majority of the research groups are focused on [9-12]. Some minor differences exists among the current robotic systems, but majority of these systems consist of robotic orthosis for gait locomotion assistance, pelvic mechanism for pelvic motion assistance, apparatus for body weight support, and treadmill as the medium for foot progression. However, the assistance of pelvic motion is restricted by the body weight support apparatus used in those robotic systems.

Figure 1-1: Four major fields of study of robotic assisted gait rehabilitation

Matching template approach is used for gait pattern planning in majority of current robotic gait rehabilitation systems. A template matching the anatomical parameters of the targeted subject will be used as the gait pattern to predefine the motion of the robotic system during rehabilitation. The research team of WalkTrainer proposed an idea for the generation of pelvic motion based on walking velocity. The work provides a general solution for the pelvic motion generation (based on an average pelvic motion collected from several subjects), which do not create a motion specifically for a targeted subject. In this research work, a new methodology for gait pattern generation is proposed. The developed methodology enables the gait pattern to be generated specifically for the targeted subject.

Robotic assisted gait rehabilitation commonly provides passive gait rehabilitation, which is also known as position control strategy. The robotic system provides gait locomotion assistance regardless of the patient effort. Enabling active participation with control strategy is the relatively new research field in robotic assisted gait rehabilitation. Robotic strategies are developed for the implementation of a unique feature of robotic assisted gait rehabilitation. Patient's participation level (or patient effort) determination is one of the key topic in this field. The determination of patient's participation can be obtained from bio-feedback. Two control strategies have been proposed in this research work. Passive control strategy is proposed for overall control of the robotic gait rehabilitation system through position control. Another control strategy is proposed for active pelvic motion assistance and adaptive body weight support.

Bio-feedback serves three main purposes in robotic assisted gait rehabilitation, (1) feedback for control purposes and model validation of controller (part of control strategy implementation), (2) feedback for visual display of rehabilitation status to patient and clinical practitioner, and (3) as a source for offline assessment of the patient's condition and therapeutic outcome. This research work does not focus in this area.

1.1 Motivation of Research

Therapist assisted body weight support gait rehabilitation has clear limitation due to the requirement of high labor cost. On the other hand, robotic assisted body weight support gait rehabilitation reduced the number of therapists required for training each patient, but the performance of these robotic systems does not ideally match the original methodology of gait locomotion training. Majority of these systems, train the patient on treadmill. Walking on treadmill is hypothesized different than over-ground walking.

Several essential features, such as unrestricted pelvic motion, over-ground walking experience during rehabilitation, body weight shifting, are not found on those developed robotic gait rehabilitation systems. Pelvic movement, which contributes to body weight shifting, is not commonly found on robotic gait rehabilitation systems. Robotic gait rehabilitation system, such as the WalkTrainer, provides pelvic movement assistance but the motion is constrained by the body weight support apparatus.

This research work is motivated to develop a robotic system for the provision of unrestricted pelvic motion and allows body weight shifting, actuated assistance to the lower limb joints (hip, knee, and ankle), body weight support, and functional over-ground walking. With the provision of pelvic motion, the body weight shifting can be promoted during gait rehabilitation. The proposed methodology provides an alternative approach for gait rehabilitation, in contrast to existing BWS approach, and provides a more specific training task (actual walking on ground), better and more complete gait motion guidance, and minimize the alteration of gait pattern [13].

Template based matching for the gait pattern used in robotic assisted gait rehabilitation is not desired. The number of template available could be limited and it is not possible to match every individual perfectly. In view of this limitation, this research work is motivated to develop individual-specific prediction model for gait pattern generation. The gait pattern is specifically generated for the targeted subject, based on anatomical parameters of the subject, without the need of matching template.

In order to provide unique body weight support feature together with pelvic motion assistance, this research work is motivated to explore a control strategy, which make the idea possible on the developed gait rehabilitation system in this work.

1.2 Objectives and Scope

The objective of this work is to propose a gait rehabilitation methodology incorporating gait pattern planning, pelvic motion assistance, and mobility training in the context of over-ground walking and investigate it with a robotic system. Towards this end, this thesis presents the work in three major areas in robotic assisted gait rehabilitation, aiming to provide a better clinical experience for patient in robotic assisted gait rehabilitation.

The scope for this research work covers the three areas:

Study of Human Gait Motion

- Acquisition of the reference trajectory of gait pattern for robotic system through human gait motion study.
- Develop a prediction model of individual-specific gait pattern generation, for the generation of gait pattern without the need of matching template.

- Predict natural gait parameters and generate waveforms of gait pattern based on selected anatomical parameters of the targeted subject.

Robotic Walking Rehabilitation System

- Design, build and subject trial of robotic system for the implementation of the proposed gait rehabilitation methodology.
- Design and incorporate the necessary modules on robotic system for the provision of the key features of lower limb assistance, pelvic movement assistance, and body weight support needed for gait rehabilitation, in the context of over-ground walking.
- Design of a module for the robotic system for the provision of unrestricted pelvic movement assistance with body weight support capability.

Gait Locomotion Control Schemes for Rehabilitation

- Implement passive control strategy for the provision of robotic assisted gait rehabilitation through the proposed robotic system.
- Propose and study of control strategy for adaptive body weight support and active pelvic motion assistance.

1.3 Outline of Thesis

The thesis consists of seven chapters and four appendices. This chapter provides the background of, and motivation for, the research in body weight support gait rehabilitation. The outline of this thesis is structured as follows:

Chapter 2: Review on Clinical and Engineering Aspects of Gait Rehabilitation

provides the literature review on the clinical and engineering aspects in gait rehabilitation. State of the art in robotic gait rehabilitation systems is reviewed and pro and cons of these systems are summarized. Concluding remarks are provided at the end of this chapter on the gap in the current research. This chapter highlights the basis of this research work leading to the proposed research work and methodology for robotic assisted gait rehabilitation.

Chapter 3: Individual Specific Gait Pattern Planning for Robotic Assisted Gait Rehabilitation introduces a model of gait pattern generation for individuals. The gait pattern generation approach replaces the need of a matching template. The developed model, *GaitGen*, is able to predict gait parameters and the waveforms of gait trajectories specifically for individuals based on their anatomical parameter. The gait pattern generated by this methodology will be compared to gait pattern obtained by motion capture experiment.

Chapter 4: Accomplishment of Proposed Robotic Assisted Gait Rehabilitation Features has the details on the engineering realization for the features of robotic assisted gait rehabilitation proposed in this research work. Based on the findings of literature reviewed and the suggestion provided by doctors and therapists from Tan Tock Seng hospital of Singapore, several features for robotic assisted gait rehabilitation are proposed. The proposed features are achieved with the introduction of the developed robotic gait rehabilitation system, *NaTure-gaits*. The design, key features, kinematics, and controller hardware of the *NaTure-gaits* are described. The robotic gait rehabilitation is tested with human subject together with the passive control strategy presented in Chapter 5. Results of the test are presented in Chapter 6.

Chapter 5: Control Strategy for Robotic Assisted Gait Rehabilitation presents the two control strategies for robotic assisted gait rehabilitation. Passive control strategy is discussed as this is the most basic control strategy for a robotic gait rehabilitation system. This strategy is explained and eventually implemented on the developed robotic gait rehabilitation system for the subject trial presented in Chapter 6. Active body weight shifting and active pelvic motion assistance is an active assistance control strategy developed for the system.

Chapter 6: Gait Locomotion Assistance with *NaTure-gaits* on Human Subject gives an insight for the trials, which carried out to verify and validate the work done in this research. The developed robotic gait rehabilitation system is tested for the gait rehabilitation features it is designed for, together with the passive control strategy discussed in

Chapter 5. Electromyography signals are recorded during the trials to study the effect of robotic assisted locomotion on the subjects.

Chapter 7: Conclusion and Future Work synthesizes the results and contributions of this work. Suggestions for future work are also discussed.

Chapter 2

REVIEW ON CLINICAL AND ENGINEERING ASPECTS OF GAIT REHABILITATION

Rehabilitation with robotic system has a long and venerable history, whereby the first research dated year 1997 [14]. The earlier research focused on the applications to assist the disabled. In the early eighties [15], robotic system is introduced to supplement and improve the clinical rehabilitation settings. This chapter provides introduction on spinal cord injury and stroke, and literature review on the clinical and engineering aspects in gait rehabilitation. The conclusion and gap in the current research are identified from the literature survey. This information is combined with the suggestion and opinion gathered from doctors and therapists of Tan Tock Seng hospital in Singapore, formed the basis and the need of carrying out the research work presented in this thesis. Figure 2-1 provides an overview of the key elements reviewed in this chapter.

Stroke and spinal cord injury are the two leading causes of the gait disability. The basic information about stroke and spinal cord injury is provided at the beginning of this chapter. Subsequently the gait deficit related to the stroke and spinal cord injury patients is discussed. These patients require gait rehabilitation in order to recover the walking ability.

The methods of gait rehabilitation for these patients are introduced. The review covers the clinical aspects of gait rehabilitation. Three general principles of rehabilitation are introduced. Discussion is made on the advantage of modern model of rehabilitation over the traditional model of rehabilitation. Therapist assisted body weight supported gait rehabilitation is reviewed and the advantages and disadvantages of this method are discussed.

Robotic assisted gait rehabilitation was introduced to address the disadvantage of therapist assisted body weight support gait rehabilitation, mainly to reduce the laborious demand placed on therapist. This chapter reviews the state of the art in robotic assisted gait rehabilitation. Lokomat, HapticWalker, WalkTrainer, and KineAssist are robotic

gait systems that have been reviewed as a complete system in this chapter. Subsequently, essential features (lower limb motion assistance, body weight support approach, pelvic motion assistance, and over-ground/treadmill walking) of developed robotic gait rehabilitation systems are reviewed and comparisons are made for these features. Gait pattern generation and control strategies of these robotic gait rehabilitation systems are reviewed subsequently.

Figure 2-1: Review of clinical and engineering aspects for gait rehabilitation in this chapter

Towards the end of chapter, the gap of current research for gait rehabilitation is presented to conclude the literature review on both clinical and engineering aspects for gait rehabilitation. Overview of research approach is then proposed to fill up the gap of current research.

2.1 Stroke, Spinal Cord Injured Patient and Associated Gait Deficits

This research project has identified two patient groups, which are Spinal Cord Injured (SCI) and post-stroke patients. These two groups of patients accounted for the major causes of gait disability around the world. Loss of walking ability is a debilitating outcome in post-stroke, with more than 50% of the post-stroke patients demonstrating persistent walking deficits. On the other hand, the condition of SCI patients often results in a partial loss of both sensory and motor control of the lower limbs, depending on the extent of the injury level on the spinal cord. Stroke and SCI patients often experience poor walking capabilities, forcing them to rely on wheelchairs or other walking assisting devices to achieve functional mobility.

Although the gait rehabilitation methodology proposed might be applicable to other group of patient, the discussion of this research project will mainly focus on gait rehabilitation methodology study for stroke and spinal cord injury. This section provides a brief introduction to stroke and spinal cord injury and the gait deficit after the illness/lesions. More details on stroke and spinal cord injury can be found in Appendix - A. The brief introduction equips reader with the background of stroke and spinal cord injury, which facilitates the understanding of the concept of this research project and the developed gait rehabilitation methodology.

2.1.1 Stroke

Stroke or ‘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. This leads to part of the brain not getting the blood/oxygen it needs and the brain cells will start to die. Eventually that part of brain will become 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.

Apart from classified stroke effects based on the location of the stroke occurring in the brain, there are some common ways in which stroke can affect a person. The effects include physical effects, communication effects and behavioral effects. The physical effects are the important aspect to be examined in gait rehabilitation and will be further discussed below.

The physical effects brought on by stroke include spasticity, balancing issues, toe curl, central pain syndrome (CPS), etc. *Spasticity* is one of the effects of stroke that causes muscles to involuntarily contract when the patients try to move their limbs, creating stiffness and tightness. The tendons and soft tissues surrounding the muscle become tight when it cannot complete its full range of motion. In the long run, the muscle freeze permanently into an abnormal and often, painful position. For an arm that has spasticity effect, the fist will always close tight, with the elbow bent and the arm pressed against the chest. Stiff knee, pointed foot and curling of the toes will be found on a patient with leg spasticity.

Stroke survivors often face problem in *balancing*. Strokes are likely to affect the central system which causes the stroke patient to experience dizziness and imbalance, and increasing the chances of falling. *Central pain syndrome (CPS)* is not common among stroke patients. The pain is an extremely persistent, intractable type of pain which can be quite debilitating and depressing to the patient.

2.1.2 Spinal Cord Injury (SCI)

Human motor signal flows from brain to the peripheral nerves to activate the muscle in order to complete the intended movement. However, this operation will be disrupted if the signal transmitting pathway, i.e. the spinal cord is injured. Most spinal cord 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 Asso-

ciation (ASIA) [16] 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.

Figure 2-2: Gait disability of SCI patient

The most obvious impact of SCI is the loss of control over the physical functions. A person's loss of control over some or all his/her limb and trunk depends on the level and completeness of injury. This dramatically changes the living pattern of a person, or worse, the person might even lose the control of bowel and bladder movement and face an alteration in sexual function.

Statistics [17] shows that majority of spinal cord injured patients (77.98%) are injured at the level of C5 and below. The injury at C5 level or below causes different level of gait disability as shown in Figure 2-2. It is noticeable that high spinal cord injury level caused greater gait disability compare to low spinal cord injury level.

2.1.3 Gait Deficit of Stroke and SCI Patients

Often, the disability caused by stroke and SCI are related to the type of stroke, and level of SCI. Most of the effects discussed are basic function (breathing, swallowing, and etc.), or single muscle (elbow flexion, foot drop, and etc.) related. Walking involves the coordination of several joints and muscle activation of the lower limb, as well as sensory feedback of several organs. Gait deficit usually depends on the severity of the muscle and sensory disability, and the lower limb muscle and sensory affected by the illness/lesions. The following general gait deficit comments associated with the stroke and SCI are concluded based on the observation at local hospital, literature review, and discussion with experienced rehabilitation therapists.

Figure 2-3: Stroke related gait deficits

In general, stroke patient usually suffers disability on one side of the body, either left or right side. The unaffected side's function is no different from a healthy person, but normal walking is not possible since walking requires coordinated movements of two legs. Some gait deficit after stroke might be related to disability of a specific muscle.

For example, disability of ankle dorsiflexors leads to drop foot syndrome. The gait deficits related to stroke are summarized in Figure 2-3.

Gait disabilities of complete SCI usually are symmetrical for both lower limbs. Muscle and sensory of lower limb affected by SCI depends on the injury level of spinal cord. The muscle and sensory function below the injury level will be totally lost after a complete SCI. Most of the SCI occurred at C-level, which will affect the muscle of lower limb and result in the loss of walking ability. Incomplete SCI may exhibit a more complex gait deficit situation, depends on how much the spinal cord is injured. The lost of function would be much different among SCI patients even with the same spinal cord injury level. The general information of the gait related issues due to SCI is summarized in Figure 2-4.

Figure 2-4: Gait related issues after spinal cord injury

2.2 Review on Clinical Aspect of Gait Rehabilitation

When patients are asked for their priorities of rehabilitation, patients name the restoration of a functional gait as one of their most important aspects [6]. This is understandable if one considers the value of walking and standing that is placed by the

society. The society's attitudes of different postures are reflected in our language, for example, 'stand up and take it like a man' reflects the attitude that society has towards standing.

Clearly, the main priority of rehabilitation for most patients is to regain the ability of functional walking, but, it is also an area of conflict among many health professionals. Research and experience have shown that the patients are likely to opt for wheelchair mobility instead of using their ambulation skills after rehabilitation [18]. Nevertheless, gait rehabilitation should not be abandoned. Opportunity of going through gait rehabilitation should be made available to those patients, who wish to walk again. It has been reported that psychological benefits may be received by the patient from restoring their gait ability [19].

In addition, there are some known health issues of prolonged usage of wheelchair, such as no adequate bone loading of the lower extremities (which leads to bone demineralization and osteoporosis [20]) and development of contracture. Lack of usage of lower extremities increase the rate of muscles degrading. Long term inactivity of the joint causes stiffness to the joint and damage to the articular cartilage. Furthermore, moving around public locations using wheelchair faces many obstacles, such as small gap on the ground, elevated ground level, stair case and etc., which is partially due to the fact that such facilities are mainly designed for bipedal walking.

Figure 2-5: Patient having over-ground ambulation training with therapist assistance

In typical clinical setting, a patient undergoes gait rehabilitation with the help of a therapist. The therapist will provide support to the patient, and help the patient to complete the gait locomotion (photo shown in Figure 2-5). Since both hand of the therapist are used to support and balance the patient, the swing phase of gait can only be completed by kicking the foot of patient. This method has one major limitation. It is only applicable to those patients, who are able to stand.

2.2.1 Three General Principles of Rehabilitation

Motor learning of neurological injury rehabilitation relies on three principles: practice, specificity, and effort [21, 22]. With all other things being equal, more practice will result in more learning [23]. The clear benefits of intensive repetitive therapy have been demonstrated by Kwakkel *et al.* [24]. Specificity is the second principle, which highlights that the best way to improve performance of a motor task is to execute that specific motor task [25]. The importance of task specificity in gait rehabilitation has been highlighted in reference [26]. The last principle, effort, indicates that a high degree of participation and involvement of individuals is required to facilitate motor learning [27, 28].

2.2.2 Modern Model of Rehabilitation vs. Traditional Model of Rehabilitation

Traditional model of rehabilitation aims to promote maximum capacity of the muscle strength, range, and endurance in those muscles that can be voluntarily-activated after the lesion. It is assumed that Central Nervous System (CNS) cannot recover after lesion. Thus, traditional model compensates for non-remediable deficits of paralysis, weakness, and balance by using parallel bars, standing frames, tilt tables, braces, and assistive devices. It teaches new strategies or skills for ambulation and mobility, for example, walking with clutches, wheelchair ambulation, walker ambulation, and etc.

The modern model of rehabilitation is inspired by the principle of neuroplasticity. Neuroplasticity refers to the ability of the central nervous system to remodel itself. How CNS remodels itself will not be explored in this research, additional information on neuroplasticity can be found in reference [29].

For stroke patients, it is possible that with appropriate rehabilitation, neuronal circuits that have been affected by brain lesion can potentially be retrained, resulting in

improved motor control. Unaffected regions of the brain, which normally control contralateral or synergistic muscles, may be trained to take over in controlling the affected extremity. Studies have shown evidences of the benefit gained from intensive motor training for those patients [30-33].

Dietz *et al.* [34] and Nicol *et al.* [35] have published that spinal cords have their own locomotion pattern generators, which are capable of inducing basic gait motion patterns when fed with necessary sensory input. It is hypothesized that the pattern generators can be activated, even if there is no voluntary control of these motor segments. Several researches on animals have shown that a spinalized animal can be trained to step again [36, 37]. For stroke or spinal cord injury patients, it has been demonstrated that locomotor training in different forms could be more powerful than conventional treatment using more traditional concepts [26, 38].

2.2.3 Therapist Assisted Body Weight Supported Treadmill Training (TABWS)

The method of suspending a human over a treadmill for gait rehabilitation was first reported by Barbeau *et al.* in year 1987 [5]. This is a relatively new method that originated from basic science research on the neural control of the vertebrate locomotion, also known as modern model (recovery model) of rehabilitation. This method is developed based on the observation of a spinalized cat that can be trained to step with its hind limbs on a treadmill when its weight is partially supported [36, 37]. This method enables non-ambulatory patients to be trained for repetitive practice of complex and complete gait cycles.

Typically, therapist assisted body weight supported gait rehabilitation is carried out by putting modified parachute harnesses on neurologically impaired subjects. The stepping motion on a treadmill is manually assisted by therapists, while body weight of the patient is unloaded by the parachute harnesses. The level of body weight support is adjusted to the maximum sustainable load of the patient, without the buckling of the knee during stance phase. To carry out the stepping training, one therapist is required for each leg. One hand of the therapist will be placed at the anterior surface of the leg, just below the patella. The other hand will be placed on either foot or ankle to assist the toe clearance during the swing phase as well as heel strike during the beginning of stance phase. The third therapist will be maintaining the trunk of the patient in an upright

position by stabilizing the pelvis. Figure 2-6 shows the situation of how therapist assisted body weight supported gait rehabilitation is performed.

Figure 2-6: Therapist assisted body weight supported gait rehabilitation [39]

Studies have been carried out to investigate the performance of body weight support gait rehabilitation and have demonstrated that this method can improve walking in patients [40-44]. It has been found out that 80% of wheelchair-bounded patients with chronic incomplete spinal cord injury gained functional walking ability after training [40, 45, 46].

One of the recognized limitations of therapist assisted body weight supported gait rehabilitation is the significant laborious demand it places on the therapists during training sessions. The labor intensive involvement of the therapists requires a considerable amount of physical effort from the therapists [47]. Worst still, they have to work in a non-optimal postures. Overtime, the therapists will be overexerted and might not be able to provide an optimum gait training pattern for the patient. The amount of practice is significantly reduced due to this limitation.

Therapist assisted body weight supported treadmill training originated from recovery model. Apart from the three general principles of rehabilitation, the following cues are found critical for inducing stepping after SCI [40]. These sensory cues are taken into consideration and used to help identify the gap of current research. These sensory cues are:

1. Generating stepping speeds approximating normal walking speeds (0.75-1.25 m/s)
2. Providing the maximum sustainable load on the stance limb
3. Maintaining an upright and extended trunk and head
4. Approximating normal hip, knee and ankle kinematics of walking
5. Synchronizing timing of extension of the hip in stance and unloading of limb with simultaneous loading of the contra lateral limb
6. Avoid weight bearing on the arms and facilitating reciprocal arm swing
7. Facilitating symmetrical inter-limb coordination
8. Minimizing sensory stimulation that would conflict with sensory information associated with locomotion

2.3 State of the Art in Robotic Gait Rehabilitation

Therapist assisted body weight supported treadmill training requires intensive labor involvement of therapist. To address this limitation, robotic systems that can assist in stepping training have been developed by several research groups. These robotic systems are capable in reducing the laborious involvement of therapist in stepping training and increasing the patient's training duration.

This section examines the state of the art in robotic gait rehabilitation, surveying a selected set of exemplary systems. There are, of course, a large number of relevant examples of robotic gait rehabilitation system emerged in recent years. A complete discussion of all robotic gait rehabilitation systems is not possible, this review focuses on those robotic gait rehabilitation systems correspond to the research direction, in particular, the BWS approach, robotic assisted function/motion, and gait pattern planning are the main features reviewed in this section.

2.3.1 Lokomat

Lokomat is a robotic gait rehabilitation system developed and distributed by Hocoma AG, Switzerland [9]. It is first introduced in 1999 and became a commercialized product in 2001. Lokomat comes with robotic orthosis, body weight offloading module, computer system, and treadmill. Studies have been shown that Lokomat is effective in

improving the walking ability of individuals with incomplete spinal cord injury [48, 49]. Figure 2-7(a) shows a Lokomat in operation.

Lokomat provides motion assistance to the hip and knee joints, helping the patient to complete gait locomotion on treadmill. The motion assistance is achieved by strapping patient to the robotic orthosis at the trunk, pelvis, and both lower limb, using adjustable cuffs with Velcro straps. In place of therapists, a normal physiological gait pattern has been programmed to the system to provide passively guided, symmetrical lower-extremity trajectories to the patient during gait training. The trajectories control is generated by two personal computers and a current controller. The system can be monitored by therapist and assistance can be adjusted as necessary. To ensure toe clearance during swing phase, elastic straps were fitted around the patient's footwear during the training. The weight of the robotic orthosis is taken care of by spring-loaded counterweight module.

Patient's body weight is partially supported by the harness-counterweight system. This system consists of a dynamic low inertia suspension module, allows a precise unloading of the patient. A more advanced BWS module, Lokolift, is made available by the developer of Lokomat in 2006. Similar to other BWS module, the patient is secured in a harness, and connected to the BWS module via a polyester rope. A control system that consists of two drives and two springs is connected to a dynamic roller. This control system provides the counterforce to unload the desired patient's bodyweight, by adjusting the length of the spring and force produced by the drive. The schematic of Lokolift mechanical setup is depicted in Figure 2-7(b). Lokolift has been tested on healthy subjects, and it is found out that the mean error and maximum error of unloading force are significantly lower than other BWS systems [50].

Figure 2-7: (a) Lokomat [51] (b) Schematic of Lokolift Mechanical Setup [50]

The motion of the hip and knee joints of Robotic Orthosis in Lokomat is programmed with average leg kinematics of walking gained from healthy subjects. The motion is modified for greater foot clearance during the swing phase. The pelvis and trunk are firmly attached to a parallelogram. The pelvis and trunk is stabilized and movement is allowed in vertical direction only. The parallelogram is not actuated, but rather it is supported by a gas spring. The setup allows passive movement of pelvis and trunk in the vertical direction in coordination with the lower limb kinematics.

One of the biggest advantages of Lokomat is that it is a commercialized system. Being a commercialized product, it is available for clinical use and has been tested in several studies [47, 52, 53]. In addition, the robotic system, software, and other accessories of Lokomat are developed for ease of use in clinical setting. The entire system is designed to be operated with only one therapist. In recent years, developer of Lokomat also looking into augmented feedback [54], patient cooperative control strategy [55], and adaptive body weight support [56].

One of the distinct limitations with the Lokomat is the restriction the harness places on the user's trunk and pelvis. This limitation is mainly caused by the body weight support apparatus. Lokomat firmly compresses the participant's pelvis in the frontal plane just above the greater trochanters, with sagittal plane restraints using a posterior pad with adjustable straps. As such, during walking with the robotic device, pelvis movement is restricted. These restrictions prevent pelvic rotation and body weight shifting, which then translate to limited movement of the trunk [47, 52]. Due to the

limitations of the robotic systems, the movement of pelvis is altered and it is no longer physiologically resembling the movement of pelvis during normal walking.

Other Similar Robotic Systems

In addition to Lokomat, there are many examples of similar robotic gait rehabilitation systems [57-61]. Certain characteristics are common among these systems: body weight support module, robotic orthosis, and treadmill. Active leg exoskeleton (ALEX) was developed in University of Delaware, Newark [62], focuses on the assist-as-needed paradigm for rehabilitation. ALEX has actuated hip and knee joints in the sagittal plane. All other available DoFs are passively held by springs. LOPES (developed in University of Twente, The Netherlands) is a device consists of a cable driven exoskeleton for the lower limb motion assistance and an end effector robot for the pelvis motion assistance. It provides actuation to mediolateral motion and anteroposterior motion of the pelvis, adduction and flex-extension of the hip joint, and flex-extension of the knee joint [60]. PAM (pelvic assist manipulator) and POGO (pneumatically operated gait orthosis) are two devices developed by University of California, USA. PAM provides a five DoFs actuation (actuated by pneumatic cylinder) to the pelvis and POGO has actuated hip and knee joints in the sagittal plane [61]. The BWS module in conjunction with PAM and POGO is the overheard BWS system developed by Robomedica Inc. [63]. ALEX and LOPES did not highlight the features of the BWS module incorporated in their system.

2.3.2 HapticWalker

HapticWalker is a robotic walking simulator design for gait rehabilitation. It was developed by Charite University Hospital, Berlin, Germany in collaboration with the Technical University of Berlin, Germany [11]. The feet of the patient are attached to the footplates, as shown in Figure 2-8b. The training motion is induced to the patient through the two footplates. Each footplate is connected to end effector of a three DoFs robotic arm. The robot arm moves the footplate linearly in the sagittal plane and rotates the foot plate with respect to the axis perpendicular to the sagittal plane. A photo showing the entire structure of HapticWalker is available in Figure 2-8a.

With two programmable footplates, any foot trajectories can be programmed for the rehabilitation session, thus the rehabilitation training is no longer restricted to level

ground walking. Arbitrary foot trajectories like walking up or down stairs can be trained with the programmable footplates. The foot trajectory can be fully adjusted according to individual patient's needs. These profiles are constructed based on real motion tracking data. At current stage, the pelvic and trunk of the patient are passively supported. The feet trajectories profile are modified for a constant pelvic position during the training.

HapticWalker has several control modes. Position controlled mode allows the machine to provide full guidance to the patient, which is required for the initial rehabilitation phase. In compliant foot plate behavior mode, the machine gradually decreases its support based on the patient's learning success. The machine can be set to zero-force controlled mode for swing phase so that the machine will be fully driven by the patient during the swing phase.

Figure 2-8: (a) HapticWalker [64] (b) Foot binding on the footplate [65]

The main advantage of HapticWalker is that the training profile is not limited to over-ground walking. While the machine is able to provide arbitrary foot trajectory training, the trajectory is altered due to the limited movement of the pelvic. The disadvantage and effect of pelvis fixation on the training effect is discussed in Section 2.4.3 of this thesis. Furthermore, researchers group of HapticWalker is aware of this issue.

In contrast, the floor area requirement of HapticWalker is relatively larger than any other available robotic gait rehabilitation systems. A simplified version of HapticWalker is available. Mechanized gait trainer, GT II, provides fixed foot trajectory of walking based on the footplate principle [66]. The motion is achieved by a double crank and rocker gear system.

2.3.3 WalkTrainer

The WalkTrainer (WalkTrainer™ is a commercial trademark of the Swiss Foundation of Cyberthosis [67]) is composed of a deambulator, a pelvic orthosis, a pair of leg orthoses, and body weight support module [68, 69]. This system provides body weight support, and lower limb and pelvic motions guidance in the context of over-ground walking. The WalkTrainer is shown in Figure 2-9.

Figure 2-9: (a) Concept of WalkTrainer [68] (b) WalkTrainer tested on human [69]

The deambulator is the mobile frame of WalkTrainer, which follows the patient during the exercise. It is also the structure that carries all the sub-components. Two motorized wheels are equipped on the deambulator. The two motorized wheels are controlled differentially, for omni-directional control of the deambulator. The pelvic orthosis is an Artigue parallel structure, provides six actuated DoFs to assist the pelvic motion. The robotic orthosis provides motion assistance to the thigh, shank, and foot in the sagittal plane. The body weight support feature is achieved by using a preloaded spring and a DC motor is used to compensate for small disturbances.

The motion of the pelvis is planned by taking reference from the captured pelvic motion of healthy subject. Fourier series is used to model the captured pelvic motion as a function of the test subject parameters, such as gender, size, lower limb length, pelvic width, and walking speed. The function is then used for generation of tailored trajectory for patient. Details of the pelvic trajectory planning of the WalkTrainer are in reference [70]. The leg orthoses of the WalkTrainer uses the Clinical Gait Analysis data [10] as the

motion input. However, the planning of pelvic and lower limb motions as one complete gait locomotion is not discussed. It is unclear that how pelvic and lower limb motions are integrated into the WalkTrainer.

Performance of pelvic motion guidance with the pelvic orthosis together with cable-harness body weight support is not published till date. The body weight support approach is similar to what Lokomat is using, and the similar limitation (the pelvic motion will be constrained by cable harness body weight support apparatus) is unavoidable. The feature of over-ground walking is briefly described as “the speed of deambulator is synchronized with the walking speed of the leg orthosis” [70]. The synchronization methodology is not clearly specified by the research group.

2.3.4 KineAssist

KineAssist is a robotic device for gait and balance training developed by Chicago PT LLC (renamed to Kinea Design, LLC) [12]. This research work focuses on the balancing control for gait rehabilitation instead of robotic assistance for leg motion. The main objective of this device is to assist clinicians in maintaining patient safety, while the patient is performing walking, and balancing exercises. The photo of KineAssist in operation is shown in Figure 2-10.

Figure 2-10: KineAssist in operation [12]

KineAssist consists of an omni-directional mobile base, which was developed based on Cobot Technology [71]. The mobile platform provides forward and turning motion in order to follow the subject, and to turn around corners. Motion intention of the patient is detected by a combination of passive sliders and integrated force sensors, which are

incorporated into the body weight support structure at the pelvic part of the patient. KineAssist can support a patient's weight of up to 150 pounds in the vertical direction. The support force is provided by vertical column, which is powered to provide force continuously. The vertical column allows motions of the pelvis and torso at the same time.

KineAssist includes a trunk and pelvis mechanism to allow bending motions for left-right, forward-backwards, and rotations about a person's transverse axis, as well as the hip rotation about the forward axis. Among all the DoFs of KineAssist highlighted in Figure 2-11, only two DoFs are actuated (forward bending of the trunk and lateral bending of the trunk). The collapse of the trunk is prevented by having a torso mechanism attached at the chest level. These mechanisms allow the patient to walk naturally while the safety of the patient is ensured. Falls are likely to happen when patients walk. KineAssist is designed to slow and stop the fall of the patient without inducing pain. It can safely bring a patient to a stop after a few inches of fall. The threshold distance for identifying and stopping a fall can be selected by therapist.

Unlike the systems reviewed in previous session, KineAssist is designed as a tool for therapist, the system itself does not provide physical therapy. The major advantage of this system is the realization of body weight support module. Instead of using cable-harness body weight support concept, KineAssist provides body weight support through the robotic arm. Simultaneously, the robotic arm provides the necessary freedom for natural pelvic and trunk movement. Compared to cable-harness body weight support, KineAssist does not have the constraint of pelvic motion imposed by the body weight support setup.

Since KineAssist is developed as an assistive tool for therapist, the main purpose of the system is to maintain the balance of patient during walking. The robotic arm design is not capable of providing assistance to pelvis to complete pelvic motion training for walking.

Figure 2-11: DoFs in (a) Trunk control mechanism; (b) Pelvis control mechanism [71]

2.4 Ability and Limitation of Existing Systems in Functional Walking Provision

Section 2.3 provides review of the robotic gait rehabilitation system in general perspective. In order to gain insight of the ability and limitation of the existing robotic gait rehabilitation systems in providing the functional walking, we must first consider the basis of human walking. Three required elements for gait training are (1) maintaining equilibrium during locomotion, (2) reciprocal stepping, and (3) body weight shifting. First two elements are the requirement of gait locomotion [40], and body weight shifting is included as suggested by doctors and therapists of Tan Tock Seng Hospital in Singapore. The three elements are illustrated in Figure 2-12.

If we consider one of the rehabilitation principles, specificity [25, 26], to regain walking ability, the patient has to be trained to walk. Based on this principle, body weight supported robotic gait rehabilitation systems should be developed to facilitate gait rehabilitation and allow the patient to walk, and experience the elements of gait locomotion. Reciprocal stepping and balance are also the essential elements highlighted for gait locomotion [5, 72]. The three elements are illustrated in Figure 2-11. Reciprocal stepping consists of lower limb and pelvic motions. Balance is the requirement of maintaining equilibrium during the entire walking process. Lastly, body weight shifting is the shifting of body weight from previous supporting leg to next supporting leg, to facilitate the swing phase for the previous supporting leg.

Figure 2-12: Three elements required for gait training

Reciprocal stepping of the lower limb received the most attention in the field of robotic assisted gait rehabilitation. Robotic lower limb orthosis is a module commonly found in robotic gait rehabilitation systems, provides the assistance to complete the reciprocal stepping of lower limb. The pelvic motions are assisted by robotic gait rehabilitation system with the incorporation of pelvic orthosis or similar mechanism setup. Study [57] showed that provision of pelvic motion control in gait rehabilitation also facilitates the stepping process. To promote body weight shifting, pelvic motion has to be assisted or allowed (passive degree of freedom).

The balance of the subject during gait rehabilitation is maintained by the harness of body weight support apparatus suspending the subject [6]. The trunk of the patient is kept upright during gait rehabilitation by the harness. Alternatively, the balance can be actively controlled by keeping center of mass (COM) within the support polygon. Lastly, the walking environment (progression method) for gait rehabilitation is either on treadmill, footplate induced, or over-ground.

Figure 2-13: Overview of gait rehabilitation features realization with module of the developed robotic gait rehabilitation

To summarize, three elements of gait training are implemented in robotic assisted gait rehabilitation with six features. These six features are (1) reciprocal stepping of the lower limb, (2) pelvic motion, (3) body weight shifting, (4) body weight support, (5) balance and trunk stability, and (6) progression method. The realization of these features with module of the developed gait rehabilitation systems is outlined in Figure 2-13.

Four key modules found in robotic gait rehabilitation systems with respect to their intended function are reviewed in this section. Four modules and corresponding function in gait rehabilitation are I) lower limb orthosis for reciprocal stepping of the lower limb, II) body weight support apparatus for supporting body weight and balance control, III) pelvic orthosis for pelvic motion assistance, and IV) progression method/medium.

2.4.1 Lower Limb Orthosis for Reciprocal Stepping

The clear disadvantage of therapist assisted body weight supported treadmill training is the high laborious demand on therapist. Three therapists are needed to complete the training, whereby two of them are positioned at unfavorable posture to assist the recipro-

cal stepping of the patient. Robotic lower limb orthosis has been developed to provide the reciprocal stepping assistance to the patient during the robotic assisted gait rehabilitation.

Figure 2-14: Robotic orthosis of treadmill based systems, from left to right, ALEX [73], Lokomat [74], LOPES [75], and POGO [61]

Figure 2-15: From left to right, robotic orthosis of the WalkTrainer (Over-ground based system) [10], and footplate of HapticWalker (Footplate based system)

Of all the reviewed systems, KineAssist does not consist of lower limb orthosis. KineAssist does not include reciprocal stepping as design criteria, as it is not designed to provide gait rehabilitation.

ALEX is developed with DC motor actuated hip and knee joints (total four degrees of freedom). Robotic orthosis of ALEX is equipped with more passive joints as compared to Lokomat. Passive movement for abduction and adduction of hip joint, plantarflexion and dorsiflexion of ankle joint, and limited inversion and eversion of ankle joint are incorporated in the robotic orthosis [62].

Similarly, Lokomat uses DC motor to power their robotic orthosis, providing actuation to the hip and knee joints (total four degrees of freedom, for left and right side). The actuation is provided in the sagittal plane. To prevent foot drop during swing phase, the patient's feet are suspended with elastic band. The robotic orthosis is designed to be back-drivable, therefore allowing the system to be used to capture the locomotion pattern on healthy subject, and the captured pattern can be used for rehabilitation [76].

Lower extremity powered exoskeleton (LOPES) consists of joints actuated with series elastic actuators. Actuator is detached from the exoskeleton, by transmitting the power of actuator to the intended joint via Bowden cable. LOPES has eight degrees of freedom, which includes (1) forward/backward movement of the pelvis, (2) horizontal movement of the pelvis, (3) abduction and adduction of hip joint, (4) flexion and extension of hip joint, (5) flexion and extension of knee joint. Movements (3) to (5) are available for left and right side [77].

Pneumatically operated gait orthosis (POGO) consists of two pneumatic cylinders for the motion assistance of one side of the leg. Flexion and extension of hip and knee joint is actuated [78]. The same setup might be available for both legs although it is not mentioned. It is designed to work in conjunction with Pelvic Assist Manipulator (PAM). Pneumatic cylinder is low cost and relatively lightweight as compared to electric motor.

The WalkTrainer incorporated a pair of leg orthoses to provide lower limb motion assistance. Motion assistance is provided to thigh (flexion and extension of hip joint), shank (flexion and extension of knee joint), and foot (plantarflexion and dorsiflexion of ankle joint) in the sagittal plane [10]. Passive joint is made available for inversion and eversion movement of the ankle joint.

With a very much different approach, HapticWalker provide reciprocal stepping for rehabilitation using footplate. Instead of moving the thigh and shank of the patient, HapticWalker induce gait locomotion through the ankle. Two robotic arms with three degrees of freedom each are used to provide the desire motion for the footplates. The footplate can move in the sagittal plane, and rotate with respect to the axis perpendicular to the sagittal plane.

The human leg has six primary degrees of freedom, three at the hip joint, one at the knee joint, and two at the ankle joint. Determination of the degree of freedom necessary

in a robotic orthosis depends on the tasks required. To achieve relatively normal gait locomotion, it is necessary to control at least hip and knee flexion [79]. Thus, it is a norm that most robotic orthosis of the rehabilitation systems provide guidance to hip and knee joint, for the sagittal plane motion. Abduction and adduction of hip joint is incorporated (either actuated or passive joint), if pelvic motion guidance is provided.

If forward propulsion is not to be provided by the upper body (e.g. with crutches) then plantar flexion of the ankle is also required [79]. Most of the robotic rehabilitation systems provide forward propulsion (or rather backward propulsion of foot) by treadmill, so these systems do not include ankle joint assistance. The WalkTrainer provides over-ground walking experience and ankle joint assistance is provided.

Table 2-1: Features of Lower Limb Orthosis in Robotic Gait Rehabilitation Systems

	<u>Treadmill</u>				<u>Over-Ground</u>	<u>Footplate</u>
	<i>ALEX</i>	<i>Lokomat</i>	<i>LOPES</i>	<i>POGO</i>	<i>WalkTrainer</i>	<i>HapticWalker</i>
Type of Actuator	DC motor	DC motor	Cable driven	Pneumatic	DC motor	Robotic Arm
Actuated Joint						
Sagittal	Hip and Knee	Hip and Knee	Hip and Knee	Hip and Knee	Hip, Knee, and Ankle	Translation in sagittal plane and rotation about the axis perpendicular to sagittal plane
Frontal	-	-	Hip	-	-	
Transverse	-	-	-	-	-	
Total DoF	Four	Four	Six	Four	Six	Six
Passive Joint						
Sagittal	Ankle	-	-	-	-	-
Frontal	Hip and ankle	-	-	-	Ankle	-
Transverse	-	-	-	-	-	-
Total DoF	Three	-	-	-	One	-

Field marked with - indicates information is not available

Although in gait rehabilitation, the foot is not going to provide significant power for propulsion, it has been found that most post-stroke patients, who have problems in their plantar-flexor muscles, usually are unable to control their ankle and foot, thus affecting their gait pattern [80]. To develop a normal walking dynamics for patients, it is extremely beneficial and essential for them to have ankle assistance during gait training [22].

Table 2-1 shows the summary of the features that are discussed above. The snapshot of lower limb orthosis is provided in Figure 2-14 for treadmill based systems and Figure 2-15 for other systems.

2.4.2 Review of Body Weight Support Approach

Body weight support apparatus is one of the key modules in body weight supported gait rehabilitation system. In general, BWS concepts in gait rehabilitation system can be classified as cable-harness body weight support (cBWS) concept and structural body weight support (sBWS) concept.

Cable-harness BWS concept uses a harness system, which worn by the patient. To offload the patient's body weight, the harness system is connected to a counterpoise system through cable. Cable-harness BWS is the most common body weight support concept found in most of the gait rehabilitation systems, for example, Lokomat [9], POGO [61], WalkTrainer [10], and Haptic Walker [11]. Body weight support apparatus are not made clear by research group of ALEX and LOPES.

In contrast, sBWS concept uses an extended robotic mechanism to support the patient's body weight by holding either at the waist level or at the back. This concept is less popular compared to cBWS. KineAssist [81] is one rehabilitation system that uses this concept. Among all systems, only the detail information on body weight support module of Lokomat has been clearly published.

Figure 2-16: Body weight support module, from left to right, Lokolift [82], POGO (produced by Robomedica Inc., POGO is not shown in the photo) [63], KineAssist [12], and WalkTrainer [69]

Lokolift is incorporated for the advance version of Lokomat, with the static offloading capacity of 150 kg and dynamic body weight support capacity of 80 kg. Patient is

secured in a harness, and connected to Lokolift via 8 mm diameter polyester rope. A winch connected to the rope keeps the system in an optimal working range by adjusting the length of the polyester rope. A control system that consists of two drives and two springs is connected to a dynamic roller. The control system provides the counterforce to unload the desired patient's bodyweight by adjusting the length of the spring and force produced by the drive. Lokolift has been tested on healthy persons. The mean error and maximum error of unloading force provided by Lokolift are significantly lower than that of the other body weight support approaches [50].

POGO system uses overhead body weight support system developed by Robomedi-ca Inc. [63]. The body weight support system unloads the body weight of patient by controlling the tension of the cable that suspends the patient [61]. The tension of the cable is controlled by air cylinder and the maximum support force is 130 kg. The continuous control of body weight support levels and modulation of ground reaction force can be achieved.

The WalkTrainer provides body weight support with a motor controlled preload spring. Additional motor is incorporated in the module to achieve a precise control of the unloading force during the walking process. A passive three DoFs mechanism connects the body weight support controlling module to the harness wore by the subject. The mechanism reduces the constrain imposed by the body weight support module on the subject during walking [83].

The arm of KineAssist holds the trunk and pelvic mechanism that allows the patient's bending motions both left/right, forward/backwards, rotations about a patient's transverse axis, and hip rotations about the forward axis. The information on the control of the arm and the body weight unloading condition is not available except that it can unload up to 150 lbs (70 kg) and it allows the vertical motions of the pelvis and the torso. The sBWS approach appears to be the best method in order to allow natural pelvic motion.

Table 2-2: Features of Body Weight Support Module in Robotic Gait Rehabilitation Systems

	<u>cBWS</u>			<u>sBWS</u>
	<i>Lokomat</i>	<i>PAM/POGO</i>	<i>WalkTrainer</i>	<i>KineAssist</i>
Name/Brand	Lokolift	Robomedica	-	-
Unloading Capacity				
Static	150 kg	130 kg	-	159 kg
Dynamic	80 kg	-	-	68 kg
Unloading Approach	Body weight support is provided by cable-harness connects to counterpoise module			Support force is provided by mechanical arm
Other Features				
Unloading Mechanism	Motor and Spring	Air Cylinder	Motor and Spring	Motor
Control System	Robust Control System	Active Control System	Active Control System	-

Field marked with - indicates information is not available

Limited information is available for body weight support module of majority robotic gait rehabilitation systems. Those body weight support modules reviewed are able to provide constant and desired body weight support if a control system is incorporated. However, cBWS has an obvious limitation. The typical setting of cBWS imposes movement constraint on the pelvis since the pelvis is rigidly secured in place by the harness. In contrast, sBWS appears to provide promising solution to avoid the constraint. Only one system adopted such approach, whereby conclusive remark cannot be drawn based on the review on the single system. The essential features of reviewed body weight support module of robotic gait rehabilitation systems are summarized in Table 2-2.

2.4.3 Pelvis Motion Provision with Pelvic Orthosis

Human gait locomotion consists of six determinants, which are pelvic rotation, pelvic tilt, stance phase knee flexion, knee mechanisms, foot mechanisms, and lateral displacement of the pelvis [84]. Three out of six of these determinants are pelvic related movements, which highlight the importance of pelvic control during gait rehabilitation.

As mentioned, the pelvic motion significantly affects human's gait. The lack of pelvic control on patient will result in alteration of gait pattern, which should be avoided [85].

The Importance of Pelvic Motion in Human Walking

The pelvic, also known as hip is not a single piece of bone. It consists of the two hip bones. Pelvic forms the bridge between the two lower limbs, whereby the proximal end of femur articulates with the acetabulum of the hip bone and forms the hip joint. Being the mobile link between the two lower limbs, pelvic allows the swing to be initiated since the pelvic lateral displacement allows the center of mass and body mass to transfer to the stance foot, thus allowing the contra lateral foot to swing [86, 87]. Apart from allowing the transferring of forces from the lower extremities to the trunk, pelvic contributes to the forward progression of the body and to the vertical support of trunk [88].

Review of Pelvic Orthosis and Pelvic Mechanism

In general, pelvis can be modeled as a body with six DoFs (the DoFs are highlighted in Figure 2-17), which consists of translational and rotational movements in three axes. Three translational movements consist of mediolateral movement (left/right), anteroposterior movement (front/back), and superior-inferior (up/down) movement. The rotational movements are pelvic rotation, pelvic tilt, and pelvic rotation about mediolateral axis. During gait locomotion, the pelvic rotates at transverse axis, tilts at frontal plane, and shifts laterally. Among the robotic gait rehabilitation systems in the current review, PAM and WalkTrainer provide assistance to the pelvis. In contrast, LOPES provide partial actuated assistance and passive movement for pelvic movement. ALEX, Lokomat and KineAssist do not provide any actuated assistance to the pelvic motion. Haptic-Walker do not allow pelvic motion and the pelvis is fixed at a position throughout the gait rehabilitation process.

