DEVELOPMENT AND EVALUATION OF A NOVEL OVER-GROUND WALKING DEVICE: A ROBOTIC WALKER

MUN KYUNG-RYOUL

(MS. In Biomedical Eng., Konkuk University)

A THESIS SUBMITTED

FOR THE DEGREE OF DOCTOR OF PHILOSOPHY

DEPARTMENT OF BIOMEDICAL ENGINEERING

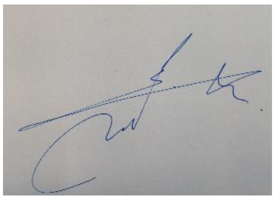
NATIONAL UNIVERSITY OF SINGAPORE

2015

DECLARATION

I hereby declare that this thesis is my original work and it has been written by me in its entirety. I have duly acknowledged all the sources of information which have been used in the thesis.

This thesis has also not been submitted for any degree in any university previously

MUN KYUNG-RYOUL		2015.08.14
Name of Student	Signature of Student	Date

Name: Mun Kyung-Ryoul

Degree: Ph.D.

Dept: Department of Biomedical Engineering

Thesis Title: Development and Evaluation of a Novel Pelvic Motion Supported Over-ground Walking Device: A Robotic walker

Abstract

This study presents a novel robotic over-ground walking device (the robotic walker or the walker) with pelvic motion support that is developed for gait rehabilitation of neurologically challenged patients. The walker is comprised of an omni-directional mobile platform, an intuitive human-machine interface, pelvic and trunk motion support brace, and an active body weight support (BWS) unit. This walker not only supports six degrees of freedom (DoFs) of pelvic motion, but also provides natural and realistic gait patterns. In addition, various functions such as BWS, assistance, and resistance trainings were systematically implemented into the walker for effective and successful gait rehabilitation to improve gait performance in people with neurological disorders. The detailed insight of design description and evaluation of the various functions of the walker such as pelvic motion facilitation, body weight support, and resistance training are dealt with in this study. A clinical test with neurologically challenged patients will be conducted in the next stage of our research.

Keyword: Robotic gait rehabilitation, Over-ground walking, Pelvic motion support, Body weight support, Assistance and resistance gait training

ACKNOWLEDGEMENTS

First of all, I would like to express my sincere gratitude to my supervisor Dr. Yu Haoyong for the continuous support of my Ph.D study and related research, for his patience, motivation, and immense knowledge. His guidance helped me in all the time of research, and my research could not be accomplished without his great supervision and patience. Besides my supervisor, I also would like to extend my heartfelt gratitude to Dr. Sangho Kim for his valuable advise whenever I fell into deep depress and distress.

I am deeply indebted to my colleagues, Dr. Guo Zhao, Li Cheng, Chen Gong, Francisco, Page Solenne, Subin Lim, and Brandon for all their help. The cherishing memories we have had together are never forgettable and I will remember the memories through all of my life time.

I will forever be thankful to Prof. Baeg GyeongHun, who is my badminton partner, flatmate, and great mentor of mine as a researcher, for his help, advice, and encouragement. My sincere thanks also goes to my friends Jaehyun, Jeongbum, bumseok, Jaesung, DK Ha, Inyong, Hanjoon, Sanghoon, Seungmin, and Seunghyun. I also need to extend my deepest appreciation to Prof. GaeRye Tack, Prof. SoonChul Chung, and Prof. BongSu Lee for their great guidance during the time when I was in Konkuk University for my Master degree.

Last but not least, special thanks to my family and my life partner Jihye. Words cannot express how grateful I am for all of the sacrifices that they have made on my behalf. Without their sacrifice and unwavering love, it was definitely not possible to accomplish my research. They has been a continuous source of love and encouragement during the 4 years of research.

CONTENTS

SUMMARY

As an alternative to traditional gait training, robot-assisted devices for gait rehabilitation have become more popular. Nevertheless, the robot-based therapy demonstrates a lack of clear predominance over a manual therapist-assisted training, and it consequently has led to some grade of dissatisfaction and uncertainty. It is clear that the mechanical design and control of the robotic devices influence the effectiveness of gait rehabilitation and gait dynamics in various ways. Therefore, this dissertation aims to develop a novel robotic walker with over-ground walking capability, and to evaluate its biomechanical effects in terms of kinematics, gait descriptive parameters, and muscle activation patterns.

Firstly, gait kinematic differences of primary (ankle, knee, and hip) and secondary (pelvis) joints as well as their associations in gait performance between normal individuals and stroke survivors are investigated. It is shown that the range of motions (RoMs) of the primary joint motions are significantly reduced while the secondary joint motions are significantly increased in stroke group. Additionally, it is noticed that, for healthy subjects, primary joint kinematics is the main factor, but secondary joint motion is the major factor in stroke patients to ensure gait performance. Therefore, this study suggests that there is a strong need to support pelvic motions during gait rehabilitation in order to achieve better rehabilitation outcomes.

Secondly, the conceptualized design and prototype of a novel robotic walker is developed and described using an omni-directional mobile platform, a pelvic and trunk motion support brace with an active body weight support (BWS) unit,

and an intuitive human-machine interface with force/torque (FT) sensor. With this robotic walker, over-ground walking with six DoFs pelvic motion facilitation is achieved in an intuitive and natural way. The results show that gait with the walker strongly resembles free over-ground walking without alteration of normal gait dynamics, indicating that pelvic motion facilitation with the walker can elicit correct afferent sensory input and provide better functional outcomes following gait rehabilitation.

Finally, the detailed insight of biomechanical effects on pelvic motion facilitation, BWS, and resistance force in backward direction generated by the robotic walker are investigated in terms of joint kinematics and muscle activation patterns to indicate gait differences caused by the respective experimental protocols. We concluded that 1) the pelvic restriction significantly alters normal gait dynamics, thus inhibiting the efficacy of gait rehabilitation; 2) the BWS training with the walker provides an important indication of reduced step-to-step transition (SST) cost and energy expenditure, and increased lateral body balance with greater stabilization during gait; 3) the resistance function can cause a larger number of motor unit activations (increased amplitude) with lower firing rates (decreased frequency) indicating that this type of resistance training can improve the muscular strength and endurance in a task-specific manner, especially for knee and hip joint motions. This research will further stretch a clinical test with neurologically challenged patients to determine the effects of various rehabilitation protocols such as assistance, resistance, and BWS, in the next stage of our research.

LIST OF FIGURES

LIST OF TABLES

CHAPTER I. INTRODUCTION AND BACKGROUND

1. Neurological Disorder and Stroke

A neurological disorder is defined as a disease of the central and peripheral nervous systems. Structural, biochemical, or electrical abnormalities in the brain, spinal cord, or other nerves can result in a range of symptoms including paralysis, muscle weakness, poor coordination, loss of sensation, seizures, confusion, pain, and altered levels of consciousness [1, 2]. Among the varieties of neurological disorders, stroke is a leading cause of disability, and in the United States (US), an estimated 6.6 million people over 20 years of age have had a stroke [3]. It is also the third most frequent cause of death worldwide and the leading cause of permanent disability in the US and Europe [3-5].

Stroke is caused by the interruption of the blood supply to the brain, generally due to a burst blood vessel or lodged clot [2]. Neurological impairment followed by stroke frequently leads to hemiparesis or partial paralysis of one side of the body which affects the ability to perform activities of daily living (ADL) such as walking, speaking, and eating.