Figure 2-17: (a) Movement definition of the pelvis (b) Coordination assignment for left side (balloon L) and right side (balloon R) of the pelvis

Figure 2-18: (a) PAM prototype (b) Schematic illustration of PAM [89]

PAM consists of a pair of tripods. Each tripod is constructed by three pneumatic cylinders. The two tripods are coupled together with a belt piece. PAM is a five DoFs pelvic orthosis, providing actuated assistance to three translations (mediolateral movement, anteroposterior movement, and superior-inferior movement) and two rotations (pelvic rotation and pelvic tilt). Pelvic rotation about mediolateral axis is made passive, and stabilized by the cBWS. The range of motion for rotational motion of PAM is designed as $\pm 40^\circ$ for pelvic rotation, $\pm 15^\circ$ for pelvic tilt. The designed translational motion of PAM is 25 cm for mediolateral motion, 25 cm for anteroposterior motion, and

20 cm for superior-inferior motion [61]. A prototype photograph and schematic of PAM is shown in Figure 2-18.

Figure 2-19: (a) Close up of pelvic orthosis on WalkTrainer [68] (b) CAD model of the pelvic orthosis [83] (the arrows indicate motors placement and actuated movements)

The pelvic orthosis of WalkTrainer is a six DoF robot. The robot is actuated by six DC motors. The pelvic orthosis provides motion assistance in three translations (mediolateral movement, anteroposterior movement, and superior-inferior movement) and three rotations (pelvic rotation, pelvic tilt, and pelvic rotation about mediolateral axis) for the pelvis. The prototype and CAD model of the pelvic orthosis for WalkTrainer are shown in Figure 2-19.

LOPES provide actuated pelvic assistance for mediolateral movement and anteroposterior movement. The movement in the vertical direction is made passive with parallelogram. Pelvic rotation, pelvic tilt, and pelvic rotation about mediolateral axis are locked. Lokomat and ALEX do not provide pelvic assistance. In order to allow some movement of the pelvis and trunk, passive movement in the vertical direction is allowed with the use of a parallelogram structure. The parallelogram structure ensures the trunk is upright at all time. The pelvis mechanism of KineAssist allows passive translational movement in the mediolateral and anteroposterior axis and all passive rotational movements (pelvic rotation, pelvic tilt, and pelvic rotation about mediolateral axis) of the pelvis.

Figure 2-20: (a) Pelvic movement provided by LOPES [60] (b) Parallelogram of Lokomat [90] (c) Pelvic mechanism of KineAssist [12]

Table 2-3: Pelvic Movement Provision by Different Robotic Gait Rehabilitation Systems

	<u>Active Assistance</u>				<u>Passive Movement</u>	
	<i>Human Walking</i>	<i>PAM</i>	<i>WalkTrainer</i>	<i>LOPES</i>	<i>ALEX/Lokomat</i>	<i>KineAssist</i>
Type of Actuator	NA	Pneumatic	DC motor	DC motor	NA	NA
BWS	NA	cBWS	cBWS	cBWS	cBWS	sBWS
Translational Pelvic Movement						
Mediolateral	Required	Active	Active	Active	Fixed	Passive
Antero-posterior	-	Active	Active	Active	Fixed	Passive
Superior-inferior	-	Active	Active	Passive	Passive	Fixed
Rotational Pelvic Movement						
Pelvic rotation	Required	Active	Active	Fixed	Fixed	Passive
Pelvic tilt	Required	Active	Active	Fixed	Fixed	Passive
Pelvic rotation about medio-lateral axis	-	Fixed	Active	Fixed	Fixed	Passive

Active – Actuated movement; Passive – Not actuated but free movement allowed; Fixed – No movement allowed
HapticWalker not included in this table, since all pelvic movements are fixed

One of the major gaps in gait rehabilitation research is the natural pelvic control. Due to the incorporation of cBWS in majority of the robotic gait rehabilitation systems, the implementation of natural pelvic control is limited by the cBWS. PAM and WalkTrainer provide the most complete DoF to achieve natural pelvic movement for walking. But the full potential of these systems are limited by the body weight support

module. LOPES provides limited DoF for the pelvis, allowing some movement of the pelvis, but does not lead to a natural pelvic movement for walking. ALEX, Lokomat, and KineAssist allow certain extend of passive movement for the pelvis only. There is no attempt of any robotic gait rehabilitation system to provide natural pelvic motion with sBWS. The pelvic motion assistance available for robotic gait rehabilitation systems are summarized in Table 2-3.

2.4.4 Over-Ground or Treadmill Training?

In reality, over-ground walking is the mode that human moves from one location to another. A treadmill is a machine designed for running or walking while staying in one place. Body weight supported treadmill training was first invented for walking training on a treadmill. Subsequently, there are other progression approaches, footplate and over-ground, made available on robotic gait rehabilitation. The main advantage of treadmill is to allow walking to be carried out stationary. This advantage could be the main consideration when the therapist assisted body weight supported gait rehabilitation was proposed. The nature of the training preferred the whole process to be carried out stationary (else three therapists and the body weight support apparatus have to move together).

Based on the original idea of body weight supported gait rehabilitation, most of the robotic gait rehabilitation systems are developed to provide gait training on treadmill. Systems using treadmill are ALEX, PAM/POGO, Lokomat, and LOPES. Instead of treadmill, HapticWalker uses two moving footplate. WalkTrainer and KineAssist are systems providing gait training in the context of over-ground walking. Limited study is available to highlight the difference between footplate “walking” and over-ground walking. Review will not be carried out on footplate “walking”. The difference between over-ground and treadmill walking are highlighted in this section.

The difference between over-ground and treadmill walking has been studied in many researches. There are opposing views on the differences between over-ground and treadmill walking [91]. In contrast, some studies concluded that there are differences between over-ground and treadmill walking. Differences are found in term of spatial-temporal gait parameters [92], thorax and pelvis movement [93], joint kinematics, and muscle activation. A faster cadence and shorter stride length tends to be used by normal

subjects on the treadmill than during over-ground walking [94]. More details on the studies on the differences between over-ground and treadmill walking are available as part of literature in references [92-98].

Reviews revealed two contradicting views on the acceptance of treadmill to replace over-ground walking [91, 92]. Nevertheless, it is noteworthy that the ultimate goal of gait rehabilitation is to return the patient to over-ground walking, and not walking on a treadmill. Practicing gait in functional over-ground contexts, as opposed to treadmills, is widely desirable [81]. Treadmill systems may not be effective in rehabilitation because walking on a moving belt is not ecological [99].

2.5 Summary on Review of State of the Art

In general, robotic gait rehabilitation systems reviewed in this thesis can be categorized into three groups. First group is the treadmill based robotic gait rehabilitation system, in this group, Lokomat [9] is studied and reviewed, and other similar systems in this group are introduced. Second group is the end effector, also refers to as footplate based group. This group includes the Gait Trainer II [100] and the Haptic Walker [11], both developed by the same research group. The systems in this group used footplate to induce the desired gait locomotion, and treadmill is not required. Last group consists of robotic gait rehabilitation systems, which provide gait rehabilitation in the context of over-ground walking. The WalkTrainer [68] and KineAssist [71] are under this group. Although KineAssist is not classified as gait rehabilitation system by its designer, it is reviewed for the unique feature of its body weight support approach.

In general, current treadmill based robotic gait rehabilitation systems provide the most promising solution for gait rehabilitation training. Originated from therapist assisted body weight supported treadmill training, these robotic systems replace the laborious involvement of therapist by robot. The reciprocal stepping is completed with the help of lower limb orthosis. However, due to the nature of the body weight support setup, the movement of pelvis is constrained. Body weight shifting is discouraged with the limited pelvic movement allowed. In addition, it is unclear how well is the treadmill training can be translated to over-ground walking. The overview of features available in this group is illustrated in Figure 2-21.

- 1: Not available on ALEX and LOPES, these two systems do not incorporate BWS module
 2: Only available on PAM and LOPES

Figure 2-21: Overview of features found on treadmill based robotic gait rehabilitation systems

Figure 2-22: Overview of features found on footplate based robotic gait rehabilitation systems

On treadmill based machine, patient can be trained for walking only. The development of footplate based robotic gait rehabilitation systems allows the training of different lower limb locomotion. This group has the same limitations as treadmill based system, as the pelvic motion of the patient is fixed, when training using the footplate based systems. The body weight shifting is not possible with fixation of pelvis. Using footplate as the training environment might introduce artificial walking environment for the patient. The overview of features of footplate based robotic gait rehabilitation system is illustrated in Figure 2-22.

Figure 2-23: Overview of features found on robotic gait rehabilitation systems provides gait rehabilitation in the context of over-ground walking

The last group of robotic gait rehabilitation system allows the gait rehabilitation to be practiced in the context of over-ground walking. This approach provides the best rehabilitation environment representing the actual human walking. Gait rehabilitation in the context of over-ground walking obviously offered a more specific (specificity is one of the three general principles of rehabilitation) training environment compared to the two other groups. Pelvic motion is actuated or permitted by those systems in this group. However, the ability of the WalkTrainer in pelvic movement assistance is limited by the cable harness type of body weight support setup. KineAssist allows unrestricted pelvic motion passively. The actuated pelvic motion assistance is not available on KineAssist. Body weight shifting is not made available in both WalkTrainer and KineAssist. The overview of features for robotic gait rehabilitation system providing over-ground walking gait rehabilitation is illustrated in Figure 2-23.

To conclude the review on state of the art, the limitations of each grouping are highlighted as follows:

Group 1 - Overhead Cable-Harness Body Weight Supported Treadmill Training Systems

- Restriction of pelvic rotation, pelvic obliquity, and horizontal translation of the pelvis
- Treadmill might not translate to actual functional walking patterns
- Huge space, heavy weight, non-portability and prohibitively high price

Group 2 – Systems with End Effector/Footplate Concept

- Fix pelvic position during training
- Presents an artificial training environment for walking training

Group 3 – Systems Provides Rehabilitation in the Context of Over-Ground Walking

- No motion guidance for the pelvis (KineAssist)
- Motion guidance is limited/constrained by the body weight support setup (WalkTrainer)
- Requires relatively robust and complicated control algorithm in order to provide unconstrained hybrid force and position control

It is obvious that limitation of most of the robotic rehabilitation systems is linked to the pelvic motion. Those systems providing pelvic motion guidance have a in natural pelvic motion provision, due to the setup of body weight support module. Due to the limitation in the provision of natural pelvic motion, body weight shifting is discouraged.

Clearly, there is a need to provide unrestricted pelvic motion guidance, in order to encourage the training of body weight shifting. In addition, attention should be diverted to the design of new body weight support approach. The body weight support approach shall not hinder the provision of pelvic movement. Although there is no clear indication of over-ground walking is better than treadmill walking, nevertheless, we hypothesize that gait rehabilitation in the context of over-ground walking provides a physiological resembling walking environment. Lastly, balance of the patient during gait rehabilitation is secured by the body weight support setup. Patient is not able to maintain balance with own effort. The practice of balance control can be included in robotic gait rehabilitation, utilizing the Center of Mass concept.

2.6 State of the Art from Clinical Perspective

Previous section provides the information and review on several state of the art robotic gait rehabilitation systems. One question which remained unanswered is, how well is the developed systems fit to clinical use, with respect to the three general principles and eight sensory cues of gait rehabilitation presented in Section 2.2. The robotic systems are

compared and concluded in this section with respect to these two sets of basis. Therapist assisted body weight supported treadmill training approach is included in the comparison, as the benchmark for robotic gait rehabilitation systems.

This section provides a comparison of features available on the robotic gait rehabilitation viewing from the perspective of the principles and sensory cues for gait rehabilitation. Based on this comparison, the gap of research for the field of robotic gait rehabilitation can be identified from the clinical perspective.

2.6.1 Three Principles of Gait Rehabilitation and Robotic Gait Rehabilitation Systems

The three principles of gait rehabilitation are used to verify and justify the effectiveness of robotic gait rehabilitation systems, with the Therapist Assisted Body Weight Supported Treadmill Training (TABWS TT) used as the benchmark. In term of practice, robotic systems clearly have the upper hand. Robotic systems allow longer practice duration compared to therapist assisted body weight support gait rehabilitation, since robotic system does not feel tire like human (therapist) do.

In view of specificity, the three elements of gait training are used as the criteria for comparison. Robotic system is capable of executing any pre-defined trajectories precisely. Thus, the gait locomotion is able to be induced by robotic system precisely on the patient. In contrast, the gait locomotion provided by therapist has significant deviation [101], which is deemed as poorly controlled in this review. Pelvic motion is deemed as limited control for robotic systems, due to the constraint of the pelvic motion by the body weight support apparatus. Body weight shifting is only available on therapist assisted body weight supported gait rehabilitation. Both approaches use the similar progression method.

* Limited systems offer complete over ground walking
[#] Research on active participation started, however, it is still in research phase

Figure 2-24: Comparisons of therapist and robotic assisted body weight supported gait rehabilitation for three principles of gait rehabilitation

The major weakness of robotic system is the lack of interaction with patient. Patient can remain totally passive during the process of rehabilitation. This greatly reduced the effort of participation of the patient. In comparison, therapist can feel the patient and encourage the patient to participate in the process of rehabilitation. This is the major advantage of therapist assisted body weight supported gait rehabilitation. Although active participation is being researched intensively in those robotic system, but the therapist input for this segment can be hardly replaced [102].

In general, robotic system shows more advantages than therapist assisted body weight supported gait rehabilitation. The robotic system will be even better, if the active participation for robotic system has significant breakthrough. The key points of the comparison are illustrated in Figure 2-24.

2.6.2 Realization of Eight Sensory Cues in Existing Robotic Gait Rehabilitation Systems

Therapist assisted body weight supported treadmill training originated from recovery model. As indicated in section 2.2.3, apart from three general principles of rehabilitation, there are other eight sensory cues found critical for inducing stepping [40]. These sensory cues are:

1. Generating stepping speeds approximating normal walking speeds (0.75-1.25 m/s)
2. Providing the maximum sustainable load on the stance limb
3. Maintaining an upright and extended trunk and head
4. Approximating normal hip, knee and ankle kinematics of walking
5. Synchronizing timing of extension of the hip in stance and unloading of limb with simultaneous loading of the contra lateral limb
6. Avoid weight bearing on the arms and facilitating reciprocal arm swing
7. Facilitating symmetrical inter-limb coordination
8. Minimizing sensory stimulation that would conflict with sensory information associated with locomotion

The eight sensory cues can be grouped into individual category to better relate it to function of robotic systems. The groupings are as follows:

Gait locomotion related cues

1. Approximating normal hip, knee and ankle kinematics of walking
2. Synchronizing timing of extension of the hip in stance and unloading of limb with simultaneous loading of the contra lateral limb
3. Facilitating symmetrical inter-limb coordination
4. Minimizing sensory stimulation that would conflict with sensory information associated with locomotion

Body weight support related cues

1. Providing the maximum sustainable load on the stance limb
2. Maintaining an upright and extended trunk and head
3. Avoid weight bearing on the arms and facilitating reciprocal arm swing

Other cues

1. Generating stepping speeds approximating normal walking speeds (0.75-1.25 m/s)

Robotic system has no issue in providing the sensory cues grouped as gait locomotion related cues. Precise motion play back is the clear advantage of a robotic system. With the use of body weight support, patient can be trained to walk without the need of using clutch or supporting frame, which effectively avoided the weight bearing on the arms. However, due to the incorporation of robotic mechanism to provide motion assistance to lower limb, the natural swinging of arm is prohibited by the mechanism. This is a common limitation found on robotic gait rehabilitation system. Other sensory cues grouped to body weight support related cues have no problem to realize on existing robotic systems. Lastly, the realization of stepping speeds of 0.75 – 1.25 m/s is not an issue.

2.7 Reference Gait Pattern of the Developed Robotic Gait Rehabilitation System

In order for robotic gait rehabilitation system to carry out gait training, a set of pre-defined trajectory is required. To maximize therapeutic outcome, it is crucial to induce a gait pattern that resembles natural human gait pattern during gait rehabilitation [36, 40]. In the context of practicing correct kinematic gait patterns in a repetitive manner, little attention has been placed to the study of gait pattern used in various gait rehabilitation system. In manual assisted BWS treadmill gait rehabilitation, the therapist moves the patient leg and pelvic using visual feedback and “feel”. The assistance provided can vary greatly between therapists and between training sessions [101].

In comparison, robotic orthosis found in gait rehabilitation systems adopt simplified approach to replicate the leg kinematics [57, 103]. Lokomat took a step further by allowing the gait pattern to be set according to the patient’s height and range of motion of the lower limb joint [52]. The BWS apparatus restricts pelvic motion, causes the modification of pelvic motion planning. The restricted pelvic motion are planned according to the lower limb motions and desired foot trajectory [58]. Apparently, most of the motion planning is template based, whereby a motion template of subject with anatomy

parameters similar to the targeted subject is required. Template based planning needs a wide range of template collection, which is not practically feasible.

2.8 Review of Control Strategies for Robotic Gait Rehabilitation

During the early stage of motor skill acquisition, use of passive guidance could be beneficial [28]. Although studies have shown that passive guidance during practice of a voluntary motor task is less effective in enhancing motor learning than an unrestricted practice [27, 28, 104]. In addition, for patients who are not able to, or are too weak to perform voluntary movements, passive movements in functionally relevant contexts can be used to replace active training [28]. Several studies reported that passively elicited movement activates the similar cortical regions compared to voluntary movement [105-107]. Robotic assisted BWS training can be optimized as a gait rehabilitation strategy, which theoretically allows practice of correct kinematic gait patterns [48]. In general, most of the robotic rehabilitation systems used in clinical populations provides passive assistance.

In the recent advancement in robotic gait rehabilitation, researches have been slowly focused on enabling active assistance for robotic assisted gait rehabilitation. If a less severe patient is trained with passive assistance strategy, the patient tends to use reduced activity of muscles and metabolism. For example, people with incomplete SCI, train in a robotic gait training system with position control consumed 60% less energy than in traditional therapist assisted therapy [108]. On the other hand, if patient attempts to walk actively while the robotic rehabilitation system provides passive assistance, the patient will work against the resistance of the system, which result in abnormal muscle activation patterns [109].

To address those limitations of passive assistance strategy, active assistance, also known as patient-cooperative control strategy is being developed by many research groups [55, 62, 110-113]. Patient cooperative control strategies are designed to recognize the movement intention and muscle effort of the patient. The biofeedback information will be displayed to the patient and therapist. The strategy will adapt to the patient's contribution during training session and adjust to robotic assistance accordingly. The ideal control strategy for gait rehabilitation should do the same as an experienced

therapist. It will assist the patient movement only as much as needed and suggest the patient how to use voluntary muscle efforts and motion strategy [114].

Patient cooperative control strategies are expected to stimulate active participation for the patient. Often, patient cooperative control strategies are combined with biofeedback. The strategies have the potential to motivate the patient because the progress of the patient can be feedback to the patient instantly on a visual display.

2.8.1 Principles of Patient Cooperative Control Strategy

The term cooperative in the patient cooperative control strategy has the principle of interactivity, adaptivity, and flexibility. The meaning of these three principles is:

Interactivity

A bidirectional exchange of information between the robotic rehabilitation system and patient during the training process shall exist. The information from patient feed to robot, such as ground reaction force (GRF), electromyography (EMG), kinematic information, and etc. can be captured by sensors. On the other hand, the information such as ankle trajectory, gait events, lower limb joint angles, and etc. can be displayed instantly through screen to patient, which will help him/her to be more involved in the training process.

Adaptivity

Based on the information flow between the robotic system and the patient, the controller of the robot shall be able to determine the best way and the amount of assistance it will support the patient. In other words, the robotic system shall be able to adapt to the patient's movement and based on the bio-information obtain from patient.

Flexibility

The design of the robotic system shall be flexible and can react to the patient's movement softly and gently.

2.8.2 State of the Art of Patient Cooperative Control Strategy

There are several developed patient cooperative control strategies in the field of gait rehabilitation. The states of the art are:

Automatic Gait Pattern Adaptation Algorithm

Lokomat research team focused on the development of the algorithms for automatic gait pattern adaptation, to allow patient walks actively during the training process [115]. Three different approaches are developed by the team. The first algorithm estimates the interaction torques between the human and robot, and adjusts the angle trajectories in a way that lead to reduction in the interaction torques. The second algorithm estimates the interaction torques and translates the interaction torques into the desired change in the trajectory accelerations. As for third algorithm, an impedance control is adopted to specify the relationship between the interaction force and the allowed position deviation.

Adaptive Impedance Control Algorithm

This is another patient cooperative control strategy developed by the Lokomat research team [55]. For this algorithm, instead of having the gait pattern change after several steps, this approach works with a constant reference trajectories with impedance controller parameters adaptation. The controlled impedance (stiffness, damping, and inertia) is set high when there is a little effort made by the patient. The impedance magnitude will be reduced as soon as an increased patient effort is detected. The lower the impedance magnitude, the greater deviations from the reference trajectory is possible. The behavior of the algorithm is that it guides the patient along the fixed reference trajectories when the patient can't make any effort, and allows for larger deviations when the patient tries to walk in his/her pattern. Adaptive controller is incorporated in the impedance controller, which allows the modification of the impedance parameters based on the patient's performance during the training course.

Patient Driven Motion Reinforcement (PDMR) and Hybrid Force-Position Control Algorithm

Basically, two different approaches were presented. First one is the adaptive control algorithm, which is applied to adapt the reference gait pattern by minimizing the human-robot interaction torque/force. The second approach uses an impedance controller with

adjustable parameters. Both approaches allow the patient to have a certain extent of influence on the reference gait pattern when the patient is making some effort. However, the limitation is that the patient cannot have any influence on the walking speed. Thus the motivation for the patient to generate active force is limited. Supportive control architecture was presented by Lokomat research team [110]. With the algorithm, the robotic system provides a certain amount of assistance for the patient to walk. There is no fixed reference trajectory of gait pattern and the patient can walk arbitrarily at his own speed and his own desired gait pattern.

A dynamic model of the lower extremities of the patient is required for the implementation of PDMR algorithm. Lokomat research team decided not to construct the dynamic model for the stance phase. The team uses position control mode for the stance phase. A hybrid control structure has been applied for the switching between of force and position control. The hybrid control algorithm uses a gait event detector to separate the swing phase and the stance phase. The controller switches from force control to position control when swing phase change to stance phase.

Virtual Force/Velocity Field Control Strategy

Two algorithms were developed for Assist-as-Needed (AAN) strategy for gait rehabilitation, which is named as “Band” and “Window” [116]. The first strategy creates two fixed boundaries: an inner boundary and outer boundary, forms a “band” surrounding 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 specify 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 “Window” strategy is basically analogous to the “Band” strategy, but based on moving “window” geometry. In this approach, a circular “window” moves along the desired trajectory to provide the loose timing control. Within the “window”, the ankle is guided by a small constant velocity field, which is tangent to the desired trajectory, but does not enforce a specify timing inside the “window”. Outside of the “window”, a radio velocity field points inward to put the ankle back to the window.

Path Control

Path control is a new algorithm introduced by Lokomat research team in 2010 [117]. This strategy generates a virtual tunnel around the reference trajectory by an adjustable dead band of width. Within the tunnel, the controller is in “free run” mode, the output of the impedance controller is zero and the gravity and friction of the robot are compensated. The patient will be able to move freely at their own timing as long as they stay within the tunnel. Adjustable supportive torques can be added to the controller output for helping the patient to overcome their lower limb weakness. The directions of the supportive torque vectors are tangential to the movement of the path in joint space.

The impedance controller will provide assistance to correct the leg posture of the patient when it moves outside the tunnel. An optional “moving window” moving at the velocity specified by the therapist along the trajectory can be added to correct the velocity of the leg during swing phase. When the patient moves faster or slower than the predefined velocity, the patient will reach the “front wall” or the “back wall” of the window, and the patient will be put back to the window.

Virtual Model Control (VMC)

The research group of LOPES rehabilitation system has introduced a different gait rehabilitation methodology [118]. 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.

2.8.3 Summary of Control Strategies for Robotic Gait Rehabilitation

Patient Cooperative Control Algorithm

Automatic gait pattern adaptation algorithm for active control which updates the reference trajectories after several gait cycle can allow the patient to contribute their effort during the training process. This approach is good for those who are strong enough to make their effort. However, if the patients are weak and can't make their effort, this strategy will become a position control. This strategy required several gait cycles for the estimation of the human-robot interaction torque and the updating of the trajectories. Therefore, the response of this controller could be a concern.

Adaptive Impedance Control Algorithm

Impedance control provides deviation around the reference trajectory by generating virtual mechanical impedance. It allows the patient to make use of own effort. The adaptive controller is added for the modification of the impedance parameters based on the patient's performance during the training session. This approach is well satisfied for encouraging the patient to be participated. However, the patient can only affect the reference trajectory in spatial. The patient will be enforced by the robot since the patient cannot interfere the trajectory in temporal.

Patient Driven Motion Reinforcement (PDMR) and Hybrid Force-Position Control Algorithm

PDMR controller only supports the patient with a certain amount of force for walking. This is a patient centered approach instead of robot centered. The patient can move in any pattern and the robot will follow. Since no reference trajectory is provided to the system, proper gait locomotion might not be induced to the patient. This could results in improper gait pattern trained. This phenomenon is against the gait rehabilitation principle, specificity.

Virtual Force Velocity Field Control Algorithm

Virtual force/velocity field control creates a virtual "tunnel" around the reference trajectory of ankle position. The patient can move his/her foot freely as long as the foot stays within the "tunnel". The reference trajectory is flexible in both spatial and temporal. However, after the virtual "tunnel" is defined, the width of the tunnel is fixed for

the entire training process. The controller should be able to perform better if there are some other parameters taken into account to modify the wall of the “tunnel” based on the performance of the patient.

Path Control

Path control is the most recent control strategy, which is introduced in 2010. It is similar to force field control but the trajectory in the joint space is controlled rather than the trajectory in Cartesian space. One of the limitations is that the width of the “tunnel” can’t be modified during the training process. The reference lower limb angles allow the patient to have some deviations, which the patient can move freely inside. The tradeoff is that the heel strike cannot be warranted. Therefore, the width of the “tunnel” has to be tight enough for the motion to be resembled to a natural gait pattern. There is an impedance controller outside of the virtual wall of the “tunnel”, which will drive the patient back into the “tunnel”. The parameters for the impedance controller are however unchangeable. The deviation provided by the virtual mechanical impedance is remained unchanged regardless of the patient’s performance.

Virtual Model Control

Virtual Model Control (VMC) was developed by LOPES research group in 2007. This control method is mainly for subtask level (i.e. foot clearance, balance control) training. There is no fixed reference trajectory required. Therefore, the therapeutic outcome of the training process such as natural gait pattern and natural foot trajectory can’t be guaranteed.

2.9 Gaps in the Current Research

Review of the state of the art and discussions with doctors and therapists of Tan Tock Seng Hospital in Singapore reveals that there are several issues that needed to be addressed in the field of robotic assisted gait rehabilitation.

2.9.1 Gait Pattern Generation for Robotic Assisted Gait Rehabilitation

As described previously, most of the gait rehabilitation systems use template based approach for gait pattern planning. The available template is limited and does not cover a wide spectrum of gait patterns for different walking speed, cadence, and stride length.

Thus, it is believed that these gait rehabilitation systems use the limited set of gait pattern in their gait training. Lokomat system uses template based approach to obtain the gait pattern for the robotic system. This approach matches gait pattern for the patient from a template pre-recorded from another subject with similar anatomical features.

Obviously, the issue of gait pattern generation for robotic assisted gait rehabilitation is not well addressed. A methodology of gait pattern planning is required since current approach cannot provide a gait pattern for a specific subject with any given walking speed, cadence, and stride length.

2.9.2 Pelvic Motion Assistance and Body Weight Shifting

It is clear that pelvic motion plays a crucial role in human walking. However, many current gait rehabilitation systems do not provide pelvic motion assistance. Among all, complete work for pelvic motion assistance is found addressed in detail on WalkTrainer and PAM. However, the pelvic motion is restricted by the cable-harness body weight support apparatus used by both systems.

Body weight shifting is a training process of conventional over-ground gait rehabilitation. Due to the lack of unrestricted pelvic motion assistance in robotic gait rehabilitation systems, body weight shifting is not a feature available on those systems.

2.9.3 Over-Ground Walking for More Realistic Walking Experience

Majority of the developed gait rehabilitation systems provide gait training on treadmill. Two contradicting views are revealed on the acceptance of treadmill to replace over-ground walking. Nevertheless, the ultimate goal of gait rehabilitation is to return the patient to over-ground walking, rather than walking on treadmill.

2.9.4 Control Strategy

Control strategies are designed by respective research group to achieve specific objective for active assistance during robotic assisted gait rehabilitation. Substantial works have been carried out to develop the control strategy for the active assistance of the lower limb motion. However, not much has been done on the active assistance of the pelvic motion.

2.10 Concluding Remarks and Overview of Research Approach

In this chapter, reviews of clinical and engineering aspect in gait rehabilitation are presented. Clearly, body weight supported gait rehabilitation enable the disabled patient to train for walking. This is previously not possible with the traditional training approach. Commonly, the body weight supported gait rehabilitation is carried out on treadmill. Over the last 10 years, studies have shown that SCI and stroke subjects who receive body weight supported treadmill training demonstrate improved EMG activation patterns, more natural walking characteristics, increment in weight bearing on their legs, and demonstrate functional improvement in walking ability. Review of these evidences can be found in [7].

Robotic system for body weight supported gait rehabilitation is introduced to address the limitation of the therapist assisted body weight supported gait rehabilitation. Most of the robotic systems provide lower limb assistance in the similar manner with little attention focused on the pelvic motion guidance. It could be partially due to the body weight support apparatus used in majority of the robotic systems, which inhibit the assistance of unrestricted pelvic motion. Cable-harness body weight support apparatus secures pelvis rigidly in place, disallow natural pelvic movement during gait rehabilitation. With this restriction, body weight shifting is made impossible.

To conclude this chapter, it is found out that there is a need to address the issue on natural pelvic motion guidance, which also promotes body weight shifting. It is unclear that how well treadmill training can be translated to over-ground walking, but over-ground walking may provide more physiological resembled gait training for the patient. Lastly, active participation of patient in rehabilitation is a hot research topic of robotic assisted gait rehabilitation. Active participation ensures the patient uses own effort as much as possible during the gait training. According to the principle of gait rehabilitation, a high degree of participation and involvement of individuals are required to facilitate motor learning [27, 28]. However, it is noteworthy that robot will never be able to replace the interpersonal contact of therapist and patient, which is so necessary for rehabilitation. It is only an additional option to intensify therapy [102].

2.10.1 Research Approach

The literature review in both clinical and engineering aspects provides valuable background information, forms the basis of the research methodology proposed in this research work. With the research requirements laid down in the research methodology, engineering realization convert the idea into robotic gait rehabilitation. The engineering realization do considers the opinion given by doctors and therapists, as a form of clinical requirements for the engineering realization. This research work categorized targeted patients according to their physical condition after lesion, and the robotic rehabilitation system are designed for the selected groups of patient.

Figure 2-25: Overview of research approach

The developed robotic rehabilitation system undergoes two trial cycles. Prototype trial is arranged to test the system on healthy subject. The feedback collected during prototype trial will be used to improve the system. After the improvement, clinical trial will be arranged for testing the robotic system with patients in a local hospital. The feedback collected during clinical trial will be analyzed for future improvement of the robotic system. The overview of the research approach is provided in Figure 2-25.

The main focus for this research work is the creation of new methodology for robotic assisted gait rehabilitation, the realization of the proposed features in the methodology via engineering approach, and eventually the trials of the robotic gait rehabilitation. The methodology for gait rehabilitation and engineering realization are further categorized into four areas:

1 Gait Study

- a. The main aim of this gait study is to fill the gap of current research in the gait pattern planning. A prediction model for individual-specific gait pattern generation is introduced. Gait parameters and waveform of the lower limb joint angles will be generated based on the some input parameters. Details are discussed in Chapter 3.
- b. The motion capture data will be used as the predefined motion input (reference trajectories) for the passive control strategy of the proposed robotic system.
- c. The range of motion for every joint involved in the locomotion training will be carefully studied. Hardware limits and virtual constraints will be implemented to ensure the gait rehabilitation system will not travel beyond the patient's acceptable range of motion during the gait rehabilitation session.

2 Robotic System

The design and development of robotic system for the realization of gait locomotion assistance, pelvic motion assistance, and body weight support, in the over-ground context for targeted patient.

3 Control Strategy

- a. An overall control system will be developed, consists of programmed controller for controlling each motion of the system, and provide a Graphical User Interface (GUI) to allow the user (doctor/therapist) to make necessary change to the gait rehabilitation process (for example, walking velocity for the training). The control system can be programmed to enable different rehabilitation control strategies based on specific rehabilitation requirement.
- b. This project will study the feasibility of implementing active assistance for gait rehabilitation using biological-signals (for example, EMG) and gait parameters.

4 Bio-feedback

Bio-feedback provides the essential bio-signal feedbacks, e.g. ground reaction force, electromyography signal, gait events, and etc. The bio-feedback serves two main objectives: visual feedback for assessment, and feedback for implementation of control algorithm. Bio-feedback is not subject of study in this work. The concept proposed in Figure 2-26 on bio-feedback is based on basic knowledge on bio-feedback. It is for the completion of the overall research approach.

Figure 2-26: Realization of the research approach of this research work

Chapter 3

INDIVIDUAL SPECIFIC GAIT PATTERN PLANNING FOR ROBOTIC ASSISTED GAIT REHABILITATION

A gait pattern is required for robotic system to define the correct motion, which will be induced by the system to the patient during rehabilitation. Motion capture experiment was carried out in this research work to acquire the required motion for the developed robotic gait rehabilitation system (the details are presented in Chapter 5). However, the acquisition of gait patterns catering different walking speed, cadence, and stride length combination specifically for all targeted subject is not practical.

To maximize therapeutic outcome, it is crucial to induce a gait pattern that resembles natural human gait pattern during gait rehabilitation [36, 40]. In the context of practicing correct kinematic gait patterns in a repetitive manner, little attention has been devoted to the study of gait pattern generation for various gait rehabilitation system. In manual assisted BWS treadmill gait rehabilitation, the therapist moves the patient leg and pelvic using visual feedback and “feel”. The assistance provided can vary greatly between therapists and between training sessions [101].

In comparison, robotic orthosis found on those robotic gait rehabilitation systems adopt simplified approach to replicate the leg kinematics [57, 103]. Lokomat took a step further by allowing the gait pattern to be set according to the patient’s height and range of motion of the lower limb joint [52]. In another research work, the BWS apparatus restricts pelvic motion, causes the modification of pelvic motion planning. The restricted pelvic motion are planned according to the lower limb motions and desired foot trajectory [58].

Most of the research groups using CGA data as reference trajectory in their gait pattern planning [62, 83, 103, 117, 123]. CGA data is commonly taken from publication of Winter [124]. The CGA data is walking motion of one human subject recorded at one walking speed. It is unclear that how well the CGA data can fit into different individuals. Nevertheless, we believed that gait pattern of each individual should be different due to

the anthropometry characteristics. Furthermore, CGA data is limited to the walking speed of the motion when it is recorded and disallowed the changing of walking speed in gait pattern planning.

Every individual has a certain amount of peculiarities superimposed on the basic pattern of bipedal locomotion during walking. Hence, gait pattern generation plays an important role in robotic gait rehabilitation. The importance of subject specific gait pattern is obvious, as the patients learn the gait pattern imposed to them during the gait rehabilitation session. Improper gait pattern imposed by the robotic system during gait rehabilitation session could result in unwanted actual gait pattern learning outcome. In view of the issue, a methodology is developed to systematically generate a desired gait pattern for robotic gait rehabilitation.

As summarized in the literature review, little research work have been done on gait pattern generation, this chapter presents a methodology, which address the need of gait pattern generation for robotic assisted gait rehabilitation. This methodology facilitate the generation of gait pattern for a patient, without the need of a matching template. The objective of the proposed methodology is to generate a pattern in the context of gait rehabilitation specifically for each subject, based on anatomical parameters of the subject. The methodology can generate gait pattern for given walking speed and cadence, as long as it is physically feasible for the targeted subject. This work could be useful for other researches that need to plan gait pattern. The gait pattern generation methodology aims to generate a gait pattern for a given subject resembling to the pattern obtained from motion capture experiment, for the same gait parameters.

3.1 Gait Pattern Generator – *GaitGen*

Robotic assisted gait rehabilitation system has been designed to induce gait pattern to patient with gait disability during rehabilitation. A few examples of these systems are Lokomat (Hocoma AG, Switzerland) [9], WalkTrainer (Swiss Foundation of Cyberthosis) [67], HapticWalker (Technical University of Berlin) [11], and KineAssist (Kinea Design, LLC) [12]. For robotic assisted gait rehabilitation, walking patterns for the targeted subject must be planned to provide a natural gait-like movement during the gait locomotion training session. To maximize the therapeutic outcome, it is crucial to induce

a gait pattern that resembles natural gait pattern of the targeted subject during gait rehabilitation [36, 40].

In view of this limitations stated in the beginning of this chapter, *GaitGen*, a model for planning of gait pattern, was proposed [125]. The proposed model will generate a gait pattern specifically for each targeted subject. The model consists of three stages as illustrated in Figure 3-1. The first stage works on the prediction of basic gait parameters: walking speed, stride length, and cadence, for each individual. The work on this stage is presented in Section 3.2. Second stage explores the generation of gait pattern based on basic gait parameters. The work on second stage can be found in Section 3.3. Third stage works on translating and coordinating the gait pattern for gait locomotion playback on robotic gait rehabilitation system. With these three stages, the proposed model can generate gait pattern for any targeted individual, for any desired walking speed.

Figure 3-1: Overview of proposed model, *GaitGen*, for gait pattern planning in robotic assisted gait rehabilitation

3.2 Prediction of Individual-Specific Basic Gait Parameters for Over-Ground Walking

Every individual displays certain personal peculiarities superimposed on the basic pattern of bipedal locomotion during walking. Gait pattern is affected by basic gait parameters, such as walking speed, stride length, and cadence [119-122]. Thus, an approach to predict the basic gait parameters of walking for individuals shall be established to facilitate the planning of gait pattern. Preliminary study reveals that walking speed, stride length, and cadence are related to their age, body height, body weight, and lower limb anthropometric parameters. The prediction of gait parameters will be carried out for subject's self-selected slow and normal walking speed.

The objective of this work described in this section is to train a set of Generalized Regression Neural Networks (GRNNs) for the prediction of slow and normal walking speed and corresponding stride length of any given individual. This work will provide a quicker way of obtaining a reference set of basic gait parameters of any individual for

therapists or doctors to facilitate gait pattern planning during gait rehabilitation. Gait parameters database of 50 subjects is used to train the neural network. The accuracy of this work is compared to normalized gait parameters formula [87]. The formula is the only related work for the prediction of gait parameters for normal walking.

3.2.1 Preliminary Study for Verification of Normalized Gait Parameters Equation

The preliminary study is arranged to derive the normalized gait parameters equation. The normal gait data are collected using GAITRite system [126]. Thirty healthy subjects (15 males and 15 females) are recruited from Nanyang Technological University for the study. The subjects are ages ranging 22 to 51 years (mean 33.2 ± 10.0 years), height ranging 150 – 189 cm (mean 167.1 ± 9.9 cm), and weight ranging 37 – 82 kg (mean 58.9 ± 12.9 kg). Gait data collected include walking speed, cadence, and stride length. Experimental data are provided in Appendix - F.

Experiment is carried out to examine if the equation (3-2) was applicable for Asian population. Thirty subjects are instructed to walk barefooted on 6 m long GAITRite at normal walking speed that they are comfortable with. To ensure a natural walking on GAITRite, an additional space of 2.5 m is allocated at the two ends of GAITRite mat. The additional distance also caters for development phase at the start of the walk and decay phase at the end of each walk. Three walk data have been collected for each subject. Data are recorded and computed to obtain the normalized index. As derived from the normalized gait equation, the index is defined as [87]

$$Index = \frac{stride_length}{cadence \times body_height} \quad (3-1)$$

The unit of the parameters is: stride length (cm); cadence (steps/min); body height (cm). The index is calculated with equation (3-1). A new index, L-index, is introduced to study the difference between the indices based on the body height and the leg length. The L-index has been introduced by modifying equation (3-1). By changing the body height in equation (3-1) to the leg length, while keep other parameters the same, the formula is now modified for the new index as

$$L-index = \frac{stride_length}{cadence \times leg_length} \quad (3-2)$$

For differentiation, the index in equation (3-1) is renamed to B-index. L-index is introduced to study the significance of leg length in the normalized gait formula.

The original index and L-index of male and female subjects are presented graphically in Figure 3-2 and Figure 3-3. The indices are plotted with respect to several key parameters. A linear trend line is generated to each graph to illustrate the trend of the index with respect to the corresponding parameter.

Based on the calculated B-index and L-index, the following observations are drawn:

- The cadence (steps/min) for female subjects appears wider in range if compared to the cadence for male subjects (the range of cadence for female subject is 95 – 135 steps/min, while the range of cadence for male subjects is 100 – 120 steps/min.)
- It is found that the B-index is 0.0071 for both male and female subjects, by fitting the data with a linear trend line intercepting at the origin of the graph shown at (a) of Figure 3-2 and Figure 3-3. The associated L-index is 0.014, which is twice of B-index, which is shown in (b) of Figure 3-2 and Figure 3-3.

Figure 3-2: Gait parameters equation verification for 15 male subjects

Figure 3-3: Gait parameters equation verification for 15 female subjects

3.2.2 Methods

This work allows the therapists or doctors to obtain a reference set of suitable gait parameters for the targeted subject by just specifying the desired speed state (normal or slow), and providing the basic information: age, gender, height, and weight, of the targeted subject. Suitable gait parameters will be predicted by the functions with the given information. Subject's self-selected slow and normal walking speeds are considered in the training database of this work. Fast walking speed is not considered, because gait rehabilitation does not involve fast walking speed.

The procedure of the gait parameters prediction is provided in Figure 3-4. To start, the state of walking speed has to be chosen by doctor/therapist. Corresponding trained GRNN of walking speed module will predict a walking speed suitable for the targeted subject at the selected speed mode. Human do not walk at a constant speed at self-

selected walking pace. Usually the walking speed fluctuates within a range. Walking speed of the 50 subjects ranges from 0.25% to 18.84% of the average walking speed of each individual. The cause of the variation is not studied in this work, but we would like to suggest the variation of walking speed can be set at 5% for slow walking and 3.8% for normal walking of each individual's average walking speed. The suggested value is the average of the variation of the 50 subjects. The walking speed module is trained to predict the average walking speed for targeted subject.

Figure 3-4: Overall concept of gait pattern generation in robotic assisted gait rehabilitation

Range of predicted walking speed by *GaitGen* is formulated as

$$walking_speed = predicted_walking_speed \pm suggest_variation \quad (3-3)$$

A desired walking speed can be then selected by doctor/therapist from the range of predicted walking speed, and the selected walking speed will become the input of the stride length prediction module. Stride length prediction module predicts a suitable stride length based on the selected walking speed and the height of the targeted subject. Last

step of the gait parameters prediction is the computation of cadence based on the selected walking speed and mapped stride length. The cadence is computed by

$$cadence = \frac{2 \times 60 \times walking_speed}{stride_length} \quad (3-4)$$

Experimental Setup

The gait parameters were acquired by carrying out an experiment on 50 subjects. 50 subjects (26 male and 24 female) with ages ranging from 17 to 51 years (mean 28 ± 10 years), height 150 to 189 cm (mean 166.2 ± 9.2 cm), and weight 37 to 85 kg (mean 59.6 ± 11.5 kg) were recruited from the population of Nanyang Technological University (NTU), Singapore. The general information of the recruited subjects is presented in Table 3-1. These subjects have neither neurological injury nor gait disorder. Each subject was instructed to walk six times, three times at their self-selected slow walking speed, and three times at their self-selected normal walking speed. GAITRite was used to record walking speed, and stride length of these subjects during their walk. An additional space of 2.5 meters was provided at the two ends of the GAITRite, to provide room for the subject to enter the recording zone at rhythmic walking stage.

Table 3-1: Information of Subjects and Numbers of Walks

Numbers of subjects	Age	Height (cm)	Weight (kg)	Numbers of normal walks	Numbers of slow walks
50	17-51	150-189	37-85	150	150

Design of the Neural Networks

Neural network is a prevalent analysis tool used in gait data studies in the last 10 years [21]. Neural network allows gait data to be analyzed with a highly flexible, non-linear modeling approach. The study of complicated gait variable relationships, which have been difficult to be modeled analytically can be facilitated by the non-linear property of neural network. In this work, GRNNs were trained to predict walking speed and stride length.

Based on existing literatures [3-6, 22] and preliminary analysis, we observed that age, gender, body height, and weight are the key factors influencing the basic gait parameters. This work only considers the factors that can be assessed quantitatively. Descriptive factor (e.g., mood, environment, and foot wear) is not considered, even though these factors can influence the basic gait parameters. The inputs of the GRNNs trained for walking speed modules are age, gender, body height, and weight. Walking speed and height are the inputs to the GRNNs trained for stride length module. These inputs were selected to be the same as that found in the normalized gait parameters formula.

To verify the concept proposed in this work using the relatively small sample size, the GRNNs for average walking speed prediction are trained 50 times using different subject in each training as validation data. The trainings are carried out by using 49 subjects as training data and one subject as validation data. With 50 trainings, the accuracy of GRNN can be validated using all 50 subjects. For GRNNs trained for predicting stride length, it is trained 150 times with all walks data, with every training using different walk data as the validation data.

The smoothing factor plays an important role in function approximation [23, 24]. The function found within GRNN is steep for smoothing factor with low value; therefore, neuron with weight vector closest to the input vector will give a much larger output than other neurons. As smoothing factor becomes larger, several neurons can respond to an input vector so that the function becomes smoother but the neural network becomes less selective. Therefore, optimal values of smoothing factors are required for GRNN. In this study, the smoothing factor was evaluated and selected based on the least value of the average Mean Square Error (MSE) of the predicted outcome. The average MSE are computed for GRNNs designed with smoothing factor values of 0.01 to 1. The smoothing factors selected for each function are summarized in Table 3-2. The example on the effect of smoothing factor measured in term of MSE is given in Figure 3-5.

Table 3-2: Smoothing Factor Selected for Corresponding GRNNs

	Walking Speed	Stride Length
Normal	0.250	0.065
Slow	1.000	0.090

Figure 3-5: Effect of smoothing factor represented by MSE of predicted outcome for normal walking speed

Table 3-3: Mean Square Error of Predicted Outcome

	<u>Walking Speed</u>		<u>Stride Length</u>		<u>Stride Length (Index)</u>	
	Slow	Normal	Slow	Normal	Slow	Normal
Average	270.45	223.82	31.63	19.96	84.58	56.34
Standard Deviation	393.63	334.81	60.12	27.35	112.50	100.00
Maximum	1937.84	1725.52	407.78	145.54	794.23	768.86
Minimum	0.09	0.10	0.00	0.00	0.03	0.02

3.2.3 Results

This session summarizes the results of the outcome predicted by the GRNNs. The MSE of predicted outcome for gait parameters of the 50 subjects is tabulated in Table 3-3. The MSE of stride length computed by using the normalized gait parameters formula is tabulated in Table 3-3 too for comparison with the predicted outcome of stride length by

trained GRNN. Predicted outcomes of the trained GRNNs with highest MSE, least MSE, and average MSE are presented in Table 3-4. The complete result of the predicted outcome is lengthy and is included as Appendix G. The accuracy of trained GRNNs is summarized at Table 3-5.