As the most basic form of human locomotion, gait comprises an intricate network between the neurophysiological network and the musculoskeletal system. The control of this network involves a constant communication between the efferent signals from the central command and afferent signals from sensory feedback [6, 7]. Any damage to the communication network is the reason that many stroke patients lose their basic ability to walk [8-10]. Abnormal gait patterns can be characterized by significantly reduced gait speed, shortened step length, an inability to maintain balance, and gait asymmetry; these are observed in the majority of individuals with post-stroke hemiplegia [11, 12]. The abnormalities are correlated with other parameters such as balance, use of walking aids, number of falls, and ability to perform activities of daily living. In addition, this loss of mobility greatly affects patients as they lose their sense of independence and this adversely impacts their social lifestyle [13]. Therefore, the main purpose of gait therapy lies in improving and restoring gait patterns through a proper gait rehabilitation regimen that include balance training, weight bearing exercises, strength training, and the use of electrical stimulation [14]. The restoration of gait is for not only improving the ability to walk, but also for restoring the quality of walking, thus improving their overall quality of life.

2. Gait Rehabilitation for Stroke Survivors

Despite the defective motor functions caused by stroke, brain has an interesting characteristic called brain plasticity. Brain plasticity or neuro-plasticity refers to alterations in neural pathways and synapses which are due to changes in behavior, environment, and neural processes. The neural pathway can be reconnected and reorganized by repetitive and persistent stimulation. Based on this concept, the impaired motor function can be restored [15-17], hence, there is a strong need for therapeutic interventions that can reduce the long-term need for physical assistance.

2.1. Conventional gait therapy

The conventional therapy usually includes active joint mobilization and conventional gait and balance training. The participants perform active joint mobilization of the lower limbs, including ankle, knee, and hip joint in the supine position. Then, conventional gait and balance therapy are performed with the instruction of therapists. The therapist usually facilitates pelvic motion to improve control and mobility of the patients. Finally, the patients undergo gait training with/without aids or orthoses and with manual assistance from the physical therapist, depending on the individual subjects abilities [18]. In addition to certain routines, it can be varied according to therapist’s experience and decision.

It has been shown that chronic, non-ambulatory hemiparetic patients improved their gait ability through the conventional rehabilitation process [19, 20]. Additionally, it has also been reported that the conventional treadmill therapy with body weight support was more effective than without it in subacute, non-ambulatory stroke patients [21]. However, conventional rehabilitation is limited in terms of availability, duration, and training session frequency due to excessive and exhaustive physical efforts of therapists in assisting the gait of severely affected subjects, setting the paretic limb, and controlling the trunk and pelvic movements. As therapists often have to lift up the body-weight of patients, or work in ergonomically unfavorable postures, there is the possibility that the therapist will suffer from back injury. This fact imposes an enormous burden on the health care system, hence limits its clinical acceptance. In addition, the quantification of gait performance is challenging because training sessions are largely subjective and depend on therapists’ individual experiences. Furthermore, the repetitive treatment for stroke patients may be very costly due to the increasing number of stroke patients as a result of extended life spans, increasing physiotherapy costs, and the limited number of therapists. In this regard, there is a strong need to reduce the excessive labor of therapists and to increase the quality of the gait rehabilitation process.

2.2. Robotic gait rehabilitation

As an alternative to conventional gait rehabilitation, robot-assisted gait rehabilitation (RAGR) devices have gained popularity, and are expected to serve as a solution to automated training for gait rehabilitation. RAGR devices can replace or facilitate the physical training effort of a therapist, allowing more intensive and repetitive motions, delivering therapy at a reasonable cost, and quantitatively assessing the motor recovery level by measuring gait kinematic and kinetic patterns [22]. Alessandro et al [18] conducted a study which investigated whether a rehabilitation program with RAGR is more effective than conventional one, and determined that the mean gait velocity improved more in the rehabilitation group with RAGR device, showing improved walking ability in patients with Parkinson disease. Werner et al. conclude that the robotic gait device was at least as effective as conventional treadmill therapy while requiring less input from therapist [23]. Nevertheless, robot-based therapy demonstrates a lack of clear predominance over the manual therapist-assisted training, and it consequently has led to some degree of dissatisfaction and uncertainty [11]. It is clear that the mechanical design and control of the robotic device influences the effectiveness of gait rehabilitation and gait dynamics in many different ways. Therefore, a careful consideration of requirements for successful RAGR devices, based on reviews of currently available robotic gait devices, should be conducted. The next section will discuss the reviews of currently available RAGR devices to adequately account for important parameters for developing effective RAGR devices.

3. Reviews on the Types of Robotic Gait Rehabilitation Devices

There are several types of robotic systems for gait rehabilitation. These systems can be grouped according to the rehabilitation principle they follow; i) treadmill-based devices, ii) foot-plate-based devices, iii) stationary and ankle rehabilitation systems, iv) over-ground gait devices [24]. The review of type i), ii), and iv) is discussed in the sub-sections below.

3.1. Treadmill-based devices

Treadmill-based devices are the most prevalent and clinically well evaluated robotic rehabilitation methods. Current treadmill-based devices, such as Lokomat (Fig. I-1) [25-27], LOPES (Fig. I-2) [28-31], ALEX (Fig. I-3) [32-34] and LokoHelp [35], are built on a treadmill-based walking platform with active or passive exoskeleton to support movements of lower limbs in combination with overhead harness body weight support (BWS) unit. However, the use of treadmills in gait rehabilitation is still widely controversial as there is ambiguity in the assumption that walking on a treadmill could represent an actual over-ground gait in terms of sensorimotor feedback and proprioceptive input. Additionally, it has been shown that walking on a treadmill gives an indication of greater cadence, smaller stride length and stride time, as well as reduced joint angles, powers, and pelvic rotation excursion compared to over-ground walking [36, 37]. Furthermore, these systems provide only a single anatomical plane of movement (sagittal) with a pre-determined path resulting in constraints to the patients along a fixed platform. This constrained and pre-determined gait may lead to less satisfactory functional outcomes and lack of cycle-to-cycle variation which reduce sensory responses, and may ultimately impair the motor learning process [38, 39].

Moreover, the restriction of pelvis caused by robotic devices are a great concern in the field of RAGR. Overhead harness BWS-systems often restrict pelvic rotation and lateral movements which contribute to the optimization of energy consumption, and also affect the aesthetics of the walking pattern [11]. The fixation of the pelvic lateral and rotational motions greatly affects gait dynamics by shortening step width and reducing the coronal trunk rotation, while increasing the step length and sagittal trunk rotation [40]. Therefore, when designing a RAGR device, consideration should be given to the pelvic movements to obtain a more realistic and aesthetic locomotion upon post-rehabilitation.