Table 3-4: Predicted Outcomes with Highest MSE, Least MSE, and Average MSE for the Trained GRNNs

	Age	Gender	Height (cm)	Weight (kg)	Speed (cm/s)	Stride Length (cm)	Cadence (step/min)	Predicted Outcome	MSE
<i>Prediction Outcome of Normal Walking Speed</i>									
A	48	Female	158	43	85.17	109.18	93.60	126.71	1725.52
B	44	Male	168	67	111.09	121.15	110.03	126.00	222.26
C	25	Male	172	63	116.37	132.48	105.47	116.70	0.10
<i>Prediction Outcome of Slow Walking Speed</i>									
D	31	Female	165	65	45.68	105.54	51.77	89.70	1937.84
E	24	Male	185	75	100.63	138.66	87.10	82.73	320.67
F	26	Female	168	53	88.08	122.28	86.43	87.79	0.09
<i>Prediction Outcome of Stride Length (Normal Walking Speed)</i>									
G	25	Male	175	70	121.69	127.54	114.50	139.60	145.54
H	22	Male	162	60	106.49	127.79	100.00	132.31	20.42
I	22	Female	155	59	106.21	119.34	106.80	119.41	0.00
<i>Prediction Outcome of Stride Length (Slow Walking Speed)</i>									
J	48	Male	168	75	91.42	101.11	108.50	121.30	407.78
K	20	Female	172	58	60.50	93.32	77.80	98.72	29.15
L	22	Female	155	59	78.47	105.10	89.60	105.09	0.00

3.2.4 Discussion

For normal walking speed prediction, 32% of the predicted outcomes are at least 95% accurate and 56% of the predicted outcomes are at least 90% accurate. Among the 50 predicted outcomes for slow walking speed, 24% of the predicted outcomes are at least 95% accurate and 45% predicted a outcome which is at least 90% accurate. A factor that could affect the accuracy of predicted outcome for slow walking speed is the instability during slow walking. The variation in walking speed during slow walk compared to normal walk is larger. We also need to remind several subjects to walk at slower speed as they tends to walk at normal walking speed, some of the experimental data might not reflects the real slow walking speed of some subjects.

Table 3-5: Accuracy of Trained GRNNs in Predicting the Basic Gait Parameters

Predicted outcome		Accuracy		
		$\geq 95\%$	$\geq 90\%$	$\geq 80\%$
GRNN	Normal Walking Speed	32%	56%	90%
	Slow Walking Speed	24%	46%	81%
	Stride Length (Normal)	85%	100%	100%
	Stride Length (Slow)	74%	93%	100%
Index	Stride Length (Normal)	63%	95%	100%
	Stride Length (Slow)	39%	79%	98%

The stride length is predicted using walking speed and height as the inputs for the trained GRNNs. The performance of the trained GRNNs is compared to normalized gait parameters formula. The normalized gait parameters formula can be used to compute stride length, if walking speed is known. The trained GRNNs is more accurate in the prediction of stride length compared to normalized gait parameters formula, because normalized gait parameters formula relates stride length to height and walking speed by linear regression. Our result suggests that the relationship between stride length, height and walking speed shall not be modeled using linear regression.

In general, GRNN has poor accuracy in predicting an outcome for target not following the general trend/pattern of the database. By referring to the predicted outcomes summarized in Table 3-4, those subjects with highest MSE of predicted outcome seem to be not following the general pattern of the population. For example, Subjects ‘A’ and ‘D’ are walking at an extremely slow speed, compared to the other subjects walking at the same state of walking speed.

To summarize, 56% and 46% of the predicted outcomes of GRNNs are minimally 90% accurate slow and normal walking speed. The accuracy of GRNNs trained for stride length is higher, 100% and 93% predicted outcomes are at least 90% accurate.

3.3 Individual-Specific Gait Pattern Prediction Model via Generalized Regression Neural Network

Most of the research groups using CGA data as reference trajectory in their gait pattern planning [13-17]. CGA data is commonly taken from publication of Winter [18]. The CGA data is walking motion of one human subject recorded at one walking speed. It is unclear that how well the CGA data can fit into different individuals. Nevertheless, we believed that gait pattern of each individual should be different due to the anthropometry characteristics. Furthermore, CGA data is limited to one recorded walking speed and it does not allow the change of walking speed in gait pattern planning.

The objective of this work is to develop an individual-specific gait pattern prediction model to generate walking gait patterns based on gait parameters (stride length and cadence) and anthropometric data of any targeted subjects. Motion capture experiments of human walking were carried out and gait patterns in one gait cycle were represented by Fourier coefficient vectors. A database which consists of gait parameters, anthropometric data, and Fourier coefficient vectors was constructed. Gait parameters and Fourier coefficient were computed from experimental data. From the database, a non-linear mapping function was designed and implemented to predict the Fourier coefficient vectors from given gait parameters and anthropometric data of target subjects. Finally, the model applies inverse Fourier transform to generate the gait patterns using the predicted Fourier coefficient vectors.

3.3.1 Methods

Each lower limb joint angle waveform in one gait cycle was decomposed into one Fourier coefficient vector which consisted of eleven elements. A database which consisted of gait parameters, anthropometric data and Fourier coefficient vectors was constructed from the experimental data. Subsequently, a gait pattern prediction model for any individual was created based on this database. Non-linear mapping functions were applied to learn the relationship between the input vectors (gait parameters and anthropometric data) and Fourier coefficient output vectors. When the mapping functions are available, predicted Fourier coefficient vectors can be obtained from given gait parameters selected by therapist or doctor, and anthropometric data measured from a specific subject. Inverse Fourier transform is applied to generate lower limb joint angle

waveforms for the subject. In this work, Generalized Regression Neural Network was implemented as the non-linear mapping function. The model of gait pattern prediction for specific subject is illustrated in Figure 3-6.

Figure 3-6: Individual-specific gait pattern prediction model for robotic assisted gait rehabilitation

Experimental Setup

Seventy healthy male subjects with ages ranging from 18 to 33 years (mean 23.6 ± 2.2 years), height 156 - 192 cm (mean 171.1 ± 6.2 cm) and weight 49 - 120 kg (mean 64.3 ± 11.5 kg), were recruited from the general student population at the Nanyang Technological University (NTU), Singapore. These subjects have neither neurological injury nor gait disorder. Each subject was instructed to walk ten times at their self-paced slow and normal walking speeds on a twelve-meter linear walkway. The gait data was collected using an eight-camera motion analysis system (Eagle, Motion Capture Analysis Corporation) and the reflective marker placement scheme for motion capture is illustrated in Figure 3-7 which is the same as in [25]. The capture volume was positioned at the middle of the walkway so that the subjects entered the capture volume at a rhythmic walking stage and data from three complete gait cycles can be obtained. Anthropometric data of each subject was also measured.

Figure 3-7: 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)

Positions of the lower limb joint centers were estimated using the positions of the reflective markers and recorded anthropometric data. Subsequently a segment reference frame in each lower limb segment was defined by the positions of lower limb joint centers and reflective markers. Joint angles are defined as a rotation of the distal segment frame relative to the proximal segment frame [25]. In this work, the lower limb joint rotations in the sagittal plane (flexion-extension) were used to verify the proposed gait pattern prediction model.

Computation of Gait Pattern

A total of fifteen markers are placed on subject's lower limb as per the anatomical positions labeled in Table 3-6 (the anatomical positions are depicted in Figure 3-7). Positions of joint centers and segmental endpoints are estimated with these markers. Subsequently the joint center positions are used to generate segment reference frames.

Joint angles are calculated based on these segmental reference frames [129]. Anthropometric data of all subjects are recorded for the joint angles computation. The respective denotations of the anthropometric data are listed in Table 3-7. The computation of lower limb and pelvic motion will be explained in this section.

Table 3-6: Numbering and Position of the Motion Capture Markers

Marker Number	Denotation	Anatomical Position of the Markers
M1	$\mathbf{P}_1$	Right metatarsal head II
M2	$\mathbf{P}_2$	Right heel
M3	$\mathbf{P}_3$	Right lateral malleolus
M4	$\mathbf{P}_4$	Right mid-shank
M5	$\mathbf{P}_5$	Right femoral epicondyle
M6	$\mathbf{P}_6$	Right mid-thigh
M7	$\mathbf{P}_7$	Right ASIS [#]
M8	$\mathbf{P}_8$	Left metatarsal head II
M9	$\mathbf{P}_9$	Left heel
M10	$\mathbf{P}_{10}$	Left lateral malleolus
M11	$\mathbf{P}_{11}$	Left mid-shank
M12	$\mathbf{P}_{12}$	Left femoral epicondyle
M13	$\mathbf{P}_{13}$	Left mid-thigh
M14	$\mathbf{P}_{14}$	Left ASIS [#]
M15	$\mathbf{P}_{15}$	Sacrum

ASIS - anterior superior iliac spine

Details for the angle calculation are provided in detail in Appendix - C. Four examples of computed lower limb joint angles are presented in Figure 3-8. Three markers placed at right ASIS, left ASIS, and sacrum are used to compute the trajectory of hip joint center (HJC). The position of the HJC ($\mathbf{P}_{HJC,r}$ is the position of right HJC and $\mathbf{P}_{HJC,l}$ is the position of left HJC) can be computed by

$$\mathbf{v}_{Pelvis} = (\mathbf{P}_{14} - \mathbf{P}_7) / |\mathbf{P}_{14} - \mathbf{P}_7| \quad (3-5)$$

$$\mathbf{w}_{Pelvis} = \frac{(\mathbf{P}_7 - \mathbf{P}_{15}) \times (\mathbf{P}_{14} - \mathbf{P}_{15})}{|(\mathbf{P}_7 - \mathbf{P}_{15}) \times (\mathbf{P}_{14} - \mathbf{P}_{15})|} \quad (3-6)$$

$$\mathbf{u}_{Pelvis} = \mathbf{v}_{Pelvis} \times \mathbf{w}_{Pelvis} \quad (3-7)$$

$$\mathbf{P}_{HJC,r} = \mathbf{P}_{15} + 0.598A_2\mathbf{u}_{Pelvis} - 0.344A_2\mathbf{v}_{Pelvis} - 0.290A_2\mathbf{w}_{Pelvis} \quad (3-8)$$

$$\mathbf{P}_{HJC,l} = \mathbf{P}_{15} + 0.598A_2\mathbf{u}_{Pelvis} + 0.344A_2\mathbf{v}_{Pelvis} - 0.290A_2\mathbf{w}_{Pelvis} \quad (3-9)$$

where $\mathbf{u}, \mathbf{v}, \mathbf{w}$ are the unit vector defined for the pelvis, based on position of left ASIS ($\mathbf{P}_{14}$), right ASIS ($\mathbf{P}_{13}$), and sacrum ($\mathbf{P}_{15}$). A_2 is the breadth of ASIS. A set of computed pelvic motion during walking are presented in Figure 3-9.

Table 3-7: Denotations of Anthropometric Data Recorded

Parameter Name	Denotation	
	Left	Right
Waist breadth		A_1
ASIS breadth		A_2
Thigh length	A_3	A_4
Mid-thigh circumference	A_5	A_6
Calf length	A_7	A_8
Calf circumference	A_9	A_{10}
Knee diameter	A_{11}	A_{12}
Foot length	A_{13}	A_{14}
Malleolus height	A_{15}	A_{16}
Malleolus width	A_{17}	A_{18}
Foot breadth	A_{19}	A_{20}

Figure 3-8: Lower limb joint angles during walking (at slow speed), data obtained via motion capture experiment, four walks (indicates by W1, W2, W3, and W4) of same subject

In general, the computed trajectories 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 respect to the anteroposterior axis.

Figure 3-9: Trajectories of pelvic motion during normal gait locomotion, obtained via motion captured experiment. Motion of pelvis in the: (a) x -axis (Anteroposterior direction); (b) y -axis (Mediolateral direction); (c) z -axis (Superior-inferior direction). (Solid line – right side, dashed line – left side)

Extraction of Fourier Coefficient Vectors

Prior to the extraction of Fourier coefficients of the lower limb joint angle waveforms, the cut-off frequency of the data must be determined. Power spectral density analysis was conducted on the collected experimental gait data and the average values are shown in Figure 3-10. The results revealed that over 96.2% of the power of the hip joint, knee joint, and ankle joint angle waveform are contained below 6 Hz. Thus, the cutoff frequency of these signals was chosen to be 6 Hz. The result is agreed with previous studies, which suggested that the cutoff frequency for waking data should be set at about 6 Hz [18, 26].

Figure 3-10: Integral of power of lower limb joints angle waveforms

Each lower limb joint angle waveform in one gait cycle can be decomposed to sinusoidal components as follow:

$$f(t_n) = \frac{1}{2}a_0 + \sum_{p=1}^{N/2} \left[a_p \cos\left(\frac{2\pi pn}{N}\right) + b_p \sin\left(\frac{2\pi pn}{N}\right) \right] \quad (3-10)$$

$n = 0, 1, \dots, N$

where a_p and b_p are the Fourier coefficients, and N is the samples taken for one cycle.

Fourier coefficients can be computed with the following formulas:

$$a_p = \frac{2}{N} \sum_{n=1}^N f(t_n) \cos\left(\frac{2\pi pn}{N}\right), p = 0, 1, \dots, \frac{N}{2} - 1 \quad (3-11)$$

$$a_{N/2} = \frac{1}{N} \sum_{n=1}^N f(t_n) \cos(\pi n)$$

$$b_p = \frac{2}{N} \sum_{n=1}^N f(t_n) \sin\left(\frac{2\pi pn}{N}\right), p = 0, 1, \dots, \frac{N}{2} \quad (3-12)$$

Since the sinusoidal components above 6 Hz are insignificant, the lower limb joint angles for any sample point, n , in one gait cycle can be represented by:

$$f(t_n) = \frac{1}{2}a_0 + \sum_{p=1}^5 \left[a_p \cos\left(\frac{2\pi pn}{N}\right) + b_p \sin\left(\frac{2\pi pn}{N}\right) \right] \quad (3-13)$$

Using the cutoff frequency as 6 Hz, Fourier coefficients vector is formed as:

$$\mathbf{u}_j^i = (a_0^i, a_1^i, a_2^i, a_3^i, a_4^i, a_5^i, b_1^i, b_2^i, b_3^i, b_4^i, b_5^i)_j^T \quad (3-14)$$

where $i = 1, 2, 3, \dots, 700$ stands for the walking trial number and j is the joint index (j = hip, knee, or ankle).

Design of GRNNs

Neural network is a highly flexible and nonlinear modeling approach and has been a prevalent analysis tool used in gait studies in the last 10 years [27]. In this work, generalized regression neural networks (GRNNs) were applied to predict the Fourier coefficient vectors when the gait parameters (stride length and cadence) and the lower limb anthropometric data of the subjects are provided as input vector.

Based on existing literatures [3, 4, 28, 29], we summarize that stride length, cadence, ASIS breadth, thigh length, shank length and foot length are the key factors influencing the lower limb joint angles waveforms. Gait parameters (stride length and cadence) were defined by therapists or doctors while the lower limb anthropometric data of the subject (ASIS breadth, thigh length, shank length, and foot length) were measured by the experimenters.

GRNN is often used as a powerful regression tool because it has a high training speed and a strong non-linear mapping capability [23]. The network consists of an input layer, a radial basis layer and a linear output layer. Each neuron in the hidden layer has a radial basis function to perform the nonlinear transformation from the input space to the hidden space.

Experimental data of all 700 walking trials was divided into a training set and a test set as shown in Table 3-8. The input vector of the GRNN ($\mathbf{p}_j^i$) consists of gait parame-

ters and anthropometric data and the output vector ($\mathbf{q}_j^i$) is a Fourier coefficient vector. GRNNs were trained with input vectors $\mathbf{p}_j^i$ and output vectors $\mathbf{q}_j^i$ from the training set.

Table 3-8: Division of Experimental Data for GRNN

	Number of walks	Input vector	Output vector	Predicted by GRNNs
Training set	600	$\mathbf{p}_j^i$	$\mathbf{q}_j^i$	$\tilde{\mathbf{q}}_j^i$
Test set	100	$\mathbf{v}_j^i$	$\mathbf{w}_j^i$	$\tilde{\mathbf{w}}_j^i$

In this study, the optimal smoothing factors were determined by comparing the distances between the prediction and the actual Fourier coefficient vectors in the training set using cross validation. The obtained optimal smoothing factors for GRNN_{Hip}, GRNN_{Knee}, and GRNN_{Ankle} are respectively to 25.6, 30.1 and 35.6.

3.3.2 Results

The reconstructed and actual lower limb joint angle waveforms associated with the 700 walking trials are described in Figure 3-11. The plots show that the reconstructed waveforms using Fourier coefficient vectors closely fit the actual waveforms. Assessment parameters (ρ , MAD and TAD) were also calculated to measure how close the reconstructed waveforms were to the actual waveforms (Table 3-9). The results show very high correlations (0.99 – 1.00), low MAD ($\leq 0.8^\circ$), and small TAD ($\leq 0.1\%$).

Figure 3-11: Actual lower limb joints angle waveforms and reconstructed waveforms with the Fourier coefficient vectors associated with 700 walks

Table 3-9: Mean (STD) of Assessment Parameters from 700 Walks

	ρ	MAD (Degree)	TAD (%)
Hip	1.00(0.00)	0.0(0.0)	0.0(0.0)
Knee	0.99(0.02)	0.5(0.2)	0.0(0.3)
Ankle	0.98(0.03)	0.8(0.3)	0.1(0.7)

Validation of Gait Pattern Prediction Model

The actual and predicted joint angle waveforms associated with 100 walking trials are shown in Figure 3-12, which indicates that the predicted waveforms have similar pattern as the actual waveforms. In general, the average predicted lower limb joint angle waveforms falls within one standard deviation of actual waveforms. Variations from predicted waveforms exceed one standard deviation of actual waveforms at some instant but the values are less than 4.4° .

Assessment parameters (ρ , MAD and TAD) were calculated to verify the quality of the gait pattern prediction model and to compare the proposed model with the CGA approach [18]. Mean and standard deviation of assessment parameters from 100 walks in test set were calculated and presented in Table 3-10. Based on the values computed, in term of correlation coefficient, the proposed approach score a higher value compared to CGA approach.

Table 3-10: Mean (STD) of Assessment Parameters from 100 Walks in Test Set

	ρ		MAD (Degree)		TAD (%)	
	Gait pattern prediction model	CGA from Winter [18]	Gait pattern prediction model	CGA from Winter [18]	Gait pattern prediction model	CGA from Winter [18]
Hip	0.98(0.03)	0.87(0.07)	3.73(1.64)	7.66(1.78)	29.64(23.13)	64.62(14.73)
Knee	0.97(0.04)	0.85(0.11)	5.41(2.01)	9.28(3.07)	44.94(23.19)	62.29(15.21)
Ankle	0.92(0.07)	0.58(0.34)	3.58(1.44)	6.03(1.68)	24.64(20.47)	47.35(16.86)

Figure 3-12: Actual lower limb joints angle waveforms and predicted waveforms from gait pattern prediction model associated with 100 walks

MAD values highlighted that the average deviation of predicted waveforms from the actual ones ranged from 3.73 degrees to 5.41 degrees. For CGA, the average value of *MAD* is as high as 9.28 degrees, while the lowest is 6.03 degrees. When using gait pattern prediction model, the maximum *TAD* (i.e. 44.94%) and minimum *TAD* (i.e. 24.64%) percentages were found for the knee joint and ankle joint, respectively. These values are much lower compared to those from CGA approach, which ranged from 47.35% to 64.62%.

3.3.3 Discussion

An individual-specific gait pattern prediction model for generating lower limb joint angle waveforms of human walking is presented in this work. The model involves two major components. First, FFT was used to simplify the lower limb joint angle waveforms in one gait cycle. The simplification is able to facilitate the analysis process, whereby each waveform in one gait cycle is significantly reduced to eleven Fourier coefficients. After applying inverse Fourier transforms, more than 96.2 % of the original data was preserved. We felt that it is adequate to present the waveform. The second component of the proposed subject-specific gait pattern prediction model is the non-linear mapping function which predicts Fourier coefficient vectors from given gait parameters and anthropometric data of the subject. The angles waveforms of the hip, knee, and ankle joints can be generated from the predicted Fourier coefficient vector.

The accuracy of the model in predicting gait pattern is assessed by using the correlation coefficient, *MAD* and *TAD*. The three assessment parameters showed that gait pattern prediction model is able to generate gait patterns closer to the actual patterns compared to gait patterns selected from CGA. Specifically, compared to the CGA approach, the proposed prediction model is associated with a higher correlation coefficient, lower *MAD* and lower *TAD*. These findings support that the proposed model is able to generate a gait pattern better than does the CGA approach in terms of resembling the actual gait pattern of the targeted subject.

Individual factors, such as individual anthropometric dimensions, were taken into account when developing the model. Thus, one of the strength of the proposed model is that it is individual specific and able to address inter-individual differences. Figure 3-12

shows that the model presented in this work predicted individual-specific pattern as desired.

In robotics gait rehabilitation, different walking speeds used in the robotics systems can be required at different stages of the training process. For example, slow walking speeds are usually used for training at the beginning stage. When the condition of patient is improved, faster walking speeds can be planned to provide a better therapeutic outcome. Walking speed is one of the key factors that influence the walking gait pattern [28, 29]. Therefore, when different walking speeds are required for the training purpose, suitable gait patterns must be planned for the robotics systems.

The walking speeds can be related to gait parameters (stride length and cadence). Since gait parameters are included in the input vectors of the trained GRNNs, the proposed model can be implemented for planning gait patterns of targeted patient for various walking speed at slow or normal walking (i.e. by changing the stride length and cadence inputted to the trained GRNNs). In this work, the walking speed of 100 walks in the test set is ranging from 73 cm/s to 127.3 cm/s.

In this work, GRNNs were used as regression tool because it has a high training speed and a strong non-linear mapping capability. The proposed model allows prediction of individual-specific gait pattern for various walking speeds. However, GRNNs do not explicitly describe the relationship between input vectors and output vectors. The relationship could be model if an explicit non-linear mapping function can be implemented for the proposed model instead of using GRNNs.

Future work is suggested for exploring this approach for pelvic motion generation. Similar approach is believed working on pelvic motion. Last challenge of this work will be the coordinating the gait patterns (lower limb and pelvic motions) for play back on robotic gait rehabilitation system.

3.4 Concluding Remarks

GaitGen is proposed to address the need of gait pattern generation for robotic assisted gait rehabilitation in this chapter. Two stages have been designed to achieve the aims of the proposed concept. Stage-I of *GaitGen* provides natural gait parameters prediction

specifically for an individual. Stage-II of *GaitGen* predicts the Fourier coefficients for generating the waveform of lower limb joints angle.

GRNNs have been trained for the prediction of basic gait parameters. This work shows that walking speed of an individual can be predicted with trained GRNNs using age, gender, height, and weight as the input. 100% accurate prediction is not possible, since this work is not able to consider all factors that affect gait parameters. Furthermore, some of these factors (e.g., mood, environment, and foot wear) cannot be measured quantitatively. Nevertheless, this work proposed an individual-specific model for prediction of gait parameters, which will allow the prediction of walking speed for individuals. Approximately 86% of the predicted outcome for walking speed is 80% accurate and 97% of the predicted outcome for stride length respectively, is 90% accurate.

The proposed method offers a quick way of obtaining a reference set of gait parameters for each individual. Previously, therapist or doctor can only guess or take a reference set from the available gait data. These gait data are often published for groups, which is often not accurate or suitable for individual. The accuracy of the predicted outcome of the trained GRNNs could be further improved by enlarging the database used to train the GRNNs. However, it should be noted that the accuracy can only be improved to a certain limit because there are still some factors affecting gait parameters which cannot be measured or accounted for.

Stage-II of *GaitGen* is proposed for the prediction of waveform of gait pattern specifically for an individual, given the individual's anthropometry parameters and desired walking speed and cadence. The concept has been verified and compared with the experimental data comprising 700 walks of 35 subjects. The waveform of gait pattern (250 data points) is first decomposed into eleven Fourier coefficients using Fourier Transform. The decomposition effectively reduces the amount of data needed to deal with. GRNNs are trained to correlate the Fourier coefficients and the input factors (i.e. anthropometry parameters and gait parameters).

The generated waveforms from predicted Fourier coefficient vectors were compared to the actual waveforms and Clinical Gait Analysis (CGA) waveforms to illustrate the advantage of the proposed gait pattern prediction model. Three assessment parameters

used for comparison are correlation coefficient, mean absolute deviation (*MAD*) and threshold absolute deviation (*TAD*). The results show that lower limb joint angles waveforms generated by gait pattern prediction model are closer to actual waveforms compared to the CGA waveforms.

A concept for gait pattern generation has been proposed in this chapter. The concept of gait pattern generation for a specific individual based on anatomical parameters is achieved by *GaitGen*. A gait pattern similar to a gait pattern obtained by motion capture experiment can be generated with *GaitGen*. The gait pattern generated by *GaitGen* is comparable to gait pattern obtained by motion capture experiment. *GaitGen* can replace the need of matching template for the aforementioned robotic gait rehabilitation system, which relying on template based gait pattern planning.

The application of *GaitGen* for robotic assisted gait rehabilitation is presented in Section 6.4. CGA data and *GaitGen* generated gait patterns are compared by carrying out test with the *NaTure-gaits*.

This page is intentionally left blank

Chapter 4

ACCOMPLISHMENT OF PROPOSED ROBOTIC ASSISTED GAIT REHABILITATION FEATURES

In Chapter 2, the discussions on existing gait rehabilitation practice in clinical and robotic assisted gait rehabilitation suggested the need for a better gait rehabilitation methodology. With the findings obtained from the review of clinical and engineering aspects, a methodology for robotic gait rehabilitation is proposed to address the limitation and fill in the gap of the current research.

Four essential features are proposed for robotic assisted gait rehabilitation in the methodology. The features are pelvic motion assistance (and body weight shifting), body weight support, over-ground walking, and lower limb motion assistance. A unique feature is proposed for the provision of pelvic motion assistance and body weight support with the same mechanism. As discussed, the pelvic motion assistance in current robotic gait rehabilitation systems is constrained by the body weight support apparatus. Providing both features with one mechanism reduce the motion conflicts between the two.

This chapter describes a new robotic system for the fulfillment of the features proposed for robotic assisted gait rehabilitation. The kinematic model of each module found on the robotic system is presented.

4.1 Proposed Essential Features for Robotic Assisted Gait Rehabilitation

The following essential features (based on conclusion drawn in Chapter 2 and suggestion obtained from doctors and therapists of Tan Tock Seng Hospital) for robotic assisted gait rehabilitation are proposed:

Pelvic Motion and Body Weight Shifting

- Unrestricted pelvic movement plays an important role in normal locomotion and thus it is included as one of the essential features in this work

- Promote body weight shifting during gait rehabilitation (only possible if unrestricted pelvic movement is provided)

Body Weight Support

- Unique body weight support approach, without the restriction of pelvic movement

Progression Method – Over-Ground Walking

- Practicing gait in functional over-ground walking, as opposed to treadmills

Reciprocal Stepping of Lower Limb

- Active assistance to hip, knee and ankle (sagittal plane motion) in order to provide actual functional over-ground walking.

To realize these features, a new robotic gait rehabilitation system is proposed. A robotic gait rehabilitation system is developed for the provision of unrestricted pelvic motion (allows body weight shifting), actuated assistance to the lower limb joints (hip, knee, and ankle), body weight support, and over-ground walking. With the provision of unrestricted pelvic motion, the body weight shifting can be promoted during gait rehabilitation. The proposed features provide an alternative gait rehabilitation methodology, in contrast to cBWS method, and provide a more specific training task of actual walking on ground, better and more complete gait motion guidance, and minimize the alteration of gait pattern [13]. Balance and trunk stability will be maintained by limiting the trunk movement of the subject.

4.1.1 Overview of Proposed Features and Accomplishment of the Features

Each of the proposed features can be achieved by a module on the robotic system. These modules need to synchronize together throughout the operation to achieve the desired gait locomotion. In order to provide over-ground walking, a mobile platform (MP) is incorporated in the system. The mobile platform follows the patient during gait rehabilitation and it carries all other modules with it. A pair of robotic arms for Pelvic Assistance (PA) mechanism is attached on the mobile platform. The end (end effector) of the arm connects to the waist of patient, to provide BWS and pelvic control. Lastly, a lower limb robotic orthosis (RO) is attached to the end of the parallelogram arm, which induces gait locomotion to the patient during training. The conceptual idea of the mod-

ules contained in the robotic gait rehabilitation system is shown in Figure 4-1. The robotic system is named as *NaTure-gaits*, as it allows natural and tune-able gait locomotion during rehabilitation. The overall design of *NaTure-gaits* is provided in Figure 4-2. Details of each module will be explained in the subsequent sections.

Figure 4-1: Conceptual idea of the gait rehabilitation methodology and its realization
(* pelvic motion and body weight shifting)

NaTure-gaits is designed for the anthropometric data listed in Table 4-1. The anthropometric data are referenced from the 5th to 95th percentile of anthropometric data of Singapore population. The subsequent design will be based on these anthropometric requirements. The more specific requirement of each module will be discussed in the following sections.

Table 4-1: Design Requirement based on Anthropometric Data of Targeted User Group

	<i>Minimum</i>	<i>Maximum</i>
Body Weight	-	80kg
Waist Height	840 mm	1000 mm
Shank Length	330 mm	400 mm
Thigh Length	330 mm	400 mm
Hip Breadth	250 mm	550 mm

Figure 4-2: Main modules on *NaTURE-gaits* and corresponding fourteen actuated motions

Figure 4-3: Planes assignment on human body [131]

Coordinates and Plane Assignment

There are three planes assigned to human body. Each of the planes is at right angles, or perpendicular to the other two planes. Sagittal plane divides the body into left and right side. Frontal plane divides the body into front and back parts. Transverse plane also known as horizontal plane, which divides the body into top and bottom parts. Apart from the plane assignment, general coordination assignment adopted in this work is, x and z axes form the sagittal plane, y and z axes form the frontal plane (as shown in Figure 4-3).

Local Coordinates of NaTure-gaits

Three main modules are found on *NaTure-gaits*. Robotic orthosis is the module that provides lower limb assistance, using the lower limb gait locomotion computed in Section 0 as the motion of the robotic mechanism. The reference pelvic motion computed is the motion for the pelvic assistance mechanism and mobile platform.

Figure 4-4: Local coordinates for each module of *NaTure-gaits*

To facilitate the localized planning of gait locomotion, each module is assigned with a reference position/coordinate. The mobile platform is the base for the entire system and the reference position of mobile platform is assigned with respect to the global coordinate. Global coordinate is fixed to the environment. Pelvic assistance mechanism is positioned on top of mobile platform, and the reference position for pelvic assistance mechanism is assigned with respect to the local coordinate of mobile platform. Lastly, robotic orthosis is connected to the end effector of pelvic assistance mechanism. The reference position of the robotic orthosis is assigned at the end effector of pelvic assistance mechanism. The respective reference positions are marked in Figure 4-4. The relative distance between $\mathbf{P}_{MP}$ and $\mathbf{P}_{PA}$ are fixed. The position of $\mathbf{R}_{Hip,g}$ is attached to the end effector of the pelvic assistance mechanism, which is changing with respect to $\mathbf{P}_{PA}$. The end effector position with respect to $\mathbf{P}_{PA}$ is given in Section 4.2.1.

4.2 Realization of Unrestricted Pelvic Motion Assistance and Body Weight Support

Pelvic motion assistance of the current robotic gait rehabilitation is restricted by the cable-harness body weight support apparatus. In order to provide unrestricted pelvic motion, the body weight support method has to be changed. The body weight support feature in this work is similar to what KineAssist is providing. The support force is

provided by the robotic system, which also provides the pelvic motion assistance at the same time.

Pelvis has six movements (the description of the pelvic movements have been made available in Figure 2-17), but only three movements are significant during gait locomotion. The three movements are pelvic rotation, pelvic tilt, and translational mediolateral movement. Based on these requirements, the PA mechanism is designed to provide actuated assistance for **five** pelvic movements. The pelvic rotation about the mediolateral axis is not actuated, but passive movement is allowed. Complete and natural pelvic movements are made possible with these movements.

4.2.1 Pelvic Assistance Mechanism (PA) – Mechanism Design and Kinematic Model

The PA mechanism consists of a pair of robotic arms. Each robotic arm consists of three actuated motions. Each of this motion is achieved by a linear sliding mechanism, actuated by a DC servo motor. The layout and configuration of the robotic arm at the right side are illustrated in Figure 4-5. The left robotic arm is mirror image of the right robotic arm, in term of layout and configuration. The two robotic arms hold the subject at both sides of pelvis as shown in Figure 4-6. The subject is secured to the robotic arm at the attachment plate with a special designed harness.

With the two robotic arms, the three translational movements and two rotational movements of the pelvis can be achieved. The pelvic rotation about the mediolateral axis is made passive, with the range of motion restricted by the harness to $\pm 5^\circ$. The PA mechanism is mounted on a mobile platform. The pelvic motion in the *x-axis* is provided by the combined motions of PA mechanism and mobile platform in the axis.

Figure 4-5: PA module (a) Robotic arm providing pelvic motion in the sagittal plane (b) Lateral shift mechanism of the right side (Arrows indicates the actuated motion)

Figure 4-6: Configuration of robotic arms of pelvic assistance mechanism

Actuators 1 and 2 formed a sub-mechanism which provides the motion in the sagittal plane. Actuator 3 is connected to the end effector of the sub-mechanism (balloon 1 in Figure 4-5) that provides the lateral shifting motion as shown in Figure 4-5. The position of the end effector are given as

$$\mathbf{P}_{PA,g3} = \begin{bmatrix} x_{g3} & y_{g3} & z_{g3} \end{bmatrix} \quad (4-1)$$

The kinematic chain of the sub-mechanism is depicted in Figure 4-7. The sagittal motion (x_{g3}, z_{g3}) of pelvic is provided by the linear motion of s_{g1} and s_{g2}

$$\begin{aligned} s_{g2} &= \sqrt{x_{g3}^2 + z_{g3}^2} \\ \theta_{g2} &= \sin^{-1}(z_{g3} / s_{g2}) \\ s_{g1} &= \sqrt{(x_{g2} - l \cos \theta_{g2} - x_{g0})^2 + (z_{g2} - l \sin \theta_{g2} - z_{g0})^2} \end{aligned} \quad (4-2)$$

and $g = l, r$. The PA variables for the right side is denoted by $g = r$, and for the left side is denoted by $g = l$. The reference point for PA ($\mathbf{P}_{PA}$) is placed at the middle of right and left PA, which is given by

$$\mathbf{P}_{PA} = \frac{\mathbf{P}_{r0} + \mathbf{P}_{l0}}{2} \quad (4-3)$$

$$\mathbf{P}_{g0} = [x_{g0} \quad y_{g0} \quad z_{g0}]^T \quad (4-4)$$

The initial position of PA is at

$$\begin{aligned} \theta_{g1} &= 90^\circ \\ \theta_{g2} &= 0^\circ \end{aligned} \quad (4-5)$$

The lateral movement in y -axis is directed provided by actuator 3 through a rack and pinion mechanism.

Figure 4-7: Kinematic representation of PA sub-mechanism

Figure 4-8: Detailed design of mechanism for lateral pelvic movement actuated by actuator 3, located at the right side of pelvis (Arrow indicates the actuated motion)

PA mechanism consists of six actuators. All six actuators are DC brushless motor, whereby the four motors of the sub-mechanism (actuators 1 and 2) are rated 250 watts and actuator 3 is rated 70 watts. The range of motion for actuators 1 and 2 are 100 mm, and the range of motion for actuator 3 is 80 mm. The range of motion for the actuators is carefully selected to ensure that it fits to the range of motion of pelvic during gait locomotion. The effective and designed workspaces for the sub-mechanism are plotted in Figure 4-9. The designed workspace is 80 mm in the x direction and 100 mm in the z direction. The workspace is created using the end effector position as origin.

Figure 4-9: Effective and designed workspace for sub-mechanism of PA mechanism, with the end effector set as reference coordinates ($x = 0$, $z = 0$ at end effector)

4.2.2 Provision of Pelvic Motion

Pelvis holds by the pelvic assistance mechanism at the side as illustrated in Figure 4-10. The pelvic motion is induced by the combined movements of the left and right modules of the pelvic assistance mechanism. The pelvic orientation and position (consider the center of the pelvis) can be controlled by the position of the two end effectors of the pelvic assistance mechanism, which is given by

$$\theta_{tilt} = \tan^{-1} \left(\frac{z_{l3} - z_{r3}}{y_{l3} - y_{r3}} \right) \quad (4-6)$$

$$\theta_{rot} = \tan^{-1} \left(\frac{y_{l3} - y_{r3}}{x_{l3} - x_{r3}} \right) \quad (4-7)$$

$$\mathbf{P}_{Pelvis} = \frac{1}{2} \begin{bmatrix} x_{r3} + x_{l3} \\ y_{r3} + y_{l3} \\ z_{r3} + z_{l3} \end{bmatrix} \quad (4-8)$$

θ_{tilt} is the angle of pelvic tilt, and θ_{rot} is the angle of pelvic rotation. $\mathbf{P}_{Pelvis}$ denotes the center of the pelvis.

Figure 4-10: Illustration of pelvic motion provision by the pelvic assistance mechanism

4.2.3 Body Weight Shifting and Body Weight Support

During gait locomotion, human constantly shift the body weight from previous supporting leg to the next supporting leg, to facilitate the swinging of the previous supporting leg. Body weight shifting is part and parcel of the pelvic motion. The lateral movement of the pelvic shifts the Center of Mass (CoM) of the body to the supporting leg. The provision of the lateral movement by the proposed pelvic assistance mechanism provides body weight shifting.

This research work suggests an adaptive body weight support approach. Rather than providing a constant body weight support force, a control strategy is designed to alter the magnitude of body weight supporting force, based on the effort of the subject. The control strategy will be discussed in Chapter 5.

4.3 Realization of Over-Ground Mobility for Robotic Assisted Gait Rehabilitation

Mobile platform is the base of *NaTure-gaits*. The main function of the mobile platform is to provide the progression for the entire system and the subject during gait rehabilitation. Mobile platform also serves as the carrier of the robotic orthosis, pelvis assistance mechanism, controller, power source, and any other electronics components of *NaTure-gaits*. Two motorized wheels are mounted at the rear of mobile platform. Each wheel is independently controlled, and it is possible for the mobile platform to follow straight or curved path. Thus the gait rehabilitation is not limited to progression in a straight line only. Figure 4-11 shows the CAD model of mobile platform.

The speed of the mobile platform is designed and customized based on several considerations, such as comfortable normal walking speed of a healthy person and patient. The average of natural walking speed for a healthy adult is 1.5 m/s (Aged 20 – 65 years) [4]. Our gait study of 30 healthy subjects of sampled population (Aged 22 – 51 years) shows average natural walking speed of 1.22 m/s, and slowest natural walking speed of 0.75 m/s is observed.

Based on the feedback from doctors and therapists from Tan Tock Seng hospital, the patient usually admitted to rehabilitation center with average walking speed of 0.1 m/s. By completing balance and weight transfer training, patient are able to walk at 0.3 m/s and patient's walking speed is 0.4 m/s when they discharge, about two months later. The

above information serves as the guideline for the decision of the system's speed. The mobile platform is designed for maximum speed of 0.7 m/s. The system operating speed will be around 0.2 – 0.4 m/s. *NaTure-gaits* does not target high walking speed during training, unlike treadmill based gait rehabilitation systems. Slower walking speed during training allow the practice of weight transfer and postural balancing in the actual walking of the patient.

Figure 4-11: CAD model of mobile platform

4.3.1 Kinematic Model of the Mobile Platform

The kinematic model formulates the key position of the mobile platform with respect to global coordinates (environment). The coordinate system assignment follows the same convention as per previously described in Section 4.1.1.

The variables with subscript '*r*' are subscript assigned for variables of right wheel and variables with subscript '*l*' are subscript assigned for variables left wheel. The denotations for the mobile platform kinematics model are provided in Figure 4-12. Assume no slippage on the wheel

$$c_r = \frac{d\theta_r}{2} \quad (4-9)$$

$$c_l = \frac{d\theta_l}{2} \quad (4-10)$$

where c is the trajectory travelled by wheel based on the angle, θ , rotated. Diameter of the wheel is denoted by d .

The two trajectories, c_1 and c_2 have the same instantaneous center of curvature, $\mathbf{P}_{ICC}$. The radius of the trajectories with respect to the rotational center is denoted as r_1 and r_2 respectively, where

$$r_r = \frac{c_r}{\varphi} \quad (4-11)$$

$$r_l = \frac{c_l}{\varphi} \quad (4-12)$$

Figure 4-12: Denotations for kinematic modeling of mobile platform

The distance between two wheels, l , is constant and can be described as a function of r_1 and r_2

$$\begin{aligned}
l &= r_l - r_r \\
&= \frac{d}{2\varphi}(\theta_l - \theta_r)
\end{aligned} \tag{4-13}$$

The pivoting angle, φ can be computed by

$$\varphi = \frac{d(\theta_l - \theta_r)}{2l} \tag{4-14}$$

With equations (4-9) to (4-14), r_r and r_l can be related to rotational angle of wheel, θ , by

$$r_r = \frac{\theta_r l}{(\theta_l - \theta_r)} \tag{4-15}$$

$$r_l = \frac{\theta_l l}{(\theta_l - \theta_r)} \tag{4-16}$$

Figure 4-13: Time function description for key variable of mobile platform
(The coordinate system is assigned as per Figure 4-3)

In order to obtain the positions for mobile platform as a discrete time function, the mobile platform is modeled for three states in Figure 4-13 for explanation purpose. Each state has a discrete and fix time interval with respect to the previous state. The z -axis component is excluded from the kinematic model, since the wheels of mobile platform do not have any effect on z -axis component, during over-ground movement. Set n as zero, the position of rear wheel on the right side at state t_1 , can be computed by

$$\mathbf{P}_r(t_0) = [y_{r0} \quad x_{r0}]^T \quad (4-17)$$

$$\mathbf{P}_r(t_1) = [y_{r1} \quad x_{r1}]^T$$

$$\begin{bmatrix} -y_{r1} \\ x_{r1} \\ 1 \end{bmatrix} = \begin{bmatrix} \cos \varphi(t_0) & -\sin \varphi(t_0) & r_r(t_1) \cdot (1 - \cos \varphi(t_1)) \\ \sin \varphi(t_0) & \cos \varphi(t_0) & r_r(t_1) \cdot \sin \varphi(t_1) \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} y_{r0} \\ x_{r0} \\ 1 \end{bmatrix} \quad (4-18)$$

The position of rear wheel on the right side at state t_2 , can be computed by

$$\mathbf{P}_r(t_2) = [y_{r2} \quad x_{r2}]^T \quad (4-19)$$

$$\begin{bmatrix} -y_{r2} \\ x_{r2} \\ 1 \end{bmatrix} = \begin{bmatrix} \cos \varphi(t_1) & -\sin \varphi(t_1) & r_r(t_2) \cdot (1 - \cos \varphi(t_2)) \\ \sin \varphi(t_1) & \cos \varphi(t_1) & r_r(t_2) \cdot \sin \varphi(t_2) \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} y_{r1} \\ x_{r1} \\ 1 \end{bmatrix} \quad (4-20)$$

Simplify equations (4-18) and (4-20), which yield

$$\begin{bmatrix} -y_{r1} \\ x_{r1} \\ 1 \end{bmatrix} = \mathbf{T}_{r0} \begin{bmatrix} y_{r0} \\ x_{r0} \\ 1 \end{bmatrix} \quad (4-21)$$

$$\begin{bmatrix} -y_{r2} \\ x_{r2} \\ 1 \end{bmatrix} = \mathbf{T}_{r1} \begin{bmatrix} y_{r1} \\ x_{r1} \\ 1 \end{bmatrix} \quad (4-22)$$

and

$$\begin{bmatrix} -y_{r2} \\ x_{r2} \\ 1 \end{bmatrix} = \mathbf{T}_{r1} \mathbf{T}_{r0} \begin{bmatrix} y_{r0} \\ x_{r0} \\ 1 \end{bmatrix} \quad (4-23)$$

The general form for identify the position of rear wheel on the right side at any state j can be computed by

$$\begin{bmatrix} -y_{rj} \\ x_{rj} \\ 1 \end{bmatrix} = \mathbf{T}_{r(j-1)} \mathbf{T}_{r(j-2)} \dots \mathbf{T}_{r0} \begin{bmatrix} y_{r0} \\ x_{r0} \\ 1 \end{bmatrix} \quad (4-24)$$

and

$$\mathbf{T}_{rk} = \begin{bmatrix} \cos \varphi(t_k) & -\sin \varphi(t_k) & r_r(t_{k+1}) \cdot (1 - \cos \varphi(t_{k+1})) \\ \sin \varphi(t_k) & \cos \varphi(t_k) & r_r(t_{k+1}) \cdot \sin \varphi(t_{k+1}) \\ 0 & 0 & 1 \end{bmatrix} \quad (4-25)$$

Similarly the position for the left wheel can be computed by

$$\begin{bmatrix} -y_{lj} \\ x_{lj} \\ 1 \end{bmatrix} = \mathbf{T}_{l(j-1)} \mathbf{T}_{l(j-2)} \dots \mathbf{T}_{l0} \begin{bmatrix} y_{l0} \\ x_{l0} \\ 1 \end{bmatrix} \quad (4-26)$$

and

$$\mathbf{T}_{lk} = \begin{bmatrix} \cos \varphi(t_k) & -\sin \varphi(t_k) & r_l(t_{k+1}) \cdot (1 - \cos \varphi(t_{k+1})) \\ \sin \varphi(t_k) & \cos \varphi(t_k) & r_l(t_{k+1}) \cdot \sin \varphi(t_{k+1}) \\ 0 & 0 & 1 \end{bmatrix} \quad (4-27)$$

and reference position of mobile platform ($\mathbf{P}_{MP}$) is given by

$$\mathbf{P}_{MP} = \frac{\mathbf{P}_r + \mathbf{P}_l}{2} \quad (4-28)$$

An exceptional case is when both wheels rotates the same angle, where

$$\begin{aligned} \theta_r &= \theta_l = \theta \\ r_r &= r_l = \infty \\ \varphi &= 0 \end{aligned} \quad (4-29)$$

and equations (4-25) and (4-27) becomes (g is the variables to identify left wheel, l , or right wheel, r)

$$\mathbf{T}_{gk} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & r_g(t_{k+1}) \cdot \sin \varphi(t_{k+1}) \\ 0 & 0 & 1 \end{bmatrix} \quad (4-30)$$

and

$$\begin{aligned}
r_g(t_{k+1}) \cdot \sin \varphi(t_{k+1}) &= \frac{(\theta_g)l}{(\theta_r - \theta_l)} \cdot \frac{(\theta_l - \theta_r)d}{2l} \\
&= \frac{(\theta_g)d}{2} \\
&= \frac{(\theta)d}{2}
\end{aligned} \tag{4-31}$$

for $\varphi = 0$, $\sin \varphi = \varphi$. The reference position of mobile platform ($\mathbf{P}_{MP}$) for both wheel traveling with the same rotational angle, can be found by

$$\mathbf{P}_{MP}(t) = \frac{(x_r + x_i + 2\theta d)i + (-y_r(t) - y_l(t))j}{2} \tag{4-32}$$

4.4 Provision of Reciprocal Lower Limb Stepping with Robotic Orthosis (RO)

The provision of reciprocal lower limb stepping is commonly found in most of the current robotic gait rehabilitation. In this research work, a similar robotic orthosis has been implemented for the provision of reciprocal lower limb stepping, with actuated ankle joint. Most of the developed system, such as Lokomat, LOPES, ALEX, and POGO do not incorporate actuated ankle joint into their robotic orthosis.