Figure I-1. Lokomat

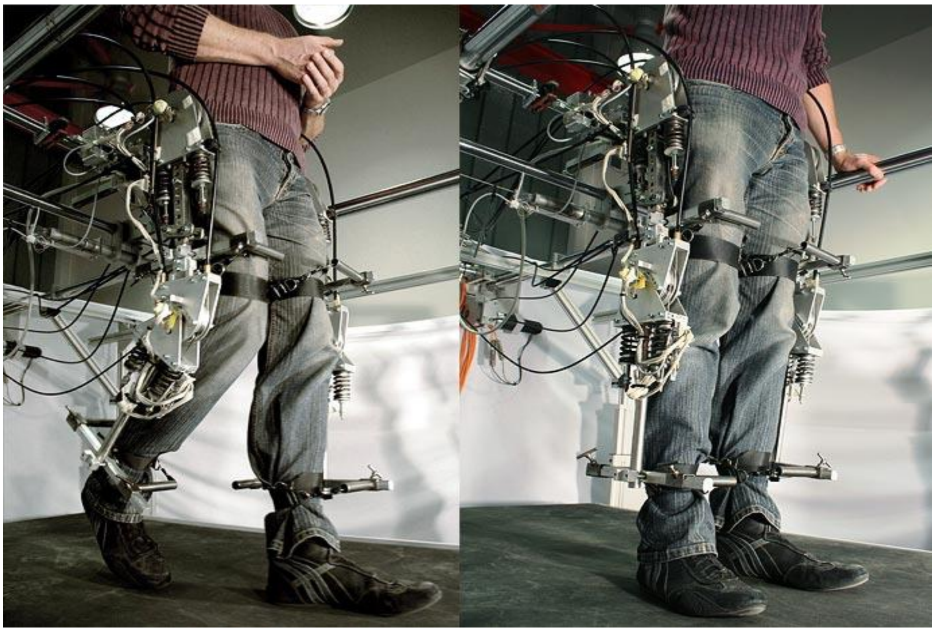


Figure I-2. LOPEZ

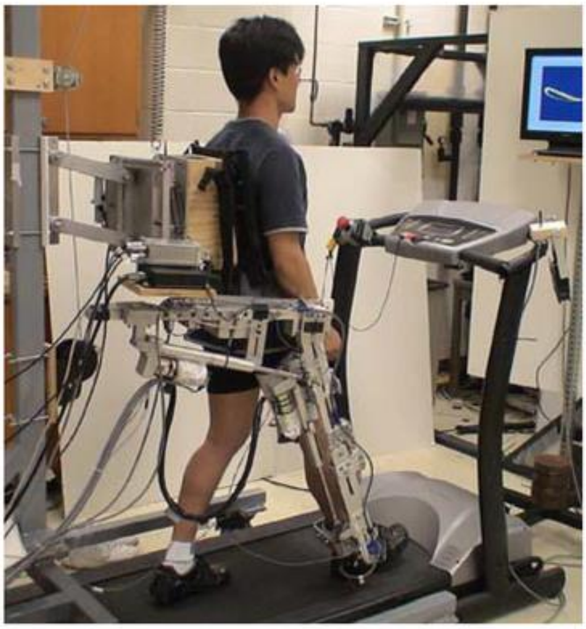


Figure I-3. ALEX powered leg orthosis

3.2. Foot-plate-based gait trainer

The foot-plate-based gait trainer is considered as one of the pioneering robotic systems for gait rehabilitation. The GT I (Reha-Stim) and GaitMaster5 (GM5) are commercialized devices for the foot-plate-based gait trainer [41]. Similar to treadmill gait trainers, the GT I is at least as effective as manual treadmill therapy but requires less input from the therapist [23]. However, this type of device provides different movement from actual gait patterns; therefore, the functional outcomes after intervention could be limited due to lack of appropriate afferent sensory input. As a result, the gait performance improved as much as conventional gait rehabilitation provided, showing no significant superiority.

Figure I-4. GT I foot-plate based gait trainer

3.3. Over-ground gait platform with pelvic motion support

From the perspective of effective and successful gait rehabilitation, the facilitation of pelvic lateral and rotational movements have been accentuated with an over-ground walking platform to provide better functional outcome following gait training. Several over-ground rehabilitation devices with pelvic motion support, such as KineAssist (Kinea Design LLC) (Fig. I-5) [42], WalkTrainer (Swortec SA) (Fig. I-6) [43], MLLRE device [44], NaTUre-gait (Univ. Nanyang Tech) (Fig. I-7) [45], and a BWS system with a pelvic holding mechanism [46], have been developed recently.

These devices are mobile gait training robotic systems with actuated trunk and pelvic support mechanisms, which allow users to control the movement of the platform, rather than depending on the device to move patients' lower limbs through a predetermined movement path [47]. In addition, these devices were designed to enable realistic walking patterns with proper sensory input from the ground and to increase active patient participation. However, the pelvic lateral and rotational movements are passively implemented in Kineassist, while the trunk motions such as lateral and forward bending were actively supported. In consequence, the pelvic motion cannot be actively supported by Kineassist. The WalkTrainer and NaTUre-gait were designed to actively support six DoFs of pelvic motion. However, many additional actuators to accommodate pelvic motions were required. As a result, the mechanical structures and control of pelvic motions are highly complicated, thus may not be sufficient to provide correct afferent sensory input, task-specific motor training, and higher participation and intensity of practice due to system complexity [11]. On the other hand, DoFs for pelvic motion might be insufficient in the RGR trainer (Fig. I-7) [47] and a robotic device designed by Watanabe [46]. These devices are mainly designed to support one DoF of pelvic motion, which is pelvic obliquity or lateral displacement, respectively. The limited DoFs may not be sufficiently applicable for successful and comprehensive gait rehabilitation.

Figure I-5. Kineassist

Figure I-6. WalkTrainer

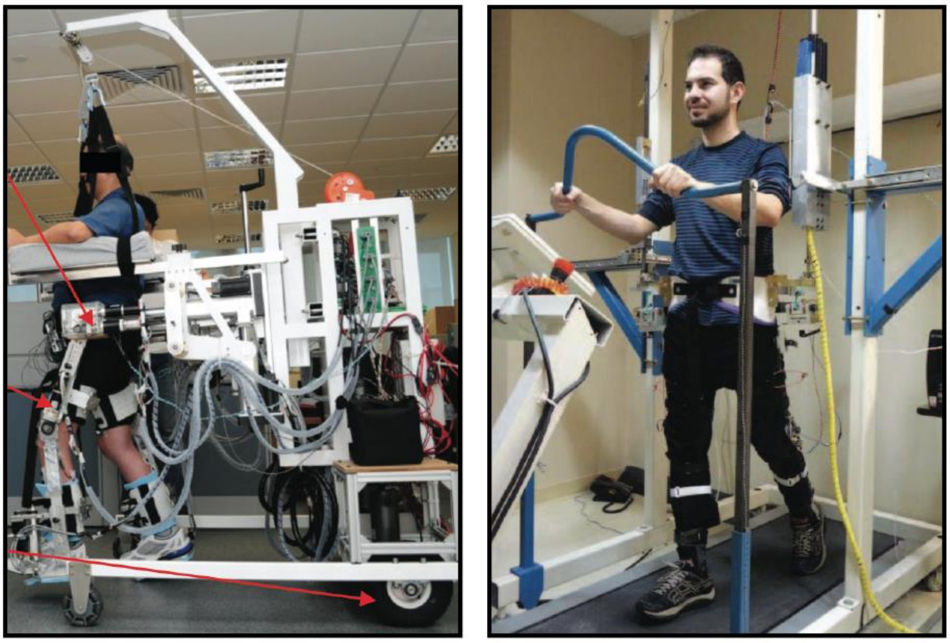


Figure I-7. NaTUre-gait and RGR trainer

In summary, as there are many limitations in the currently available RAGR devices mentioned above, there is strong need to develop a more effective robotic system for successful gait rehabilitation that can benefit the broad spectrum of patients. As reviewed, the design considerations of a novel robotic gait rehabilitation should be focused on the facts that; 1) the over-ground walking platform is required for the proper sensory input and feedback to elicit the appropriate sensory feedback; 2) the device should provide multi-plane movements comprising forward-backward, lateral, and rotational mobility without any restriction; 3) The six DoFs pelvic motions should be supported during gait; 4) The desired gait training system should not only be able to relieve the therapists from mechanical work, but also be able to sense joint position and/or muscle activation in relation to the gait cycle and provide assistance only as needed during defined periods of the gait cycle; 5) additionally, given the fact that the robotic device always interacts with and is controlled by the patients, the biomechanical effectiveness of the device should be evaluated through analysis of the gait patterns during body weight support and pelvic motion support situations.