4.4.1 Robotic Orthosis (RO)

Robotic orthosis is the module in *NaTure-gaits* providing actuated assistance to complete the gait locomotion of the lower limb. Six primary degrees of freedom (DoFs) are found on human leg, three at the hip, one at the knee, and two at the ankle. The determination of the DoFs required for an orthosis depends on the task requirement. To achieve a relatively normal walking locomotion, the hip and knee flexion are necessary. If the forward propulsion is to be provided by foot alone (not by upper body with clutch or other walking aids), the plantar flexion of the ankle is also required [79]. For this study, after literature review and discussion with clinical practitioners, the robotic orthosis is designed to provide actuated assistance in the sagittal plane, for hip, knee, and ankle joints.

Table 4-2: The Designed Range of Motion (ROM) for Robotic Orthosis
(Presented with Permissible ROM of Human and ROM of Human during Walking)

		<i>Range of Motion (Sagittal plane)</i>		
		<i>Permissible</i> [132]	<i>Walking</i> [133]	<i>Designed</i>
Hip	Flexion	122°	20°	50°
	Extension [#]	10°	11°	15°
Knee	Flexion	134°	60°	80°
	Hyperextension	0°	0°	0°
Ankle	Dorsiflexion	13°	9°	10°
	Plantarflexion	56°	18°	20°

[#] The permissible range of hip extension is 10° as recorded in reference [132]. As recorded in reference [134], the permissible range has a range of 10° to 30°. The discrepancies of the value could be due to measuring method. The designed ROM for hip extension is 15°.

The robotic orthosis has a total of six degrees of freedom, three on the right side and three on the left side. The length of the robotic orthosis for the thigh and shank can be adjusted to cater to different subjects. The conceptual design of the robotic orthosis is shown in Figure 4-16. The range of motion (ROM) designed for each joint is provided in Table 4-2. The ROM is limited by hardware limiter. A particular joint will not travel beyond the designed ROM.

The robotic orthosis is divided into three sub-modules for each side of the lower limb: hip module, knee module, and ankle module. The sub-modules for left side are the mirror image of the right side, in term of the design. The description provides subsequently is applicable for both sides. The hip module (cad model displayed in Figure 4-14a) provides actuated assistance to the hip joint, by moving the thigh. The hip module is powered by DC brushless motor rated 70 watts. The knee module (cad model displayed in Figure 4-14b) provides actuated assistance to the knee joint, by moving the shank. The DC brushless motor that powers the knee module is rated 70 watts too. Hip and knee modules are driven by the motor through a bevel gear, and the axis of rotation is aligned to the respective joint of the user. The hip, knee and ankle modules are connected by telescopic linkage, which allow the distance between hip joint and knee joint, and distance between knee joint and ankle joint to be adjusted.

Figure 4-14: CAD model of robotic orthosis
(a) Hip module, (b) Knee module, (c) Ankle module

Ankle module (cad model displayed in Figure 4-14c) consists of the lower half of shank and the foot plate. The foot plate is separated into two portions, and held together by spring hinges at the separation. This design is to mimic the metatarsal joint of the foot, which can allow a more natural toe off motion. The actuation of the ankle joint is

achieved by moving the foot plate via a linear mechanism. The linear mechanism is powered by a 200 watts DC brushless motor. With the specially designed mechanism, the ankle joint angle is related to the length of the linear actuator by

$$\begin{aligned} l_3 &= \sqrt{(l_1)^2 + (l_2)^2} \\ s_2 &= \sqrt{(l_4)^2 + (s_1)^2} \end{aligned} \quad (4-33)$$

$$\begin{aligned} \theta_1 &= \tan^{-1} \left(\frac{l_1}{l_2} \right) \\ \theta_2 &= \cos^{-1} \left[\frac{(l_3)^2 + (l_5)^2 - (s_2)^2}{2(l_3 \times l_5)} \right] \\ &= \cos^{-1} \left[\frac{(l_1)^2 + (l_2)^2 + (l_5)^2 - [(l_4)^2 + (s_1)^2]}{2 \left(\left[\sqrt{(l_1)^2 + (l_2)^2} \right] \times l_5 \right)} \right] \end{aligned} \quad (4-34)$$

$$\theta_{ankle} = 90^\circ - \theta_1 - \theta_2 \quad (4-35)$$

where the denotation are given in Figure 4-15. θ_{ankle} is the angle of ankle joint, positive value indicates dorsiflexion and negative value indicates plantarflexion.

Figure 4-15: Denotation of parameters for ankle module kinematics

Figure 4-16: Conceptual design and features of robotic orthosis for *NaTure-gaits*

4.5 *NaTure-gaits* – Completed Prototype

The three main modules, robotic orthosis, mobile platform, and pelvic assistance mechanism are developed and assembled. The complete prototype is shown in Figure 4-17. Extra space is provided at the rear of mobile platform, to house the control systems and batteries. A safety structure is made available on *NaTure-gaits* for safety consideration. The safety structure is used to secure the safety rope, connected to the harness. The safety rope can stop the patient from falling, as a preventive measure. *NaTure-gaits* is attached to subject as shown in Figure 4-18.

Figure 4-17: Prototype of *NaTure-gaits*

Safety rope prevent as extra measure to prevent the subject from falling

Pelvis of subject is hold by pelvic assistance mechanism through a special design harness, pelvic motion is provided as described in Section 4.2

Locomotion of lower limb is provided by the robotic orthosis by strapping the subject's lower limb to robotic orthosis at thigh and foot

Forward progression provided by mobile platform

Figure 4-18: Attachment of *NaTure-gaits* to subject

4.5.1 Control Modules and Power Source

The control modules of *NaTure-gaits* are grouped as control system of PA, control system of MP, and control system of RO. These three systems are synchronized and controlled by host computer via a wireless communication. The controller of PA is developed in-house and powered by microchip processor. It controls a total of six DC-brushless servo motors. Galil motion control card is used as controller of MP and controller of RO. For MP, the Galil motion control card controls two DC brushed motors. For RO, the Galil motion control card controls six DC brushless motors. Figure 4-19 shows the basic formation of the control systems on *NaTure-gaits*.

Two set of batteries are placed in *NaTure-gaits*. The first set has four lead acid batteries, which provides 48 volts. This set of battery powers twelve motors of the PA and RO. The second set of batteries is 24 volts, provided by two batteries connected in series. The second set of battery powers the two motors of MP and controls systems. The batteries are lead acid batteries, and each battery is rated 12 volts.

Figure 4-19: Control system of pelvic assistance mechanism, mobile platform, and robotic orthosis

4.6 Concluding Remarks

As the first stage of this research work, *NaTure-gaits* is developed for the realization of proposed gait rehabilitation. Features such as unrestricted pelvic motion assistance, lower limb motion assistance, body weight support, body weight shifting, and over-ground walking experience are made possible with the developed modules. The developed robotic system is completed with necessary modules and control systems for the implementation of control strategy in the future. The motion play back ability of the robotic system is validated based on the visual inspection. Human subject test will be presented in Chapter 6.

4.6.1 Features of *NaTure-gaits*

Three key modules have been developed for *NaTure-gaits*, which are given name as robotic orthosis, mobile platform, and pelvic assistance mechanism. These three modules enable the developed robotic gait rehabilitation system to provide body weight supported gait rehabilitation, with the assistance to hip and lower limb movement, in the context of

over-ground walking. The unique feature found on *NaTure-gaits* is the provision of pelvic motion assistance and body weight support with the pelvic assistance mechanism. *NaTure-gaits* is one out of the two robotic systems, which provide gait rehabilitation in the context of over-ground walking.

Table 4-3: Pelvic Movement Assistance of *NaTure-gaits* Compared to Pelvic Movement Assistance Available on other Robotic Gait Rehabilitation Systems

		<u>Active Assistance</u>			<u>Passive Movement</u>	
	<i>NaTure-gaits</i>	<i>PAM</i>	<i>WalkTrainer</i>	<i>LOPES</i>	<i>ALEX/Lokomat</i>	<i>KineAssist</i>
Type of Actuator	DC motor	Pneumatic	DC motor	DC motor	NA	NA
BWS	sBWS	cBWS	cBWS	cBWS	cBWS	sBWS
Translational Pelvic Movement						
Mediolateral	Active	Active	Active	Active	Fixed	Passive
Antero-posterior	Active	Active	Active	Active	Fixed	Passive
Superior-inferior	Active	Active	Active	Passive	Passive	Fixed
Rotational Pelvic Movement						
Pelvic rotation	Active	Active	Active	Fixed	Fixed	Passive
Pelvic tilt	Active	Active	Active	Fixed	Fixed	Passive
Pelvic rotation about medio-lateral axis	Passive	Fixed	Active	Fixed	Fixed	Passive

Active – Actuated movement; Passive – Not actuated by free movement allowed; Fixed – No movement allowed

HapticWalker is not included in this table, since pelvic movement is not allowed

A pelvic assistance mechanism is developed for pelvic motion assistance and body weight support. The mechanism is a five DoF robotic arms provide the essential movements for the pelvis. Without the restriction of cable harness body weight support, natural pelvic motion can be induced by pelvic assistance mechanism and also promoting body weight shifting. A six DoF robotic orthosis is developed for the lower limb motion assistance. It has the similar features of lower limb orthosis found on most robotic gait rehabilitation system. The lower limb joints' (hip, knee, and ankle) movement in the sagittal plane is assisted by the robotic orthosis. For a quick glance,

comparison is made for the two modules with the similar modules found on current gait rehabilitation systems. The comparison tables are presented in Table 4-3 to Table 4-5.

Table 4-4: Body Weight Support Feature of *NaTure-gaits* Compared to State of the Art Gait Rehabilitation System

	<i>NaTure-gaits</i>	<i>Lokomat</i>	<i>WalkTrainer</i>	<i>KineAssist</i>
Name/Brand	PA mechanism	Lokolift	-	-
Unloading Capacity				
Static	120 kg	150 kg	-	159 kg
Dynamic	80 kg	80 kg	-	68 kg
Other Features				
Unloading Mechanism	Motor	Motor and Spring	Motor and Spring	Motor
Control System	Robust Control System	Robust Control System	Active Control System	-

Field marked with - indicates information is not available

HapticWalker do not have body weight support feature incorporated

Table 4-5: Features of Robotic Orthosis of *NaTure-gaits* Compared to Lower Limb Orthosis of State of the Art Gait Rehabilitation System

	<i>NaTure-gaits</i>	<i>Lokomat</i>	<i>WalkTrainer</i>	<i>HapticWalker</i>
Type of Actuator	DC motor	DC motor	DC motor	Robotic Arm
Actuated Joint				
Sagittal	Hip, Knee and Ankle	Hip and Knee	Hip, Knee, and Ankle	Translation in sagittal plane and rotation about the axis perpendicular to sagittal plane
Frontal	-	-	-	
Transverse	-	-	-	
Total DoF	Six	Four	Six	Six

Field marked with - indicates information is not available

KineAssist is excluded since it has no lower limb orthosis

This page is intentionally left blank

Chapter 5

CONTROL STRATEGY FOR ROBOTIC ASSISTED GAIT REHABILITATION

The proposed robotic gait rehabilitation system, *NaTure-gaits*, consists of robotic orthosis, pelvic assistance mechanism, and mobile platform. These three modules have to work together in a synchronized manner to ensure the desired gait pattern is provided to the patient. Position control is the most basic control strategy used for all these modules working together, which is presented in the first part of this chapter. Position control cannot provide the optimized therapeutic outcome. A clear limitation of position control is that patient is forced to comply with the predefined gait pattern. Patient can remain totally passive when having training with this control strategy. This greatly reduces the therapeutic outcome, as no effort is contributed by the patient during the rehabilitation. As such, the second part of this chapter explores the research work on encouraging active assistance of patient motion during rehabilitation.

An overall concept of control strategies is proposed for patient grouped in this research work based on their physical conditions. Lower limb condition and upper trunk control ability are the main considering factors when dividing the patient into groups. In general, the grouping facilitates the assignment of control strategies.

This research work focuses on the development on a control strategy for active pelvic motion assistance with adaptive body weight support. The concept of the control strategy is explained subsequently. The performance and stability of the developed control strategy will be tested by implementing on a preliminary experimental setup.

5.1 Patients Oriented Control Strategies

Stroke and spinal cord injury patients exhibit a complex gait deficit, depending very much on the severity of the lesion. Patients are categorized at a general level based on their lower limb condition after the lesion, which subsequently a robotic assisted gait rehabilitation control strategy are suggested according to the categories.

5.1.1 Categorization of Patients

Stroke and SCI are the greatest cause of gait deficit. Categorizing these two groups of patient into a specific group for gait rehabilitation protocol planning according to their illness type and injury level may not be the most efficient method. This conclusion is drawn based on discussion with experienced therapists and literature review findings.

At a general level, the physical condition of stroke, SCI, and any patient with gait deficit can be divided into symmetric and asymmetric gait condition. This serves as general guideline that the physical condition of the patient is suitable for symmetric or asymmetric gait rehabilitation pattern.

The range of motion (ROM) of the patient's lower limb joints will be tested under no load condition. If the patient retains the ROM as per a healthy person, the patient will be categorized as symmetric gait. Symmetric gait indicates that the left and right gait pattern is repeatable in each gait cycle, with insignificant difference.

If the patient exhibit spasticity, frozen joint, or any physical constraint that prevent the lower limb joints from achieving normal ROM, they will belong to asymmetric gait group. Asymmetric gait is the opposite of symmetric gait, whereby left and right gait pattern is significantly not similar. This group of patient will be excluded from the targeted patient group for this research project. The asymmetric gait condition can be further divided into compensatory gait and severely altered gait.

For example, a hemiplegia patient will exhibit asymmetric gait pattern if he is walking using own effort with walking aids, since one side of the body is paralyzed. However, if the patient lower limb does not exhibit any joint constraint resulting from sequela of the illness/lesions, like spasticity, the patient can be classified as symmetry gait under the classification algorithm. Symmetry rehabilitation walking protocol should be arranged.

Compensatory gait indicates that the physical condition allows the patient in walk in compensatory gait pattern only. Frozen of knee joint requires the patient to lift the pelvic higher in order to clear the ground during swing phase. This is a typical example for compensatory gait group. Severely altered gait group walk with significantly altered gait pattern, for example, scissors gait.

Under symmetric gait condition, it is eventually categorized into four groups, based on the lower limb condition and ability of trunk control. Two groups were defined for lower limb condition. Weak, self-support-able group is the group of patient, who are able to support their own body weight with or without any walking aids. However, they are not able to carry out complete gait cycle due to weakness of the lower limb. Weak, cannot self-support group is the group that cannot support their own body weight with or without any walking aids.

Trunk control condition is divided into three groups, from good trunk control, moderate trunk control to no trunk control. This grouping highlights the trunk control ability of the patient.

The four final groupings are:

G1 – No trunk control and cannot self-support due to weaken lower limb muscle

G2 – Moderate trunk control and cannot self-support due to weaken lower limb muscle

G3 – Good trunk control and cannot self-support due to weaken lower limb muscle

G4 – Good trunk control and can support own body weight for quiet standing

It should be highlighted that the severity of gait disability decreases with the increment of grouping number. The overall algorithm of patient categorization is shown in Figure 5-1. Rehabilitation protocol for the corresponding categorized groups will be discussed subsequently. The group of patient in the symmetric gait group is the targeted users for this research work.

Figure 5-1: Patient categorization algorithm

5.1.2 Rehabilitation Strategies for Patients

Rehabilitation strategies are designed for the different groups of patient, considering their gait deficit after lesion. As discussed, the patient can be briefly divided into four groups. Rehabilitation strategies are created for these four groups based on their physical conditions. It is noteworthy that in real life, patients' actual condition could be more complex to be grouped into these four groups only.

Considering the three key functions provided by *NaTure-gaits*, lower limb assistance, pelvic motion assistance, and body weight support, the rehabilitation strategies are planned with these functions. The over-ground walking is coupled with pelvic motion assistance in rehabilitation strategies planning. In general, each of these functions can

have two basic rehabilitation strategies, namely, passive assistance and active assistance. The combination of these basic rehabilitation strategies of each function creates an overall rehabilitation strategy.

The basic idea of the basic rehabilitation strategies for each function is described as follow:

- *Passive assistance for lower limb motion*

Predefined gait locomotion will be induced to the patient, and patient has to follow the motion completely.

- *Active assistance for lower limb motion*

The robotic orthosis will follow the patient motion, and will not hinder the voluntary movement of the patient. The robotic orthosis will provide assistance only if the patient cannot complete the desired gait locomotion

- *Passive assistance for pelvic motion*

Similar to passive assistance of lower limb motion, predefined pelvic motion will be provided to the patient.

- *Active assistance for pelvic motion*

The active assistance of pelvic motion promotes the active participation of the patient. A reference trajectory is provided but the patient can have a certain extent of influence to the reference trajectory, thus the pelvic motion will not be restricted by the reference trajectory.

- *Passive assistance for body weight support*

Support force will be provided in a percentage of body weight to the patient. The support force will be kept constant throughout the training.

- *Active assistance for body weight support*

The support force will be varied based on the posture of the patient. The support force will be gradually increased when the patient is deviated too much from the reference trajectory. This will encourage the patient to move toward the reference trajectory.

As a basic guideline, the rehabilitation strategies for the four groupings of patient are planned based on their physical condition. It is illustrated in Figure 5-2.

Figure 5-2: Proposed possible rehabilitation strategies for the patients grouped according to this research work
(Solid line indicates passive assistance and dashed line indicates active assistance)

The totally passive rehabilitation strategy is proposed for patients in G1 and G2. This control strategy is also known as position control strategy. This is the most basic and common control strategy and it has been implemented with *NaTure-gaits* and presented in Section 5.2. This control strategy is especially useful for those patients with much severe gait deficit. Active pelvic assistance control strategy provides active assistance for the pelvic motion, and body weight support and lower limb motion are passive assistance. This control strategy is recommended for the patients in G3, since patient effort of pelvic motion is required for active assistance to work.

Various Control strategies had been proposed for active assistance (review are presented in Section 2.8. Those control strategies are designed based on the need of

different approaches of active assistance (detailed literature review is provided in Section 2.8). For this research work, a control strategy is developed for the provision of active pelvic motion assistance with adaptive body weight support. The control strategy is designed for the patients in G4. The active assistance of lower limb is not explored in this research work.

5.2 Passive Control Strategy

Majority of the patients are weak and passive control strategy is crucial for this group of patient. This section provides the explanation on the provision of passive control strategy for gait locomotion assistance for walking in a straight line. Passive control strategy also refers to as position control strategy, since robotic system provides the motion assistance based on pre-defined trajectories by position control. The predefined trajectories for *NaTure-gaits* are obtained by carrying out motion capture experiment. The process of converting the motion capture data to the motion for *NaTure-gaits* is discussed.

5.2.1 Human Gait Locomotion Acquisition via Motion Capture Experiment

NaTure-gaits provides gait training for over-ground walking, specifically, the assistance of lower limb and pelvic. The previous section discussed about the hardware of *NaTure-gaits*. In order to realize the gait locomotion on *NaTure-gaits*, motions of lower limb and pelvic during over-ground walking are required. The basic information for human walking is provided in Appendix - B, the basic terminology of human walking will be used frequently in the subsequent sections.

Lower limb and pelvic trajectories can be obtained from literature or publication [4, 135]. However, the available motion of lower limb and pelvic trajectories is not discussed together in the literature or publication. Furthermore, the pelvic motions are often described as pelvic rotation, pelvic tilt, and pelvic rotation about the mediolateral axis. Whereby in this work, the PA mechanism is designed to support and provide actuated pelvic motion assistance at two sides of the pelvis. In other words, the pelvic movements for both sides are required for this work.

Thus, motion capture experiment is carried out on human subject to obtain the reference walking motion for *NaTure-gaits*. 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 are explained in Section 0.

Gait rehabilitation sessions usually vary in duration from several minutes to half an hour, subject to the physical ability of the patient. For a ten minutes gait rehabilitation session, it is equivalent to 240 gait cycles if each gait cycle is 2.5 seconds (gait cycle of a healthy person is approximately 1.2 seconds). Technically, the feasibility of capturing the entire walking motion as the input for robotic gait rehabilitation system is ruled out.

To obtain a repetitive locomotion in robotic gait rehabilitation, the motion of one cycle for pelvic and lower limb can be duplicated for several repetitions as desired. Walking locomotion during rhythmic stage is remarkably consistent (example is shown in Figure 3-8), thus motion of one gait cycle obtained during rhythmic stage can be reproduced to several consecutive gait cycles. Even though the walking motion is remarkably consistent, it does not follow one single motion cycle exactly during the rhythmic phase. The slight change in landscape, stride length, cadence, and other factors can cause small variation in the walking motion.

Lower Limb Locomotion

The acquisition of lower limb locomotion by motion capture experiment has been explained in Section 0. Robotic orthosis provides the lower limb locomotion. The locomotion is trajectory of the hip, knee, and ankle joints. 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 obtained via motion capture experiment (from one of the walk). It is noticeable that the same gait cycle does not connect smoothly inter cycle, if it is duplicated to form several consecutive gait cycles. Figure 5-3 shows three gait cycles constructed using the same gait cycle, by duplicating it. It is most visible for hip joint angle, where 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.

Figure 5-3: Several gait cycles constructed by duplication of same gait cycle ('A' marked the interconnection between two cycles, which does not connect smoothly)

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. The filter does not affect the data at 0% and 300% in Figure 5-3. 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 and displayed in Figure 5-4. The final constructed lower limb joint angles can be used as reference trajectories for robotic orthosis of *NaTure-gaits*.

Figure 5-4: Post processed lower limb joint angles constructed for the right side (three gait cycles), original data obtained via motion capture experiment

Pelvic Motion

The trajectories presented in Section 0 are computed for the hip joint center (HJC). PA mechanism is designed to hold at both sides of the pelvis, and induce the motion to the patient through the holding points. The computed HJC trajectories are needed to transfer to the side of the pelvis as shown in Figure 5-5. The computed HJC trajectories can be translated to the point, where PA is holding at, by

$$\begin{aligned}
 \mathbf{u}_{Hip} &= \frac{\mathbf{P}_{HJC,l} - \mathbf{P}_{HJC,r}}{|\mathbf{P}_{HJC,l} - \mathbf{P}_{HJC,r}|} \\
 \mathbf{P}_{Hip,r} &= \mathbf{P}_{HJC,r} - \left(\frac{A_1 - |\mathbf{P}_{HJC,l} - \mathbf{P}_{HJC,r}|}{2} \right) \cdot \mathbf{u}_{Hip} \\
 \mathbf{P}_{Hip,l} &= \mathbf{P}_{HJC,l} + \left(\frac{A_1 - |\mathbf{P}_{HJC,l} - \mathbf{P}_{HJC,r}|}{2} \right) \cdot \mathbf{u}_{Hip}
 \end{aligned} \tag{5-1}$$

$$\begin{aligned}
 \mathbf{P}_{Hip,r} &= \begin{Bmatrix} x_{Hip,r} & y_{Hip,r} & z_{Hip,r} \end{Bmatrix}^T \\
 \mathbf{P}_{Hip,l} &= \begin{Bmatrix} x_{Hip,l} & y_{Hip,l} & z_{Hip,l} \end{Bmatrix}^T
 \end{aligned} \tag{5-2}$$

where $\mathbf{u}_{Hip}$ are the unit vector defined for this translation. Position of holding point of PA mechanism at right side is denoted by $\mathbf{P}_{Hip,r}$, and at left side is denoted by $\mathbf{P}_{Hip,l}$. A_1 is the breadth of waist.

Figure 5-5: Motion assistance provided by PA mechanism. The relationship between motion assistance provided by PA mechanism and computed HJC trajectories are highlighted

Using the same approach discussed for lower limb joint angles construction, to obtain a repetitive locomotion in robotic gait rehabilitation, the motion of one cycle will be repeated for a desired repetition. In this study, we assumed that the pelvic motion of the left and right side are symmetrical. 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 assistance mechanism of *NaTure-gaits*, are shown in Figure 5-6. The reference trajectories are given as

$$\begin{aligned} \mathbf{R}_{Hip,r} &= \left\{ x'_{Hip,r} \quad y'_{Hip,r} \quad z'_{Hip,r} \right\}^T \\ \mathbf{R}_{Hip,l} &= \left\{ x'_{Hip,l} \quad y'_{Hip,l} \quad z'_{Hip,l} \right\}^T \end{aligned} \quad (5-3)$$

Figure 5-6: Post processed pelvic motion constructed for the right side (three gait cycles), original data obtained via motion capture experiment

Figure 5-7: Reference trajectories constructed for both lower limbs
(Solid – right lower limb, dashed – left lower limb)

5.2.2 Lower Limb Locomotion Play Back with Robotic Orthosis

The reference gait locomotion of lower limb is constructed for the right lower limb. The lower limb joint angles for the left side can be constructed using the trajectories of the right side with 50% phase lag. The constructed reference gait locomotion for both lower limbs is presented in Figure 5-7. The constructed reference trajectories for both lower

limbs are converted to input for controller of the robotic orthosis. Robotic orthosis of *NaTure-gaits* are designed to provide the lower limb joint angles directly, except for the ankle joint. The joint angle of hip and knee joints do not need kinematic model to convert the joint angles to controller motion input. Hip and knee joints of robotic orthosis are directly driven by motor through connecting gear head. Ankle joint is driven by linkages. Kinematic model is used to convert the ankle joint motion to input for the controller. The kinematic model is presented in Section 4.4.1. The flow chart presented in Figure 5-8 describes the process of the conversion of human gait motion to motion inputs for the controller of robotic orthosis.

Figure 5-8: Overview of conversion of constructed reference trajectories to input for the controller of robotic orthosis

5.2.3 Pelvic Motion Play Back with Pelvic Assistance Mechanism and Mobile Platform

For pelvic motion to be assisted by *NaTure-gaits*, we have to look from different perspective. The reference trajectories ($\mathbf{R}_{Hip,r}$ and $\mathbf{R}_{Hip,l}$) of pelvis constructed in Section 5.2.1 are the movement at the side of pelvis. The final constructed trajectories of pelvis are given as

$$\mathbf{R}_{Hip,g} = \left\{ x'_{Hip,g} \quad y'_{Hip,g} \quad z'_{Hip,g} \right\}^T \quad (5-4)$$

whereby the variables for PA at the right side are denoted with $g = r$, and variables for PA at the left side are denoted with $g = l$. The relative movement of PA is (with respect to starting position of PA)

$$\mathbf{P}_{PA,g} = \begin{Bmatrix} x_{PA,g} & y_{PA,g} & z_{PA,g} \end{Bmatrix}^T \quad (5-5)$$

The reference trajectories are induced by *NaTure-gaits* through the following method. The constant velocity of pelvic movement in the direction of x -axis can be predicted by

$$\dot{x}'_{Hip,g} = \frac{x'_{Hip,g}(1) - x'_{Hip,g}(0)}{T} \quad (5-6)$$

whereby

$\dot{x}'_{Hip,g}$: Average velocity of pelvic movement in the direction of x -axis

$x'_{Hip,g}(1)$: Position of hip in the direction of x -axis at 100% gait cycle

$x'_{Hip,g}(0)$: Position of hip in the direction of x -axis at 0% gait cycle

T : Total time taken to complete the gait cycle

Motion of mobile platform is given by

$$x_{MP} = \dot{x}'_{Hip,g} \times GC \quad (5-7)$$

whereby GC is the percentage of gait cycle, $0\% \leq GC \leq 100\%$. The motion of mobile platform is calculated to rotational angle of the wheels of mobile platform by equation (4-32). The motion for PA in the x -axis direction is given by

$$x_{PA,g} = x'_{Hip,g} - x_{MP} \quad (5-8)$$

The conversion of pelvic motion in the x -axis direction is illustrated in Figure 5-9.

The pelvic motion along y -axis for PA mechanism is computed with

$$y_{PA,g} = y'_{Hip,g} - \bar{y}'_{Hip,g} \quad (5-9)$$

and pelvic motion along z -axis for PA mechanism is given by

$$z_{PA,g} = z'_{Hip,g} - z_{Hip} \quad (5-10)$$

where z_{Hip} is the height of hip joint when standing.

The pelvic motion for the left side is created using the motion of the right side with 50% phase lag (y -axis motion is mirrored, as explained). The final trajectories of pelvic motion processed for play-back on PA mechanism are presented in Figure 5-10. The

same procedures as presented in Figure 5-8 are used to convert the final trajectories into the motion input for PA controller.

Figure 5-9: Conversion from pelvic motion to motion for mobile platform and pelvic assistance mechanism

The motion of the pelvis in the z -axis is found to be always lower than the hip height of standing. The combined motion for the PA sub-mechanism (the movement of pelvis in the sagittal plane) is presented in Figure 5-11. The left and right side of PA sub-mechanism have the same trajectories.

Figure 5-10: Final constructed motion for PA mechanism (a) Motion in the direction of x -axis; (b) Motion in the direction of y -axis; (c) Motion in the direction of z -axis (Solid line – right side, dashed line – left side)

Figure 5-11: Trajectories of pelvic motion input to PA sub-mechanism (motion in the x and z axes, which is the sagittal plane)

Figure 5-12: Overview of conversion of final constructed pelvic motion to input for the controller of robotic orthosis

The final constructed motion for pelvic can be computed for the input for controller of pelvic assistance mechanism. The trajectories presented in Figure 5-11 are computed to motion input for the sub mechanism of pelvic assistance mechanism by kinematic model of the sub mechanism presented in Section 4.2.1. For the lateral pelvic shift motion (Figure 5-10b), it can be converted to input for the controller by multiply the corresponding gear ratio of the later shifting mechanism. The overview of the conversion is presented in Figure 5-12.

5.2.4 Improvement of Foot Contact during Walking via Pelvic and Lower Limb Motions Coordination

Gait cycle is made up of a complex series of joint rotations. These joint rotations are coordinated as a whole and provide a smooth forward progression of the COM [1]. The coordination of the joint rotations also ensures the smooth foot contact with the environment. Central pattern generators within the spinal cord play an important role in the production of the coordinated movement in human walking [1, 2]. Although the central pattern generators are able to produce stereotyped human walking patterns, sensory

feedback from the periphery allow the rich variation in walking patterns and adaptability to task and environmental conditions.

For robotic gait rehabilitation, the walking pattern provided is symmetric and controlled environment is assumed (i.e. leveled ground). Human walking motion can be obtained from motion capture experiment or by *GaitGen*, and the walking motion is in a form of coordinated pelvic motion and lower limb joints rotation angle. The coordination of the motions required adjustment in order to facilitate their use for driving the robotic gait rehabilitation system. The adjustment is essential for the following reason.

Experimental errors. The motion obtained from motion capture experiment contains minor errors such as: human error during the placement of the reflective markers (e.g. the placement of sacrum marker slightly offset from the correct position vertically will create reference plane with 1 ~ 2 degrees error), the human error in the measurement of anthropometry parameters, and the human subject might not stand at an absolute perfect posture during initialization. The anthropometry parameters are measured for the computation of pelvic and lower limb motion from data of motion capture experiment. The initialization posture requires the subject to stand-still with lower limb keeps at neutral position (i.e. joint angle stays at zero degree with respect to origin). These experimental errors cannot be eliminated entirely. Joint to joint distance is estimated because it is measured extrinsically. It is impossible for any subject to stand at a posture that meet the initialization requirement perfectly. For example, if hip joint angle has a constant error of one degree in flexion, and other factors are assumed to be correct, the foot position based on experimental result will be always higher than the actual foot position when the hip is in flexion. Assume at the instance of heel strike, hip angle is 20 degrees and contains of 2 degree of error due to marker placement (i.e. measured 22 degrees during experiment), for a lower limb length measured 800 mm, the heel will offset from the actual position for 10 mm. *GaitGen* has the same error because it is modeled based on the data of motion capture experiment.

This work aims to improve the foot contact of subject during *NaTure-gaits* assisted walk through the adjustment. This work also aims to minimize the changes on the walking motion caused by the adjustment and maintain the smoothness and continuity of the motions.

Foot Contact during Human Walking

During normal human walking, the only interaction with the environment is the foot contact with ground. The foot contact is the main consideration when adjusting the motion for use on *NaTure-gaits*. This subsection introduces the foot contact of human walking. The foot contact at each phase of gait cycle is:

- The heel touches the ground at *heel strike* (HS), allow foot to pivot at heel until *foot flat* (FF)
- The sole of the foot comes into contact with ground during *foot flat*, and maintains contact throughout
- The toe remains on the ground from *heel off* (HO) to *toe off* (TO), and the motion of foot is pivoting at toe joint
- During swing phase, the foot shall not come into contact with the ground

Ankle reference angle indicates the orientation of foot with respect to the ground surface. The foot is parallel to the ground when the ankle reference angle is zero, and a positive ankle reference angle indicates that toe is higher than heel, and a negative ankle reference angle indicates that toe is lower than heel. Based on the foot contact at each phase of gait cycle, the ankle reference angle is positive within heel strike to foot flat, and remain zero degree from foot flat to heel off and from heel off to toe off it is negative. An ideal ankle reference angle is given in Figure 5-13.

Figure 5-13: Ideal ankle reference angle during human walking

Problem Statement

Due to the experimental errors, it is expected that the foot contact won't be correct if the walking motion is used on *NaTure-gaits* without any adjustment. Using the kinematic model of *NaTure-gaits* and the motions of pelvic and lower limb obtained by motion capture experiment, the trajectory of heel of ten subjects is computed and presented in Figure 5-14. A clear indication of incorrectness is that the heel of some subjects did not touch the ground for the instance of heel strike at 0% of gait cycle. Furthermore, negative value of heel position indicates that the heel travelled below the ground level, which is not possible in reality.

Figure 5-14: Heel trajectories in the direction of Z-axis, computed from lower limb and pelvic motion obtained by motion capture experiment

Experimental errors can also be observed by inspecting the ankle reference angle. Ankle reference angle of ten subjects are computed from motion captured walking motion and presented in Figure 5-15. Ideally, during the phase from foot flat to heel off, the ankle reference angle shall be zero degree. However, this is not observed on every ankle reference angles presented in the figure. It indicates that the lower limb joint angles contained experimental error, resulted in the computed ankle reference angle does not reflect the real motion of human walking.

These problems highlighted the need of adjusting the motion for its use on *NaTure-gaits*. Without adjustment, the walking motion obtained from motion capture experiment or *GaitGen* is not suitable for playback on *NaTure-gaits*.

Figure 5-15: Ankle reference angle computed from data of motion capture experiment

Adjustment Model for Improvement of Foot Contact

The pelvic and lower limb motions are the inputs of the *NaTure-gaits*, we need to setup an analytical requirement for adjusting the motions, and ensure that the adjusted motions are smooth and comply with the actual foot placement during walking at each phase of gait cycle.

Through preliminary analysis, this work identified that by adjusting lower limb joints angle (hip, knee, and ankle), the ankle reference angles and heel position can be improved to a more realistic state. Adjusting pelvic trajectory is not preferred because adjustment carried out on the trajectory rendered a non-smooth motion.

The physical requirements of human walking are:

- I. Heel position for the period from heel strike to heel off needed to adjust to ground level.

- II. Ankle reference angles during foot flat to heel off period needed to offset to zero degree.

However, adjusting the motion to match the requirements exactly is not practical. Forcing the motion to match the requirements through adjustment cannot maintain the original shape and smoothness of the lower limb and pelvic motions, which is not desire for human motion. This work proposes the following modifications by adjusting lower limb joints angle:

- I. Heel position for the period from heel strike to heel off is adjusted to ground level. The vertical distance of heel position measured from ground will be kept to the minimum. Final step is to offset the entire heel trajectory so that the lowest position of heel is at ground level. This modification improves the heel contact to more resemble the real heel contact of human walking.
- II. Minimum ankle reference angle during foot flat to heel off period is shifted to zero degree through the adjustment. Physically, ankle reference angle cannot have negative value within this phase because the toe cannot travel below ground level. Toe is lower than heel when the ankle reference angle is negative, and heel is at ground level at this phase. This step attempt to minimize the initial offset of the lower limb joints angle.

The adjustment is carried out based on the occurrence of gait events. The occurrence of gait events can be estimated from the ankle reference angle based on the orientation of foot during human walking, as shown in Figure 5-16.

Figure 5-16: Identification of foot flat and heel off using ankle reference angle: Balloon “1” marked the foot flat event (The angle changing from +ve to 0); Balloon “2” marked the heel off event (The angle changing from 0 to -ve)

The main modification is the adjustment of the heel position to ground level from heel strike to heel off. The objective of the modification is to minimize the distance of heel trajectory measured from ground, through adjusting the lower limb joints angle. Let the heel trajectory before adjustment to be Z_{heel} and heel trajectory after adjustment to be Z'_{heel} . The heel trajectories can be computed by

$$Z_{heel}(t) = Z_{hip}(t) - l_t \cos \theta_h(t) - l_s \cos(\theta_k(t) - \theta_h(t)) - l_f \cos \theta_a(t) - l_{hl} \sin \theta_a(t) \quad (5-11)$$

$$Z'_{heel}(t) = Z_{hip}(t) - l_t \cos \theta'_h(t) - l_s \cos(\theta'_k(t) - \theta'_h(t)) - l_f \cos \theta'_a(t) - l_{hl} \sin \theta'_a(t) \quad (5-12)$$

according to the kinematic model illustrated in Figure 5-17.

$$\begin{aligned} \theta'_h &= \theta_h(t) + \Delta\theta_h(t) \\ \theta'_k &= \theta_k(t) + \Delta\theta_k(t) \\ \theta'_a &= \theta_a(t) + \Delta\theta_a(t) \end{aligned} \quad (5-13)$$

The objective function for this modification is

$$f(\Delta\theta_h, \Delta\theta_k, \Delta\theta_a) = \frac{1}{T_{HO}} \int_0^{T_{HO}} \left(Z'_{heel}(t) - z_{heel,min} \right)^2 dt \quad (5-14)$$

where T_{HO} is the instance of heel off (HO) measured in percentage of one gait cycle and $Z_{heel,min}$ is the lowest heel position in z -axis before adjustment.

The objective function has two constraints. The first constraint is imposed by the modification of ankle reference angle. The ankle reference angle during the foot flat to heel off phase is offset by the minimum value, $\theta_{f,min}$, of ankle reference angle found within the phase. The modified ankle reference angle

$$\theta_{f,mod} = \theta_f - \theta_{f,min} \quad (5-15)$$

Physically, ankle reference angle is obtained by:

$$\theta_f = \theta_h - \theta_k + \theta_a \quad (5-16)$$

θ_h is angle of hip joint, θ_k is angle of knee joint, and θ_a is angle of ankle joint. Inspecting Equations (5-15) and (5-16), to obtain the modified ankle reference angle

$$(\Delta\theta_h - \Delta\theta_k + \Delta\theta_a) = -\theta_{f,min} \quad (5-17)$$

In other word, the offset of ankle reference angle can be achieved by modifying hip, knee, and ankle joints angle. This is the first constraint for the objective function.

The second constraint is the physical limitation of hip, knee, and ankle joints' range of motion. The limit of hip, knee, and ankle joints range of motion is set as:

$$\begin{aligned} -20^\circ &\leq \theta_h \leq 30^\circ \\ 0^\circ &\leq \theta_k \leq 80^\circ \\ -20^\circ &\leq \theta_a \leq 10^\circ \end{aligned} \quad (5-18)$$

Figure 5-17: Denotations for variables of PA and RO

Solving the Model using Generic Algorithm

The generic algorithm was implemented to solve the proposed model considering the two constraints. Generic algorithm is a tool that can be used to solve both constrained and unconstrained optimization problems based on natural selection. The generic algorithm repeatedly updates a population of individual solutions. At each step, the algorithm selects the individuals based on selection rules from the current population to be parents

and uses them to produce the children for the next generation. Over successive generations, the population evolves toward an optimal solution.

The generic algorithm is implemented using global optimization toolbox of Matlab. The details of the generic algorithm applied for solving the proposed model is summarized as follows:

1. The algorithm starts by creating a random initial population, which is a set of lower limb adjustment angles. In this work, the population size was chosen to be 100 and the change of angles is bounded within $\pm 5^\circ$.
2. The algorithm creates a sequence of new populations by performing the following steps.
 - a. Scores each member of current population by computing its fitness values using the objective function as shown in Equation (5-14).
 - b. Scales the raw fitness scores to convert them into a more usable range of values.
 - c. Selects members, call parents, based on their fitness.
 - d. Some of the individuals in the current population that have lower fitness are chosen as elite. These elite individual are passed to the next population. At least four elites were selected to pass in this work. The more number of elites required, the more versatile the population but the optimization process is slow.
 - e. Produces children from the parents. Children are produced either by making random changes to a single parent (mutation) or by combining a pair of parents (crossover).
 - f. Form the next generation
3. The algorithm stops when one of the stopping criteria is met. In this work, three stopping criteria were used.
 - a. Number of generations reaches the maximum value which was chosen to be 60.
 - b. Average cumulative change in value of the objective function less than 0.001.
 - c. Fitness value of objective function is smaller than 0.001.

Results and Discussion

An example of ankle reference angle before and after adjustment is presented in Figure 5-18. For the same set of motion adjustment, the coordinated motions of pelvic and lower limb joint show an improved foot contact in Figure 5-19. Those physically not possible positions have been improved. The adjusted lower limb joints angle are presented in Figure 5-20.

The adjusted walking motion based on the data of motion capture experiment is presented in Figure 5-20. The walking motion after the adjustment is resemble more to the actual movement. In particular the foot contact has been improved.

Figure 5-18: Ankle reference angle: before and after adjustment

Figure 5-19: Heel position in the *Z-axis* direction: before and after adjustment

Figure 5-20: Lower limb joints angle: Before and after adjustment

A model has been proposed for improving foot contact during walking by adjustment of lower limb joints angle. The ankle reference angle is modified to follow the physical foot orientation during walking. This is achieved by adjusting the three lower limb joints angle. Simultaneously, the heel position in the z -axis direction is modified for a better foot contact by adjusting the three lower limb joints angle. Three constraints have been imposed to ensure the adjustment of the angles does not exceed physiologically permitted range of motion.

The model presented does not adjust the trajectories of heel and the ankle reference angle to follow exactly the phenomenon described, (i.e. heel trajectory is adjust to exactly 0 mm during foot flat period). Refers to Figure 5-19, heel position in the z -axis direction is not at 0 mm throughout the foot flat period. Theoretically heel position should stay at 0 mm during that period, but mathematically adjust it to such position disrupted the smoothness of the lower limb joints angle. In this work, the first consideration of the adjustment is to maintain smoothness of the motion, because human walking shall be carried out smoothly.

To summarize, a model has been proposed and tested for the improvement of foot contact. This model has been proposed to translate the walking motion obtained from

motion capture experiment for its use on robotic gait rehabilitation system, *NaTure-gaits*. It has been shown that by modify hip, knee, and ankle joints angle, the foot contact can be improved compared to the original motion.

Figure 5-21: Adjustment of walking motions based on the data of motion capture experiment: before and after (Red triangle representing the foot, graph unit is mm)

5.2.5 Implementation of Passive Control Strategy on *NaTure-gaits*

The computed inputs from the reference trajectory are programmed into the controller of respective modules. All the controllers are synchronized by the host computer as explained in Section 4.5.1. *NaTure-gaits* is tested without subject (subject trial will be discussed in Chapter 6). For this validation, the main aim is to illustrate the ability of *NaTure-gaits* in playing back the reference trajectory, and observe the contact of the foot pad with the ground. The overall movement of *NaTure-gaits* is observed and the scenario of the gait locomotion play back is presented in Figure 5-22.

Based on the visual inspection, *NaTure-gaits* successfully played back the reference trajectory. The heel contact with ground is observed at the start of gait cycle, and the key gait events (foot flat, heel off, and toe off) are observed.

Figure 5-22: Scenario of *NaTure-gaits* gait locomotion play back, (a) – (c): Heel strike to foot flat of the right limb, (d): Heel strike of contra lateral limb and heel off of right limb, (e) Foot flat of control lateral limb, (f) Heel strike of the right limb and completion of one gait cycle

5.3 Active Pelvic Motion Assistance with Adaptive Body Weight Support for Robotic Gait Rehabilitation

In this research work, the provision of pelvic motion assistance and body weight support with one robotic module are proposed and realized. Pelvic assistance mechanism has been built to realize this feature. Control strategy is designed to control the pelvic assistance mechanism to achieve the desired active assistance. To facilitate the understanding of the need of adaptive body weight support, the function of body weight support is first explained. The concept of the control strategy is introduced, follows by the discussion on the design and implementation of the control algorithm. The control strategy is validated with a preliminary experiment setup with a one DoF linear mechanism.

5.3.1 Function of Body Weight Support

To achieve bipedal walking, human must be able to support his body weight using the supporting leg at all times, while swinging the contra-lateral leg in the direction of progression. Apart from supporting body weight, net muscle force along the motion progression direction is produced to achieve the intended body movement.

Figure 5-23: Illustration of dynamic equilibrium of walking (a) Without external force component, (b) With external force components, BWS (Forces location indicate in this figure does not necessary denotes the actual location of force acting)

Free body diagram of human walking showing contact with environment is illustrated in Figure 5-23. The interaction between human body and the environment (Figure

5-23a) is the contact between the feet and the ground during walking. The contact forces are commonly referred as ground reaction force (GRF). The dynamic equilibrium during walking can be denoted as

$$\sum F_x = ma_x, \sum F_y = ma_y, \sum F_z = ma_z + w_{body} \quad (5-19)$$

Figure 5-24: Dynamic equilibrium during walking (a) without external support (b) with body weight support

which the dynamic equilibrium analysis based on the interaction of human body and the environment. Within the human body, dynamic equilibrium of hip joint in Figure 5-24 is achieved by

$$\begin{aligned} \mathbf{M}_{hip} &= (m \cdot \mathbf{a}) \times \mathbf{d}_1 + \mathbf{w}_{body} \times \mathbf{d}_2 + \mathbf{I} \cdot \boldsymbol{\alpha} \\ \mathbf{M}_{hip}, \mathbf{a}, \mathbf{w}_{body}, \mathbf{d}_1, \mathbf{d}_2, \boldsymbol{\alpha} &\in \mathbb{R}^3 \end{aligned} \quad (5-20)$$

where $\mathbf{M}_{hip}$ is the vector of hip moment, $\mathbf{a}$ is the vector of center of mass (COM) acceleration, m is the effective mass at COM, $\mathbf{w}_{body}$ is the vector of body weight, $\mathbf{I}$ is the 3×3 inertia matrix, and $\boldsymbol{\alpha}$ is the vector of angular acceleration. $\mathbf{d}_1$ and $\mathbf{d}_2$ is the moment arm

vector of $m \cdot \mathbf{a}$ and $\mathbf{w}_{body}$ respects to the hip joint center. The joint moment vector ($\mathbf{M}_{joint}$) is related to the force vector of the muscle strength ($\mathbf{F}_m$) by

$$\begin{aligned}\mathbf{M}_{joint} &= \mathbf{F}_m \times \mathbf{d} \\ \mathbf{M}_{joint}, \mathbf{F}_m, \mathbf{d} &\in \mathfrak{R}^3\end{aligned}\tag{5-21}$$

and vector, $\mathbf{d}$, is the moment arm distance between the force vector and joint center.