4. Requirements for a Novel Robotic Gait Rehabilitation Device

The requirements for effective and successful gait rehabilitation can be extracted from the literature review performed in section 3. Figure I-8 depicts a summary of the requirements for the novel robotic gait rehabilitation device.

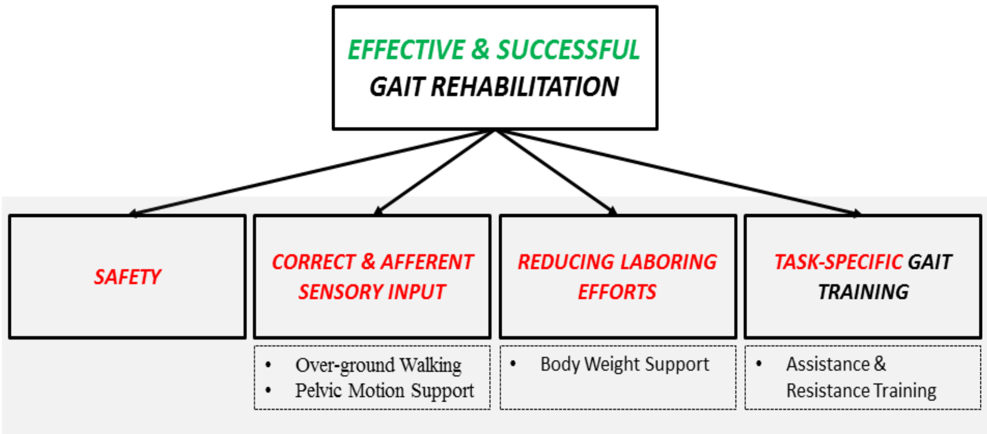


Figure I-8. Design considerations for effective and successful gait rehabilitation

1) Safety: Safety is of utmost importance in robotic gait rehabilitation for neurologically challenged patients. For example, falling is the most serious problem for the user and the fear of falling can increase anxiety in patients as well as decrease intervention effectiveness (Appendix A).

2) Correct afferent sensory input: A lack of sensory input acts a barrier to motor learning and reorganization. The normal sensorimotor input and feedback with the proportionately scaled motor responses are required for effective and successful gait rehabilitation [48]. Among the several types of gait rehabilitation devices, the over-ground gait assistive devices allow users to move the platform under their own control rather than depending on the device to move the patients

through a predetermined movement path. Thus, it may elicit the proper and appropriate sensory motor input and feedback, which is critical for successful gait rehabilitation. In addition, six DoFs pelvic motion facilitation will provide a natural and aesthetic gait pattern for neurologically challenged patients.

3) Reducing laboring efforts of therapists: Conventional gait rehabilitation methods are limited in terms of availability, duration, and frequency of training sessions due to the excessive and exhaustive physical effort of therapists because of the need to physically move the patient’s limbs, or even to support their body weight [11, 49]. The robotic gait rehabilitation device should facilitate the laboring efforts of the therapists and provide correct and afferent sensory input with simultaneously reduced muscle activation.

4) Task-specific gait training: To improve gait functionality and muscle strength, assistance and resistance training have been recommended and widely adopted in gait training with positive results [50]. Especially, strength training has been shown to improve neural adaptations such as motor unit activation and synchronization, thus leading to higher muscular strength and control [51]. A recent review indicated that many studies that conduct strength training on neurological patients do not show positive outcomes with gait outcomes due to the lack of task-specificity [52]. In this regard, it is important that the robotic gait rehabilitation device have a muscle strengthening function combined with task specific training.

5. Thesis Objectives and Structure

Based on the above review of the currently available RAGR devices, it becomes obvious that most of the devices may have both beneficial and unsatisfactory effects on gait performance. Additionally, despite the purported benefit of the robotic devices, the efficiency of the devices has not been proven. Thus, there is a strong need to develop a more effective neuro-rehabilitation robotic system that can benefit the broad spectrum of patients. Given the fact that the robotic device always interacts with the patient, the biomechanical effectiveness of the device is evaluated through analysis of user gait patterns in terms of kinematics, gait descriptive parameters, and muscle activation in various practices with the developed device.

Therefore, this dissertation aims to develop a novel robotic over-ground walking device for neurologically challenged patients, which will provide a more natural and realistic gait pattern, and to provide the detailed insight into the biomechanical effects of the developed device.

The study is divided into VII Chapters. After an abstract, Chapter I provides the introduction and background of this study. Chapter II explains gait abnormalities of the stroke survivors and essential considerations for their gait rehabilitation. In Chapter III, the design description, control, and functions of the developed device are described. Chapter IV, V, and VI highlight the biomechanical effects of the developed device in terms of pelvic motion facilitation, body weight support (BWS), and task-specific resistance training, respectively. Finally, the conclusions and recommendations for future works are discussed in Chapter VII.

CHAPTER II. PRIMARY AND SECONDARY GAIT DEVIATIONS OF STROKE SURVIVORS AND THEIR ASSOCIATION WITH GAIT PERFORMANCES

1. Introduction

Damaged descending neural pathways can cause abnormal movements including abnormal gait pattern after stroke [15]. A stroke consequently alters kinematic, kinetic, and muscle activation patterns of survivors, and it can frequently cause spasticity at a certain joint. Antagonist groups with weakened muscle and spasticity tend to suffer more from passive stiffness and joint contracture [53-55]. Although specific figures vary between studies, it is reported that 38-60% of stroke survivors suffer from muscle weakness and spasticity, and stroke survivors with spasticity are more prone to be functionally impaired than those without spasticity [56].

Particularly, weakened and spastic muscles at the knee joint interfere with voluntary knee movements. Various symptoms can be developed into such as crouch gait, stiff-knee gait (SKG), genu recurvatum with altered excursion and reduced range of motion (RoM) of lower limb joints due to the pathology [57-61]. Crouch gait is defined as an excessive knee flexion in the initial stage of gait phase derived from weak hip extensors, knee extensors, or ankle plantar flexors [59, 62, 63]. The SKG is defined as significant diminished and delayed knee flexion during swing-phase and is mainly caused by spasticity of the rectus femoris muscle, weakness in the hip flexor muscle, and over-activity of the ankle plantar flexor leading to inefficient gait pattern [61, 64, 65]. Genu recurvatum, which is defined as knee hyperextension during stance phase, is caused by quadriceps malfunction [58, 60]. All these symptoms substantially increase the energy expenditure during walking and can eventually lead to excessive pain and chronic joint degeneration if not corrected.

With dysfunction of voluntary knee movements, compensatory pelvic motions have been employed to avoid foot drop to compensate the effects of gravity acting on the body, during swing and stance phase [66]. The pelvic movements during gait have begun to receive attention with knee and foot mechanisms ever since Saunders [67] proposed the concept of the six determinants of gait, which are the six factors responsible for minimizing displacement of center of gravity [68-70]. Many studies have shown that stroke survivors have not only increased anterior pelvic tilt but also have contralateral pelvic drop in the coronal plane. Furthermore, the affected side of pelvis is retracted in transversal plane [71-74]. In addition, Karen and his colleagues reported that the pelvic lateral displacement in patients with acute hemiparetic stroke was significantly increased to keep the body balanced during walking [70].

Stroke survivors affected by voluntary knee movement dysfunction with abnormal pelvic motion have shown remarkably deteriorated gait performances characterized by significantly reduced gait velocity, step and stride length in their gait patterns [65]. The difference in gait velocity, step and stride length identifies stroke survivor gait performances, which is the major goal of the gait rehabilitation [75, 76], so it is this identification that enables therapists to plan and provide proper guidance for gait rehabilitation. Therefore, identifying the mechanisms to achieve the gait performance such as gait velocity and step or stride length, could be crucial to provide better and quantitative clinical interventions.

However, to our best knowledge, little work has been done to identify the underlying mechanisms that affect the gait performances of stroke survivors with voluntary knee movement dysfunction in the context of their primary (ankle, knee, and hip joint) and secondary (pelvis) deviations. Therefore, the aim of this study is to identify the kinematic differences of primary and secondary joint between normal individuals and stroke survivors. Once these differences are ascertained, further efforts are then made to identify their associations with gait performances in stroke survivors. The hypothesis of this study is that there would be increased pelvic motion with decreased ankle, knee, and hip joint RoMs in stroke survivors compared to controls. Additionally, the increased pelvic motion would be associated with the gait performances in stroke survivors. It is hoped that these findings can serve as a guideline for designing improved clinical interventions, which aim to rectify any dysfunction of voluntary knee movements and provide an indication of rehabilitation-oriented gait pattern improvements.

2. Methods

2.1. Participants

This study conducted on five healthy young males (age: 29 ± 2.88 years, height: 1714 ± 50.66 mm, and weight: 66.6 ± 5.77 kg) and five stroke patients (age: 61.2 ± 9.98 years, height: 1616 ± 50mm, and weight: 64.94 ± 7.28kg) who were once admitted to National University Hospital in Singapore. The five participants were community-dwelling stroke survivors chosen from those who had experienced ambulatory hemiparesis for longer than two years caused by either right or left supratentorial ischemic stroke and intracerebral hemorrhage. All survivors showed impaired gait. Participants with movement disorders or orthopedic diseases that could influence their gait such as arthrosis or total hip joint replacement were excluded from the experiment. This study was conducted with prior consent of all subjects, and no human rights of subjects were violated throughout the study. This research work was approved by the NHG Domain Specific Review Board.

2.2. Experimental Design

With 16 reflective optical markers attached to the body, all subjects were instructed to walk at self-selected speed along a 10m walkway in a gait lab. The markers were attached to anatomic landmarks located on pelvis and lower extremities in accordance with the Plug-in-gait marker set. During the experiments, subjects were not allowed to wear an orthosis nor were they provided with any weight support. Eight high speed optical cameras (Vicon, Oxford, UK) captured the 3D positions of the reflective markers with the sampling rate of 100Hz, and the gait kinematics were calculated based on the positions of each marker. Subjects were asked to repeat the trial, as long as they were not tired, until five successful trials were achieved.

2.3. Data Processing and Analysis

Customized software provided by Vicon motion capture system (Nexus, Oxford, UK) was used for pre-processing of the raw kinematic data. Marker data were low-pass filtered using a zero-lag fourth order Butterworth filter with cut-off frequency of 6Hz. All gait related parameters were grouped into three categories: control for healthy subjects, unaffected and affected limb for the stroke survivors. Gait phases such as heel strike, toe off time, as well as the kinematic profiles of ankle, knee and hip flexion/extension angles, and pelvic tilt, obliquity, and rotation were extracted from the software. For further analysis, the gait phase, gait kinematic parameters, and three-dimensional markers data were mounted into MATLAB (MathWorks, Natick, MA). Range of motions (RoMs) of the kinematic profiles as well as step length, stride length, gait velocity, and lateral displacement of the pelvis were computed by the customized Matlab program.

Since each subject had different RoMs, which varied kinematic profiles, the kinematic profiles were normalized with the corresponding RoMs of each profile for clearer comparison of the kinematic profiles and to eliminate the effects of different RoMs caused by severity of the pathology.

2.4. Statistical Analysis

SPSS program (SPSS Inc., Chicago, IL) was used to analyze the data. The differences of kinematic profiles between groups were compared by using the Pearson product-moment correlation. One-way ANOVA test was used to investigate between group differences for the RoM of the kinematic profiles and gait performances. In case that any statistically significant results were detected through the one-way ANOVA, Tukey's post hoc test was performed to compare the differences. All significance levels were set as p=0.05. To identify the association between gait performance and other gait kinematic parameters, the Pearson product-moment correlation method was used, i.e. step length, stride length, and gait velocity were compared with RoM of kinematic profiles. Correlation criterion r value suggests that .9 to 1 is very high, .7 to .9 is high, .5 to .7 moderate, .3 to .5 low, and .0 to .3 is little if any correlation.

3. Results

The stroke survivors were divided into four groups according to the symptoms of knee voluntary movement dysfunction. Two were identified as a combination of crouch gait with stiff-knee gait, one was identified as crouch gait, one was stiff-knee gait, and one had genu recurvatum.

3.1. Gait differences in kinematics and gait performance between groups

The kinematic profiles are shown in Figure II-1, and the RoMs of the kinematic parameters and gait performance are presented in Table 1. Figure II-1, a), b), and c) shows the normalized ankle, knee, and hip (AKH) profiles, respectively, which are the primary joint motions. The stroke survivors' AKH profile shows similar patterns with controls but the RoMs of AKH joints are significantly reduced. The correlation coefficient of the kinematic profiles between controls and unaffected limb is 0.402 (p<0.001) for ankle, 0.650 (p<0.001) for knee, and 0.885 (p<0.001) for hip. The correlation coefficient between controls and affected limb is slightly higher, 0.678 (p<0.001) for ankle, 0.8955 (p<0.001) for knee, and 0.9660 (p<0.001) for the hip. However, the RoMs of the AKH profiles are significantly reduced at unaffected and affected limbs compared with controls (Table 1).

Figure II-1, d) shows the normalized pelvic tilt profiles. The unaffected side pelvic motion (USPM) shows excessively anterior tilted pattern at the initial contact (IC) and the anterior tilt is decreased during the mid-stance (MST), while the affected side pelvic motion (ASPM) shows relatively reduced anterior tilt at IC but it reached maximum anterior tilt at the mid swing (MS). There is no correlation between control and USPM (r=-0.0241, p=0.4465), and between control and ASPM (r= -0.0835, p=0.008). The USPM is highly but negatively correlated with ASPM (r= -0.7663, p<0.01). There is a significant difference in RoM of pelvic tilt among the groups (p<0.01). The RoMs of the pelvic tilt of unaffected and affected side are significantly increased compared to the controls by 5.54 and 6.04 times at the USPM and ASPM, respectively.