SCI and stroke usually results in the weakness of the leg, disallow the patients from producing adequate muscle force to maintain the dynamic equilibrium as described in equation (5-20). The use of BWS compensates for the weakness and maintains the dynamic equilibrium as illustrated in Figure 5-24b. The dynamic equilibrium is altered by the force provided by body weight support as (based on free body diagram of Figure 5-24b)

$$\begin{aligned}\mathbf{M}_{hip} + \mathbf{F}_{BWS,r} \cdot \mathbf{d}_r - \mathbf{F}_{BWS,l} \cdot \mathbf{d}_l &= (m \cdot \mathbf{a}) \times \mathbf{d}_1 + \mathbf{w}_{body} \times \mathbf{d}_2 + \mathbf{I} \cdot \boldsymbol{\alpha} \\ \mathbf{M}_{hip}, \mathbf{F}_{BWS,r}, \mathbf{F}_{BWS,l}, \mathbf{w}_{body}, \boldsymbol{\alpha}, \mathbf{a}, \mathbf{d}_1, \mathbf{d}_2, \mathbf{d}_r, \mathbf{d}_l &\in \mathfrak{R}^3\end{aligned}\tag{5-22}$$

where $\mathbf{F}_{BWS,r}$ and $\mathbf{F}_{BWS,l}$ are the vector of body weight supporting force for the right and left respectively, $\mathbf{d}_r$ and $\mathbf{d}_l$ are the vector of distance measured from the position of the body weight supporting force acting at to the position of hip joint center (HJC marked in Figure 5-24b). All other denotations remain unchanged.

5.3.2 Concept of the Proposed Control Strategy

Control strategy is proposed for active pelvic motion assistance with adaptive body weight support [139]. Instead of providing a constant force for body weight support throughout the rehabilitation session, the control strategy is designed to adjust the amount of the force provide by pelvic assistance mechanism based on the performance of the patient. Body weight support is usually provided at vertical (z -axis) direction to offload partial of patient's body weight. The forces along x -axis and y -axis are not controlled under normal circumstances. The pelvis is firmly secured and restricted by the BWS apparatus, whereby pelvic motion is not allowed in these two directions.

From previous section, if the body weight support force is maintained constant, the muscle force ($\mathbf{M}_{hip}$ in equation (5-22)) exerted by the patient is not optimized. Judging from the clinical observation, we hypothesized that a constant body weight support is not preferred (in fact, therapists/helpers do not provide constant support, but supporting

force is changing according to the posture of the patient). Thus, we propose a control strategy for adaptive BWS, which is able to provide BWS similar to the support provided by the helpers during gait rehabilitation. The requirements for the proposed posture adaptive BWS control strategy are

- Allow acceptable deviation from reference trajectory, instead of restricting the motion to follow the reference trajectory
- Control of force magnitude of BWS based on the extent of postural deviation

The control strategy adapts the patient's pelvic motion, and increases the amount of BWS proportional to the deviation between the patient's trajectory and desired trajectory. This allows patient to move at their own desired motion, and resistance force will be gradually increased to encourage the patient to follow the desired trajectory. By such, it allows active participation of the patient, by allowing the patient to put in own effort to adjust the desired motion.

5.3.3 Design of Controller for Adaptive Body Weight Support with Natural Pelvic Motion Assistance

The main objective of the proposed control strategy is to optimize the weight bearing on the lower limb. Instead of a constant body weight support force provision, the strategy vary the amount of the force provides by the robotic system based on the patient's posture and performance.

Ideally, when a patient's pelvis moves at the desired trajectory, q_d , pelvic assistance mechanism provides only the desired body weight support force, f_r . In this work, the desired BWS force is set equal to zero. BWS force can be set to null if the patient is able to follow the desired trajectory, which in turn maximizes the weight bearing on the lower limbs. The actual pelvic position, q_a , is obtained from the encoder feedback of pelvic assistance mechanism. Pelvic position error, q_e , is the difference between the desired position and the actual position, given by

$$q_e = q_d - q_a \quad (5-23)$$

If there is a pelvic position error, a vertical interaction force shall exist between the patient and the robotic arm. The vertical interaction force is denoted as f_v . The pelvic position error is used in the adaptive controller and the Virtual Model Control (VMC)

generator to create a virtual spring with spring stiffness, k . The spring stiffness is the impedance parameter and it can be updated as every step.

The allowable deviation of pelvic position from the reference trajectory is denoted as q_f . It can be obtained from the impedance controller if a vertical interaction force exists. The pelvic position command, q_c , is calculated from q_d and q_f and send to PA block in Figure 5-26. PA block is the position controller of the pelvic assistance mechanism, which convert q_c to the motion of the two motors found on pelvic assistance mechanism (the conversion is done with the kinematic model described in Section 4.2.1). The maximum allowable deviation of pelvic position from the reference trajectory is denoted as $\hat{q}_f$. The difference between the current and the previous pelvic position error is given by

$$\Delta q_{e,i} = q_{e,i} - q_{e,i-1} \quad (5-24)$$

The pelvic position is moving further from the reference trajectory if $\Delta q_{e,i} > 0$.

The denotations used in this controller are given in Figure 5-25. The control algorithm for the implementation of the control strategy is presented in Figure 5-26. Some of the denotations will be explained subsequently.

Figure 5-25: Denotations use for the design of controller (dotted line is the reference trajectory for the pelvic motion; dash line is the maximum allowable deviation for pelvic motion; figure is for illustration purposes only and do not necessary denote the actual pelvic motion)

Figure 5-26: Control algorithm of the proposed control strategy

Virtual Model Control

To achieve the proposed control algorithm, a Virtual Model Control (VMC) technique is applied. This control technique make uses of virtual mechanical components, such as springs, dampers, masses, or any other imaginable components in the creation of control model for the generating of actuator torques/forces [140]. VMC technique allows the generation of virtual reaction force at any point along the desired trajectory. For the application in this work, a virtual spring is created along the reference pelvis trajectory. The parameters of the virtual spring are varied based on the pelvic position. The virtual is designed with four criteria as follows:

- I. If the actual pelvic position, p_a , is equal or higher than the reference trajectory ($q_a > q_d$), the virtual spring is turned off. The controller will be switched to position control mode.
- II. When the actual pelvic position, p_a , is lower than the reference trajectory ($q_a < q_d$), an interaction force, f_v , between the subject and the PA will occur. This interaction force is measured with force sensor. When this happened, the virtual spring is turned on and the controller is switched to compliance mode so that deviation of pelvic position from the reference pelvic trajectory is possible. This deviation is designed to change smoothly from zero to the maximum allowable deviation, $\hat{q}_f$, which is specified by the therapist.
- III. The virtual spring allows the patient to have an extent of own initiated movement, deviates from the pelvic reference trajectory. However, the patient's pelvis still moved by the PA and the own initiated movement is somehow restricted.

The controller can be further improved by making the virtual spring behaves as if it follows then intention of the patient's movement. The virtual spring is designed with its stiffness decreased when the patient's pelvic position is moving further from the reference trajectory, as $\Delta q_{e,i} > 0$.

- IV. When $q_e > 0$, and $\Delta q_{e,i} = 0$ for a certain preset period of time, T , the pelvic position is passively driven along the path that offset from the desired trajectory by the amount of q_f . In this situation, the controller needed to force the patient to move back to the reference trajectory. The virtual spring is required to increase the supportive force by increasing the stiffness, which generate a higher force to move the patient back to the reference trajectory. This can be achieved by implementing an adaptive control to increase the stiffness of the stiffness of the virtual spring.

In order to fulfill the first and the second criteria of the virtual spring design, the bipolar sigmoid function, $\gamma(f_v)$, is used to generate a smooth transition in the range of $[-1, 1]$ from the vertical force, f_v . For the proposed controller, the pelvis will be supported on in the vertical direction, $f_v > 0$ and $1 \geq \gamma(f_v) \geq 0$. The bipolar sigmoid function is shown in Figure 5-27. The allowed deviation for pelvic position can be obtained from the bipolar sigmoid function

$$\gamma(f_v) = 2 \left(1 + e^{-\delta f_v} \right)^{-1} - 1 \quad (5-25)$$

$$q_f = \hat{q}_f \gamma(f_v) \quad (5-26)$$

where δ is the slope of the bipolar sigmoid function. The vertical supportive force, f_v , can be derived as

$$f_v = -\frac{1}{\delta} \ln \left[2 \left(\frac{q_f}{\hat{q}_f} + 1 \right)^{-1} - 1 \right] \quad (5-27)$$

$$f_v = -\frac{1}{\delta} \ln \left[2 \left(\frac{q_f}{\hat{q}_f} + 1 \right)^{-1} - 1 \right] \quad (5-28)$$

Figure 5-27: Bipolar sigmoid function

The stiffness, k , of the virtual spring is derivative of f_v with respect to q_f and given by

$$k = \frac{d(f_v)}{d(q_f)}$$

$$k(\delta, q_f) = \frac{-2}{\delta \left(\frac{q_f}{\hat{q}_f} \right)^2 - \delta} \quad (5-29)$$

The value of the virtual spring's stiffness is proportional to the allowable deviation of pelvic position, q_f , and inversely proportional to the slope of the bipolar sigmoid function. The stiffness function $k(\delta, q_f)$ is plotted in Figure 5-28.

Figure 5-28: Stiffness value of virtual spring as a function of slope of the sigmoid function and allowable deviation for pelvic position

Adaptive Control for Slope of Sigmoid Function Update

The adaptive control is developed to update the value of the slope of sigmoid function (δ), based on the performance of the patient. The function of the adaptive control is given by

$$\delta_{j+1} = f\delta_j + g\Delta\tilde{q}_{e,j} \quad (5-30)$$

$$\Delta\tilde{q}_{e,j} = \tilde{q}_{e,j} - \tilde{q}_{e,j-1} \quad (5-31)$$

where δ_j is the slope of sigmoid function (slope) at the iteration j ($0 < \delta_j$). $\Delta\tilde{q}_{e,j}$ is the difference between the average of pelvic position error in j and $j-1$ iteration, g is the gain of $\Delta\tilde{q}_{e,j}$, and f is the forgetting factor ($0 < f < 1$).

When the patient tends to move away from the reference trajectory, the slope will be increased proportionally and vice versa. Therefore, the virtual spring's stiffness is adapted during the training process to follow the intended pelvic movement (as described in third design criteria). When $\Delta\tilde{q}_{e,j} = 0$, the slope will be reduced and the stiffness of the virtual spring increases to move the pelvis back to the reference trajectory (as described in fourth criteria).

The average of the pelvic position error in j iteration is computed with the moving average approach

$$\tilde{q}_{e,j} = \frac{1}{N} \sum_{i=N(j-1)}^{N,j} q_{e,i} \quad (5-32)$$

where N is the number of samples taken for one iteration.

5.3.4 Experiments and Results

For experimental stage, the controller performance is studied with a single DoF mechanism. A linear moving platform is built to study and verify the response of the proposed control strategy. The linear moving platform consists of one DC brushed motor, a ball screw, and a movable block attached to the ball screw. The setup for experiment is depicted in Figure 5-30. The experiment is conducted to obtain the initial result for the designed control algorithm. FlexiForce sensor is used to obtain the interaction force, also refers to as the vertical interaction force denoted in the control algorithm (component f_v in Figure 5-29). PA is replaced by the position controller of the linear moving platform. The control algorithm for the experiment is presented in Figure 5-29.

Figure 5-29: Control algorithm of the proposed control strategy

xPC Target is used as the platform to realize the proposed control algorithm. Two computers are used for the operation of xPC Target, one is the host PC and the other is the target PC. One Sensoray 626 card from Sensoray [141] is plugged in PCI slot of the target PC. Force signal and motor encoder feedback are acquired by ADC channel and counter of Sensoray 626 respectively. These signals are transmitted to the host PC via RS 232 communication. On the host PC, the control algorithm is designed using Matlab Simulink. Position command is generated in host PC and transmitted to target PC. At

target PC, the control signal is converted into analog signal at DAC channel of Sensoray 626, and sent to the amplified. The overview of the hardware and software configuration of the controller is given in Figure 5-31.

Figure 5-30: Linear moving platform built for the controller verification

Figure 5-31: Hardware and software configuration of the controller

Software Implementation for the Controller

xPC Target is a Matlab toolbox, which provides a high performance host-target environment that enables us to connect the Simulink and Stateflow models to physical system and execute them in a real time manner on PC hardware. It supports rapid prototyping, hardware in the loop (HIL) testing, and application deployment in open hardware architecture. The main advantageous features of xPC target are the support provided for more than 300 commercial I/O boards, dealing with low language programming, and tuning system parameters online.

A system model is first created using Simulink control blocks on host PC. The model is run in simulation mode to check for the response of the controller. Once the controller is successfully simulated, the interface blocks from the xPC Target block library will be added to the Simulink model. For example, the Sensoray 626 AD block and the Sensoray 626 Encoder block are used to read the interaction force and encoder value respectively. Sensoray 626 DA block will output the command signal to the amplifier for the DC brushed motor. Sensoray 626 INIT block declares the necessary settings for the Sensoray 626 card, and it is not required to connect to any other of the blocks. The complete layout of the controller design in Simulink is presented in Figure 5-32.

Figure 5-32: Control block layout in Simulink for the controller design

The updated model is interpreted to executable codes by using host PC with Real Time workshop and a C/C++ compiler. The executable code is downloaded from host PC to target PC, which is running xPC Target Real Time kernel. The kernel is booted by

CD created with Matlab software. The physical system can be tested in real time, after the executable code is downloaded to the target PC.

Experimental Results

A sinusoidal wave with the amplitude of 360 units (unit-less) is used as the desired trajectory in this experiment. Sinusoidal wave is used since it is resembled to an actual pelvic movement in the direction of the z -axis. The maximum allowed deviation from the reference trajectory is set to 100 units. The fundamental sampling time of Simulink is two milliseconds. Experiments are carried out for 10 seconds and 100 samples are taken for updating the slope of the bipolar sigmoid function (δ). The minimum value of the slope of the bipolar sigmoid function is 0.1.

The response and stability of the controller is tested and verified with two scenarios. In the first scenario, (I), the controller is provided with two different values of forgetting factor, $f = 1$ (scenario I-a) and $f = 0.8$ (scenario I-b), and tested with a constant vertical interaction force (f_v). The purpose of the experiment for scenario (I) is to evaluate the response of the controller for the four criteria mentioned in Section 5.3.3, and illustrates the effect of the forgetting factor in the controller.

In real rehabilitation scenario, the interaction force between patient and robotic cannot be constant over time. Therefore, in the second scenario (II), a force is varied randomly to test the response of the controller. The responses of the controller for scenario (I) are presented in Figure 5-33 and Figure 5-34. The values of the slope (δ) and stiffness of the virtual spring (k) in scenario (I) are shown in Figure 5-35 and Figure 5-36. The interaction force and the response of the controller for scenario (II) are presented in Figure 5-37.

Result Discussion

The experimental results confirmed that the designed controller fulfilled the design criteria. Some extent of deviation from the desired trajectory is allowed when the vertical interaction force exists. The actual trajectory is smooth and it is limited to movement not beyond the maximum allowable deviation (Figure 5-33 and Figure 5-34). The allowable deviation can be defined in the controller. In the actual operation scenario, the interaction force will exist when the patient do not follow the desired trajectory.

When this happened, the designed controller reacts and allows the patient to deviate from the desired trajectory. The maximum allowable deviation is specified such that the patient's pelvis will not deviate too significantly from the desired movement during the training process. The deviation allows the patient to have a certain degree of participation in the process. The first and second criteria are thus satisfied.

Figure 5-33: Response of the controller for scenario (I-a)
(solid line-actual trajectory, dotted line-desired trajectory, and dashed line-maximum allowed deviation)

Figure 5-34: Response of the controller for scenario (I-b)
(solid line – actual trajectory, dotted line – desired trajectory, dashed line – maximum allowed deviation)

The third criterion is the response of the controller towards the movement intention of the patient. The experimental results show that the slope (δ) is increased to modify the stiffness of the virtual spring when the patient starts to deviate from the desired trajectory. If the patient remains deviated from the desired trajectory for a period of time, the stiffness of the spring is reduced since the controller consider the patient wanted to move

at the deviated trajectory. This will allowed the patient to have greater flexibility in influencing the desired trajectory of the system.

Figure 5-35: Slope δ and the stiffness k of the virtual spring in scenario (I-a)
(solid line – stiffness (unit, N/mm), dotted line – slope (no unit))

The forth criterion states that if the patient remained passive for a duration (the duration can be defined in the controller), the controller will increase the supportive force to gradually move the patient back to the desire trajectory. In scenario (I-a), the forgetting factor is set to 1 (the effect of forgetting factor is turned off), thus even the patient remained totally passive (in the experiment, the interaction force is remained at a value), the system will continue to move at the maximum allowable deviation (Figure 5-33). In scenario (I-b), the forgetting factor is included. The controller will move the actual trajectory back to the desired trajectory after a period of time if the interaction force is remained at a value (Figure 5-34). The effect of the forgetting factor is clearly highlighted with scenarios (I-a) and (I-b). As for scenario II, it tries to mimic the actual scenario of the rehabilitation session. The controller has fulfilled all four criteria specified for the controller design.

Figure 5-36: Slope δ and the stiffness k of the virtual spring in scenario (I-b)
(solid line – stiffness (unit, N/mm), dotted line – slope (no unit))

Figure 5-37: Response of the controller for scenario (II)
(solid line – actual trajectory, dotted line – desired trajectory, dashed line – interaction force (unit, mN))

5.4 Concluding Remarks

At the beginning of this chapter, stroke and spinal cord injury patients are classified into four groups, based on their gait ability. Overview of rehabilitation strategies applicable to the groupings is suggested. Out of the suggested possible rehabilitation strategies, passive control strategy and one of the active control strategies are implemented in this research work. Active control strategy implemented is active pelvic motion assistance with adaptive body weight support, which provides adaptive body weight support based on the pelvic postural deviation of the subject.

5.4.1 Passive Control Strategy Implementation on *NaTure-gaits*

A robotic system would not be able to perform without the input of correct motion. In order to obtain the gait locomotion for *NaTure-gaits*, motion capture experiment is

carried out. Pelvis movement and lower limb joint angles are computed from the data obtained in motion capture experiment. It is post-processed to make the gait locomotion to be played back on *NaTure-gaits* seamlessly. The motion acquired in motion capture experiment has been successfully transferred to as motion input for *NaTure-gaits*. The passive control strategy is successfully implemented on *NaTure-gaits*.

The adjustment of lower limb joint angles to achieve the desired heel contact during support phase improved the foot contact. By using optimization approach to minimize the joint angles change, the adjustment of joint angles is kept to minimum.

5.4.2 Active Pelvic Motion Assistance with Adaptive Body Weight Support

Virtual Model Control (VMC) approach is used to create a virtual spring at the body weight support direction. The supporting force of the body weight support is created by adjusting the spring constant of the virtual spring. An adaptive controller is created to update the spring constant based on the deviation between the actual position of the patient and the desired position. Experiment shows that proposed control strategy allowed certain extent of deviation from the desired trajectories, which is the intended aim of the designed control strategy. The experiment also shows that the supporting force is increased with the increment in deviation between actual and desired trajectories.

The performance and stability of the developed control strategy are tested by implementing on a preliminary experimental setup. The control strategy can be transferred to *NaTure-gaits* without major modification of the controller design, only the system block in the control algorithm needed to be updated, as explained in Section 5.3.4. The system block is the kinematic model of the targeted mechanism, in this case, pelvic assistance mechanism. The kinematic model has been presented in Section 4.2.1. The actual performance and stability of the developed control strategy on *NaTure-gaits* will be carried out after the clinical trials.

Chapter 6

GAIT LOCOMOTION ASSISTANCE WITH *NATURe-GAITS* ON HUMAN SUBJECT

The research work proposed and presented in this thesis is put to test by carrying out trials. Trials are carried out with healthy subject and patient to verify the basic functions and safety of the proposed robotic system, *NaTure-gaits*, and the overall system control is implemented with passive control strategy. Complete description of *NaTure-gaits* is provided in Chapter 4, and the passive control strategy has been explained in Section 5.2. The protocol of the trials will be described in the subsequent section. Four healthy subjects and one patient have been recruited for the trials. Muscle activity is studied with the EMG (electromyography) signals recorded during the trials. Survey is carried out to obtain the feedback and walking experience from the subjects.

6.1 Protocol of Trials

Each recruited subject for the trials will go through two set of sub-trials. Pre-trial is carried out right before the two sub-trials. Pre-trial is arranged to acquire EMG signals for the respective subject during normal over-ground walking. The EMG signals serve as reference for the EMG signals recorded during the trials.

The first set of sub-trial (trial – T1) is carried out by suspending the subject off the ground, such that the subject's feet will not come into contact with the ground, while the robotic system induces the predefined gait locomotion. This test is to study the muscle activity of the subjects, attempting to isolate the contribution of the muscle in supporting body weight.

Second set of trial (trial – T2) allow actual over-ground walking experience whereby feet of the subject will come into contact with the ground during the trial. With the ground contact, the body weight will be partially supported by the subject, and thus the muscle activity could be different than the muscle activity of the same subject in Trial – T1.

During pre-trial, EMG electrodes are placed on the respective skin surfaces of the muscle, and the subject is asked to walk at several speeds over GAITRite for recording of EMG signals and corresponding gait parameters. The EMG electrodes are not removed and subject will continue with the trials (trial – T1 and trial – T2).

At the beginning of the trials, the subject is strapped to *NaTure-gaits* by a specially designed harness and attachment straps. The robotic system is adjusted to match the body of the subject. A safety rope connected to the harness provides additional safety to catch the subject from falling if any unexpected event happened. Once the strapping is done, *NaTure-gaits* is started to induce gait locomotion to the subject. Subjects are asked to remain completely passive and relax during trial – T1 and trial – T2. Survey is carried out at the end of each trial as part of the outcome measures. The setup of the trials is shown in Figure 6-1.

Figure 6-1: Snap shot of the set up for trials

The human walking motion for the trial is acquired through motion capture experiment (motion capture experiment has been presented in Section 0). The motion is processed into reference gait pattern for use in passive control strategy. During the trials, *NaTure-gaits* is controlled using passive control strategy (details of passive control

strategy are available in Section 5.2). All modules (pelvic assistance mechanism – PA, robotic orthosis – RO, and mobile platform – MP) of *NaTure-gaits* are controlled by the passive control strategy and provide the gait locomotion assistance to the subject.

The walking speed is lowered to 0.03 m/s for better observation of the process and also safety consideration. The overview of the protocol is illustrated in Figure 6-2. Each subject will go through two set of trials, which consist of five gait cycles for each sub-trial.

Figure 6-2: Overview of prototype trial protocol

6.1.1 EMG Signal Acquisition

In the trials, EMG (electromyography) signals are recorded from the subjects to study the activity of the muscles during the robotic assisted gait rehabilitation. The analysis of EMG signals will provide insight to the level of assistance provided by *NaTure-gaits*.

Figure 6-3: Anatomical position of the selected muscle and electrode placement sites

EMG signals are acquired with Noraxon TeleMyo 2400 G2 Telemetry System [142]. EMG signals of five muscles for each limb (a total of ten muscles for both legs) are recorded during the trials. The five muscles are Vastus medialis (VM), Tibialis anterior (TA), Gluteus maximus (GM), Biceps femoris (BF), and Gastrocnemius lateral (GL). These five muscles provide the flexion/extension of the hip, knee, and ankle joints. Surface EMG electrodes (gelled Ag/AgCl electrodes) are used. The EMG electrodes are

placed on the surface of skin of the corresponding muscle. The locations of the electrodes placement are marked in Figure 6-3. The electrodes passed through preamplifiers via compliant cables and lined to a wearable data acquisition unit, secured at the waist level. The wearable data acquisition unit transmits the EMG signals to the host computer through wireless.

The EMG signals are trimmed into gait cycles by using video recording to determine the heel contact of each gait cycle. EMG signals are full-wave rectified, smoothed using a 50 ms root-mean-square (RMS) algorithm, and low-pass filtered at 6 Hz. The post-process approach of the EMG signals is similar of what is described in the references [109, 143-145]. The EMG signals are time normalized and presented in one unity gait cycle (0% - 100% of gait cycle) in the subsequent discussion.

6.2 Healthy Subjects' Trials

The main objective of healthy subjects' trials is to verify the functionality and safety of *NaTure-gaits* and the passive control strategy, before carrying out the clinical trials with the patient recruited from a local hospital. Apart from the main objective, the sub-objectives are:

- Verify the ability of *NaTure-gaits* in provision of locomotion assistance through EMG signals analysis
- Study the comfort level of the system during the rehabilitation process

Four healthy subjects are recruited for the prototype trials. Their anatomical parameters are tabulated in Table 6-1.

Table 6-1: Information of Subjects Participated in the Trials

	Age	Gender	Body Height	Weight
Subject 1 (S1)	23	Male	160 cm	70 kg
Subject 2 (S2)	23	Male	173 cm	64 kg
Subject 3 (S3)	24	Male	168 cm	60 kg
Subject 4 (S4)	26	Male	170 cm	55 kg

6.2.1 Observation and Subjects' Survey

NaTure-gaits has successfully provided the gait locomotion assistance on the four healthy subjects recruited for the trials. No safety related issues were observed during the trials. EMG signals are recorded and processed as discussed in Section 6.1.1. The analysis of the EMG signals will be provided in later part.

The four subjects were given a simple survey after the trials, the questions focused on the comfort level and walking experience of using *NaTure-gaits*. The subjects mentioned that the use of the special harness hinder their thigh movement, as they are strapped in the harness. When the gait locomotion is provided by *NaTure-gaits*, they felt little constraint on the thigh. In general, they do not have any uneasy feeling during the trials. All four subjects expressed that they are able to walk normally.

In order to have a quantitative measure for the locomotion assistance and the alteration the system might made to the gait pattern of the subjects, EMG signals of five lower limb muscles are acquired and analyzed.

6.2.2 Muscle Activity during Robotic Assisted Gait Locomotion

EMG signals of five selected muscles during slow walking speed (0.17 m/s for subject S1, 0.16 m/s for subject S2, 0.25 m/s for subject S3, and 0.19 m/s for subject S4) are recorded in pre-trial as reference signals for comparison in the subsequent analysis. Activities of the five muscles are recorded for five complete gait cycles in trial – T1 and trial – T2.

Vastus Medialis (VM)

Vastus medialis extends shank at knee joint, which is contributed to the movement of knee joint extension. The EMG signals acquired during the trials are presented in Figure 6-4. It is noticeable that four subjects do not exhibit similar gait pattern at slow walking speed. For subjects S2, and S3, it is visible that EMG signals have higher muscle activation level, when the gait locomotion assistance is provided with ground contact, compared to gait locomotion assistance that is provided without ground contact. The magnitude of the EMG signals is highly related to the muscle force contribution of the corresponding muscle. Significant reduction in the magnitude of EMG signals are exhibited in subjects S2 and S4, indicates that *NaTure-gaits* are reducing the effort

contribution of the muscle in the completion of gait locomotion. However, this observation is not found for subjects S1 and S3. For subject S1, the muscle activation level in all trials is significantly lower than other, which could be due to the poor contact of the EMG electrodes.

Figure 6-4: Average EMG signals of Vastus Medialis (VM) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Tibialis Anterior (TA)

Tibialis anterior is the muscle dorsiflexes foot at ankle joint and inverts foot at intertarsal joints. The EMG signals acquired during the trials are presented in Figure 6-5. Four

subjects do not exhibit a common gait pattern at slow walking speed. EMG signals for tibialis anterior during gait locomotion assistance with ground contact has higher muscle activation level compared to normal walking, in particular for subjects S2, S3, and S4. This could be extra muscle effort has been used trying to balance within *NaTure-gaits*.

Figure 6-5: Average EMG signals of Tibialis Anterior (TA) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Gluteus Maximus (GM)

The extension of thigh at hip joint and medially rotation of the thigh at hip joint is provided by gluteus maximus. The EMG signals acquired during the trials are presented

in Figure 6-6. Significant reduction in muscle contribution is found during gait locomotion assistance without ground contact. However, for subjects S3 and S4, a high muscle activation level was found on 28% of gait cycle and 20% of gait cycle respectively.

Figure 6-6: Average EMG signals of Gluteus Maximus (GM) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Biceps Femoris (BF)

Biceps femoris flexes shank at knee joint and extends thigh at hip joint. The EMG signals acquired during the trials are presented in Figure 6-7. For this muscle group, only subject S2 exhibits a significant reduction in muscle activity when gait locomotion are

provided with no ground contact. Four subjects exhibit different muscle activation pattern. A significantly high muscle activation level is shown for subjects S1, S3, and S4 at approximately 20% gait cycle during off ground trials. Off ground trials shall not exhibit high muscle activation level, as the muscle activity is minimized. More studies are required to investigate this observation.

Figure 6-7: Average EMG signals of Biceps Femoris (BF) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Gastrocnemius Lateral (GL)

Gastrocnemius lateral is the muscle provides the movement of foot plantarflexion at ankle joint and shank flexion at knee joint. The EMG signals acquired during the trials are presented in Figure 6-8. Subjects S2, S3, and S4 exhibit similar EMG signals pattern. Muscle activity is reduced to a minimum for both trials, off ground and over-ground. This provides a general indication that, the muscle stays at a minimum activation level when the gait locomotion is provided by *NaTure-gaits*.

Figure 6-8: Average EMG signals of Gastrocnemius Lateral (GL) of the right leg for five walks of normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

General Remarks

In general, the EMG signals of four subjects do not exhibit a similar pattern in the trials and walking at a very slow speed (0.17 m/s for subject 1, 0.16 m/s for subject 2, 0.25 m/s for subject 3, and 0.19 m/s for subject 4). When human walking at very slow speed, lower limb muscle are contributing more effort on maintaining the postural balance than on the propulsion of body weight forward [143]. This is clearly exhibited by the EMG signals presented. No similar muscle activity is found among the four subjects, since human maintained the postural balance rather differently. Nevertheless, the EMG signals do provide a basic idea that, muscle activity level is significantly reduced with the gait locomotion assisted by *NaTure-gaits*. Further study is required to investigate the potential alteration of muscle activation pattern the system made to a subject during gait locomotion assistance.

6.3 Patient Trial

To verify the ability of the proposed robotic system and the passive control strategy in providing gait rehabilitation for patient, one patient has been recruited for trial. The objectives of patient trial are focused on:

- Verify the ability of locomotion assistance provided by *NaTure-gaits* on patient through EMG measurement
- Study the acceptance of the patient toward the comfort level of the system during the rehabilitation process
- Illustrate the significant reduction in manpower requirement to rehabilitate a patient for over-ground walking

The patient is spinal cord injured (ASIA D, injury level C5). The same protocol of subject trials as described in Section 6.1 is applied for the patient trial.

6.3.1 Pre-Trial

The pre-trial is carried out to obtain information as the reference for outcome measure of robotic assisted gait rehabilitation. In addition, pre-trial for the patient also demonstrated the difficulty of carry out over-ground walking with the manual assistance of trainers. With the help of the two trainers, over-ground walking training is then possible to be carried out for the recruited patient in this trial. The over-ground walking training of pre-

trial is carried out on GAITRite (for recording of gait parameters), and EMG signals are recorded simultaneously.

During the over-ground walking training, the stronger side (left side) of the patient is balanced with a clutch, while the right side is supported by one trainer. The gait locomotion of the left side can be voluntarily completed by the patient, the gait locomotion of the right side has to be completed by another trainer by lifting the foot and place it one step forward. Proper gait locomotion training can only be administrated with the help of two persons. This clearly demonstrated the great requirement of man power to provide over-ground gait locomotion training for the patient. Furthermore, substantial laborious effort is required from both of the trainers, and one of the trainers is working at an ergonomically unfavorable posture. Figure 6-9 shows the scenario of the pre-trial for the patient.

Figure 6-9: Over-ground walking training of the patient at pre-trial (a) The patient walk with the help of two trainers (b) The right leg of the patient is moved forward with the help of one trainer (the trainer is working at an ergonomically unfavorable posture)

6.3.2 Observation and Discussion

Gait locomotion has been successfully provided for the patient by *NaTure-gaits* during the trial. Passive control strategy is programmed to *NaTure-gaits* for the gait locomotion playback during the trial.

During the trial, it is observed that 20 minutes is needed to strap the patient to the system. One trainer is needed to maintain the patient in standing position, while two other helpers suspend the patient off ground with a winch (as shown in Figure 6-10). Once the patient is suspended off ground, the rest of the attachment can be done by only one helper.

The patient tolerated 20 minutes of gait locomotion training in *NaTure-gaits*, completed five gait cycles (suspended off ground), and ten gait cycles in the context of over-ground walking. The patient does not felt uncomfortable during walking within the *NaTure-gaits* based on his feedback.

Figure 6-10: Suspending the patient into *NaTure-gaits*

Through the patient trial, *NaTure-gaits* is further confirmed in its ability to provide gait locomotion assistance to a patient. The comfort level of the system during the rehabilitation process is acceptable for the patient. It is also illustrated that man power is significantly reduced for the provision of over-ground gait locomotion training for the patient. Although two persons are required for the strapping process before the gait locomotion training, only one person is required for the control of *NaTure-gaits* during the gait locomotion training. No laborious involvement is required for that person as the person only control *NaTure-gaits* through computer. The scenario of the gait locomotion training of the patient is shown in Figure 6-11.

Figure 6-11: Gait locomotion assistance for the patient

Survey with the Patient

A short survey is carried out with the patient at the end of the trials. The survey attempt to gather the feedback from the patient's perspective as the final targeted user for the proposed system is patients. Two questions were posted to the patient:

1. Sense of security when using the system
2. Comfort when strapped to the harness

The patient expressed that he felt confidence using the system. However, he suggested that when strapping or preparing the subject for the system, conversation with the

subject is crucial as the conversation reduces the sense of nervous of the subject. As for the comfort level, he expressed that the harness strapping is not totally comfort. He gave a rating of 80% – 90% comfort and the feeling of being strapped is bearable. He felt totally secured when strapped and offloaded. The feeling is clearly shown as he even let go of both hand from the hand support/arm rest, when strapped and suspended.

Muscle Activity of Patient during Robotic Assisted Gait Locomotion

EMG signals of the five muscles at patient preferable speed are recorded just before the trials as reference signals for comparison purpose. Activities of five muscles are recorded as a form of EMG signals during both trials. The EMG signals are time normalized and presented in a unity gait cycle. The post-processed EMG signals for the five muscles are presented in Figure 6-12 – Figure 6-16.

In general, all muscles remain relatively low activation level throughout the two trials, as compared to the muscle activation level during normal walking. Small burst was found for tibialis anterior, at 10% of gait cycle for over-ground trial, and at 20% of gait cycle for off ground trial. This could be due to the voluntary movement of the patient at the beginning of gait cycle, when the feet dorsiflexed to heel strike. This phenomenon is not observed for the right leg, since the right leg does not have the ability to voluntary dorsiflex the feet. Muscle activation level of biceps femoris is found higher during over-ground trial compared to off ground trial. However, the muscle is contracted throughout the gait cycle as the EMG signals are flat, which could be involuntary contraction of the muscle.

The low muscle activation level indicates high level of motion assistance is provided by *NaTure-gaits* in the completion of gait locomotion. However, this is not beneficial to the patient, as the patient participation is greatly minimized. The EMG signals can be incorporated into control strategy to encourage the active participation of the patient during gait rehabilitation using *NaTure-gaits*. The similar muscle activation level (in term of magnitude) as of normal walking shall be maintained during robotic assisted gait rehabilitation to maximize the therapeutic outcome.

Majority of the muscles displayed very low activation level and the leveled signals, due to assistance of the system to the patient. The result is not conclusive enough to be

commented on the alteration in muscle activation pattern caused by *NaTure-gaits*. Further study is required to identify the alteration of muscle activation pattern.

Figure 6-12: Average EMG signals of Vastus Medialis (VM) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Figure 6-13: Average EMG signals of Tibialis Anterior (TA) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Figure 6-14: Average EMG signals of Gluteus Maximus (GM) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Figure 6-15: Average EMG signals of Biceps Femoris (BF) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

Figure 6-16: Average EMG signals of Gastrocnemius Lateral (GL) of both leg for normal walking (normal), gait locomotion assistance without ground contact (off ground), and gait locomotion assistance with ground contact (over-ground)

6.4 Clinical Trials

Clinical trials have been carried out for three subjects, two male SCI subjects (P1 and P2) and one female stroke subject (P3) in rehabilitation center of Tan Tock Seng Hospital. Details of the subjects are:

P1 is a 32 year old Asian male, with a complete C6 spinal cord injury. He 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.

Their description is summarized in Table 6-2.

Table 6-2: Detailed Information of SCI/Stroke Subjects

Subject Number		P1		P2		P3	
Gender		Male		Male		Female	
Age		32		67		73	
Years of injury		5		10		3	
Impairment		SCI		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	Full	Full
	Knee	-	-	Full	Full	Full	Full
	Ankle	-	-	20-40	30-40	NA*	0-20
Spasticity of limbs	Hip	-	-	-	-	-	-
	Knee	-	-	2	1+	-	1
	Ankle	-	-	-	-	-	-

* not available due to the block of AFO

Blood pressure and heart rate of the subjects were measured normal during the trials. Average time for harness setup is three minutes, for *NaTure-gaits* strapping is ten minutes (Figure 6-17). All subjects have gone thru four cycles of ten meters walk with *NaTure-gaits*, without any disruption during the walks. The subjects spent 16 – 20 minutes in the system for the four walks. Photos taken during the trials are provided in Figure 6-18. The protocol of the clinical trials is provided in Appendix - D.

A survey (survey form is attached in Appendix - E) has been carried out with the subjects after the trial. The comfort level of being strapped within the specially designed harness is acceptable by all the subjects. The strapping on *NaTure-gaits* is reported comfortable as well, except one subject felt minor tightness on the right thigh. All subjects felt comfortable of being assisted by the system to complete the gait locomotion.

Through the clinical trial, *NaTure-gaits* is further confirmed in its ability to provide gait locomotion assistance to patients. Although two persons are needed for the strapping process prior before the gait locomotion training, only one person is required for the control of *NaTure-gaits* during the training. No laborious involvement is required for that person as the person only control the movement of *NaTure-gaits* through computer.

Figure 6-17: Pre-setting for system trials: (a) Harness setup for one of the SCI subjects; (b) and (c) *NaTure-gaits* strapping for the same SCI subject

Figure 6-18: Snapshot of clinical trials of: (a) SCI subject – I, (b) SCI subject – II, and (c) Stroke subject

6.5 Practical Implementation of *GaitGen* and Motions Coordination on *NaTure-gaits*

The following procedures have been applied for implementing *GaitGen* generated motion on *NaTure-gaits* for robotic assisted gait locomotion training. The first step is to obtain the basic information and measure the anthropometry parameters of the targeted user. Basic information are the age, gender, height, and weight. Anthropometry parameters are ASIS breadth, thigh length, shank length and foot length. These data are the inputs of *GaitGen*. Subjects data are summarized in Table 6-3, S1-S5 indicates subject number one to subject number five.

Table 6-3: Basic Information and Anthropometry Parameters of the Trial Subjects

	Age	Gender	Height (cm)	Weight (kg)	ASIS Breadth (mm)	Thigh Length (mm)	Shank Length (mm)	Foot Length (mm)
S1	24	Male	173	68	238	460	420	250
S2	33	Male	168	61	254	405	403	243
S3	24	Male	175	62	272	500	430	251
S4	22	Male	173	66	238	445	450	266
S5	23	Male	160	70	233	332	391	223

Range of walking speeds suitable for the targeted user is predicted by *GaitGen* stage I. A desired walking speed for the training is selected from the suggested range. The walking speed is selected to match the walking speed of pelvic motion template, because a matching pelvic motion will be selected from database to complete the walking motion. A suitable pelvic motion template is not available for the walking speed of subject S4 predicted by *GaitGen*. It is believe that *GaitGen* is not able to predict a suitable walking speed for this subject. The walking speed of a selected pelvic motion template is used for this subject instead. *GaitGen* stage I will provide corresponding stride length for the selected walking speed. The predicted and selected basic gait parameters are listed in Table 6-4.

Table 6-4: Selected Basic Gait Parameters for the Trial Subjects Based on the Predicted Values

	Predicted Walking Speed (cm/s)	Selected Walking Speed (cm/s)	Predicted Stride Length (cm)
S1	116.60 ± 4.43	113.07	126.67
S2	119.28 ± 4.53	114.58	130.27
S3	115.91 ± 4.50	113.01	123.35
S4	114.69 ± 4.36	97.76	123.29
S5	113.25 ± 4.30	115.45	116.28

Reference gait pattern is predicted based on the selected walking speed and the corresponding stride length predicted, and the anthropometry parameters of each individual. The reference gait pattern is the desired gait pattern for patient. For clinical application, the rehabilitation of patient can be planned using the reference gait pattern as the training target. Initially, the patient is trained with a gait pattern close to the patient's pre-training gait pattern. Patient's pre-training gait pattern can be obtained via motion capture experiment or recorded on any robotic system with back-drive ability. Back-drive ability of robotic system refers to the ability of the robotic system can be driven voluntary by the subject with little or no constrain imposed by the robotic system [15].

When the rehabilitation progresses and the patient improve in his walking ability, the gait pattern is scaled closer to the desired gait pattern. Toward the end of the training program, the gait pattern uses will be desired gait pattern. Let the patient's pre-training

gait pattern be $\mathbf{M}_p$ and the reference gait pattern be $\mathbf{M}_r$. The following weighted average was used to plan the gait training pattern $\mathbf{M}_t$:

$$\mathbf{M}_t = (1 - \eta)\mathbf{M}_p + \eta\mathbf{M}_r \quad (6-1)$$

$$\mathbf{M} = \{x_{hip} \ y_{hip} \ z_{hip} \ \theta_{hip} \ \theta_{knee} \ \theta_{ankle}\}^T \quad (6-2)$$

For $\eta = 0$, the gait pattern is the pre-training pattern of the patient. The gait pattern is the desired gait pattern for $\eta = 1$. Any value of η between 0 and 1 will be a scaled gait pattern.

The gait training pattern for the trial is designed using $\eta = 1$. Modification of the gait pattern is carried out as discussed in Section 5.2.4. The modified walking motion of the five subjects recruited for the trial is provided in Appendix G. The modified walking motion is programmed to the motion controller of *NaTure-gaits*. The process of computing the input for motion controller based on the modified walking motion was discussed in Section 5.2 of the submitted thesis. The motions of pelvic and lower limb joints are synchronized by the motion controller based on percentage of gait cycle.

6.5.1 Subject Trials

Two gait patterns: gait training pattern generated by *GaitGen* and CGA (clinical gait analysis) pattern, were planned for the five subjects. Four walks were arranged for the subject. The two gait patterns were randomly arranged to be used equally within the four walks. Each walk covered a distance of ten meters, approximately eight gait cycles depending on the stride length of the pattern. At the end of each walk, the subject is asked to rate the gait pattern induced, using 1 to 5 scale. 1 for not resemble his normal walking, and 5 for resemble his normal walking. The survey outcome is given in Figure 6-19.

Figure 6-19: Rating of two patterns by the subjects based on their trial experience

Three subjects have rated *GaitGen* pattern is better than CGA, one subject gave the same rating for the two patterns, and one subject rate the CGA 1 point higher than *GaitGen*. In general, most subjects felt that *GaitGen* pattern is more resemble their walking pattern. This trial also shows that individual-specific gait pattern generation would allow *NaTure-gaits* to induce a gait pattern that provide better walking experience for the subject. Snapshots of the trial are given in Figure 6-20, showing how the subject walks with the *NaTure-gaits*.

Figure 6-20: Snapshots of video taken during the trial

In summary, this section provides the procedures for the practical implementation of the works on *NaTure-gaits*. Trials were carried out for 5 subjects using the gait pattern generated for *NaTure-gaits*.

6.6 Concluding Remarks

The main objective of the healthy subject trials is to verify the functionality and the safety of the *NaTure-gaits* before clinical trials with patient. The result was positive as four recruited subjects tolerated the device well throughout the entire trials period without any previous exposure to it. Each trial took about 1 hr for each subject, which is considerable amount of time spent. The subject trials verified the ability of *NaTure-gaits* in the provision of gait locomotion, through the designed passive control strategy.

The subjects did not experienced unnatural and great restriction of motion during the trial, except the thigh movement is slightly restricted by the harness. From the analysis of EMG signals, it is clearly shown that *NaTure-gaits* is capable of providing fully assisted gait locomotion to the subjects. The comfort level is acceptable by all the participated subjects based on their feedback.

The patient trial further confirmed the ability of *NaTure-gaits*. It is tested that the *NaTure-gaits* can provide gait locomotion seamlessly on the recruited patient. The patient tolerated 20 minutes of training on *NaTure-gaits*, and do not feel uncomfortable, even though the patient do not has any prior experience using the robotic system. The significant reduction on the laborious demanding for over-ground gait training is demonstrated by the patient trial.

From the recorded EMG signals, it is clearly shown that the muscle activation level is significantly reduced compared to trainer assisted over-ground walking training. This also shows that the patient is more actively participated in trainer assisted over-ground walking training compared to the robotic assisted gait training. Future work is suggested for development of control strategy, which encourages the active participation of patient during robotic assisted gait rehabilitation. More experimental results are required for EMG signals to determine the potential alteration of muscle activation pattern caused by *NaTure-gaits*.

Clinical trials have been successfully carried out for three subjects in Tan Tock Seng Hospital. All the subjects felt comfortable being assisted by *NaTure-gaits*. The subjects have walked for approximately 20 – 30 minutes with the assistance from the system. The laborious reduction advantage of robotic assisted over-ground gait rehabilitation is clearly demonstrated.

GaitGen is used to predict gait parameters and gait pattern for 5 subjects. Adjustment on the predicted gait pattern has been made in accordance to the work presented in Section 5.2.4. The final gait pattern is programmed to the motion controller of the *NaTUre-gaits*. The motion controller has playback the gait pattern (pelvic and lower limb motions) in a synchronized manner and guided the subjects to walk.

This page is intentionally left blank

Chapter 7

CONCLUSION AND FUTURE WORK

The work presented in this thesis focuses on the development of gait rehabilitation methodology for robotic assisted body weight supported gait rehabilitation. To realize the methodology, a robotic gait rehabilitation system, *NaTure-gaits*, is developed to fulfill the gait rehabilitation features proposed. This work also focuses on the natural gait pattern generation tailored for a specific subject. Effort has been placed on the development of passive control strategy for robotic assisted gait rehabilitation. Many of these works are implemented and demonstrated in the trials. Some works conducted have been documented and the list of publication is provided in Appendix - I.

7.1 Summary of Contributions

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

7.1.1 Individual-Specific Gait Pattern Generation

GaitGen has been designed for gait pattern generation for robotic assisted gait rehabilitation. Two stages have been created in *GaitGen*. Natural gait parameters can be predicted in the Stage-I of *GaitGen*, given the basic information (i.e. age, gender, weight, and height) of a specific individual. The Stage-II of *GaitGen* predicts the waveforms of gait pattern for lower limb joint angles during walking, based on the gait parameters and anthropometry data of the individual.

Stage-I of *GaitGen* are able to predict stride length and walking speed specifically for an individual, with the given basic information of the subject. Cadence can be then calculated with the stride length and walking speed. The accuracy of the trained GRNNs is validated by comparing with the actual gait parameters of the corresponding subject obtained by experiment. Approximately 86% of the predicted outcome for walking speed is 80% accurate and 97% of the predicted outcome for stride length respectively, is 90% accurate. A comparison is made with the normalized gait parameters equation and the clear advantage of this proposed methodology is shown.