The profile of the normalized pelvic obliquity is shown in Figure II-1, e). The USPM drops at the initial contact with further reduction at the loading response and terminal stance, and is elevated during swing phase. On the other hand, the ASPM rises at IC, but is more elevated during mid-swing (MS), which is a hip hiking pattern to counter foot drop [73]. There is mild correlation between controls and USPM (r=0.4502, p<0.01) and no correlation between controls and ASPM (r=0.2743, p<0.01). No significant difference is found in terms of RoM of pelvic obliquity between control and stroke group.

Figure II-1, f) illustrates the normalized pelvic rotation profile. The USPM is retracted at the IC. However, the profiles are moderately or highly correlated with each other; correlation between controls and USPM is 0.4266 ($p<0.01$), and 0.666 (p= $p<0.01$) between controls and ASPM, and 0.9464 ($p<0.01$) between USPM and ASPM. However, the RoM of pelvic rotation is significantly increased on both side of limbs for stroke survivors.

Other important gait parameters are shown in the Table 1. The lateral displacement of pelvis, which is the indicator of postural performance during gait, is significantly increased in the stroke patients ($p<0.01$). Additionally, the gait performances such as normalized step length, stride length, and gait velocity are significantly reduced for both unaffected and affected limb of stroke group ($p<0.01$).

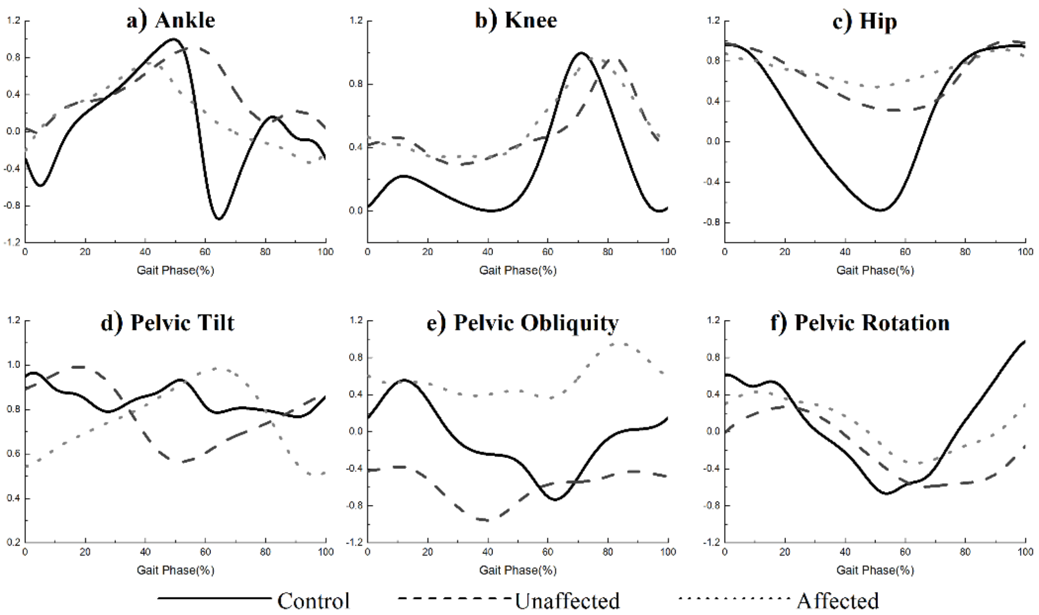


Figure II-1. Profiles of ankle, knee, and hip joint angles for normal individuals and stroke survivors.

Table 1. The Range of Motions of Primary and Secondary Joints and the Important Gait Parameters.

	Controls	Stroke Patients	
		Unaffected	Affected
RoM of Ankle (◦)	30.41 ± 6.45	26.08 ± 12.34*	20.51 ± 4.09**
RoM of Knee (◦)	55.56 ± 5.55	41.25 ± 16.13*	21.27 ± 11.64**
RoM of Hip (◦)	41.78 ± 1.86	35.09 ± 14.40	19.80 ± 12.95**
RoM of Pelvic Tilt (◦)	2.54 ± 0.57	12.12 ± 7.70*	14.32 ± 8.80**
RoM of Pelvic Obliquity (◦)	7.56 ± 2.58	7.65 ± 1.87	8.66 ± 2.76
RoM of Pelvic Rotation (◦)	11.71 ± 2.32	14.44 ± 11.87*	17.42 ± 13.97**
Lateral Displacement (LD, mm)	67.30 ± 19.34	107.42 ± 27.24*	123.17 ± 27.05**,***
Normalized Step Length	0.71 ± 0.04	0.32 ± 0.23	0.39 ± 0.15
Normalized Stride Length	1.35 ± 0.07	0.67 ± 0.33*	0.71 ± 0.35**
Velocity (m/s)	1.03 ± 0.15	0.34 ± 0.22*	0.33 ± 0.20**
Stance phase duration (%)	60.00 ± 1.34	75.26 ± 5.85*	65.51 ± 11.19

* Statistical difference between control and unaffected, P<0.05

** Statistical difference between control and Affected, P<0.05

*** Statistical difference between unaffected and affected, P<0.05

3.2. Association between kinematic differences and gait performance

To evaluate the relationship between gait performances and other kinematic parameters, the correlations between RoMs of the kinematic parameters and normalized step length, stride length as well as gait velocity are calculated and presented in Table 2. As shown in Table 2, for the control group, the RoM of ankle is moderately correlated ($r=0.692$, $P<0.05$) and the RoM of the pelvic obliquity is negatively correlated with step length ($r=0.483$, $P<0.05$). In addition, the RoM of knee is slightly correlated with stride length while the hip ($r=0.593$, $P<0.01$) RoM is moderately correlated and the lateral displacement of the pelvis is negatively correlated ($r=-0.719$, $P<0.01$) with gait velocity for the controls.

In the case of unaffected limb, parameters including RoM of ankle, knee, hip, pelvic tilt and rotation are significantly and highly correlated with step length and stride length, but are moderately or highly related to gait velocity. For the affected side, RoM of knee, hip, pelvic tilt, and rotation are the main contributors of gait performance. Interestingly, the RoM of ankle joint of the affected limb does not contribute to gait performance in stroke survivors. The lateral displacement of the pelvis is negatively correlated with gait velocity for both sides.

In other words, the AKH flexion/extension RoMs are the main contributors for controls to ensure the gait outputs, while pelvic kinematic RoMs especially, pelvic tilt, and rotation combined with AKH joint motions are the main contributors to dysfunction of knee voluntary movement in stroke patients.