Stage-II of *GaitGen* is designed to predict Fourier coefficient vectors from given gait parameters and lower limb anthropometric data. Lower limb joints angle waveforms can be constructed with the predicted Fourier coefficient vectors. The generated waveforms from predicted Fourier coefficient vectors were compared to the actual waveforms and Clinical Gait Analysis (CGA) waveforms to illustrate the advantage of the proposed gait pattern prediction model. The results show that lower limb joint angles waveforms generated by gait pattern prediction model are closer to actual waveforms compared to the CGA waveforms.

A systematic approach for gait pattern generation has been proposed and validated in this research work. *GaitGen* has been created for generating a gait pattern similar to actual gait pattern obtained by motion capture experiment. The gait pattern generated from *GaitGen* has been tested with *NaTure-gaits*.

7.1.2 Realization of Robotic System – *NaTure-gaits*

NaTure-gaits is developed for the proposed gait rehabilitation features. Features such as unrestricted pelvic motion assistance, lower limb motion assistance, body weight support, body weight shifting, and over-ground walking experience are made possible with the developed modules.

Three sub-modules are built to enable the developed robotic gait rehabilitation system to provide body weight supported gait rehabilitation, with actuated assistance to hip and lower limb movement, in the context of over-ground walking. The unique feature found on *NaTure-gaits* is the provision of pelvic motion assistance and body weight support with the pelvic assistance mechanism. *NaTure-gaits* is one of the two robotic systems, which provide gait rehabilitation in the context of over-ground walking. Another system is the WalkTrainer.

Pelvic assistance mechanism is developed for pelvic motion assistance and body weight support. The mechanism is a two-five DoFs robotic arms provide the essential movements for the pelvis. Without the restriction of cable harness body weight support, natural pelvic motion can be induced by pelvic assistance mechanism and also promoting body weight shifting. A six DoFs robotic orthosis is developed for the lower limb motion assistance. It has the similar features of lower limb orthosis found on most

robotic gait rehabilitation system. The lower limb joints' (hip, knee, and ankle) movement in the sagittal plane is assisted by the robotic orthosis.

7.1.3 Control Strategy

Passive control strategy and one of the active control strategies are proposed in this research work. The proposed active control strategy is active pelvic motion assistance with adaptive body weight support, which provides adaptive body weight support based on the pelvic postural deviation of the subject.

In order to obtain the reference gait locomotion for *NaTure-gaits*, motion capture experiment is carried out. Pelvis movement and lower limb joint angles are computed from the data obtained in motion capture experiment. It is post-processed to make the gait locomotion to be played back on *NaTure-gaits* seamlessly. The passive control strategy is implemented on *NaTure-gaits* and tested in the human subject trials.

Natural pelvic motion assistance with adaptive body weight support control strategy is proposed. The control strategy is created by Virtual Model Control method. The developed control algorithm fulfilled the requirement of the control strategy, which allows the patient to influence the reference pelvic trajectory.

Virtual Model Control (VMC) approach has been incorporated into the control algorithm. A virtual spring is created at the body weight support direction. The supporting force of the body weight support is created by adjusting the spring constant of the virtual spring. An adaptive controller is created to update the spring constant based on the deviation between the actual position of the patient and the desired position. Experiment shows that the proposed control strategy allowed certain extent of deviation from the desired trajectories, which is the intended aim of the designed control strategy. The experiment also shows that the supporting force is increased with the increment in deviation between actual and desired trajectories.

The performance and stability of the developed control strategy are tested by implementing on a preliminary experimental setup. The control strategy can be transferred to *NaTure-gaits*, by changing the system block in the control algorithm. *NaTure-gaits* has been scheduled for clinical trials in February. Before that, major modification will not be carried out on the system so as not to delay the trials. The actual performance and

stability of the developed control strategy on *NaTure-gaits* will be carried out after the clinical trials.

7.1.4 Human Subject Trials

Human subject trials have been successfully carried out for four healthy subjects and one patient. The results were positive as the recruited subjects tolerated the device well throughout the entire trials period without any previous exposure to it. Each trial took about 1 hr for each subject, which is considerable amount of time spent. The subject trials verified the ability of *NaTure-gaits* in the provision of gait locomotion, through the designed passive control strategy.

The subjects did not experience unnatural and great restriction of motion during the trial, except the thigh movement is slightly restricted by the harness. The analysis of EMG signals showed that *NaTure-gaits* is capable of providing fully assisted gait locomotion to the subjects. The comfort level is acceptable by all the participated subjects based on their feedback.

The provision of gait locomotion on patient shows that the proposed robotic system and the passive control strategy are suitable for patient use. The patients tolerated 20 minutes of training on *NaTure-gaits*, and do not felt uncomfortable, even though the patient do not has any prior experience using the robotic system. The significant reduction on the laborious demanding of human helpers for over-ground gait training is also demonstrated.

7.2 Suggestion for Future Works

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

Using Support Vector Machines for Gait Parameters Prediction

The Stage-I of *GaitGen* has adopted GRNN for the function prediction of the inputs and outputs. The GRNN has good accuracy in predicting the wanted gait parameters. Nevertheless, using different function approximation approach such as support vector machine might yield better outcome. Support vector machines are introduced in year 1995, and have become one of the preeminent machine learning paradigms. Support vector ma-

chines are commonly employed in areas that range from handwriting recognition to bioinformatics to the mining of very large databases [146].

Generation of Waveforms for Pelvic Motion

The work for the waveforms of lower limb joint angles is completed in Stage-II of *GaitGen*. The same concept is applicable for the pelvic movement. Further study is suggested for identify the most appropriate factors affecting the waveform generation. Instead of using multiple linear regression model for the function approximation for the inputs and outputs, other model or approach, such as GRNN, support vector machines, are suggested for the function approximation.

Enhanced Control Strategy

The active control strategy proposed in this work at Section 5.3 was not validated on the robotic gait rehabilitation system. Future work is suggested for the implementation of the control strategy on actual rehabilitation session.

Currently, only impedance control is introduced for the pelvic movement. Future work is suggested for the incorporation of two other control algorithms. One of the algorithms is adaptive path control, which provides flexible “tunnel” and an adjustable deviation around the “tunnel”. The adaptation can be based on patient’s performance in real time. Fuzzy impedance is the other algorithm suggested to explore in future work. The fuzzy controller can be employed to modify the width of the “tunnel” and the impedance parameters. The “tunnel” and the deviation zone around the “tunnel” can be adjusted during the training process with the implementation of the fuzzy controller.

Clinical trial

More clinical trials should be arranged for more patients with different gait conditions. The extended clinical trials will provide more relevant clinical feedback for the improvement of the system.

This page is intentionally left blank

Appendix - A

Information on Stroke and Spinal Cord Injury

Nervous System

Nervous system is the highly complex mechanism in our bodies that controls, stimulates, and coordinates all other body systems. It can be anatomically divided into the Central Nervous System (CNS), the Peripheral Nervous System (PNS), and the Autonomic Nervous System (ANS). The division of the nervous system is given in Figure A-1.

Figure A-1: Nervous system of human [147]

Brain

The brain is made up of three parts, the cerebrum, brainstem and cerebellum. Cerebrum is the largest portion of the brain, which is responsible for the highest-level mental functions. Cerebrum can be further divided into four zones, Frontal Lobe, Parietal Lobe, Temporal Lobe, and Occipital Lobe. The functions of the Cerebral Lobes are given as follows:

Frontal Lobe	Thinking, planning, and central executive functions; motor execution
---------------------	--

Parietal Lobe	Somatosensory perception integration of visual and somatospatial information
Temporal Lobe	Language function and auditory perception involved in long term memory and emotion
Occipital Lobe	Visual perception and processing

Brainstem lies below the cerebrum. It can be divided into three parts, midbrain (center for visual reflexes), pons (Latin for 'bridge', located between the midbrain and medulla oblongata), and medulla oblongata (center for automatic control of respiration and heart rate).

Cerebellum is Latin for 'little brain', located in the posterior portion of the cranium behind the pons and medulla. Main functions of the cerebellum include control of muscle coordination, tone, and posture.

Spinal Cord

Spinal cord is the portion of the central nervous system, which extends from the medulla oblongata just above the foramen magnum at the base of the skull to the level of the L1 or L2 vertebra. It is housed within the vertebral canal, also called the vertebral foramen. The vertebral body and vertebral arches form a protection layer anteriorly and posteriorly for the spinal cord (Figure A-2). Below the L2 level is the collection of nerve roots running down from the spinal cord much like a horse's tail, hence the name cauda equina. It is made up of the nerve roots for L2 through S5.

Figure A-2: Fifth lumbar vertebra

The spinal cord can be classified into central gray matter and peripheral white matter by looking at the cross sectional view (Figure A-3). Gray matter is 'H' or 'Butterfly' shape area at the middle of spinal cord, which contains neuronal cell bodies and synapses. Posterior horn, which transmits sensory impulses, is positioned at the top portion of the 'H'. Anterior horn, which transmits motor impulses, is positioned at the lower portion of the 'H'. Posterior (dorsal) columns, which transmit the sensations of proprioception, pressure, and vibration, are located in the posterior medial portions of the spinal cord. Ascending (sensory) and descending (motor) fiber pathways are contained in the white matter. The pathway carries a particular type of impulse, such as touch, from and to a specific area. The pathways cross over from one side of the body to the other at different levels. This crossover phenomenon is the reason why a stroke on the left side of the brain will affect the right side of the body.

Figure A-3: Cross section of spinal cord [147]

In summary, spinal cord serves as the pathway of signal exchange from brain to sensory fibers and spinal nerves. Dermatome (Figure A-4) refers to the area of skin supplied with the sensory fibers of a spinal nerve. Often, there is an overlap of contiguous dermatome, which is the result of complete anesthesia of an area. This will not occur unless more than two spinal nerves have lost function. If only one of the spinal nerves is injured, the sensation will be decreased or altered, but it will not be absent.

The general innervations level of major muscles is shown in Figure A-5. Most muscles take innervations from more than one spinal level. Injury at one spinal level may weaken a muscle, but some of the function might be remained.

Figure A-4: Dermatomes [148]

Muscle innervation level	Spinal nerve level	Vertebra level
	C1	
Diaphragm & trapezius	C2	C1
	C3	C2
	C4	C3
Deltoid & biceps	C5	C4
Wrist extensors	C6	C5
Triceps	C7	C6
Hand intrinsic	C8	C7
	T1	T1
	T2	T2
Intercostals	T3	T3
	T4	T4
	T5	T5
	T6	T6
	T7	T7
	T8	T8
Abdominals	T9	T9
	T10	T10
	T11	T11
	T12	T12
	L1	L1
	L2	L2
Leg muscles	L3	L3
	L4	L4
Foot muscles	L5	L5
	S1	
	S2	
	S3	
Bowel & bladder	S4	
	S5	

Figure A-5: Spinal cord level and muscle innervations [147]

Stroke

Stroke or '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.

Effect of Stroke

Our humanity has been distinguished based on the ability to define the world and our place. This world-making capacity has been altered permanently by a stroke or brain attack, which changes stroke patients' world, from a comprehensible and manageable once, to a confusing, intimidating and hostile environment. The patient's most basic capabilities of interacting with the world can be robbed by a stroke.

Depending on the location of the brain damage and the severity of the damage, the specific abilities controlled by that brain area will be lost or affected. The brain is a highly complex organ and the function or ability of every individual area is not clearly understood at this moment. In general, the effects of stroke can be classified based on four primary parts of the brain.

Right-Hemisphere Stroke

Left side of our body is controlled by the right hemisphere, which also controls the analytical and perceptual tasks, such as judging distance, size, speed, position and seeing how parts are connected as a whole. Stroke that occurs at the right hemisphere often causes paralysis in the left side of the body which is known as left hemiplegia. The stroke might affect the spatial and perceptual abilities of the stroke survivor and cause them to misjudge distances which may lead to a fall. The patient might not be able to guide their hands to pick up an object, button a shirt or tie their shoes. For some patients, they cannot tell right-side up from upside-down when reading. The patient may also experience left-sided neglect, which leads them to ‘forget’ or ‘ignore’ objects or people on their left side. Some survivors will have short-term memory – they have poor memory of what has just happened but they are able to recount events that happened several years ago.

Left-Hemisphere Stroke

Movement of the right side of the body is controlled by left hemisphere, which controls speech and language abilities for most people as well. Stroke that occurs at the left-hemisphere results in paralysis of the right side of the body, which is known as right hemiplegia. Aphasia, a term that is used to describe a wide range of speech and language problems, is developed after left-hemisphere stroke. The specific component for communication of the patient might be affected, resulting in poor talking ability, but it may not affect the ability of writing, reading or understanding speech for the same patient. Unlike right-hemisphere stroke, left-hemisphere stroke patient often develop a slow and cautious behavioral style and frequent instruction and feedback may be required to complete tasks. Lastly, the patient may develop similar memory problems like those of

right-hemisphere stroke patient. The patient may have shortened retention spans, difficulty in learning new information and problems in conceptualizing and generalizing.

Cerebellar Stroke

Our reflexes and much of our balance and coordination are controlled by cerebellum. A stroke that occurs in the cerebellum causes abnormal reflexes of the head and torso, coordination and balance problems, dizziness, nausea and vomiting.

Brain Stem Stroke

Brain stem controls all of our involuntary, 'life-support' functions, such as breathing rate, blood pressure and heartbeat. It also controls abilities like eye movements, hearing, speech and swallowing. Thus, stroke that occur at brain stem are especially devastating. The signal generated in brain must travel through the brain stem to the other parts of body like arms and legs, brain stem stroke might leave the patients with paralysis in one or both sides of the body.

Common Effects

Apart from classified stroke effects based on the location of the stroke occurring in brain, there are some common ways in which stroke can affect a person. The effects include physical effects, communication effects and behavioral effects. The physical effects are the important aspect to be examined in gait rehabilitation and will be further discussed below.

The physical effects brought on by stroke include spasticity, balancing issues, toe curl, central pain syndrome (CPS), etc. Spasticity is one of the effects of stroke that causes muscles to involuntarily contract when the patients try to move their limbs, creating stiffness and tightness. The tendons and soft tissues surrounding the muscle become tight when it cannot complete its full range of motion. In the long run, the muscle freeze permanently into an abnormal and often, painful position. For an arm that has spasticity effect, the fist will always close tight, with the elbow bent and the arm pressed against the chest. Stiff knee, pointed foot and curling of the toes will be found on a patient with leg spasticity.

Stroke survivors often face problem in balancing. Strokes are likely to affect the central system which causes the stroke patient to experience dizziness and imbalance,

and increasing the chances of falling. Central pain syndrome (CPS) is a rare syndrome among stroke patients. The pain is an extremely persistent, intractable type of pain which can be quite debilitating and depressing to the patient.

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 [149]:

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.

The neurological level of injury is defined as the most caudal level of the spinal cord that exhibits intact sensory and motor functioning. To further classify injury level, two other tests are carried out. Sensation test is carried out in key areas shown in Figure A-6 and key muscle for muscles strength grading is shown in Table A-1.

Figure A-6: Key sensory areas for sensation test of SCI [149]

Table A-1: Key Muscles for Muscle Grading

Upper Limb Key Muscles		Lower Limb Key Muscles	
C5	Elbow flexors	L2	Hip flexors
C6	Wrist extensors	L3	Knee extensors
C7	Elbow extensors	L4	Ankle dorsiflexors
C8	Finger flexors	L5	Long toe extensors
T1	Finger abductors	S1	Ankle plantar flexors

The muscle strength is graded according to the standard of ASIA:

0: total paralysis

1: palpable or visible contraction

2: active movement, full range of motion, gravity eliminated

3: active movement, full range of motion, against gravity

4: active movement, full range of motion, against gravity and provides some resistance

5: active movement, full range of motion, against gravity and provides normal resistance

5*: muscle able to exert, in examiner's judgment, sufficient resistance to be considered normal if identifiable inhibiting factors were not present

NT: not testable. Patient unable to reliably exert effort or muscle is unavailable for testing due to factors such as immobilization, pain on effort or contracture

Sensory testing is carried out by using light touch and pin prick. The sensory scoring is as follows:

0: absent

1: impaired or hyperesthesia

2: intact

NT: not testable.

If the patient cannot differentiate between a prick by sharp point of dull edge, a zero is given.

Impact of Spinal Cord Injury

Spinal cord injury brings about a sudden change to a person's normal living. The injury not only affects a person physically but also psychosocially. A cord-injured person, often an active individual, may be active in many activities like swimming, driving, or earning a living. Yet the very next moment after injury, he/she is incapable of all the activities he/she is used to doing. Sooner or later, the person finds himself/herself paralyzed, immobile, confined to wheel chair, incontinent or perhaps isolated. As the person's rehabilitation progresses and he/she finally manage to venture out from the rehabilitation center, a social and physical environment that seems almost hostile in its inaccessibility is another challenge awaiting the person.

The most obvious impact of SCI is the loss of control over the physical functions. A person's loss of control over some or all his/her limb and trunk depends on the level and completeness of injury. This dramatically changes the living pattern of a person, or worse, the person might even lose the control of bowel and bladder movement and face an alteration in sexual function.

On the other hand, the community often overlooks the social impact of SCI. Often, individuals with disabilities are seen as less desirable and substantially different from able-bodied people. Discrimination of disabled person manifests itself in many ways. Person with disabilities often found themselves being treated discriminatory in many situation (seeking employment, interpersonal relationships, and etc.). Perception of the individual is likely to be dominated by the disability when an able-bodied person meets someone with a disability. The disability characteristics are always the focus rather than other qualities.

Able-bodied people are likely to be uncomfortable talking to people with disabilities and tend to avoid contact with them due to the devaluation and stereotyping of them. This results in the social isolation following spinal cord injury.

Appendix - B

Basic Information of Human Walking

Gait Terminology

Gait, is a term used to describe the particular way or manner of moving with foot. The two basic human gaits are walking and running. A better understanding of human gait is crucial for gait pattern planning. This section provides the basic information about human walking to facilitate the understanding of some terms used throughout the paper.

Locomotion is the process by which the animal moves itself from one geographic position to another. Human walking is a relatively unique form of locomotion in that it is bipedal. Locomotion includes starting, stopping, speed changing, alteration in direction, and adjustments of motion to adapt the changes in slope of the ground. In general, human walking can be distinctly differentiated into three stages:

- Development stage (from rest to a speed)
- Rhythmic stage (maintaining at some constant average velocity)
- Decay stage (going back to rest)

Rhythmic stage is the stage which most research concentrated on. The patterns of locomotion during this rhythmic stage are remarkably consistent between individuals and form the basis of the information to be presented in this section.

Gait cycle, also known as stride, is made up by a left and right step. A gait cycle has two crucial dimensions: the time taken to complete the cycle, cycle time (usually step time is a preferred parameter for gait analysis usage, left and right step time in one gait cycle constitute a cycle time), and the distance covered during one cycle (stride length). The time duration can be expressed alternatively as stride frequency.

Gait cycle consists of two periods of double support and two periods of single support. Double support is the period when both feet are touching the floor while single support is the period when only one foot is touching the floor. Non-support period, which would not happen in walking, refers to the period when no foot is in contact with the floor.

Many terms have been developed based on the traditional terminology to describe the components of walking. Among them all, the terminology, RLA terms, developed by the Gait Laboratory at Rancho Los Amigos (RLA) Medical Center has started to gain acceptance from the rest of its peers. The distinct difference between the two sets of terms is that the traditional terms refer to points in time whereas RLA terms refer to periods of time. Traditional terminology accurately pointed out the key points within the gait cycle. RLA terminology reflects the moving or dynamic nature of gait accurately. Both set of terminologies are similar with a few exceptions, but it is best to be familiar with both sets and terms. In this thesis, the gait event terms from traditional definition are used, as this work analyzes at the points in time within the gait cycle.

Gait Events

A complete gait cycle can be divided into stance phase and swing phase. Stance phase occurs when the foot is in contact with the floor. It starts with heel strike and ends with a toe off for the same foot. The time of stance phase is roughly 60% of the time of a gait cycle. Swing phase occurs when the foot is not in-contact with the floor. It begins when the foot leaves the floor and ends right before the same foot touching the floor again. It makes up the other 40% of the gait cycle.

Traditionally, stance phase can be sub-divided into five components: (1) heel strike, (2) foot flat, (3) mid-stance, (4) heel off, and (5) toe off. An illustration of stance phase components is shown in Figure B-1. Heel strike refers to the time when the heel first comes into contact with the floor after the swing phase. Ankle will be at its neutral position between dorsiflexion and plantarflexion. This is the period of initial contact in RLA. Foot flat occurs at about 7% of gait cycle, shortly after heel strike. It is the event when the foot is flat on the floor. This is comparable with the period, loading response in RLA. Midstance is the point when the body weight is transferred to the weight bearing foot. This phase usually occurs at 30% of the gait cycle. Midstance is the period between the end of foot flat and the end of midstance in RLA definition. Heel off is the point when heel rises off from the floor, usually at 40% of the gait cycle. RLA terminology defines a similar period, terminal stance, which is the period between the end of mid-stance and the end of heel off. Stance phase ends with toe off. It is the instance when the

toe lifts off from floor and signifies the start of swing phase. A similar component in RLA terms is the pre-swing, which is the period of initial contact of the contra-lateral foot and ends with ipsilateral toe off.

Figure B-1: Components of Stance Phase (Modified from [134])

Swing phase is divided into three components according to traditional terminology. The components are (1) acceleration, (2) mid-swing, and (3) deceleration. For swing phase, both traditional terminology and RLA terminology does not have any significant difference in their definition. Figure B-1 illustrates the components of swing phase.

Acceleration is the first part of swing phase. It begins right after the toe lifts off from the floor (toe off) and starts to swing. RLA defines initial swing as the period between the end of toe off and the end of acceleration. Mid-swing is the instance when the swinging leg is passing directly under the body. In RLA terminology, mid-swing starts from end of acceleration to the end of mid-swing. Deceleration occurs after mid-swing, when the leg is decelerating in preparation for heel strike. The corresponding term in RLA definition is terminal swing, which covers the period from the end of mid-swing to the event right before initial contact.

Appendix - C

Detailed Mathematics for Computation of Lower Limb Joint Angles

In this section, the computation of lower limb joint angles with the coordinate of fifteen markers is described. The reference systems, $\mathbf{uvw}$, are first computed for each segment, to predict the positions of joint centers and segment endpoints. Segment reference frames (xyz) are generated with the joint center positions and the external marker positions. The segment reference frames are embedded at the center of gravity of each segment Figure C-1.

Figure C-1: Embedment of segment reference frames (xyz) at the center of gravity of each segment

The unit vectors for the right foot are defined as

$$\mathbf{u}_{Foot,r} = (\mathbf{P}_1 - \mathbf{P}_2) / |\mathbf{P}_1 - \mathbf{P}_2| \quad (C-1)$$

$$\mathbf{w}_{Foot,r} = \frac{(\mathbf{P}_1 - \mathbf{P}_3) \times (\mathbf{P}_1 - \mathbf{P}_3)}{|(\mathbf{P}_1 - \mathbf{P}_3) \times (\mathbf{P}_1 - \mathbf{P}_3)|} \quad (C-2)$$

$$\mathbf{v}_{Foot,r} = \mathbf{w}_{Foot,r} \times \mathbf{u}_{Foot,r} \quad (C-3)$$

The position of the right ankle ($\mathbf{P}_{Ankle,r}$) and right toe ($\mathbf{P}_{Toe,r}$) can be obtained by

$$\mathbf{P}_{Ankle,r} = \mathbf{P}_3 + 0.016A_{13}\mathbf{u}_{Foot,r} + 0.392A_{15}\mathbf{v}_{Foot,r} + 0.478A_{17}\mathbf{w}_{Foot,r} \quad (C-4)$$

$$\mathbf{P}_{Toe,r} = \mathbf{P}_3 + 0.742A_{13}\mathbf{u}_{Foot,r} + 1.074A_{15}\mathbf{v}_{Foot,r} - 0.187A_{19}\mathbf{w}_{Foot,r} \quad (C-5)$$

The unit vectors for the left foot are defined as

$$\mathbf{u}_{Foot,l} = (\mathbf{P}_8 - \mathbf{P}_9) / |\mathbf{P}_8 - \mathbf{P}_9| \quad (C-6)$$

$$\mathbf{w}_{Foot,l} = \frac{(\mathbf{P}_8 - \mathbf{P}_{10}) \times (\mathbf{P}_9 - \mathbf{P}_{10})}{|(\mathbf{P}_8 - \mathbf{P}_{10}) \times (\mathbf{P}_9 - \mathbf{P}_{10})|} \quad (C-7)$$

$$\mathbf{v}_{Foot,l} = \mathbf{w}_{Foot,l} \times \mathbf{u}_{Foot,l} \quad (C-8)$$

The position of the left ankle ($\mathbf{P}_{Ankle,l}$) and left toe ($\mathbf{P}_{Toe,l}$) can be obtained by

$$\mathbf{P}_{Ankle,l} = \mathbf{P}_{10} + 0.016A_{14}\mathbf{u}_{Foot,l} + 0.392A_{16}\mathbf{v}_{Foot,l} - 0.478A_{18}\mathbf{w}_{Foot,l} \quad (C-9)$$

$$\mathbf{P}_{Toe,l} = \mathbf{P}_{10} + 0.742A_{14}\mathbf{u}_{Foot,l} + 1.074A_{16}\mathbf{v}_{Foot,l} + 0.187A_{20}\mathbf{w}_{Foot,l} \quad (C-10)$$

The unit vectors for the right calf are defined as

$$\mathbf{v}_{Calf,r} = (\mathbf{P}_3 - \mathbf{P}_5) / |\mathbf{P}_3 - \mathbf{P}_5| \quad (C-11)$$

$$\mathbf{u}_{Calf,r} = \frac{(\mathbf{P}_4 - \mathbf{P}_5) \times (\mathbf{P}_3 - \mathbf{P}_5)}{|(\mathbf{P}_4 - \mathbf{P}_5) \times (\mathbf{P}_3 - \mathbf{P}_5)|} \quad (C-12)$$

$$\mathbf{w}_{Calf,r} = \mathbf{u}_{Calf,r} \times \mathbf{v}_{Calf,r} \quad (C-13)$$

The position of the right knee ($\mathbf{P}_{Knee,r}$) can be obtained by

$$\begin{aligned} \mathbf{P}_{Knee,r} &= \mathbf{P}_5 + 0.000A_{11}\mathbf{u}_{Calf,r} + 0.000A_{11}\mathbf{v}_{Calf,r} + 0.5A_{11}\mathbf{w}_{Calf,r} \\ &= \mathbf{P}_5 + 0.5A_{11}\mathbf{w}_{Calf,r} \end{aligned} \quad (C-14)$$

The unit vectors for the left calf are defined as

$$\mathbf{v}_{Calf,l} = (\mathbf{P}_{10} - \mathbf{P}_{12}) / |\mathbf{P}_{10} - \mathbf{P}_{12}| \quad (C-15)$$

$$\mathbf{w}_{Calf,l} = \frac{(\mathbf{P}_{11} - \mathbf{P}_{12}) \times (\mathbf{P}_{10} - \mathbf{P}_{12})}{|(\mathbf{P}_{11} - \mathbf{P}_{12}) \times (\mathbf{P}_{10} - \mathbf{P}_{12})|} \quad (C-16)$$

$$\mathbf{u}_{Calf,l} = \mathbf{w}_{Calf,l} \times \mathbf{v}_{Calf,l} \quad (C-17)$$

The position of the left knee ($\mathbf{P}_{Knee,l}$) can be obtained by

$$\begin{aligned} \mathbf{P}_{Knee,l} &= \mathbf{P}_{12} + 0.000A_{12}\mathbf{u}_{Calf,l} + 0.000A_{12}\mathbf{v}_{Calf,l} - 0.5A_{12}\mathbf{w}_{Calf,l} \\ &= \mathbf{P}_{12} - 0.5A_{12}\mathbf{w}_{Calf,l} \end{aligned} \quad (C-18)$$

The unit vectors for the pelvis are defined as

$$\mathbf{v}_{Pelvis} = (\mathbf{P}_{14} - \mathbf{P}_7) / |\mathbf{P}_{14} - \mathbf{P}_7| \quad (C-19)$$

$$\mathbf{w}_{Pelvis} = \frac{(\mathbf{P}_7 - \mathbf{P}_{15}) \times (\mathbf{P}_{14} - \mathbf{P}_{15})}{|(\mathbf{P}_7 - \mathbf{P}_{15}) \times (\mathbf{P}_{14} - \mathbf{P}_{15})|} \quad (C-20)$$

$$\mathbf{u}_{Pelvis} = \mathbf{v}_{Pelvis} \times \mathbf{w}_{Pelvis} \quad (C-21)$$

The position of the right hip joint center ($\mathbf{P}_{HJC,r}$) and left hip joint center ($\mathbf{P}_{HJC,l}$) can be obtained by

$$\mathbf{P}_{HJC,r} = \mathbf{P}_{15} + 0.598A_2\mathbf{u}_{Pelvis} - 0.344A_2\mathbf{v}_{Pelvis} - 0.290A_2\mathbf{w}_{Pelvis} \quad (C-22)$$

$$\mathbf{P}_{HJC,l} = \mathbf{P}_{15} + 0.598A_2\mathbf{u}_{Pelvis} + 0.344A_2\mathbf{v}_{Pelvis} - 0.290A_2\mathbf{w}_{Pelvis} \quad (C-23)$$

With the joint center positions and external marker positions, the segment reference frames (xyz) embedded to the center of gravity of each segment can be generated.

Pelvic segment:

$$\mathbf{i}_{Pelvis} = \mathbf{w}_{Pelvis} \quad (C-24)$$

$$\mathbf{j}_{Pelvis} = \mathbf{u}_{Pelvis} \quad (C-25)$$

$$\mathbf{k}_{Pelvis} = \mathbf{v}_{Pelvis} \quad (C-26)$$

Right thigh segment

$$\mathbf{i}_{Thigh,r} = \frac{(\mathbf{P}_{HJC,r} - \mathbf{P}_{Knee,r})}{|\mathbf{P}_{HJC,r} - \mathbf{P}_{Knee,r}|} \quad (C-27)$$

$$\mathbf{j}_{Thigh,r} = \frac{(\mathbf{P}_6 - \mathbf{P}_{HJC,r}) \times (\mathbf{P}_{Knee,r} - \mathbf{P}_{HJC,r})}{|(\mathbf{P}_6 - \mathbf{P}_{HJC,r}) \times (\mathbf{P}_{Knee,r} - \mathbf{P}_{HJC,r})|} \quad (C-28)$$

$$\mathbf{k}_{Thigh,r} = \mathbf{i}_{Thigh,r} \times \mathbf{j}_{Thigh,r} \quad (C-29)$$

Left thigh segment

$$\mathbf{i}_{Thigh,l} = \frac{(\mathbf{P}_{HJC,l} - \mathbf{P}_{Knee,l})}{|\mathbf{P}_{HJC,l} - \mathbf{P}_{Knee,l}|} \quad (C-30)$$

$$\mathbf{j}_{Thigh,l} = \frac{(\mathbf{P}_{Knee,l} - \mathbf{P}_{HJC,l}) \times (\mathbf{P}_{13} - \mathbf{P}_{HJC,l})}{|(\mathbf{P}_{Knee,l} - \mathbf{P}_{HJC,l}) \times (\mathbf{P}_{13} - \mathbf{P}_{HJC,l})|} \quad (C-31)$$

$$\mathbf{k}_{Thigh,l} = \mathbf{i}_{Thigh,l} \times \mathbf{j}_{Thigh,l} \quad (C-32)$$

Right calf segment

$$\mathbf{i}_{Calf,r} = \frac{(\mathbf{P}_{Knee,r} - \mathbf{P}_{Ankle,r})}{\left\| (\mathbf{P}_{Knee,r} - \mathbf{P}_{Ankle,r}) \right\|} \quad (\text{C-33})$$

$$\mathbf{j}_{Calf,r} = \frac{(\mathbf{P}_4 - \mathbf{P}_{Knee,r}) \times (\mathbf{P}_{Ankle,r} - \mathbf{P}_{Knee,r})}{\left\| (\mathbf{P}_4 - \mathbf{P}_{Knee,r}) \times (\mathbf{P}_{Ankle,r} - \mathbf{P}_{Knee,r}) \right\|} \quad (\text{C-34})$$

$$\mathbf{k}_{Calf,r} = \mathbf{i}_{Calf,r} \times \mathbf{j}_{Calf,r} \quad (\text{C-35})$$

Left thigh segment

$$\mathbf{i}_{Calf,l} = \frac{(\mathbf{P}_{Knee,l} - \mathbf{P}_{Ankle,l})}{\left\| (\mathbf{P}_{Knee,l} - \mathbf{P}_{Ankle,l}) \right\|} \quad (\text{C-36})$$

$$\mathbf{j}_{Calf,l} = \frac{(\mathbf{P}_{Ankle,l} - \mathbf{P}_{Knee,l}) \times (\mathbf{P}_{12} - \mathbf{P}_{Knee,l})}{\left\| (\mathbf{P}_{Ankle,l} - \mathbf{P}_{Knee,l}) \times (\mathbf{P}_{12} - \mathbf{P}_{Knee,l}) \right\|} \quad (\text{C-37})$$

$$\mathbf{k}_{Calf,l} = \mathbf{i}_{Calf,l} \times \mathbf{j}_{Calf,l} \quad (\text{C-38})$$

Right foot segment

$$\mathbf{i}_{Foot,r} = \frac{(\mathbf{P}_2 - \mathbf{P}_{Toe,r})}{\left\| (\mathbf{P}_2 - \mathbf{P}_{Toe,r}) \right\|} \quad (\text{C-39})$$

$$\mathbf{k}_{Foot,r} = \frac{(\mathbf{P}_{Ankle,r} - \mathbf{P}_2) \times (\mathbf{P}_{Toe,r} - \mathbf{P}_2)}{\left\| (\mathbf{P}_{Ankle,r} - \mathbf{P}_2) \times (\mathbf{P}_{Toe,r} - \mathbf{P}_2) \right\|} \quad (\text{C-40})$$

$$\mathbf{j}_{Foot,r} = \mathbf{k}_{Foot,r} \times \mathbf{i}_{Foot,r} \quad (\text{C-41})$$

Left foot segment

$$\mathbf{i}_{Foot,l} = \frac{(\mathbf{P}_9 - \mathbf{P}_{Toe,l})}{\left\| (\mathbf{P}_9 - \mathbf{P}_{Toe,l}) \right\|} \quad (\text{C-42})$$

$$\mathbf{k}_{Foot,l} = \frac{(\mathbf{P}_{Ankle,l} - \mathbf{P}_9) \times (\mathbf{P}_{Toe,l} - \mathbf{P}_9)}{\left\| (\mathbf{P}_{Ankle,l} - \mathbf{P}_9) \times (\mathbf{P}_{Toe,l} - \mathbf{P}_9) \right\|} \quad (\text{C-43})$$

$$\mathbf{j}_{Foot,l} = \mathbf{i}_{Foot,l} \times \mathbf{j}_{Foot,l} \quad (\text{C-44})$$

The angular kinematics of lower limb during walking can be calculated with the segment reference frames. The following conventions are applicable to all six joints:

$\mathbf{i}_{Distal}$: internal/external rotation axis

$\mathbf{k}_{Proximal}$: flexion/extension axis

$\mathbf{l}_{Joint}$: abduction/adduction axis

$$\mathbf{l}_{Joint} = \frac{\mathbf{k}_{Proximal} \times \mathbf{i}_{Distal}}{|\mathbf{k}_{Proximal} \times \mathbf{i}_{Distal}|} \quad (C-45)$$

α = flexion/extension angle

β = abduction/adduction angle

γ = internal/external rotation angle

Right Hip

$$\alpha_{Hip,r} = \sin^{-1} [\mathbf{l}_{Hip,r} \cdot \mathbf{i}_{Pelvis}] \quad (C-46)$$

$$\beta_{Hip,r} = \sin^{-1} [\mathbf{k}_{Pelvis} \cdot \mathbf{i}_{Thigh,r}] \quad (C-47)$$

$$\gamma_{Hip,r} = -\sin^{-1} [\mathbf{l}_{Hip,r} \cdot \mathbf{k}_{Thigh,r}] \quad (C-48)$$

Left Hip

$$\alpha_{Hip,l} = \sin^{-1} [\mathbf{l}_{Hip,l} \cdot \mathbf{i}_{Pelvis}] \quad (C-49)$$

$$\beta_{Hip,l} = -\sin^{-1} [\mathbf{k}_{Pelvis} \cdot \mathbf{i}_{Thigh,l}] \quad (C-50)$$

$$\gamma_{Hip,l} = \sin^{-1} [\mathbf{l}_{Hip,l} \cdot \mathbf{k}_{Thigh,l}] \quad (C-51)$$

Right Knee

$$\alpha_{Knee,r} = -\sin^{-1} [\mathbf{l}_{Knee,r} \cdot \mathbf{i}_{Thigh,r}] \quad (C-52)$$

$$\beta_{Knee,r} = \sin^{-1} [\mathbf{k}_{Thigh,r} \cdot \mathbf{i}_{Calf,r}] \quad (C-53)$$

$$\gamma_{Knee,r} = -\sin^{-1} [\mathbf{l}_{Knee,r} \cdot \mathbf{k}_{Calf,r}] \quad (C-54)$$

Left Knee

$$\alpha_{Knee,l} = -\sin^{-1} [\mathbf{l}_{Knee,l} \cdot \mathbf{i}_{Thigh,l}] \quad (C-55)$$

$$\beta_{Knee,l} = -\sin^{-1} [\mathbf{k}_{Thigh,l} \cdot \mathbf{i}_{Calf,l}] \quad (C-56)$$

$$\gamma_{Knee,l} = \sin^{-1} [\mathbf{l}_{Knee,l} \cdot \mathbf{k}_{Calf,l}] \quad (C-57)$$

Right Ankle

$$\alpha_{Ankle,r} = \sin^{-1} [\mathbf{l}_{Ankle,r} \times \mathbf{i}_{Calf,r}] \quad (C-58)$$

$$\beta_{Ankle,r} = \sin^{-1} [\mathbf{k}_{Calf,r} \times \mathbf{i}_{Foot,r}] \quad (C-59)$$

$$\gamma_{Ankle,r} = -\sin^{-1} [\mathbf{l}_{Ankle,r} \cdot \mathbf{k}_{Foot,r}] \quad (C-60)$$

Left Ankle

$$\alpha_{Ankle,l} = \sin^{-1} \left[\mathbf{l}_{Ankle,l} \cdot \mathbf{i}_{Calf,l} \right] \quad (\text{C-61})$$

$$\beta_{Ankle,l} = -\sin^{-1} \left[\mathbf{k}_{Calf,l} \cdot \mathbf{i}_{Foot,l} \right] \quad (\text{C-62})$$

$$\gamma_{Ankle,l} = \sin^{-1} \left[\mathbf{l}_{Ankle,l} \cdot \mathbf{k}_{Foot,l} \right] \quad (\text{C-63})$$

Appendix - D

Clinical Trials – Protocols and Procedures

Clinical Trials

The clinical trials in Tan Tock Seng Hospital mark a milestone and achievement of four years of intensive research. The preparations for clinical trials need to be done carefully, as it involved actual patient. The developed robotic gait rehabilitation system, *NaTure-gaits*, is tested with patient recruited from local hospital. The main objective of the clinical trial is to extend the trial of the robotic system from academic research to clinical use. During the clinical trial, the key functions of *NaTure-gaits* will be tested on patient. This chapter provides the information for clinical trial, from the ethnic approval application submission to local health authority, patient recruitment, to clinical trials procedures and results.

Objective of the Clinical Trials

The objectives of the clinical trials are as follows:

1. To verify the function and safety of the machine
2. To demonstrate the man power saving
3. To verify usability of the system on patient
4. To monitor the efficiency of biofeedback
5. Verification of data recording capability

Ethnic Approval Application

An ethnic's approval is required before any clinical trial can be carried out in local hospital. The ethnic's approval application is submitted to National Healthcare Group of Singapore [150] for approval. The submitted documents include the following:

Study Protocol Form:

This form includes all the necessary details for the research project and the clinical trials.

Patient Information Sheet:

This is the form provided to the patient recruited for the clinical trials. The procedures of the clinical trials are provided to the patient in this form. The patient needs to complete this form if they agreed to participate in the clinical trials.

Patient Inclusion and Exclusion Criteria

The inclusion and exclusion criteria for the patient recruited are as follows:

Inclusion Criteria

1. Patients with Incomplete spinal cord injury and American Spinal Injury Association Impairment Scale (ASIA) as C or D with levels from C5 to L3 and at least 6 months post injury
2. Patients with Stroke at least 6 months post injury
3. Age between 21 and 65
4. Written informed consent
5. Able to stand upright for at least one hour
6. No physical constraint on hip, knee, and ankle joints
7. Limited or no marked spasticity

Exclusion Criteria

1. Patients who are not able to follow instructions, cognitively impaired
2. Patients who have concomitant neglect, ataxia
3. Patients who have severe postural hypotension, ischemic heart disease, hypertension or other medical conditions that may be aggravated by increased exertion
4. Any other known pre-existing bone diseases that might increase the risk of bone fracture or other injury from intensive conventional therapy or robot-assisted therapy
5. Patients with chronic respiratory conditions such as chronic obstructive airways disease, or with asthma that is irritated by exercise
6. Patients who at higher risk of pressure sores on the legs or pelvis including those with leg ulcers, diabetics and those with severe varicose veins
7. Patients who suffering from dysphasia or who have difficulty in talking

8. Severe spasticity limiting over-ground ambulation
9. Patients with reduced joint range of motion
10. Patients who are unable to maintain upright posture
11. Patients who are unable to fit into device
12. Those who refuse to participate

Procedures of Clinical Trials

The clinical trials are designed with the following procedures. Five sessions will be arranged for two subjects. The arrangements of the sessions are as follows:

1. Harness fitting and parameters measurement; EMG and in sole force sensor data acquisition without the gait system (3 hours for two subjects)
Location: TTSH
2. First system trial for subject **one** (3 hours)
Location: NTU
3. Second system trial for subject **one** (3 hours)
Location: NTU
4. First system trial for subject **two** (3 hours)
Location: NTU
5. Second system trial **two** (3 hours)
Location: NTU

The common procedures for each trial are described as follows:

Before trial:

- Evaluation of state of health – If there is recent illness, flu, infections, and discomfort visit will be postponed
- A review and evaluation of your motor power, muscle tone will be done.
- Blood pressure taken while sitting, and standing. (If these are abnormal or if symptoms occur, testing will be postponed)

During trial:

- Heart rate, blood pressure, and perceived level of exertion are measured before and after application of device after 15 minutes
- Evaluation of level of discomfort of limbs, level of breathlessness

- Subjective symptoms during application of robotic device, namely, pain, discomfort, unsteadiness.
- Objective assessment of side effects after application of robotic device after each assessment- Skin redness or discomfort, increased spasms.
- Subjects with any of the above will be asked to rest until the discomfort has gone away.
- Subjects with a history of high or low blood pressure will be monitored for blood pressure before and after application of the device.

First Visit of Clinical Trial

First visit is divided into two sub-sessions: (1) Harness fitting and anthropometric parameters measurement (2) EMG and ground reaction force measurement during therapist assisted walking. This trial will be carried out in Rehabilitation Centre of Tan Tock Seng Hospital. Entire trial will last for 3 hours. Doctor and therapist will be accompanying the patient during entire visit. Both patients are attending the same session for first visit.

Testing procedures – (1) Harness fitting and anthropometric parameters measurement

1. The purpose and the procedures of the test will be explained to the patient.
2. The harness will be put on the patient when the patient is lying on bed.
3. The project team will do necessary adjustment for proper fitting of the harness.
4. Anthropometric parameters will be measured and recorded.
 - a. Age
 - b. Height
 - c. Weight
 - d. Thigh length
 - e. Shank length
 - f. Waist breadth
 - g. Thigh circumference
 - h. Shank circumference

Testing procedures – (2) EMG and ground reaction force measurement during therapist assisted walking

1. The purpose and the procedures of the test will be explained to the patient.
2. A run through of the equipment - electrodes, transmitter and receiver - used in the testing. No invasive items will be are used. Wireless EMG system will be used for EMG signal acquisition.
3. Skin preparation in the form of abrasive gel will be applied on the targeted surface so as to remove the dead cells and reduce skin impedance.
4. Surface electrodes will be pasted on the skin, where the major surface muscles are located.
5. A belt with the wireless TeleMyo 2400 G2 Telemetry System will be worn. Cables will be connected from this system to the electrodes.
6. Insole force sensor will be used to acquire ground reaction force.
7. The patient will be asked to walk for a few minutes to familiarize with their own selected walking speed (with or without the assistance of the therapists).
8. For the actual experiment, you will be required to walk a distance of 6 – 10 meters.
9. A total of 3 trials will be needed, with a 5 – 10 minutes interval to rest.

This study will take 3 hours.

Second Visit of Clinical Trial

This trial will be carried out with robotic gait rehabilitation with the patient. This trial will be carried out in Project Room of Nanyang Technological University. Entire trial will last for 3 hours. Doctor and therapist will be accompanying the patient during entire visit. Each patient will be attending this visit on different day.

Testing procedures

1. The purpose and the procedures of the test will be explained to the patient.
2. A run through of the equipment - electrodes, transmitter and receiver - used in the testing.
3. A harness will be worn on the patient when the patient is lying on bed.

4. The patient will be transferred to and will be suspended within the rehabilitation system for the project team to prepare necessary adjustment.
5. The rehabilitation system will be adjusted according to the patient's anatomic parameters.
6. Once the patient is properly strapped on the rehabilitation system, the gait motion will be replayed at a very slow phase for familiarize purpose. The patient will be still suspended in the air during the familiarize stage.
7. After familiarize stage, the patient will be lower down, so that his/her feet touches the ground.
8. The rehabilitation system will be fully automated to provide gait rehabilitation to the patient. Body weight support, pelvic control and functional over-ground walking will be tested together.
9. Walking speed will be the minimum comfortable speed of the patient or 0.2 m/s, whichever is lower.
10. Once the trial is over, the patient will be offloaded from the system.

This study will take 3 hours.

Third Visit of Clinical Trial

This trial will be carried out with robotic gait rehabilitation with the patient. This trial will be carried out in Project Room of Nanyang Technological University. Entire trial will last for 3 hours. Doctor and therapist will be accompanying the patient during entire visit. Each patient will be attending this visit on different day.

Testing procedures

1. The purpose and the procedures of the test will be explained to the patient.
2. A run through of the equipment - electrodes, transmitter and receiver - used in the testing. No invasive items will be are used. Wireless EMG system will be used for EMG signal acquisition.
3. Skin preparation in the form of abrasive gel will be applied on the targeted surface so as to remove the dead cells and reduce skin impedance. (You may feel slightly uncomfortable with the application of the abrasive gel)
4. A harness will be worn on the patient when the patient is lying on bed.

5. The patient will be transferred to and will be suspended within the rehabilitation system for the project team to prepare necessary adjustment.
6. The research team will prepare for the bio-feedback sensors placement. (Detail refers to testing procedures – placement of bio-feedback sensors)
7. The rehabilitation system will be adjusted according to the patient's anatomic parameters.
8. Once the patient is properly strapped on the rehabilitation system, the gait motion will be replayed at a very slow phase for familiarize purpose. The patient will be still suspended in the air during the familiarize stage.
9. After familiarize stage, the patient will be lower down, so that his/her feet touches the ground.
10. The rehabilitation system will be fully automated to provide gait rehabilitation to the patient. Body weight support, pelvic control and functional over-ground walking will be tested together.
11. Walking speed will be the minimum comfortable speed of the patient or 0.2 m/s, whichever is lower.
12. Once the trial is over, the patient will be offloaded from the system.

Testing procedures – placement of bio-feedback sensors

1. The purpose and the procedures of the test will be explained to the patient.
2. A run through of the equipment - electrodes, transmitter and receiver - used in the testing. No invasive items will be are used. Wireless EMG system will be used for EMG signal acquisition.
3. Skin preparation in the form of abrasive gel will be applied on the targeted surface so as to remove the dead cells and reduce skin impedance.
4. Surface electrodes will be pasted on the skin, where the major surface muscles are located.
5. A belt with the wireless TeleMyo 2400 G2 Telemetry System will be worn. Cables will be connected from this system to the electrodes.
6. Insole force sensor will be used to acquire ground reaction force.

This study will take 3 hours.

The Training Program Provided by *NaTure-gaits* during Clinical Trials

For clinical trial purpose, *NaTure-gaits* is programmed to provide over-ground gait rehabilitation in a complete passive mode. In other words, it is position controlled. The movement of the robotic orthosis, pelvic orthosis, and mobile platform are pre-programmed. The motion for *NaTure-gaits* is taken from the gait locomotion acquired from motion capture study.

Appendix - E

Clinical Trial Survey Form

NMRC Clinical Trial Data Record Sheet

Patients' Particulars

Basic information

Subject ID:		Date of Examination:	
Age:		Height/Weight:	_____(m)/_____(kg)
Date of Injury:		Type of Injury:	Spinal Cord Injury / Stroke *

Limb characteristic

Thigh Length: _____ Shank Length: _____

Waist Breadth: _____

For Spinal Cord Injury, please indicate

I. Level of injury: _____ Complete/Incomplete *

II. ASIA impairment scale: A / B / C / D / E *

III. Cause of injury: trauma / medical *

For Stroke, please indicate

I. Type of stroke: Ischemic Stroke / Hemorrhagic Stroke *

II. Side of stroke: Left / Right *

Other remarks

* Please circle the appropriate option

Physical conditions

1. Joint range of motion
 - Hip _____ Knee _____ Ankle _____
2. Level of spasticity of limbs (Modified Ashworth scale) 0 / 1 / 1+ / 2 / 3 / 4 / *
 - Hip _____ Knee _____ Ankle _____
3. Ambulating: yes/ no
 - a. ambulating with aid: walking frame / quad stick / elbow crutches *
 - b. unilateral / bilateral aid *
 - c. level of assistance: max assistance / moderate / minimal assistance / supervision / independent *

* Please circle the appropriate option

Evaluation record for before/after trial

Blood pressure:

15 minutes before trial (Standing) _____ (Sitting)

15 minutes after trial (Standing) _____ (Sitting)

Heart rate:

15 minutes before trial (Standing) _____ (Sitting)

15 minutes after trial (Standing) _____ (Sitting)

Perceived level of exertion:

15 minutes before trial (Standing) _____ (Sitting)

15 minutes after trial (Standing) _____ (Sitting)

Objective assessment of side effects after application of robotic device after each assessment - Skin redness or discomfort, increased spasms:

Remarks:

--

Outcome measure during/after trial

1. Harness setup time taken: _____ mins

Remarks:

2. *NaTure-gaits* strapping time taken: _____ mins

Remarks:

3. 10 meters walk can be completed with:

No disruption ()

Minor disruption ()

Major disruption ()

Remarks:

Major disruption indicates the whole process is disrupted and needed to be restarted entirely, minor disruption is the disruption that do not required restart of the whole process and the process can be continued from the point it stop at.

4. Total time subject spent in *NaTure-gaits*: _____ mins

Remarks:

Questionnaires for the subject

5. Comfort level of the harness:

Do not felt discomfort ()

Felt minor discomfort ()

Discomfort but bearable ()

Felt very discomfort ()

Remarks:

6. Comfort level of strapping on *NaTure-gaits*:

Do not felt discomfort ()

Felt minor discomfort ()

Discomfort but bearable ()

Felt very discomfort ()

Remarks:

7. Comment on walk assisted by *NaTure-gaits*:

Felt natural ()

Can't tell the feeling ()

Felt not natural ()

Remarks:

8. Evaluation of level of discomfort of limbs, level of breathlessness:

Remarks:

9. Subjective symptoms during application of robotic device, namely, pain, discomfort, unsteadiness:

Remarks:

10. Any other comment:

Remarks:

--

Appendix - F

Experimental Data of Gait Parameters for the Fifty Subjects

Subject	Walk	Age	Gender	Speed State	Height	Weight	Walking Speed	Stride	Cadence
1	1	48	Female	Slow	158	50	83.60	98.94	101.40
1	2	48	Female	Slow	158	50	101.74	108.14	112.90
1	3	48	Female	Slow	158	50	93.21	101.32	110.40
1	4	48	Female	Normal	158	50	114.53	114.06	120.50
1	5	48	Female	Normal	158	50	119.56	116.17	123.50
1	6	48	Female	Normal	158	50	121.92	118.47	123.50
2	1	22	Female	Normal	155	59	108.40	122.90	105.60
2	2	22	Female	Normal	155	59	103.00	118.39	104.20
2	3	22	Female	Normal	155	59	106.60	119.34	106.80
2	4	22	Female	Slow	155	59	74.30	102.55	87.40
2	5	22	Female	Slow	155	59	83.20	108.91	92.30
2	6	22	Female	Slow	155	59	78.30	105.10	89.60
3	1	22	Female	Normal	163	54	126.30	134.27	113.70
3	2	22	Female	Normal	163	54	136.70	136.62	121.20
3	3	22	Female	Normal	163	54	141.00	142.04	119.40
3	4	22	Female	Slow	163	54	94.80	118.58	95.80
3	5	22	Female	Slow	163	54	103.80	127.52	97.60
3	6	22	Female	Slow	163	54	94.50	123.07	92.90
4	1	27	Female	Slow	150	37	93.27	117.95	94.90
4	2	27	Female	Slow	150	37	97.74	112.46	104.30
4	3	27	Female	Slow	150	37	100.93	113.41	106.80
4	4	27	Female	Normal	150	37	132.55	127.25	125.00
4	5	27	Female	Normal	150	37	128.42	124.78	123.50
4	6	27	Female	Normal	150	37	140.86	131.24	128.80
5	1	26	Female	Slow	168	53	87.93	122.26	86.30
5	2	26	Female	Slow	168	53	85.48	121.39	84.50
5	3	26	Female	Slow	168	53	90.85	123.18	88.50
5	4	26	Female	Normal	168	53	114.28	133.15	103.00
5	5	26	Female	Normal	168	53	111.69	135.65	98.80
5	6	26	Female	Normal	168	53	112.53	135.03	100.00
6	1	25	Female	Slow	157	43	114.83	132.11	104.30
6	2	25	Female	Slow	157	43	116.96	134.56	104.30
6	3	25	Female	Slow	157	43	115.73	133.16	104.30

6	4	25	Female	Normal	157	43	141.71	151.16	112.50
6	5	25	Female	Normal	157	43	142.91	148.61	115.40
6	6	25	Female	Normal	157	43	143.13	146.06	117.60
7	1	51	Female	Slow	152	42	92.42	114.57	96.80
7	2	51	Female	Slow	152	42	87.90	112.22	94.00
7	3	51	Female	Slow	152	42	91.42	113.33	96.80
7	4	51	Female	Normal	152	42	126.44	131.48	115.40
7	5	51	Female	Normal	152	42	126.29	131.32	115.40
7	6	51	Female	Normal	152	42	135.09	135.77	119.40
8	1	31	Female	Slow	165	65	52.08	108.13	57.80
8	2	31	Female	Slow	165	65	47.26	109.69	51.70
8	3	31	Female	Slow	165	65	37.71	98.80	45.80
8	4	31	Female	Normal	165	65	85.09	115.12	88.70
8	5	31	Female	Normal	165	65	97.68	123.52	94.90
8	6	31	Female	Normal	165	65	103.16	130.04	95.20
9	1	48	Female	Slow	158	43	72.64	106.05	82.20
9	2	48	Female	Slow	158	43	61.43	98.68	74.70
9	3	48	Female	Slow	158	43	65.54	101.75	77.30
9	4	48	Female	Normal	158	43	82.67	107.48	92.30
9	5	48	Female	Normal	158	43	85.89	109.88	93.80
9	6	48	Female	Normal	158	43	86.95	110.19	94.70
10	1	22	Female	Slow	153	45	93.04	111.98	99.70
10	2	22	Female	Slow	153	45	97.37	110.65	105.60
10	3	22	Female	Slow	153	45	95.49	108.83	105.30
10	4	22	Female	Normal	153	45	127.85	125.75	122.00
10	5	22	Female	Normal	153	45	122.66	123.69	119.00
10	6	22	Female	Normal	153	45	125.06	126.11	119.00
11	1	26	Female	Slow	156	48	83.02	107.94	92.30
11	2	26	Female	Slow	156	48	96.29	115.55	100.00
11	3	26	Female	Slow	156	48	88.07	108.95	97.00
11	4	26	Female	Normal	156	48	108.58	114.39	113.90
11	5	26	Female	Normal	156	48	116.53	119.32	117.20
11	6	26	Female	Normal	156	48	123.79	121.76	122.00
12	1	25	Female	Slow	165	52	108.19	131.54	98.70
12	2	25	Female	Slow	165	52	111.63	132.64	101.00
12	3	25	Female	Slow	165	52	101.53	125.86	96.80
12	4	25	Female	Normal	165	52	141.32	140.04	121.10
12	5	25	Female	Normal	165	52	140.48	143.96	117.10
12	6	25	Female	Normal	165	52	137.68	143.17	115.40
13	1	25	Male	Slow	175	70	86.43	103.10	100.60
13	2	25	Male	Slow	175	70	87.86	101.08	104.30

13	3	25	Male	Slow	175	70	87.62	100.81	104.30
13	4	25	Male	Normal	175	70	121.69	127.54	114.50
13	5	25	Male	Normal	175	70	112.69	122.16	110.70
13	6	25	Male	Normal	175	70	112.13	121.99	110.30
14	1	27	Female	Slow	162	50	116.35	132.10	105.70
14	2	27	Female	Slow	162	50	117.59	132.13	106.80
14	3	27	Female	Slow	162	50	110.42	129.90	102.00
14	4	27	Female	Normal	162	50	133.91	141.33	113.70
14	5	27	Female	Normal	162	50	130.18	139.35	112.10
14	6	27	Female	Normal	162	50	130.56	137.79	113.70
15	1	26	Female	Slow	174	59	88.58	105.67	100.60
15	2	26	Female	Slow	174	59	98.46	112.21	105.30
15	3	26	Female	Slow	174	59	93.09	105.19	106.20
15	4	26	Female	Normal	174	59	114.55	120.05	114.50
15	5	26	Female	Normal	174	59	136.58	132.71	123.50
15	6	26	Female	Normal	174	59	137.99	134.08	123.50
16	1	27	Male	Slow	189	77	83.86	120.09	83.80
16	2	27	Male	Slow	189	77	85.16	115.22	88.70
16	3	27	Male	Slow	189	77	91.29	120.12	91.20
16	4	27	Male	Normal	189	77	133.80	153.95	104.30
16	5	27	Male	Normal	189	77	129.67	149.19	104.30
16	6	27	Male	Normal	189	77	137.72	158.46	104.30
17	1	26	Male	Slow	170	65	99.54	121.03	98.70
17	2	26	Male	Slow	170	65	109.89	124.88	105.60
17	3	26	Male	Slow	170	65	108.70	120.90	107.90
17	4	26	Male	Normal	170	65	127.12	136.08	112.10
17	5	26	Male	Normal	170	65	124.51	131.99	113.20
17	6	26	Male	Normal	170	65	115.62	129.55	107.10
18	1	24	Male	Slow	185	75	95.82	139.38	82.50
18	2	24	Male	Slow	185	75	101.54	138.15	88.20
18	3	24	Male	Slow	185	75	104.53	138.46	90.60
18	4	24	Male	Normal	185	75	129.85	151.28	103.00
18	5	24	Male	Normal	185	75	130.81	149.92	104.70
18	6	24	Male	Normal	185	75	131.44	153.13	103.00
19	1	46	Male	Slow	178	50	104.62	121.77	103.10
19	2	46	Male	Slow	178	50	109.48	121.76	107.90
19	3	46	Male	Slow	178	50	105.94	115.26	110.30
19	4	46	Male	Normal	178	50	126.51	133.52	113.70
19	5	46	Male	Normal	178	50	127.94	133.04	115.40
19	6	46	Male	Normal	178	50	123.87	129.83	114.50
20	1	28	Male	Slow	180	82	88.84	129.23	82.50

20	2	28	Male	Slow	180	82	101.84	133.41	91.60
20	3	28	Male	Slow	180	82	90.23	126.64	85.50
20	4	28	Male	Normal	180	82	136.44	160.05	102.30
20	5	28	Male	Normal	180	82	138.52	161.39	103.00
20	6	28	Male	Normal	180	82	141.10	161.72	104.70
21	1	27	Male	Slow	174	76	79.16	112.42	84.50
21	2	27	Male	Slow	174	76	86.94	112.30	92.90
21	3	27	Male	Slow	174	76	99.26	119.12	100.00
21	4	27	Male	Normal	174	76	106.44	122.59	104.20
21	5	27	Male	Normal	174	76	107.46	122.11	105.60
21	6	27	Male	Normal	174	76	112.99	125.67	107.90
22	1	24	Male	Slow	170	60	83.14	106.14	94.00
22	2	24	Male	Slow	170	60	84.96	108.47	94.00
22	3	24	Male	Slow	170	60	83.85	108.20	93.00
22	4	24	Male	Normal	170	60	126.43	137.18	110.60
22	5	24	Male	Normal	170	60	133.03	137.86	115.80
22	6	24	Male	Normal	170	60	120.58	132.63	109.10
23	1	25	Male	Slow	172	63	79.75	117.14	81.70
23	2	25	Male	Slow	172	63	74.91	112.09	80.20
23	3	25	Male	Slow	172	63	80.44	115.19	83.80
23	4	25	Male	Normal	172	63	116.66	128.32	109.10
23	5	25	Male	Normal	172	63	115.69	134.79	103.00
23	6	25	Male	Normal	172	63	116.76	134.34	104.30
24	1	44	Male	Slow	168	67	65.79	88.91	88.80
24	2	44	Male	Slow	168	67	70.86	91.93	92.50
24	3	44	Male	Slow	168	67	70.86	91.93	92.50
24	4	44	Male	Normal	168	67	113.63	121.86	111.90
24	5	44	Male	Normal	168	67	111.18	120.96	110.30
24	6	44	Male	Normal	168	67	108.46	120.63	107.90
25	1	29	Male	Slow	165	62	79.65	116.99	81.70
25	2	29	Male	Slow	165	62	94.08	127.57	88.50
25	3	29	Male	Slow	165	62	80.80	114.48	84.70
25	4	29	Male	Normal	165	62	128.49	141.33	109.10
25	5	29	Male	Normal	165	62	122.72	134.98	109.10
25	6	29	Male	Normal	165	62	124.56	137.63	108.60
26	1	30	Male	Slow	170	60	77.08	116.20	79.60
26	2	30	Male	Slow	170	60	89.92	124.03	87.00
26	3	30	Male	Slow	170	60	87.99	119.31	88.50
26	4	30	Male	Normal	170	60	112.54	126.45	106.80
26	5	30	Male	Normal	170	60	116.46	128.10	109.10
26	6	30	Male	Normal	170	60	116.62	134.17	104.30

27	1	41	Male	Slow	174	74	69.46	103.55	80.50
27	2	41	Male	Slow	174	74	81.12	108.65	89.60
27	3	41	Male	Slow	174	74	78.35	109.46	85.90
27	4	41	Male	Normal	174	74	132.72	136.01	117.10
27	5	41	Male	Normal	174	74	136.01	137.16	119.00
27	6	41	Male	Normal	174	74	131.66	136.91	115.40
28	1	27	Female	Slow	165	52	95.60	120.88	94.90
28	2	27	Female	Slow	165	52	103.73	129.13	96.40
28	3	27	Female	Slow	165	52	99.37	123.18	96.80
28	4	27	Female	Normal	165	52	136.19	136.87	119.40
28	5	27	Female	Normal	165	52	133.30	133.97	119.40
28	6	27	Female	Normal	165	52	132.97	136.26	117.10
29	1	20	Male	Normal	160	55	90.70	119.95	91.20
29	2	20	Male	Normal	160	55	97.10	124.27	93.80
29	3	20	Male	Normal	160	55	78.60	113.42	83.30
29	4	20	Male	Slow	160	55	71.00	110.34	78.30
29	5	20	Male	Slow	160	55	66.90	113.79	72.30
29	6	20	Male	Slow	160	55	60.60	100.84	71.30
30	1	20	Male	Normal	180	85	92.30	141.19	80.00
30	2	20	Male	Normal	180	85	97.20	150.28	78.20
30	3	20	Male	Normal	180	85	99.30	148.22	80.80
30	4	20	Male	Slow	180	85	64.00	110.12	70.00
30	5	20	Male	Slow	180	85	63.00	112.74	67.60
30	6	20	Male	Slow	180	85	59.40	111.88	64.10
31	1	19	Male	Normal	170	72	109.90	135.58	97.60
31	2	19	Male	Normal	170	72	113.30	136.29	100.00
31	3	19	Male	Normal	170	72	114.70	136.31	101.70
31	4	19	Male	Slow	170	72	84.50	114.18	89.60
31	5	19	Male	Slow	170	72	93.20	121.92	92.00
31	6	19	Male	Slow	170	72	95.60	119.79	96.80
32	1	20	Male	Normal	170	64	105.80	126.32	100.00
32	2	20	Male	Normal	170	64	98.10	119.72	98.70
32	3	20	Male	Normal	170	64	102.70	126.77	97.70
32	4	20	Male	Slow	170	64	75.20	113.40	79.60
32	5	20	Male	Slow	170	64	78.20	115.02	81.70
32	6	20	Male	Slow	170	64	84.60	118.53	86.20
33	1	47	Female	Slow	166	51	81.96	116.39	84.50
33	2	47	Female	Slow	166	51	103.33	127.05	97.60
33	3	47	Female	Slow	166	51	98.41	119.65	98.70
33	4	47	Female	Normal	166	51	125.60	136.28	110.60
33	5	47	Female	Normal	166	51	122.55	139.14	105.70

33	6	47	Female	Normal	166	51	126.62	139.27	109.10
34	1	45	Male	Slow	175	78	71.30	112.44	76.10
34	2	45	Male	Slow	175	78	68.22	110.78	73.90
34	3	45	Male	Slow	175	78	70.63	110.64	76.60
34	4	45	Male	Normal	175	78	117.81	135.67	104.20
34	5	45	Male	Normal	175	78	124.95	141.86	105.70
34	6	45	Male	Normal	175	78	126.46	142.09	106.80
35	1	19	Male	Normal	169	63	89.90	116.75	92.00
35	2	19	Male	Normal	169	63	101.20	127.23	95.80
35	3	19	Male	Normal	169	63	99.30	127.10	93.80
35	4	19	Male	Slow	169	63	71.40	111.76	76.60
35	5	19	Male	Slow	169	63	74.40	112.79	79.60
35	6	19	Male	Slow	169	63	67.20	106.29	75.60
36	1	19	Male	Normal	170	53	105.30	130.58	96.80
36	2	19	Male	Normal	170	53	102.20	129.40	94.90
36	3	19	Male	Normal	170	53	96.50	126.24	91.20
36	4	19	Male	Slow	170	53	76.30	117.49	78.30
36	5	19	Male	Slow	170	53	85.80	126.38	81.70
36	6	19	Male	Slow	170	53	76.80	125.83	73.30
37	1	23	Male	Normal	168	60	90.60	119.48	91.20
37	2	23	Male	Normal	168	60	116.80	127.13	119.90
37	3	23	Male	Normal	168	60	102.80	120.98	102.00
37	4	23	Male	Slow	168	60	55.90	94.14	72.30
37	5	23	Male	Slow	168	60	56.90	93.62	73.40
37	6	23	Male	Slow	168	60	62.40	94.12	80.00
38	1	22	Male	Normal	162	60	103.80	126.63	98.70
38	2	22	Male	Normal	162	60	107.40	127.79	100.00
38	3	22	Male	Normal	162	60	104.40	135.29	93.80
38	4	22	Male	Slow	162	60	49.20	91.89	64.40
38	5	22	Male	Slow	162	60	48.00	91.25	63.50
38	6	22	Male	Slow	162	60	55.50	95.12	70.70
39	1	20	Male	Normal	169	69	144.20	143.63	121.20
39	2	20	Male	Normal	169	69	133.80	137.27	117.60
39	3	20	Male	Normal	169	69	127.90	128.42	119.40
39	4	20	Male	Slow	169	69	105.70	116.87	109.10
39	5	20	Male	Slow	169	69	104.70	119.95	104.50
39	6	20	Male	Slow	169	69	97.00	113.24	103.10
40	1	20	Female	Normal	172	58	95.80	117.71	96.80
40	2	20	Female	Normal	172	58	98.60	122.87	95.80
40	3	20	Female	Normal	172	58	98.00	124.80	94.90
40	4	20	Female	Slow	172	58	67.60	103.66	78.30

40	5	20	Female	Slow	172	58	60.30	98.48	73.60
40	6	20	Female	Slow	172	58	61.00	93.32	77.80
41	1	20	Female	Normal	153	53	95.90	118.63	97.70
41	2	20	Female	Normal	153	53	93.50	115.25	97.70
41	3	20	Female	Normal	153	53	90.00	114.81	94.90
41	4	20	Female	Slow	153	53	52.60	93.62	68.10
41	5	20	Female	Slow	153	53	50.70	92.89	64.70
41	6	20	Female	Slow	153	53	57.60	95.74	71.80
42	1	20	Female	Normal	155	48	113.80	127.28	107.10
42	2	20	Female	Normal	155	48	110.30	124.15	106.80
42	3	20	Female	Normal	155	48	116.70	128.40	109.10
42	4	20	Female	Slow	155	48	80.80	109.54	89.30
42	5	20	Female	Slow	155	48	83.60	109.83	92.30
42	6	20	Female	Slow	155	48	72.80	104.40	84.10
43	1	20	Female	Normal	156	51	111.70	118.88	113.20
43	2	20	Female	Normal	156	51	101.50	112.46	109.40
43	3	20	Female	Normal	156	51	99.50	112.66	106.20
43	4	20	Female	Slow	156	51	89.30	108.11	98.90
43	5	20	Female	Slow	156	51	78.90	103.29	91.60
43	6	20	Female	Slow	156	51	77.20	101.72	90.90
44	1	19	Female	Normal	163	64	117.20	128.86	110.30
44	2	19	Female	Normal	163	64	127.60	135.13	113.70
44	3	19	Female	Normal	163	64	120.40	132.78	109.10
44	4	19	Female	Slow	163	64	99.30	110.14	108.40
44	5	19	Female	Slow	163	64	97.20	113.24	103.10
44	6	19	Female	Slow	163	64	92.80	111.38	99.70
45	1	20	Female	Normal	152	56	101.20	118.27	103.10
45	2	20	Female	Normal	152	56	98.50	116.03	102.00
45	3	20	Female	Normal	152	56	97.60	114.14	102.00
45	4	20	Female	Slow	152	56	82.60	110.00	90.20
45	5	20	Female	Slow	152	56	92.10	112.55	98.10
45	6	20	Female	Slow	152	56	84.00	108.22	94.00
46	1	22	Female	Normal	164	49	115.30	131.47	105.70
46	2	22	Female	Normal	164	49	112.40	131.51	103.00
46	3	22	Female	Normal	164	49	114.00	129.55	105.70
46	4	22	Female	Slow	164	49	76.60	118.11	78.30
46	5	22	Female	Slow	164	49	79.80	118.46	80.90
46	6	22	Female	Slow	164	49	78.90	116.61	78.90
47	1	48	Male	Slow	168	75	80.94	98.21	98.90
47	2	48	Male	Slow	168	75	84.15	99.59	101.40
47	3	48	Male	Slow	168	75	91.42	101.11	108.50

47	4	48	Male	Normal	168	75	122.21	123.24	119.00
47	5	48	Male	Normal	168	75	121.56	121.06	120.50
47	6	48	Male	Normal	168	75	121.83	122.86	119.00
48	1	50	Female	Slow	150	44	110.71	113.36	117.20
48	2	50	Female	Slow	150	44	116.62	117.60	119.00
48	3	50	Female	Slow	150	44	107.40	108.85	118.40
48	4	50	Female	Normal	150	44	143.16	130.24	131.90
48	5	50	Female	Normal	150	44	149.28	133.58	134.10
48	6	50	Female	Normal	150	44	147.34	134.05	131.90
49	1	21	Male	Normal	180	75	116.80	148.51	95.20
49	2	21	Male	Normal	180	75	125.90	156.30	96.40
49	3	21	Male	Normal	180	75	124.50	156.41	96.40
49	4	21	Male	Slow	180	75	86.20	128.09	80.90
49	5	21	Male	Slow	180	75	86.90	128.21	80.90
49	6	21	Male	Slow	180	75	88.90	129.42	82.40
50	1	17	Male	Normal	167	65	102.60	133.31	92.70
50	2	17	Male	Normal	167	65	103.80	135.93	91.60
50	3	17	Male	Normal	167	65	103.60	135.01	92.70
50	4	17	Male	Slow	167	65	71.40	117.86	72.80
50	5	17	Male	Slow	167	65	75.50	121.15	74.40
50	6	17	Male	Slow	167	65	76.80	122.47	75.80

Appendix - G

Predicted Basic Gait Parameters using Trained GRNNs

Predicted Outcome for Normal Walking Speed

Subject	Age	Gender	Height (cm)	Weight (kg)	Speed (cm/s)	Stride length (cm)	Cadence (step/min)	Predicted Speed (cm/s)	MSE
1	48	Female	158	50	118.67	116.23	122.50	104.56	199.25
2	22	Female	155	59	105.72	120.21	105.53	104.84	0.79
3	22	Female	163	54	135.51	137.64	118.10	122.61	166.43
4	27	Female	150	37	133.95	127.76	125.77	131.07	8.28
5	26	Female	168	53	112.83	134.61	100.60	131.52	349.31
6	25	Female	157	43	142.59	148.61	115.17	120.03	508.85
7	51	Female	152	42	129.27	132.86	116.73	131.30	4.10
8	31	Female	165	65	95.31	122.89	92.93	127.54	1038.58
9	48	Female	158	43	85.17	109.18	93.60	126.71	1725.52
10	22	Female	153	45	125.19	125.18	120.00	116.39	77.44
11	26	Female	156	48	116.30	118.49	117.70	125.40	82.73
12	25	Female	165	52	139.83	142.39	117.87	125.02	219.17
13	25	Male	175	70	115.51	123.90	111.83	117.06	2.42
14	27	Female	162	50	131.55	139.49	113.17	128.17	11.40
15	26	Female	174	59	129.71	128.95	120.50	109.42	411.64
16	27	Male	189	77	133.74	153.87	104.30	130.89	8.12
17	26	Male	170	65	122.42	132.54	110.80	116.14	39.37
18	24	Male	185	75	130.70	151.44	103.57	127.76	8.65
19	46	Male	178	50	126.11	132.13	114.53	111.60	210.67
20	28	Male	180	82	138.69	161.05	103.33	116.76	480.93
21	27	Male	174	76	108.97	123.46	105.90	121.81	164.81
22	24	Male	170	60	126.68	135.89	111.83	111.52	230.01
23	25	Male	172	63	116.37	132.48	105.47	116.70	0.10
24	44	Male	168	67	111.09	121.15	110.03	126.00	222.26
25	29	Male	165	62	125.26	137.98	108.93	115.98	85.99
26	30	Male	170	60	115.21	129.57	106.73	120.71	30.29
27	41	Male	174	74	133.47	136.69	117.17	120.79	160.82
28	27	Female	165	52	134.15	135.70	118.63	127.18	48.53
29	20	Male	160	55	89.01	119.21	89.43	105.85	283.43
30	20	Male	180	85	97.29	146.56	79.67	127.72	926.35
31	19	Male	170	72	113.12	136.06	99.77	118.57	29.62
32	20	Male	170	64	102.32	124.27	98.80	112.61	105.88

33	47	Female	166	51	124.93	138.23	108.47	111.06	192.36
34	45	Male	175	78	123.07	139.87	105.57	129.56	42.06
35	19	Male	169	63	96.81	123.69	93.87	111.68	221.11
36	19	Male	170	53	101.20	128.74	94.30	107.91	44.96
37	23	Male	168	60	106.89	122.53	104.37	113.75	47.07
38	22	Male	162	60	105.47	129.90	97.50	104.51	0.90
39	20	Male	169	69	135.79	136.44	119.40	108.07	768.60
40	20	Female	172	58	97.25	121.79	95.83	126.71	867.86
41	20	Female	153	53	93.74	116.23	96.77	108.39	214.59
42	20	Female	155	48	113.61	126.61	107.67	111.91	2.88
43	20	Female	156	51	104.79	114.67	109.60	111.11	39.95
44	19	Female	163	64	122.40	132.26	111.03	117.74	21.72
45	20	Female	152	56	99.09	116.15	102.37	103.42	18.79
46	22	Female	164	49	114.26	130.84	104.80	129.93	245.34
47	48	Male	168	75	121.87	122.39	119.50	119.25	6.85
48	50	Female	150	44	146.59	132.62	132.63	122.76	567.94
49	21	Male	180	75	123.01	153.74	96.00	120.01	8.99
50	17	Male	167	65	103.68	134.75	92.33	109.77	37.07

Predicted Outcome for Slow Walking Speed

Subject	Age	Gender	Height (cm)	Weight (kg)	Speed (cm/s)	Stride length (cm)	Cadence (step/min)	Predicted Speed (cm/s)	MSE
1	48	Female	158	50	118.67	116.23	122.50	104.56	199.25
2	22	Female	155	59	105.72	120.21	105.53	104.84	0.79
3	22	Female	163	54	135.51	137.64	118.10	122.61	166.43
4	27	Female	150	37	133.95	127.76	125.77	131.07	8.28
5	26	Female	168	53	112.83	134.61	100.60	131.52	349.31
6	25	Female	157	43	142.59	148.61	115.17	120.03	508.85
7	51	Female	152	42	129.27	132.86	116.73	131.30	4.10
8	31	Female	165	65	95.31	122.89	92.93	127.54	1038.58
9	48	Female	158	43	85.17	109.18	93.60	126.71	1725.52
10	22	Female	153	45	125.19	125.18	120.00	116.39	77.44
11	26	Female	156	48	116.30	118.49	117.70	125.40	82.73
12	25	Female	165	52	139.83	142.39	117.87	125.02	219.17
13	25	Male	175	70	115.51	123.90	111.83	117.06	2.42
14	27	Female	162	50	131.55	139.49	113.17	128.17	11.40
15	26	Female	174	59	129.71	128.95	120.50	109.42	411.64
16	27	Male	189	77	133.74	153.87	104.30	130.89	8.12
17	26	Male	170	65	122.42	132.54	110.80	116.14	39.37
18	24	Male	185	75	130.70	151.44	103.57	127.76	8.65
19	46	Male	178	50	126.11	132.13	114.53	111.60	210.67

20	28	Male	180	82	138.69	161.05	103.33	116.76	480.93
21	27	Male	174	76	108.97	123.46	105.90	121.81	164.81
22	24	Male	170	60	126.68	135.89	111.83	111.52	230.01
23	25	Male	172	63	116.37	132.48	105.47	116.70	0.10
24	44	Male	168	67	111.09	121.15	110.03	126.00	222.26
25	29	Male	165	62	125.26	137.98	108.93	115.98	85.99
26	30	Male	170	60	115.21	129.57	106.73	120.71	30.29
27	41	Male	174	74	133.47	136.69	117.17	120.79	160.82
28	27	Female	165	52	134.15	135.70	118.63	127.18	48.53
29	20	Male	160	55	89.01	119.21	89.43	105.85	283.43
30	20	Male	180	85	97.29	146.56	79.67	127.72	926.35
31	19	Male	170	72	113.12	136.06	99.77	118.57	29.62
32	20	Male	170	64	102.32	124.27	98.80	112.61	105.88
33	47	Female	166	51	124.93	138.23	108.47	111.06	192.36
34	45	Male	175	78	123.07	139.87	105.57	129.56	42.06
35	19	Male	169	63	96.81	123.69	93.87	111.68	221.11
36	19	Male	170	53	101.20	128.74	94.30	107.91	44.96
37	23	Male	168	60	106.89	122.53	104.37	113.75	47.07
38	22	Male	162	60	105.47	129.90	97.50	104.51	0.90
39	20	Male	169	69	135.79	136.44	119.40	108.07	768.60
40	20	Female	172	58	97.25	121.79	95.83	126.71	867.86
41	20	Female	153	53	93.74	116.23	96.77	108.39	214.59
42	20	Female	155	48	113.61	126.61	107.67	111.91	2.88
43	20	Female	156	51	104.79	114.67	109.60	111.11	39.95
44	19	Female	163	64	122.40	132.26	111.03	117.74	21.72
45	20	Female	152	56	99.09	116.15	102.37	103.42	18.79
46	22	Female	164	49	114.26	130.84	104.80	129.93	245.34
47	48	Male	168	75	121.87	122.39	119.50	119.25	6.85
48	50	Female	150	44	146.59	132.62	132.63	122.76	567.94
49	21	Male	180	75	123.01	153.74	96.00	120.01	8.99
50	17	Male	167	65	103.68	134.75	92.33	109.77	37.07

Predicted Outcomes of Stride Length for Normal Walking Speed

Subject	Age	Gender	Height (cm)	Weight (kg)	Speed (cm/s)	Stride length (cm)	Cadence (step/min)	Predicted Stride Length (cm/s)	MSE
1	48	Female	158	50	114.54	114.06	120.50	128.67	213.36
1	48	Female	158	50	119.56	116.17	123.50	130.76	212.73
1	48	Female	158	50	121.93	118.47	123.50	132.22	189.18
2	22	Female	155	59	102.80	118.39	104.20	126.13	59.93
2	22	Female	155	59	106.21	119.34	106.80	126.36	49.28
2	22	Female	155	59	108.15	122.90	105.60	126.48	12.80
3	22	Female	163	54	127.22	134.27	113.70	135.75	2.19
3	22	Female	163	54	137.99	136.62	121.20	142.69	36.85
3	22	Female	163	54	141.33	142.04	119.40	143.36	1.75
4	27	Female	150	37	128.42	124.78	123.50	136.86	146.04
4	27	Female	150	37	132.55	127.25	125.00	139.70	154.89
4	27	Female	150	37	140.86	131.24	128.80	143.90	160.31
5	26	Female	168	53	111.69	135.65	98.80	127.01	74.58
5	26	Female	168	53	112.53	135.03	100.00	127.33	59.33
5	26	Female	168	53	114.29	133.15	103.00	128.01	26.39
6	25	Female	157	43	141.71	151.16	112.50	142.84	69.17
6	25	Female	157	43	142.91	148.61	115.40	142.86	33.02
6	25	Female	157	43	143.14	146.06	117.60	142.99	9.41
7	51	Female	152	42	126.29	131.32	115.40	135.15	14.64
7	51	Female	152	42	126.44	131.48	115.40	135.26	14.25
7	51	Female	152	42	135.09	135.77	119.40	141.10	28.40
8	31	Female	165	65	85.09	115.12	88.70	114.03	1.19
8	31	Female	165	65	97.68	123.52	94.90	124.90	1.91
8	31	Female	165	65	103.17	130.04	95.20	125.74	18.46
9	48	Female	158	43	82.67	107.48	92.30	113.94	41.72
9	48	Female	158	43	85.89	109.88	93.80	115.68	33.69
9	48	Female	158	43	86.96	110.19	94.70	116.75	42.99
10	22	Female	153	45	122.66	123.69	119.00	132.61	79.65
10	22	Female	153	45	125.06	126.11	119.00	134.37	68.20
10	22	Female	153	45	127.85	125.75	122.00	136.43	114.03
11	26	Female	156	48	108.58	114.39	113.90	126.85	155.29
11	26	Female	156	48	116.54	119.32	117.20	129.25	98.58
11	26	Female	156	48	123.79	121.76	122.00	133.52	138.33
12	25	Female	165	52	137.68	143.17	115.40	142.26	0.82
12	25	Female	165	52	140.48	143.96	117.10	143.16	0.65
12	25	Female	165	52	141.32	140.04	121.10	143.48	11.80
13	25	Male	175	70	112.13	121.99	110.30	127.59	31.39
13	25	Male	175	70	112.69	122.16	110.70	127.78	31.58

13	25	Male	175	70	121.69	127.54	114.50	131.80	18.15
14	27	Female	162	50	130.18	139.35	112.10	137.68	2.78
14	27	Female	162	50	130.56	137.79	113.70	137.99	0.04
14	27	Female	162	50	133.91	141.33	113.70	140.14	1.42
15	26	Female	174	59	114.55	120.05	114.50	128.49	71.30
15	26	Female	174	59	136.58	132.71	123.50	142.12	88.57
15	26	Female	174	59	137.99	134.08	123.50	142.80	76.10
16	27	Male	189	77	129.67	149.19	104.30	137.05	147.30
16	27	Male	189	77	133.81	153.95	104.30	139.63	205.02
16	27	Male	189	77	137.73	158.46	104.30	141.63	283.33
17	26	Male	170	65	115.62	129.55	107.10	128.60	0.90
17	26	Male	170	65	124.51	131.99	113.20	133.79	3.25
17	26	Male	170	65	127.12	136.08	112.10	135.63	0.20
18	24	Male	185	75	129.85	151.28	103.00	137.11	200.69
18	24	Male	185	75	130.81	149.92	104.70	137.80	146.97
18	24	Male	185	75	131.44	153.13	103.00	138.12	225.33
19	46	Male	178	50	123.88	129.83	114.50	133.37	12.53
19	46	Male	178	50	126.51	133.52	113.70	135.25	3.00
19	46	Male	178	50	127.94	133.04	115.40	136.30	10.61
20	28	Male	180	82	136.44	160.05	102.30	140.95	364.88
20	28	Male	180	82	138.53	161.39	103.00	141.81	383.54
20	28	Male	180	82	141.10	161.72	104.70	142.25	379.01
21	27	Male	174	76	106.45	122.59	104.20	126.27	13.51
21	27	Male	174	76	107.46	122.11	105.60	126.40	18.39
21	27	Male	174	76	113.00	125.67	107.90	127.78	4.45
22	24	Male	170	60	120.58	132.63	109.10	130.89	3.01
22	24	Male	170	60	126.43	137.18	110.60	135.10	4.33
22	24	Male	170	60	133.03	137.86	115.80	139.68	3.30
23	25	Male	172	63	115.69	134.79	103.00	128.48	39.88
23	25	Male	172	63	116.66	128.32	109.10	129.03	0.51
23	25	Male	172	63	116.76	134.34	104.30	128.90	29.63
24	44	Male	168	67	108.47	120.63	107.90	126.61	35.82
24	44	Male	168	67	111.18	120.96	110.30	127.31	40.37
24	44	Male	168	67	113.63	121.86	111.90	128.12	39.16
25	29	Male	165	62	122.72	134.98	109.10	132.34	6.95
25	29	Male	165	62	124.56	137.63	108.60	133.68	15.64
25	29	Male	165	62	128.49	141.33	109.10	136.46	23.71
26	30	Male	170	60	112.54	126.45	106.80	127.60	1.31
26	30	Male	170	60	116.46	128.10	109.10	128.96	0.74
26	30	Male	170	60	116.62	134.17	104.30	128.84	28.37
27	41	Male	174	74	131.66	136.91	115.40	138.77	3.47

27	41	Male	174	74	132.72	136.01	117.10	139.53	12.36
27	41	Male	174	74	136.02	137.16	119.00	141.62	19.85
28	27	Female	165	52	132.97	136.26	117.10	139.68	11.72
28	27	Female	165	52	133.30	133.97	119.40	139.98	36.18
28	27	Female	165	52	136.19	136.87	119.40	141.73	23.59
29	20	Male	160	55	78.73	113.42	83.30	110.78	6.96
29	20	Male	160	55	91.16	119.95	91.20	120.69	0.55
29	20	Male	160	55	97.14	124.27	93.80	124.68	0.17
30	20	Male	180	85	94.13	141.19	80.00	122.07	365.74
30	20	Male	180	85	97.93	150.28	78.20	123.82	699.97
30	20	Male	180	85	99.80	148.22	80.80	124.51	562.24
31	19	Male	170	72	110.27	135.58	97.60	126.56	81.42
31	19	Male	170	72	113.58	136.29	100.00	127.66	74.41
31	19	Male	170	72	115.52	136.31	101.70	128.37	63.09
32	20	Male	170	64	98.47	119.72	98.70	125.30	31.18
32	20	Male	170	64	103.21	126.77	97.70	125.87	0.82
32	20	Male	170	64	105.27	126.32	100.00	126.04	0.08
33	47	Female	166	51	122.56	139.14	105.70	132.11	49.47
33	47	Female	166	51	125.60	136.28	110.60	134.51	3.14
33	47	Female	166	51	126.62	139.27	109.10	135.18	16.74
34	45	Male	175	78	117.81	135.67	104.20	129.30	40.63
34	45	Male	175	78	124.96	141.86	105.70	133.87	63.89
34	45	Male	175	78	126.46	142.09	106.80	134.99	50.45
35	19	Male	169	63	89.51	116.75	92.00	119.02	5.15
35	19	Male	169	63	99.35	127.10	93.80	125.23	3.50
35	19	Male	169	63	101.57	127.23	95.80	125.66	2.48
36	19	Male	170	53	95.94	126.24	91.20	124.07	4.72
36	19	Male	170	53	102.33	129.40	94.90	125.68	13.85
36	19	Male	170	53	105.33	130.58	96.80	125.89	22.00
37	23	Male	168	60	90.80	119.48	91.20	120.32	0.71
37	23	Male	168	60	102.83	120.98	102.00	126.04	25.60
37	23	Male	168	60	127.02	127.13	119.90	135.80	75.14
38	22	Male	162	60	104.15	126.63	98.70	125.95	0.46
38	22	Male	162	60	105.75	135.29	93.80	125.75	90.99
38	22	Male	162	60	106.49	127.79	100.00	126.08	2.91
39	20	Male	169	69	127.78	128.42	119.40	136.31	62.21
39	20	Male	169	69	134.52	137.27	117.60	140.68	11.64
39	20	Male	169	69	145.07	143.63	121.20	142.53	1.20
40	20	Female	172	58	94.95	117.71	96.80	124.00	39.62
40	20	Female	172	58	98.09	122.87	95.80	125.06	4.80
40	20	Female	172	58	98.70	124.80	94.90	125.15	0.13

41	20	Female	153	53	90.80	114.81	94.90	120.73	35.06
41	20	Female	153	53	93.83	115.25	97.70	123.45	67.31
41	20	Female	153	53	96.58	118.63	97.70	124.73	37.19
42	20	Female	155	48	110.49	124.15	106.80	127.00	8.11
42	20	Female	155	48	113.60	127.28	107.10	127.94	0.44
42	20	Female	155	48	116.74	128.40	109.10	129.06	0.44
43	20	Female	156	51	99.70	112.66	106.20	125.88	174.82
43	20	Female	156	51	102.53	112.46	109.40	126.32	192.06
43	20	Female	156	51	112.14	118.88	113.20	127.69	77.68
44	19	Female	163	64	118.44	128.86	110.30	129.79	0.87
44	19	Female	163	64	120.72	132.78	109.10	130.98	3.24
44	19	Female	163	64	128.04	135.13	113.70	136.31	1.39
45	20	Female	152	56	97.02	114.14	102.00	125.10	120.15
45	20	Female	152	56	98.63	116.03	102.00	125.50	89.66
45	20	Female	152	56	101.61	118.27	103.10	125.99	59.66
46	22	Female	164	49	112.88	131.51	103.00	127.56	15.60
46	22	Female	164	49	114.11	129.55	105.70	128.06	2.23
46	22	Female	164	49	115.80	131.47	105.70	128.61	8.16
47	48	Male	168	75	121.56	121.06	120.50	131.89	117.37
47	48	Male	168	75	121.84	122.86	119.00	132.04	84.18
47	48	Male	168	75	122.21	123.24	119.00	132.30	82.03
48	50	Female	150	44	143.16	130.24	131.90	144.09	191.87
48	50	Female	150	44	147.34	134.05	131.90	142.48	71.05
48	50	Female	150	44	149.28	133.58	134.10	141.59	64.20
49	21	Male	180	75	117.82	148.51	95.20	128.92	383.71
49	21	Male	180	75	125.56	156.30	96.40	133.94	499.96
49	21	Male	180	75	125.65	156.41	96.40	134.00	502.04
50	17	Male	167	65	102.98	133.31	92.70	125.61	59.33
50	17	Male	167	65	103.76	135.93	91.60	125.58	107.06
50	17	Male	167	65	104.30	135.01	92.70	125.66	87.48

Predicted Outcome of Stride Length for Slow Walking Speed

Subject	Age	Gender	Height (cm)	Weight (kg)	Speed (cm/s)	Stride length (cm)	Cadence (step/min)	Predicted Stride Length (cm/s)	MSE
1	48	Female	158	50	83.60	98.94	101.40	107.71	76.93
1	48	Female	158	50	93.21	101.32	110.40	111.62	106.14
1	48	Female	158	50	101.74	108.14	112.90	113.18	25.35
2	22	Female	155	59	74.69	102.55	87.40	104.07	2.30
2	22	Female	155	59	78.47	105.10	89.60	105.09	0.00
2	22	Female	155	59	83.77	108.91	92.30	108.10	0.66
3	22	Female	163	54	94.67	118.58	95.80	118.08	0.25
3	22	Female	163	54	95.28	123.07	92.90	116.91	37.90
3	22	Female	163	54	103.72	127.52	97.60	122.74	22.83
4	27	Female	150	37	93.28	117.95	94.90	112.78	26.69
4	27	Female	150	37	97.75	112.46	104.30	113.89	2.06
4	27	Female	150	37	100.93	113.41	106.80	112.06	1.83
5	26	Female	168	53	85.48	121.39	84.50	112.51	78.77
5	26	Female	168	53	87.93	122.26	86.30	115.11	51.13
5	26	Female	168	53	90.85	123.18	88.50	114.91	68.35
6	25	Female	157	43	114.83	132.11	104.30	133.78	2.79
6	25	Female	157	43	115.74	133.16	104.30	133.31	0.02
6	25	Female	157	43	116.96	134.56	104.30	132.67	3.55
7	51	Female	152	42	87.91	112.22	94.00	111.57	0.42
7	51	Female	152	42	91.42	113.33	96.80	112.82	0.26
7	51	Female	152	42	92.42	114.57	96.80	112.50	4.27
8	31	Female	165	65	37.71	98.80	45.80	108.71	98.29
8	31	Female	165	65	47.26	109.69	51.70	100.68	81.16
8	31	Female	165	65	52.08	108.13	57.80	101.44	44.74
9	48	Female	158	43	61.43	98.68	74.70	101.70	9.12
9	48	Female	158	43	65.54	101.75	77.30	104.56	7.91
9	48	Female	158	43	72.64	106.05	82.20	107.27	1.49
10	22	Female	153	45	93.04	111.98	99.70	112.31	0.11
10	22	Female	153	45	95.50	108.83	105.30	112.56	13.93
10	22	Female	153	45	97.37	110.65	105.60	111.54	0.80
11	26	Female	156	48	83.02	107.94	92.30	107.01	0.86
11	26	Female	156	48	88.07	108.95	97.00	107.96	0.99
11	26	Female	156	48	96.29	115.55	100.00	107.12	71.11
12	25	Female	165	52	101.53	125.86	96.80	123.52	5.49
12	25	Female	165	52	108.19	131.54	98.70	129.29	5.08
12	25	Female	165	52	111.64	132.64	101.00	130.82	3.32
13	25	Male	175	70	86.43	103.10	100.60	105.23	4.52
13	25	Male	175	70	87.62	100.81	104.30	105.48	21.79