Table 2 The Correlation between Gait Performance such as Step, Stride length and Gait velocity and RoMs of other gait parameters

		Ankle	Knee	Hip	Pelvic Tilt	Pelvic Obliquity	Pelvic Rotation	LD
Controls	**Step Length**	**0.692****	0.273	0.236	-0.132	**-0.472***	0.434	-0.273
	Stride Length	0.443	**0.483***	0.222	-0.022	-0.170	0.341	-0.357
	Gait Velocity	0.366	-0.187	**0.593***	-0.240	-0.242	-0.095	**-0.719****
Unaffected	**Step Length**	**0.836****	**0.851****	**0.976****	**0.889****	0.196	**0.747****	-0.303
	Stride Length	**0.813****	**0.834****	**0.978****	**0.850****	0.215	**0.716****	-0.318
	Gait Velocity	**0.553****	**0.737****	**0.889****	**0.572****	0.342	**0.449****	**-0.479***
Affected	**Step Length**	-0.249	**0.855****	**0.682****	**0.753****	0.136	**0.786****	**-0.366***
	Stride Length	-0.256	**0.922****	**0.851****	**0.866****	-0.026	**0.867****	-0.215
	Gait Velocity	-0.148	**0.735****	**0.613****	**0.629****	-0.106	**0.646****	**-0.395***

* Statistically significant, P<0.05

** Statistically significant, P<0.01

4. Discussion

This study examined the differences in kinematic profiles and investigated the relationship between gait performances and kinematic parameters in stroke survivors with knee voluntary movement dysfunction. The results showed that the movement of primary joints such as ankle, knee, and hip are significantly reduced while the RoMs of secondary joint were significantly increased in the stroke survivors group. The results were consistent with previous studies which have concluded that stroke individuals have limited RoMs at primary joints and increased RoMs at secondary joints due to compensatory movements. However, there was less agreement or no consensus on the relationship between RoMs and gait performance as well as the contribution of RoMs to gait performance. Therefore, the main emphasis of this study is to identify the relationship between gait performance and the RoMs of kinematic profiles at primary and secondary joints to provide better guidance for gait rehabilitation.

4.1. Kinematic and RoMs differences between the controls and stroke survivors

The normalized kinematic profiles of primary deviation of stroke patients were moderately or highly correlated with control group but the RoMs were significantly reduced showing consensus with previous studies [57, 59-61, 77]. The reduced RoMs at primary joint can be explained by the fact that the crouch gait caused by weakness of triceps surae muscle makes the progression of the tibia during mid-stance excessive which leads to the excessive knee flexion during initial contact to mid-stance. In addition, the spasticity in the rectus femoris muscle, weakness in hip flexor, and over-activity of ankle plantar-flexor may have led to limited peak knee angle during mid-swing [65, 78].

Consequently, these malfunctions of lower limb muscles resulted in limited RoM of ankle, knee and hip flexion/extension movements in this study. With reduced RoMs, stroke survivors might actively modify gait pattern by exerting compensatory strategies in order to enable locomotion [66]. This is important because primary abnormalities have been shown to develop into secondary deviations and they are spontaneously related to each other.

Increased pelvic tilt angles have been observed in this study, which is consistent with a previous study that concluded that the majority of stroke patients increased their pelvic tilt angles 4.77 times for unaffected side and 5.63 times for affected side with excessively tilted anterior position [73]. As seen in Figure II-1 e), the unaffected side pelvis was slightly dropped at the initial contact, while affected side was ascended during mid-swing showing the hip hiking pattern as an effort to avoid foot drop during mid-swing [74]. There were no statistical differences in RoM of pelvic obliquity among the groups. In contrast, the pelvic rotation profiles of the stroke survivors were highly correlated with the controls while the RoMs of pelvic rotation were increased in the group of stroke (1.2 times at unaffected side, 1.48 times at affected side). The increased pelvic rotation in the stroke group can be attributed to a compensatory increased gait performance.

The gait performances of the stroke individuals are characterized by reduced gait velocity and step length. Restoring gait velocity and step length has been considered as a main goal of the gait rehabilitation. The results of this study show that the average gait velocity of the stroke survivors is slower (0.34m/s for unaffected limb, and 0.33m/s for affected limb) than the controls (1.03m/s). The normalized step length (0.32m for the unaffected limb, and 0.39m for the affected limb) and stride length (0.67m for the unaffected limb, and 0.71m for the affected limb) are also significantly reduced compared to the control group (0.71m for the step length, and 1.35 for the stride length).

4.2. Associations of gait kinematics to the gait performances

Although gait performance deterioration varies with the severity of pathological conditions, the association of gait kinematics to the gait performance (velocity, step and stride length) of stroke individuals remain unclear. Hsu investigated the most important parameter determining gait velocity in patients with mild to moderate stroke individual using regression analyses, and determined that the hip flexors and knee extensors were the most important determinants of comfortable and fast gait velocities [79]. Kim also examined if the magnitude or pattern of kinematic and kinetic gait parameters related to preferred gait velocity, and showed that the RoM of hip and knee moderately correlated with gait velocity for paretic limb and the RoM of ankle is correlated with unaffected limb [75]. Cruz emphasized the paretic hip extension strength as a contributor to gait velocity [76]. However, conclusions from previous studies have not been reached by comparing data of the control group who are neurologically healthy individuals, neither were the contributions of the pelvic motion to gait performance investigated. As mentioned above, the secondary deviation can be developed from the primary deviation, thus it is important to investigate the association between pelvic motion and gait performances. Our results show that, for control group, the normalized step length, stride length, and gait velocity of the controls are significant but moderately correlated with ankle, knee, and hip RoMs, respectively. The pelvic obliquity is negatively correlated with step length, while lateral displacement negatively contributed to the gait velocity. On the contrary, for the unaffected limb, all the gait parameters except for the pelvic obliquity and lateral displacement are the main contributors to gait performance. For the affected limb, the RoM of knee, hip, pelvic tilt, and rotation are the significant contributors to gait performances.

The pelvic motion, generally, plays a central role in locomotion contributing to the forward progression of the body and trunk vertical support. In addition, the vertical displacement of the human center of mass is partly reduced by the motion of the pelvis [67]. Thus, the pelvic motions may have not contributed to the gait performance for the control group. The lateral displacement of pelvis in controls was highly and negatively correlated with gait velocity because it is related to stability and balance rather than to the forward velocity during gait. In contrast to the results of the controls, it was shown that the gait performances of the unaffected limb of stroke group is accomplished by increasing the magnitude of pelvic tilt and pelvic rotation with the abnormal excursion of pelvic obliquity in order to compensate the reduced RoM of ankle, knee, and hip joints. Therefore, the pelvic movements are not only the compensatory motion to clear the foot during mid-swing caused by lack of ankle dorsiflexion and knee and hip flexion, but also the factors to achieve gait performances such as step length, stride length, or gait velocity showing the deviation from original roles of pelvic motion. Our results show that the pelvic obliquity is not the contributor, and it may only play a foot clearance role by adjusting hip hiking pattern as can be seen in Figure II-1 e). The gait performance achieved by the affected limb is similar to that of the unaffected limb but the ankle RoM is not correlated with gait performance for the affected side, showing consensus with previous studies [75]. As Kim and Eng claimed that the ankle RoM is the key kinematics factor in gait efficiency in adolescents with cerebral palsy, our results also show that ankle RoM does not contribute to any of the gait performance parameters proving the necessity of ankle and knee muscle strength training to improve lower limb RoM and gait performance. In addition, it can be suggested that gait rehabilitation should be applied to encourage the use of the ankle, knee, and hip joint movements with constrained pelvic range of motions to reduce the influence of the secondary deviation in the gait performances.

There are several limitations in this study. First, only five subjects participated, so the results from this study need to be confirmed with larger samples. However, in an effort to increase the reliability, we reduced variability by recruiting only stroke individuals with knee voluntary movement dysfunction and by increasing the number of gait trials for each subject. Second, the control group is not matched in terms of age and gait velocity because excessively slow gait speed makes the gait of controls exaggerated. Furthermore, we only evaluate the kinematic differences between two groups. Given the fact that the stroke individuals may use different mechanisms to achieve similar kinematic movement, kinematic and joint power information should be investigated in the future. Despite the limitations, we believe that this study emphasizes the importance of pelvic motion training and will serve as a guide to gait rehabilitation.