13	25	Male	175	70	87.86	101.08	104.30	105.37	18.43
14	27	Female	162	50	110.42	129.90	102.00	131.31	1.99
14	27	Female	162	50	116.36	132.10	105.70	131.86	0.06
14	27	Female	162	50	117.60	132.13	106.80	131.98	0.02
15	26	Female	174	59	88.59	105.67	100.60	105.38	0.08
15	26	Female	174	59	93.09	105.19	106.20	108.24	9.32
15	26	Female	174	59	98.46	112.21	105.30	116.70	20.17
16	27	Male	189	77	83.86	120.09	83.80	115.48	21.22
16	27	Male	189	77	85.17	115.22	88.70	120.09	23.77
16	27	Male	189	77	91.29	120.12	91.20	117.65	6.08
17	26	Male	170	65	99.55	121.03	98.70	117.52	12.29
17	26	Male	170	65	108.71	120.90	107.90	121.54	0.41
17	26	Male	170	65	109.89	124.88	105.60	119.80	25.82
18	24	Male	185	75	95.82	139.38	82.50	137.28	4.42
18	24	Male	185	75	101.54	138.15	88.20	138.64	0.24
18	24	Male	185	75	104.54	138.46	90.60	138.16	0.09
19	46	Male	178	50	104.62	121.77	103.10	119.83	3.78
19	46	Male	178	50	105.94	115.26	110.30	122.96	59.28
19	46	Male	178	50	109.48	121.76	107.90	117.92	14.76
20	28	Male	180	82	88.85	129.23	82.50	128.05	1.40
20	28	Male	180	82	90.23	126.64	85.50	128.86	4.94
20	28	Male	180	82	101.84	133.41	91.60	119.68	188.61
21	27	Male	174	76	79.16	112.42	84.50	111.25	1.37
21	27	Male	174	76	86.94	112.30	92.90	104.10	67.23
21	27	Male	174	76	99.27	119.12	100.00	111.84	52.95
22	24	Male	170	60	83.14	106.14	94.00	113.98	61.49
22	24	Male	170	60	83.86	108.20	93.00	114.12	34.99
22	24	Male	170	60	84.97	108.47	94.00	115.07	43.62
23	25	Male	172	63	74.91	112.09	80.20	114.88	7.79
23	25	Male	172	63	79.75	117.14	81.70	113.59	12.60
23	25	Male	172	63	80.44	115.19	83.80	113.70	2.22
24	44	Male	168	67	65.79	88.91	88.80	100.69	138.71
24	44	Male	168	67	70.86	91.93	92.50	107.70	248.69
24	44	Male	168	67	70.86	91.93	92.50	107.70	248.69
25	29	Male	165	62	79.65	116.99	81.70	116.51	0.23
25	29	Male	165	62	80.80	114.48	84.70	116.68	4.83
25	29	Male	165	62	94.08	127.57	88.50	118.53	81.77
26	30	Male	170	60	77.08	116.20	79.60	115.58	0.38
26	30	Male	170	60	87.99	119.31	88.50	116.81	6.27
26	30	Male	170	60	89.92	124.03	87.00	117.21	46.55
27	41	Male	174	74	69.46	103.55	80.50	110.23	44.66

27	41	Male	174	74	78.36	109.46	85.90	112.42	8.78
27	41	Male	174	74	81.13	108.65	89.60	111.51	8.21
28	27	Female	165	52	95.60	120.88	94.90	120.77	0.01
28	27	Female	165	52	99.37	123.18	96.80	121.67	2.27
28	27	Female	165	52	103.73	129.13	96.40	125.31	14.60
29	20	Male	160	55	59.92	100.84	71.30	98.45	5.71
29	20	Male	160	55	68.56	113.79	72.30	106.97	46.45
29	20	Male	160	55	72.00	110.34	78.30	110.07	0.07
30	20	Male	180	85	59.76	111.88	64.10	111.64	0.06
30	20	Male	180	85	63.51	112.74	67.60	110.69	4.18
30	20	Male	180	85	64.24	110.12	70.00	112.51	5.73
31	19	Male	170	72	85.25	114.18	89.60	114.61	0.18
31	19	Male	170	72	93.47	121.92	92.00	117.60	18.66
31	19	Male	170	72	96.63	119.79	96.80	117.88	3.63
32	20	Male	170	64	75.22	113.40	79.60	115.25	3.43
32	20	Male	170	64	78.31	115.02	81.70	115.32	0.09
32	20	Male	170	64	85.14	118.53	86.20	113.91	21.38
33	47	Female	166	51	81.96	116.39	84.50	113.49	8.40
33	47	Female	166	51	98.41	119.65	98.70	122.59	8.65
33	47	Female	166	51	103.33	127.05	97.60	125.31	3.04
34	45	Male	175	78	68.22	110.78	73.90	108.32	6.04
34	45	Male	175	78	70.63	110.64	76.60	109.16	2.19
34	45	Male	175	78	71.31	112.44	76.10	108.63	14.53
35	19	Male	169	63	66.96	106.29	75.60	97.02	86.01
35	19	Male	169	63	71.34	111.76	76.60	105.36	40.98
35	19	Male	169	63	74.82	112.79	79.60	113.70	0.82
36	19	Male	170	53	76.66	117.49	78.30	115.36	4.56
36	19	Male	170	53	76.86	125.83	73.30	114.05	138.73
36	19	Male	170	53	86.04	126.38	81.70	113.79	158.55
37	23	Male	168	60	56.72	94.14	72.30	94.23	0.01
37	23	Male	168	60	57.26	93.62	73.40	94.50	0.77
37	23	Male	168	60	62.75	94.12	80.00	94.91	0.62
38	22	Male	162	60	48.29	91.25	63.50	94.74	12.21
38	22	Male	162	60	49.31	91.89	64.40	94.32	5.91
38	22	Male	162	60	56.04	95.12	70.70	98.84	13.80
39	20	Male	169	69	97.29	113.24	103.10	119.80	43.03
39	20	Male	169	69	104.46	119.95	104.50	119.79	0.02
39	20	Male	169	69	106.25	116.87	109.10	121.72	23.52
40	20	Female	172	58	60.40	98.48	73.60	94.16	18.69
40	20	Female	172	58	60.50	93.32	77.80	98.72	29.15
40	20	Female	172	58	67.64	103.66	78.30	105.59	3.73

41	20	Female	153	53	50.08	92.89	64.70	93.82	0.86
41	20	Female	153	53	53.13	93.62	68.10	94.01	0.16
41	20	Female	153	53	57.28	95.74	71.80	93.54	4.84
42	20	Female	155	48	73.17	104.40	84.10	103.16	1.54
42	20	Female	155	48	81.52	109.54	89.30	106.69	8.13
42	20	Female	155	48	84.48	109.83	92.30	108.09	3.02
43	20	Female	156	51	77.05	101.72	90.90	104.72	9.03
43	20	Female	156	51	78.84	103.29	91.60	105.31	4.06
43	20	Female	156	51	89.10	108.11	98.90	108.42	0.09
44	19	Female	163	64	92.54	111.38	99.70	120.45	82.35
44	19	Female	163	64	97.29	113.24	103.10	119.13	34.73
44	19	Female	163	64	99.49	110.14	108.40	121.12	120.64
45	20	Female	152	56	82.68	110.00	90.20	109.04	0.92
45	20	Female	152	56	84.77	108.22	94.00	110.84	6.85
45	20	Female	152	56	92.01	112.55	98.10	113.00	0.20
46	22	Female	164	49	76.67	116.61	78.90	117.64	1.05
46	22	Female	164	49	77.07	118.11	78.30	117.08	1.06
46	22	Female	164	49	79.86	118.46	80.90	116.51	3.79
47	48	Male	168	75	80.94	98.21	98.90	114.20	255.56
47	48	Male	168	75	84.15	99.59	101.40	114.72	228.80
47	48	Male	168	75	91.42	101.11	108.50	121.30	407.78
48	50	Female	150	44	107.40	108.85	118.40	113.42	20.90
48	50	Female	150	44	110.71	113.36	117.20	110.78	6.68
48	50	Female	150	44	116.62	117.60	119.00	113.08	20.44
49	21	Male	180	75	86.35	128.09	80.90	128.42	0.11
49	21	Male	180	75	86.43	128.21	80.90	128.38	0.03
49	21	Male	180	75	88.87	129.42	82.40	127.99	2.05
50	17	Male	167	65	71.50	117.86	72.80	103.66	201.58
50	17	Male	167	65	75.11	121.15	74.40	112.67	71.88
50	17	Male	167	65	77.36	122.47	75.80	113.96	72.49

Appendix - H

Adjusted Lower Limb Joints Angle

Figure H-1: Adjusted Lower Limb Joints Angle of Subject 1

Figure H-2: Adjusted Walking Motions of Subject 1 (Red Triangle Representing the Foot, Graph Unit is mm)

Figure H-3: Adjusted Lower Limb Joints Angle of Subject 2

Figure H-4: Adjusted Walking Motions of Subject 2 (Red Triangle Representing the Foot, Graph Unit is mm)

Figure H-5: Adjusted Lower Limb Joints Angle of Subject 3

Figure H-6: Adjusted Walking Motions of Subject 3 (Red Triangle Representing the Foot, Graph Unit is mm)

Figure H-7: Adjusted Lower Limb Joints Angle of Subject 4

Figure H-8: Adjusted Walking Motions of Subject 4 (Red Triangle Representing the Foot, Graph Unit is mm)

Figure H-9: Adjusted Lower Limb Joints Angle of Subject 5

Figure H-10: Adjusted Walking Motions of Subject 5 (Red Triangle Representing the Foot, Graph Unit is mm)

Figure H-11: Adjusted Lower Limb Joints Angle of CGA Data

Figure H-12: Adjusted Walking Motions of CGA Data (Red Triangle Representing the Foot, Graph Unit is mm)

Appendix - I

List of Relevant Publications Published

1. H. B. Lim, K. H. Hoon, K. H. Low, Y. C. Soh, and Adela Tow, "Pelvic control and over-ground walking methodology for impaired gait recovery," in *Proceedings of the IEEE International Conference on Robotics and Biomimetics*, Bangkok, Thailand, pp. 282-287, 2008.
2. H. B. Lim, K. H. Hoon, Y. C. Soh, Adela Tow, and K. H. Low, "Gait planning for effective rehabilitation - from gait study to application in clinical rehabilitation," in *Proceedings of the 2009 IEEE International Conference on Rehabilitation Robotics*, ICORR 2009, Kyoto, Japan, pp. 271-276, 2009.
3. H. B. Lim, K. H. Hoon, Y. C. Soh, Adela Tow, and K. H. Low, "Effective gait planning for robotic rehabilitation - from normal gait study to application in clinical rehabilitation," in *Proceedings of the IEEE/ASME International Conference on Advanced Intelligent Mechatronics*, Singapore, pp. 1885-1890, 2009.
4. Ping Wang, A. McGregor, Adela Tow, H. B. Lim, L. S. Khang, and K. H. Low, "Rehabilitation control strategies for gait robot via EMG evaluation and quantification," in *Proceedings of the 2009 IEEE International Conference on Rehabilitation Robotics*, ICORR 2009, Kyoto, Japan, pp. 86-91, 2009.
5. H. B. Lim, Trieu Phat Luu, K. H. Hoon, and K. H. Low, "Natural Gait Parameters Prediction for Gait Rehabilitation via Artificial Neural Network," in *Proceedings of the 2010 IEEE/RSJ International Conference on Intelligent Robots and Systems*, Taipei, Taiwan, pp. 5398-5403, 2010.
6. H. B. Lim, Trieu Phat Luu, K. H. Hoon, Xingda Qu, and K. H. Low, "Body Weight Support with Natural Pelvic Motion Assistance for Robotic Gait Rehabilitation," in *Proceedings of the International Conference on Advanced Mechatronics 2010*, OSAKA, JAPAN, 2010.
7. Trieu Phat Luu, H. B. Lim, Xingda Qu, and K. H. Low, "Subject tailored gait pattern planning for robotic gait rehabilitation," in *Proceedings of 2010 IEEE In-*

- ternational Conference on Robotics and Biomimetics (ROBIO)*, Tianjin, China, pp. 259-264, 2010.
8. Trieu Phat Luu, H. B. Lim, Xingda Qu, and K. H. Low, "Pelvic motion assistance of NaTure-gaits with adaptive body weight support," in *Proceedings of 2011 8th Asian Control Conference (ASCC)*, Kaohsiung, Taiwan, pp. 950-955, 2011.
 9. H. B. Lim, Trieu Phat Luu, K. H. Hoon, Xingda. Qu, A. Tow, and K. H. Low, "Study of body weight shifting on robotic assisted gait rehabilitation with NaTure-gaits," in *Proceedings of 2011 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, San Francisco, USA, pp. 4923-4928, 2011.
 10. Trieu Phat Luu, H. B. Lim, K. H. Hoon, Xingda. Qu, and K. H. Low, "Subject-specific gait parameters prediction for robotic gait rehabilitation via generalized regression neural network," in *Proceedings of 2011 IEEE International Conference on Robotics and Biomimetics (ROBIO)*, Phuket, Thailand, pp. 914-919, 2011.
 11. Trieu Phat Luu, H. B. Lim, Xingda. Qu, K. H. Hoon, and K. H. Low, "Subject-specific lower limb waveforms planning via artificial neural network," in *Proceedings of 2011 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Zurich, Switzerland, pp. 1-6, 2011.

References

- [1] "Stroke Statistics," [Online]. Available: <http://www.americanheart.org/presenter.jhtml?identifier=4725>. Accessed on September, 2009
- [2] "Spinal Cord Injury Facts & Figures at a Glance 2009," [Online]. Available: <http://www.spinalcord.uab.edu/show.asp?durki=119513>. Accessed on September, 2009
- [3] A. L. Behrman, M. G. Bowden, and P. M. Nair, "Neuroplasticity after spinal cord injury and training: An emerging paradigm shift in rehabilitation and walking recovery," *Physical Therapy*, vol. 86, pp. 1406-1425, 2006.
- [4] J. Rose and J. G. Gamble, *Human Walking*, 3rd ed. Lippincott Williams & Wilkins, Philadelphia, 2006.
- [5] H. Barbeau, M. Wainberg, and L. Finch, "Description and application of a system for locomotor rehabilitation," *Medical and Biological Engineering and Computing*, vol. 25, pp. 341-344, 1987.
- [6] S. Hesse and C. Werner, "Connecting research to the needs of patients and clinicians," *Brain Research Bulletin*, vol. 78, pp. 26-34, 2009.
- [7] J. Hidler, D. Nichols, M. Pelliccio, and K. Brady, "Advances in the understanding and treatment of stroke impairment using robotic devices," *Topics in Stroke Rehabilitation*, vol. 12, pp. 22-35, 2005.
- [8] J. Hidler, L. F. Hamm, A. Lichy, and S. L. Groah, "Automating activity-based interventions: The role of robotics," *Journal of Rehabilitation Research and Development*, vol. 45, pp. 337-344, 2008.
- [9] "Welcome to Hocoma," [Online]. Available: <http://www.hocoma.ch>. Accessed on August
- [10] Y. Allemand, Y. Stauffer, R. Clavel, and R. Brodard, "Design of a new lower extremity orthosis for overground gait training with the WalkTrainer," in *Proceedings of the IEEE 11th International Conference on Rehabilitation Robotics*, Kyoto, Japan, pp. 550-555, 2009.
- [11] S. Hussein, H. Schmidt, S. Hesse, and J. Kruger, "Effect of different training modes on ground reaction forces during robot assisted floor walking and stair climbing," in *Proceedings of the 2009 IEEE International Conference on Rehabilitation Robotics, ICORR 2009*, Kyoto, Japan, pp. 845-850, 2009.
- [12] "KineAssist," [Online]. Available: <http://www.kineadesign.com/>. Accessed on July
- [13] H. B. Lim, K. H. Hoon, K. H. Low, Y. C. Soh, and Adela Tow, "Pelvic control and over-ground walking methodology for impaired gait recovery," in *Proceedings of the IEEE International Conference on Robotics and Biomimetics*, Bangkok, Thailand, pp. 282-287, 2008.
- [14] G. Schmeisser and W. Seamone, "An assistive equipment controller for quadriplegics," *Johns Hopkins Medical Journal*, vol. 145, pp. 84-88, 1979.
- [15] L. Leifer, "Rehabilitative robotics, The Stanford robotic AID," in *Wescon Conference Record*, San Francisco, CA, USA, vol. 25, 1981.
- [16] "Standard neurological classification of Spinal Cord Injury," 2006. [Online]. Available: <http://www.asia->

- spinalinjury.org/publications/2006_Classif_worksheet.pdf. Accessed on November, 2009
- [17] "2006 Annual Statistical Report," National Spinal Cord Injury Statistical Center, University of Alabama at Birmingham, July, 2006.
 - [18] A. W. Heinemann, R. Magiera-Planey, C. Schiro-Geist, and G. Gimines, "Mobility for persons with spinal cord injury: An evaluation of two systems," *Archives of Physical Medicine and Rehabilitation*, vol. 68, pp. 90-93, 1987.
 - [19] M. F. Somers, *Spinal Cord Injury: Functional Rehabilitation* McGraw-hill/appleton & Lange 1992.
 - [20] L. Jrgensen, B. K. Jacobsen, T. Wilsgaard, and J. H. Magnus, "Walking after stroke: Does it matter? Changes in bone mineral density within the first 12 months after stroke. A longitudinal study," *Osteoporosis International*, vol. 11, pp. 381-387, 2000.
 - [21] C. Werner, S. Von Frankenberg, T. Treig, M. Konrad, and S. Hesse, "Treadmill training with partial body weight support and an electromechanical gait trainer for restoration of gait in subacute stroke patients: A randomized crossover study," *Stroke*, vol. 33, pp. 2895-2901, 2002.
 - [22] D. P. Ferris, G. S. Sawicki, and A. R. Domingo, "Powered lower limb orthoses for gait rehabilitation," *Topics in Spinal Cord Injury Rehabilitation*, vol. 11, pp. 34-49, 2005.
 - [23] R. Schmidt and T. Lee, *Motor Control and Learning: A Behavioral Emphasis*, 3rd ed. Human Kinetics Publishers, 1999.
 - [24] G. Kwakkel, R. C. Wagenaar, J. W. R. Twisk, G. J. Lankhorst, and J. C. Koetsier, "Intensity of leg and arm training after primary middle-cerebral-artery stroke: A randomised trial," *Lancet*, vol. 354, pp. 191-196, 1999.
 - [25] F. Henry, M, *Specificity vs. Generality in Learning Motor Skill*. Englewood Cliffs, NJ: Prentice Hall, 1968.
 - [26] H. Barbeau, "Locomotor training in neurorehabilitation: Emerging rehabilitation concepts," *Neurorehabilitation and Neural Repair*, vol. 17, pp. 3-11, 2003.
 - [27] A. Kaelin-Lane, L. Sawaki, and L. G. Cohen, "Role of voluntary drive in encoding an elementary motor memory," *Journal of Neurophysiology*, vol. 93, pp. 1099-1103, 2005.
 - [28] M. Lotze, C. Braun, N. Birbaumer, S. Anders, and L. G. Cohen, "Motor learning elicited by voluntary drive," *Brain*, vol. 126, pp. 866-872, 2003.
 - [29] M. Hallett, "Neuroplasticity and rehabilitation," *Journal of Rehabilitation Research and Development*, vol. 42, 2005.
 - [30] E. Taub, N. E. Miller, T. A. Novack, E. W. Cook Iii, W. C. Fleming, C. S. Nepomuceno, J. S. Connell, and J. E. Crago, "Technique to improve chronic motor deficit after stroke," *Archives of Physical Medicine and Rehabilitation*, vol. 74, pp. 347-354, 1993.
 - [31] R. R. Roy, A. Ishihara, J. A. Kim, M. Lee, K. Fox, and V. R. Edgerton, "Metabolic and morphological stability of motoneurons in response to chronically elevated neuromuscular activity," *Neuroscience*, vol. 92, pp. 361-366, 1999.

- [32] R. B. Shepherd, "Exercise and training to optimize functional motor performance in stroke: Driving neural reorganization?," *Neural Plasticity*, vol. 8, pp. 121-129, 2001.
- [33] K. J. Sullivan, B. J. Knowlton, and B. H. Dobkin, "Step training with body weight support: Effect of treadmill speed and practice paradigms on poststroke locomotor recovery," *Archives of Physical Medicine and Rehabilitation*, vol. 83, pp. 683-691, 2002.
- [34] V. Dietz, M. Wirz, A. Curt, and G. Colombo, "Locomotor pattern in paraplegic patients: Training effects and recovery of spinal cord function," *Spinal Cord*, vol. 36, pp. 380-390, 1998.
- [35] D. J. Nicol, M. H. Granat, R. H. Baxendale, and S. J. M. Tuson, "Evidence for a human spinal stepping generator," *Brain Research*, vol. 684, pp. 230-232, 1995.
- [36] H. Barbeau and S. Rossignol, "Recovery of locomotion after chronic spinalization in the adult cat," *Brain Research*, vol. 412, pp. 84-95, 1987.
- [37] R. G. Lovely, R. J. Gregor, R. R. Roy, and V. R. Edgerton, "Effects of training on the recovery of full-weight-bearing stepping in the adult spinal cat," *Experimental Neurology*, vol. 92, pp. 421-435, 1986.
- [38] Y. Laufer, R. Dickstein, Y. Chafez, and E. Marcovitz, "The effect of treadmill training on the ambulation of stroke survivors in the early stages of rehabilitation: A randomized study," *Journal of Rehabilitation Research and Development*, vol. 38, pp. 69-78, 2001.
- [39] "Body Weight Support," [Online]. Available: <http://www.harkema.ucla.edu/bws.html>. Accessed on December 2006
- [40] A. L. Behrman and S. J. Harkema, "Locomotor training after human spinal cord injury: A series of case studies," *Physical Therapy*, vol. 80, pp. 688-700, 2000.
- [41] R. Grasso, Y. P. Ivanenko, M. Zago, M. Molinari, G. Scivoletto, V. Castellano, V. Macellari, and F. Lacquaniti, "Distributed plasticity of locomotor pattern generators in spinal cord injured patients," *Brain*, vol. 127, pp. 1019-1034, 2004.
- [42] M. Wirz, G. Colombo, and V. Dietz, "Long term effects of locomotor training in spinal humans," *Journal of Neurology Neurosurgery and Psychiatry*, vol. 71, pp. 93-96, 2001.
- [43] H. Barbeau, K. Norman, J. Fung, M. Visintin, and M. Ladouceur, "Does neurorehabilitation play a role in the recovery of walking in neurological populations?," in *Annals of the New York Academy of Sciences*, vol. 860, pp. 377-392, 1998.
- [44] A. Wernig, A. Nanassy, and S. Muller, "Laufband (treadmill) therapy in incomplete paraplegia and tetraplegia," *Journal of Neurotrauma*, vol. 16, pp. 719-726, 1999.
- [45] A. Wernig, S. Muller, A. Nanassy, and E. Cagol, "Laufband therapy based on 'rules of spinal locomotion' is effective in spinal cord injured persons," *European Journal of Neuroscience*, vol. 7, pp. 823-829, 1995.
- [46] A. Wernig, A. Nanassy, and S. Muller, "Maintenance of locomotor abilities following Laufband (treadmill) therapy in para- and tetraplegic persons: Follow-up studies," *Spinal Cord*, vol. 36, pp. 744-749, 1998.
- [47] J. Hidler, D. Nichols, M. Pelliccio, K. Brady, D. D. Campbell, J. H. Kahn, and T. G. Hornby, "Multicenter randomized clinical trial evaluating the effectiveness of

- the Lokomat in subacute stroke," *Neurorehabilitation and Neural Repair*, vol. 23, pp. 5-13, 2009.
- [48] T. G. Hornby, D. H. Zemon, and D. Campbell, "Robotic-assisted, body-weight-supported treadmill training in individuals following motor incomplete spinal cord injury," *Physical Therapy*, vol. 85, pp. 52-66, 2005.
 - [49] M. Wirz, D. H. Zemon, R. Rupp, A. Scheel, G. Colombo, V. Dietz, and T. G. Hornby, "Effectiveness of automated locomotor training in patients with chronic incomplete spinal cord injury: A multicenter trial," *Archives of Physical Medicine and Rehabilitation*, vol. 86, pp. 672-680, 2005.
 - [50] M. Frey, G. Colombo, M. Vaglio, R. Bucher, M. Jorg, and R. Riener, "A novel mechatronic body weight support system," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 14, pp. 311-321, 2006.
 - [51] "Rehabilitation Robotics," [Online]. Available: <http://web.ethlife.ethz.ch/e/articles/sciencelife/armin.html#oben>. Accessed on July, 2010
 - [52] J. Hidler, W. Wisman, and N. Neckel, "Kinematic trajectories while walking within the Lokomat robotic gait-orthosis," *Clinical Biomechanics*, vol. 23, pp. 1251-1259, 2008.
 - [53] B. Husemann, F. Muller, C. Krewer, S. Heller, and E. Koenig, "Effects of locomotion training with assistance of a robot-driven gait orthosis in hemiparetic patients after stroke: A randomized controlled pilot study," *Stroke*, vol. 38, pp. 349-354, 2007.
 - [54] L. Zimmerli, A. Duschau-Wicke, R. Riener, A. Mayr, and L. Lunenburger, "Virtual reality and gait rehabilitation: Augmented feedback for the Lokomat," in *2009 Virtual Rehabilitation International Conference, VR 2009*, Haifa, pp. 150-153, 2009.
 - [55] R. Riener, L. Lunenburger, S. Jezernik, M. Anderschitz, G. Colombo, and V. Dietz, "Patient-cooperative strategies for robot-aided treadmill training: First experimental results," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 13, pp. 380-394, 2005.
 - [56] A. Duschau-Wicke, S. Felsenstein, and R. Riener, "Adaptive body weight support controls human activity during robot-aided gait training," in *2009 IEEE International Conference on Rehabilitation Robotics, ICORR 2009*, Kyoto, pp. 413-418, 2009.
 - [57] C. Y. E. Wang, J. E. Bobrow, and D. J. Reinkensmeyer, "Dynamic motion planning for the design of robotic gait rehabilitation," *Journal of Biomechanical Engineering*, vol. 127, pp. 672-679, 2005.
 - [58] Z. Feng, J. Qian, Y. Zhang, L. Shen, Z. Zhang, and Q. Wang, "Dynamic walking planning for gait rehabilitation robot," in *2nd International Conference on Bioinformatics and Biomedical Engineering, iCBBE 2008*, Shanghai, China, pp. 1280-1283, 2008.
 - [59] "AutoAmbulator," [Online]. Available: http://www.healthsouth.com/what_we_do/inpatient_rehabilitation/rehabilitation_technology/autoAmbulator.asp. Accessed on July, 2010
 - [60] J. F. Veneman, R. Kruidhof, E. E. G. Hekman, R. Ekkelenkamp, E. H. F. Van Asseldonk, and H. Van Der Kooij, "Design and evaluation of the LOPES

- exoskeleton robot for interactive gait rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 15, pp. 379-386, 2007.
- [61] D. Aoyagi, W. E. Ichinose, S. J. Harkema, D. J. Reinkensmeyer, and J. E. Bobrow, "A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 15, pp. 387-400, 2007.
 - [62] S. K. Banala, S. H. Kim, S. K. Agrawal, and J. P. Scholz, "Robot assisted gait training with active leg exoskeleton (ALEX)," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 17, pp. 2-8, 2009.
 - [63] "Robomedica, Inc.," [Online]. Available: <http://www.robomedica.com/>. Accessed on July, 2010
 - [64] S. Hesse, H. Schmidt, and C. Werner, "Machines to support motor rehabilitation after stroke: 10 Years of experience in Berlin," *Journal of Rehabilitation Research and Development*, vol. 43, pp. 671-678, 2006.
 - [65] M. Grunwald, *Human Haptic Perception: Basics and Applications*, 1st ed. Birkhauser Basel, 2008.
 - [66] H. Schmidt, C. Werner, R. Bernhardt, S. Hesse, and J. Kruger, "Gait rehabilitation machines based on programmable footplates," *Journal of NeuroEngineering and Rehabilitation*, vol. 4, 2007.
 - [67] "Swiss Foundation for Cyberthosis," [Online]. Available: <http://www.fsc-sfc.org/>. Accessed on May
 - [68] M. Bouri, Y. Stauffer, C. Schmitt, Y. Allemand, S. Gnemmi, R. Clavel, P. Metrailler, and R. Brodard, "The WalkTrainer: A Robotic System for Walking Rehabilitation," in *Proceedings of the 2006 IEEE International Conference on Robotics and Biomimetics*, Kunming, China, pp. 1616-1621, 2006.
 - [69] "Cyberthoses," [Online]. Available: <http://lsro.epfl.ch/page64590.html>. Accessed on September, 2010
 - [70] Y. Stauffer, Y. Allemand, M. Bouri, J. Fournier, R. Clavel, P. Metrailler, R. Brodard, and F. Reynard, "Pelvic motion measurement during over ground walking, analysis and implementation on the WalkTrainer reeducation device," in *2008 IEEE/RSJ International Conference on Intelligent Robots and Systems, IROS*, Nice, pp. 2362-2367, 2008.
 - [71] M. Peshkin, D. A. Brown, J. J. Santos-Munne, A. Makhlin, E. Lewis, J. E. Colgate, J. Patton, and D. Schwandt, "KineAssist: A robotic overground gait and balance training device," in *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics*, Chicago, Illinois, vol. 2005, pp. 241-246, 2005.
 - [72] A. L. Behrman, A. R. Lawless-Dixon, S. B. Davis, M. G. Bowden, P. Nair, C. Phadke, E. M. Hannold, P. Plummer, and S. J. Harkema, "Locomotor training progression and outcomes after incomplete spinal cord injury," *Physical Therapy*, vol. 85, pp. 1356-1371, 2005.
 - [73] S. K. Banala, A. Kulpe, and S. K. Agrawal, "A powered leg orthosis for gait rehabilitation of motor-impaired patients," in *Proceedings - IEEE International Conference on Robotics and Automation*, Rome, pp. 4140-4145, 2007.

- [74] "Our Technology," [Online]. Available: <http://www.warecentre.com/images/tech/lokomat.jpg>. Accessed on July, 2010
- [75] E. H. F. Van Asseldonk, J. F. Veneman, R. Ekkelenkamp, J. H. Buurke, F. C. T. Van Der Helm, and H. Van Der Kooij, "The effects on kinematics and muscle activity of walking in a robotic gait trainer during zero-force control," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 16, pp. 360-370, 2008.
- [76] A. Duschau-Wicke, J. V. Zitzewitz, L. Lunenburger, and R. Riener, "Patient-driven cooperative gait training with the rehabilitation robot lokomat," in *IFMBE Proceedings*, Antwerp, vol. 22 IFMBE, pp. 1616-1619, 2008.
- [77] H. Vallery, J. Veneman, E. van Asseldonk, R. Ekkelenkamp, M. Buss, and H. van Der Kooij, "Compliant actuation of rehabilitation robots," *IEEE Robotics and Automation Magazine*, vol. 15, pp. 60-69, 2008.
- [78] D. J. Reinkensmeyer, D. Aoyagi, J. L. Emken, J. A. Galvez, W. Ichinose, G. Kerdanyan, S. Maneekobkunwong, K. Minakata, J. A. Nessler, R. Weber, R. R. Roy, R. De Leon, J. E. Bobrow, S. J. Harkema, and V. Reggie Edgerton, "Tools for understanding and optimizing robotic gait training," *Journal of Rehabilitation Research and Development*, vol. 43, pp. 657-670, 2006.
- [79] G. S. Virk, S. K. Bag, S. C. Gharooni, M. O. Tokhi, R. I. Tylor, S. Bradshaw, F. Jamil, I. D. Swain, P. H. Chapple, and R. A. Allen, "Robotic Walking Aids for Disabled Persons," in *Climbing and Walking Robots* Springer Berlin Heidelberg, pp. 801-811, 2006.
- [80] A. Lamontagne, F. Malouin, C. L. Richards, and F. Dumas, "Mechanisms of disturbed motor control in ankle weakness during gait after stroke," *Gait & Posture*, vol. 15, pp. 244-255, 2002.
- [81] J. Patton, D. A. Brown, M. Peshkin, J. J. Santos-Munn, A. Makhlin, E. Lewis, J. E. Colgate, and D. Schwandt, "KineAssist: Design and development of a robotic overground gait and balance therapy device," *Topics in Stroke Rehabilitation*, vol. 15, pp. 131-139, 2008.
- [82] "The Driven Gait Orthosis Lokomat," [Online]. Available: <http://www.jneuroengrehab.com/content/4/1/1/figure/F1?highres=y>. Accessed on September, 2010
- [83] Y. Stauffer, Y. Allemand, M. Bouri, J. Fournier, R. Clavel, P. Metrailler, R. Brodard, and F. Reynard, "The WalkTrainer - A new generation of walking reeducation device combining orthoses and muscle stimulation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 17, pp. 38-45, 2009.
- [84] J. B. d. M. Saunders, V. T. Inman, and H. D. Eberhart, "The major determinants in normal and pathological gait," *J Bone Joint Surg Am*, vol. 35, pp. 543-558, July 1, 1953.
- [85] J. F. Veneman, J. Menger, E. H. F. van Asseldonk, F. C. T. van der Helm, and H. van der Kooij, "Fixating the pelvis in the horizontal plane affects gait characteristics," *Gait and Posture*, vol. 28, pp. 157-163, 2008.
- [86] J. Perry, *Gait Analysis - Normal and Pathological Function*. SLACK Incorporated, 1992.

- [87] V. T. Inman, H. Ralston, and F. Todd, *Human Walking*, 1st ed. Williams and Wilkins, Baltimore/London, 1981.
- [88] R. R. Neptune, S. A. Kautz, and F. E. Zajac, "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking," *Journal of Biomechanics*, vol. 34, pp. 1387-1398, 2001.
- [89] D. Aoyagi, W. E. Ichinose, S. J. Harkema, D. J. Reinkensmeyer, and J. E. Bobrow, "An assistive robotic device that can synchronize to the pelvic motion during human gait training," in *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics*, Chicago, Illinois, vol. 2005, pp. 565-568, 2005.
- [90] "Lokomat2," [Online]. Available: <http://www.cederholm.net/lokomat2.jpg>. Accessed on September, 2010
- [91] T. Warabi, M. Kato, K. Kiriyama, T. Yoshida, and N. Kobayashi, "Treadmill walking and overground walking of human subjects compared by recording sole-floor reaction force," *Neuroscience Research*, vol. 53, pp. 343-348, 2005.
- [92] H. Stolze, J. P. Kuhtz-Buschbeck, C. Mondwurf, A. Boczek-Funcke, K. Johnk, G. Deuschl, and M. Illert, "Gait analysis during treadmill and overground locomotion in children and adults," *Electroencephalography and Clinical Neurophysiology - Electromyography and Motor Control*, vol. 105, pp. 490-497, 1997.
- [93] L. Vogt, K. Pfeifer, and W. Banzer, "Comparison of angular lumbar spine and pelvis kinematics during treadmill and overground locomotion," *Clinical Biomechanics*, vol. 17, pp. 162-165, 2002.
- [94] S. Hesse, M. Konrad, and D. Uhlenbrock, "Treadmill walking with partial body weight support versus floor walking in hemiparetic subjects," *Archives of Physical Medicine and Rehabilitation*, vol. 80, pp. 421-427, 1999.
- [95] J. L. Song and J. Hidler, "Biomechanics of overground vs. treadmill walking in healthy individuals," *Journal of Applied Physiology*, vol. 104, pp. 747-755, 2008.
- [96] P. O. Riley, G. Paolini, U. Della Croce, K. W. Paylo, and D. C. Kerrigan, "A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects," *Gait and Posture*, vol. 26, pp. 17-24, 2007.
- [97] B. Brouwer, K. Parvataneni, and S. J. Olney, "A comparison of gait biomechanics and metabolic requirements of overground and treadmill walking in people with stroke," *Clinical Biomechanics*, vol. 24, pp. 729-734, 2009.
- [98] F. Alton, L. Baldey, S. Caplan, and M. C. Morrissey, "A kinematic comparison of overground and treadmill walking," *Clinical Biomechanics*, vol. 13, pp. 434-440, 1998.
- [99] K. H. Seo and J. J. Lee, "The development of two mobile gait rehabilitation systems," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 17, pp. 156-166, 2009.
- [100] S. Hesse and D. Uhlenbrock, "A mechanized gait trainer for restoration of gait," *Journal of Rehabilitation Research and Development*, vol. 37, pp. 701-708, 2000.
- [101] J. A. Galvez, G. Kerdanyan, S. Maneekobkunwong, R. Weber, M. Scott, S. J. Harkema, and D. J. Reinkensmeyer, "Measuring human trainers' skill for the design of better robot control algorithms for gait training after spinal cord

- injury," in *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics*, Chicago, IL, vol. 2005, pp. 231-234, 2005.
- [102] S. Hesse, J. Mehrholz, and C. Werner, "Robot-assisted upper and lower limb rehabilitation after stroke," *Roboter- und gerätegestützte rehabilitation nach schlaganfall: Gehen und arm-/handfunktion*, vol. 105, pp. 330-336, 2008.
 - [103] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill training of paraplegic patients using a robotic orthosis," *Journal of Rehabilitation Research and Development*, vol. 37, pp. 693-700, 2000.
 - [104] L. G. Lippman and R. Rees, "Consequences of Error Production in a Perceptual-Motor Task," *Journal of General Psychology*, vol. 124, pp. 133-142, 1997.
 - [105] C. Weiller, M. Juptner, S. Fellows, M. Rijntjes, G. Leonhardt, S. Kiebel, S. Muller, H. C. Diener, and A. F. Thilmann, "Brain representation of active and passive movements," *NeuroImage*, vol. 4, pp. 105-110, 1996.
 - [106] C. Carel, I. Loubinoux, K. Boulanouar, C. Manelfe, O. Rascol, P. Celsis, and F. Chollet, "Neural substrate for the effects of passive training on sensorimotor cortical representation: A study with functional magnetic resonance imaging in healthy subjects," *Journal of Cerebral Blood Flow and Metabolism*, vol. 20, pp. 478-484, 2000.
 - [107] T. Mima, N. Sadato, S. Yazawa, T. Hanakawa, H. Fukuyama, Y. Yonekura, and H. Shibasaki, "Brain structures related to active and passive finger movements in man," *Brain*, vol. 122, pp. 1989-1997, 1999.
 - [108] J. F. Israel, D. D. Campbell, J. H. Kahn, and T. G. Hornby, "Metabolic costs and muscle activity patterns during robotic- and therapist-assisted treadmill walking in individuals with incomplete spinal cord injury," *Physical Therapy*, vol. 86, pp. 1466-1478, 2006.
 - [109] J. M. Hidler and A. E. Wall, "Alterations in muscle activation patterns during robotic-assisted walking," *Clinical Biomechanics*, vol. 20, pp. 184-193, 2005.
 - [110] M. Bernhardt, M. Frey, G. Colombo, and R. Riener, "Hybrid force-position control yields cooperative behaviour of the rehabilitation robot LOKOMAT," in *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics*, Chicago, IL, vol. 2005, pp. 536-539, 2005.
 - [111] S. K. Banala, S. K. Agrawal, S. H. Kim, and J. P. Scholz, "Novel gait adaptation and neuromotor training results using an active leg exoskeleton," *IEEE/ASME Transactions on Mechatronics*, vol. 15, pp. 216-225, 2010.
 - [112] A. M. Dollar and H. Herr, "Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art," *IEEE Transactions on Robotics*, vol. 24, pp. 144-158, 2008.
 - [113] H. Vallery, E. H. F. Van Asseldonk, M. Buss, and H. Van Der Kooij, "Reference trajectory generation for rehabilitation robots: Complementary limb motion estimation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 17, pp. 23-30, 2009.
 - [114] R. Riener, L. Lunenburger, and G. Colombo, "Cooperative strategies for robot-aided gait neuro-rehabilitation," in *Annual International Conference of the IEEE Engineering in Medicine and Biology - Proceedings*, San Francisco, CA, vol. 26 VII, pp. 4822-4825, 2004.

- [115] S. Jezernik, G. Colombo, and M. Morari, "Automatic gait-pattern adaptation algorithms for rehabilitation with a 4-DOF robotic orthosis," *IEEE Transactions on Robotics and Automation*, vol. 20, pp. 574-582, 2004.
- [116] L. L. Cai, A. J. Fong, L. Yongqiang, J. Burdick, and V. R. Edgerton, "Assist-as-needed training paradigms for robotic rehabilitation of spinal cord injuries," in *Proceedings - IEEE International Conference on Robotics and Automation*, Orlando, FL, vol. 2006, pp. 3504-3511, 2006.
- [117] A. Duschau-Wicke, J. Von Zitzewitz, A. Caprez, L. Lunenburger, and R. Riener, "Path control: A method for patient-cooperative robot-aided gait rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 18, pp. 38-48, 2010.
- [118] E. H. F. Van Asseldonk, R. Ekkelenkamp, J. F. Veneman, F. C. T. Van Der Helm, and H. Van Der Kooij, "Selective control of a subtask of walking in a robotic gait trainer(LOPES)," in *2007 IEEE 10th International Conference on Rehabilitation Robotics, ICORR'07*, Noordwijk, pp. 841-848, 2007.
- [119] C. Kirtley, M. W. Whittle, and R. J. Jefferson, "Influence of walking speed on gait parameters," *Journal of Biomedical Engineering*, vol. 7, pp. 282-288, 1985.
- [120] T. Oberg, A. Karsznia, and K. Oberg, "Joint angle parameters in gait: Reference data for normal subjects, 10-79 years of age," *Journal of Rehabilitation Research and Development*, vol. 31, pp. 199-213, 1994.
- [121] J. L. Lelas, G. J. Merriman, P. O. Riley, and D. C. Kerrigan, "Predicting peak kinematic and kinetic parameters from gait speed," *Gait & Posture*, vol. 17, pp. 106-112, 2003.
- [122] M. Hanlon and R. Anderson, "Prediction methods to account for the effect of gait speed on lower limb angular kinematics," *Gait & Posture*, vol. 24, pp. 280-287, 2006.
- [123] A. B. Zoss, H. Kazerooni, and A. Chu, "Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX)," *Mechatronics, IEEE/ASME Transactions on*, vol. 11, pp. 128-138, 2006.
- [124] D. A. Winter, *Biomechanics and Motor Control of Human Movement*, 3rd ed. John Wiley & Sons 2004.
- [125] H. B. Lim, Trieu Phat Luu, K. H. Hoon, and K. H. Low, "Natural Gait Parameters Prediction for Gait Rehabilitation via Artificial Neural Network," in *Proceedings of the 2010 IEEE/RSJ International Conference on Intelligent Robots and Systems*, Taipei, Taiwan, 2010.
- [126] "GAITRite Systems," [Online]. Available: <http://www.gaitrite.com/>. Accessed on September, 2009
- [127] R. J. Elble, S. S. Thomas, C. Higgins, and J. Colliver, "Stride-dependent changes in gait of older people," *Journal of Neurology*, vol. 238, pp. 1-5, 1991.
- [128] H. B. Lim, K. H. Hoon, Y. C. Soh, Adela Tow, and K. H. Low, "Effective gait planning for robotic rehabilitation - from normal gait study to application in clinical rehabilitation," in *Proceedings of the IEEE/ASME International Conference on Advanced Intelligent Mechatronics*, Singapore, pp. 1885-1890, 2009.
- [129] C. L. Vaughan, B. L. Davis, and J. C. O'Connor, *Dynamics of Human Gait*. Kiboho Publishers, 1999.

- [130] "Motion Analysis Corporation, the Motion Capture Leader," 2009. [Online]. Available: <http://www.motionanalysis.com/>. Accessed on March, 2010
- [131] "Coronal plane," [Online]. Available: http://upload.wikimedia.org/wikipedia/commons/e/e1/Human_anatomy_planes.svg. Accessed on September, 2010
- [132] D. C. Boone and S. P. Azen, "Normal range of motion of joints in male subjects," *Journal of Bone and Joint Surgery - Series A*, vol. 61, pp. 756-759, 1979.
- [133] D. A. Winter, *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*, 2nd ed. Waterloo Biomechanics, Ontario, Canada, 1991.
- [134] P. K. Levangie and C. C. Norkin, *Joint Structure And Function: A Comprehensive Analysis*, 4th ed. F A Davis Company, 2005.
- [135] V. T. Inman, "Human locomotion," *Canadian Medical Association journal*, vol. 94, pp. 1047-1054, 1966.
- [136] S. Onyshko and D. A. Winter, "A mathematical model for the dynamics of human locomotion," *Journal of Biomechanics*, vol. 13, pp. 361-368, 1980.
- [137] "Foot Levelers Orthotics," 2006. [Online]. Available: <http://www.advancedchiropractic.com/footlevelers.asp>. Accessed on December, 2009
- [138] N. Neckel, W. Wisman, and J. Hidler, "Limb alignment and kinematics inside a lokomat robotic orthosis," in *Annual International Conference of the IEEE Engineering in Medicine and Biology - Proceedings*, New York, NY, pp. 2698-2701, 2006.
- [139] H. B. Lim, Trieu Phat Luu, K. H. Hoon, Xingda Qu, and K. H. Low, "Body Weight Support with Natural Pelvic Motion Assistance for Robotic Gait Rehabilitation," in *Proceedings of the International Conference on Advanced Mechatronics 2010*, OSAKA, JAPAN, 2010.
- [140] J. Pratt, P. Dilworth, and G. Pratt, "Virtual model control of a bipedal walking robot," in *Proceedings - IEEE International Conference on Robotics and Automation*, Albuquerque, NM, USA, vol. 1, pp. 193-198, 1997.
- [141] "Sensoray | Embedded Electronics," [Online]. Available: <http://www.sensoray.com/>. Accessed on May, 2010
- [142] "TeleMyo 2400 G2 Telemetry System," [Online]. Available: http://www.noraxon.com/products/instruments/telemetry_system.php3. Accessed on Sep, 2010
- [143] A. R. Den Otter, A. C. H. Geurts, T. Mulder, and J. Duysens, "Speed related changes in muscle activity from normal to very slow walking speeds," *Gait & Posture*, vol. 19, pp. 270-278, 2004.
- [144] A. Schmitz, A. Silder, B. Heiderscheit, J. Mahoney, and D. G. Thelen, "Differences in lower-extremity muscular activation during walking between healthy older and young adults," *Journal of Electromyography and Kinesiology*, vol. 19, pp. 1085-1091, 2009.
- [145] C. Frigo and P. Crenna, "Multichannel SEMG in clinical gait analysis: A review and state-of-the-art," *Clinical Biomechanics*, vol. 24, pp. 236-245, 2009.
- [146] L. H. Hamel, *Knowledge Discovery with Support Vector Machines*. Wiley-Interscience, 2009.

- [147] L. S. Lippert, *Clinical Kinesiology and Anatomy*, 4th ed. F. A. Davis Company, 2006.
- [148] D. N. Shier, J. L. Butler, and R. Lewis, *Hole's Human Anatomy and Physiology*, 11th ed. McGraw Hill, 2007.
- [149] "American spinal injury association," [Online]. Available: <http://www.asia-spinalinjury.org/>. Accessed on July, 2006
- [150] "NHG Research & Development Office " [Online]. Available: <http://www.research.nhg.com.sg>. Accessed on October, 2010