5. Conclusion

In conclusion, we observed the mechanism differences that altered gait performance between normal and stroke patients suffering from knee voluntary movement dysfunction. It is shown that while the primary joints excursion including ankle, knee, and hip RoM are the main contributors of gait performances of the control group, the pelvic tilt and pelvic rotation also play an important role to ensure gait velocity, step, and stride length for the stroke group. Our findings suggest that gait rehabilitation should be approached as a way of reducing the influence of the pelvic motions on gait performance, and the use of primary joint motions on gait performance should be encouraged. In addition, given the fact that the pelvic motions are excessively involved in gait performances, there is a strong need to constrain and support pelvic motions and to increase the RoMs of primary joint movements during or after gait rehabilitation. It is expected that this study can draw attention to the importance of pelvic motion training in the gait rehabilitation of stroke survivors.

CHAPTER III. DESIGN AND CONTROL OF THE NOVEL OVER-GROUND ROBOTIC DEVICE FOR GAIT REHABILITATION

1. Introduction

The first motivation for developing a novel gait rehabilitation device is that current robot-assisted gait rehabilitation (RAGR) devices, which are built on a treadmill-based platform with active or passive exoskeleton robots in combination with overhead harness BWS unit, are still widely controversial as there is uncertainty on the assumption that walking on a treadmill represents an actual gait on the ground. It is also mentioned that providing movements on a single anatomical plane with pre-determined path can result in less satisfactory functional outcomes as well as pain, and the lack of cycle-to-cycle variation in the kinematics and sensorimotor feedback may cause habituation to sensory responses for locomotion, thus ultimately impair motor learning. Most of all, the over-ground walking is considered as the most natural gait pattern with actual foot contact, so over-ground walking rehabilitation devices are required. The second motivation is to avoid pelvic motion restriction during gait. The overhead harness BWS system often restricts pelvic rotation and lateral movements which contribute to energy optimization and an aesthetic walking pattern. It has been shown that the immobilization of the pelvis greatly affects gait dynamics by shortening step width and reducing the coronal trunk rotation, and by lengthening the step length and sagittal trunk rotation. In addition, the study in Chapter 2 shows that the pelvic motions are excessively involved in gait performance of the stroke survivors, and there is a strong need to facilitate pelvic motion during gait rehabilitation. The currently available pelvic motion supported over-ground walking devices, however, confront between mechanical complexity with large number of DoFs and simplicity with limited DoFs as mentioned in Chapter I. Therefore, compromise and optimization between complexity with larger DoFs and simplicity with limited DoFs for pelvic motions are needed to account for effective and successful gait training. The third motivation is to reduce laboring effort of therapists and burden on patients' lower limbs through appropriately implemented BWS unit. Finally yet importantly, it is motivated to provide various gait functions such as assistance and resistance in a task-specific manner.

Therefore, the aim of this Chapter is to present the design and control strategy of a novel robotic device for over-ground walking with pelvic motion support (hereafter the robotic walker or the walker) and to evaluate the mobility and feasibility of the robotic walker.

2. Design Description of the Robotic Walker

2.1. Overall design

Figure III-1 shows the overall concept and actual prototype of the robotic walker. The robotic walker consists of i) an omni-directional mobile platform with ASOC ; ii) a pelvic and trunk motion support brace; iii) body weight support (BWS) unit; and iv) an intuitive human-machine interface with force/torque (FT) sensor for intuitive control of the walker (Fig. III-1B). With this system, subjects are supported to move in any direction without being constrained in one plane of movement so that the therapist can facilitate the gait rehabilitation more effectively and practically with less laboring efforts. The details are described in the sub-sections below.

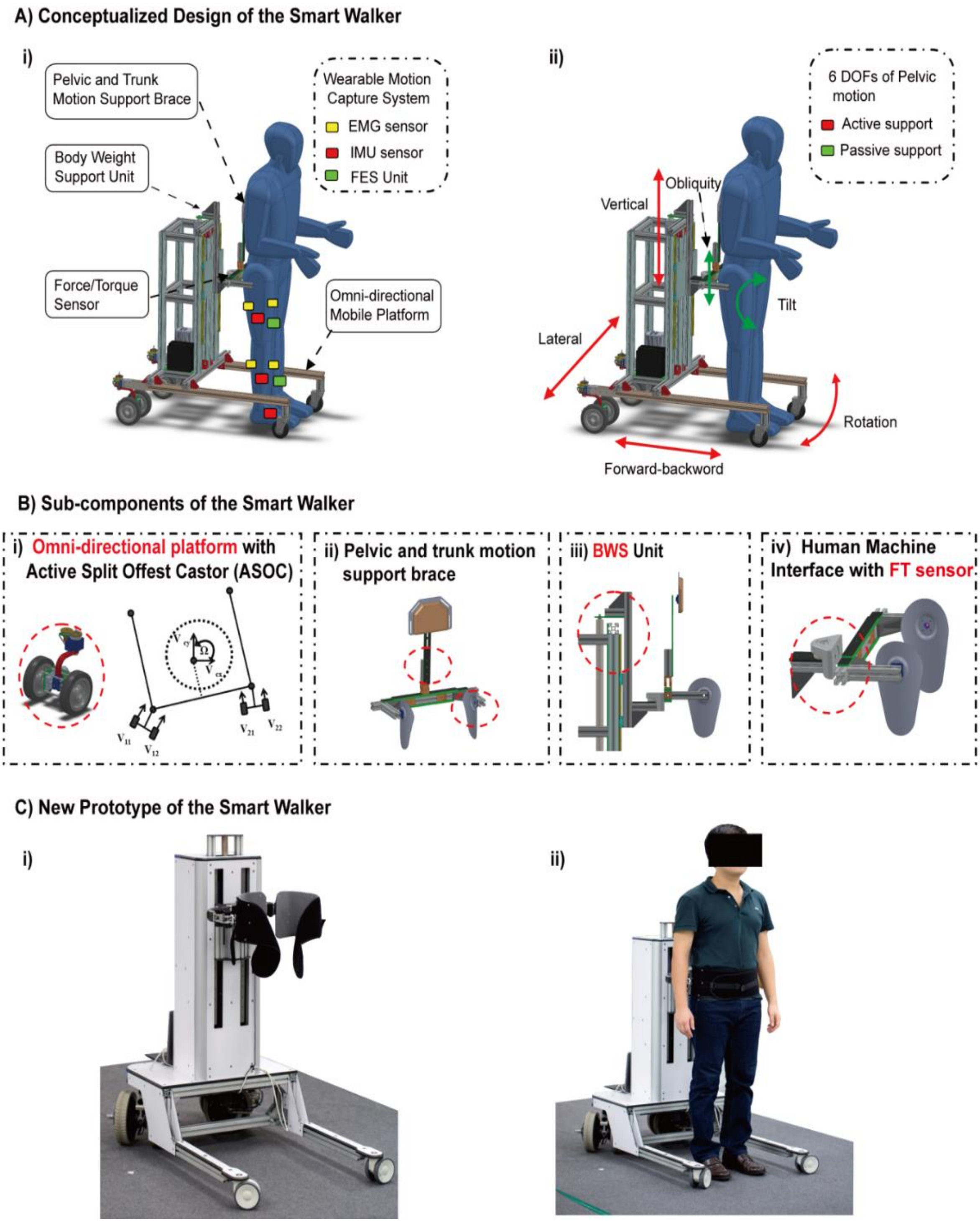


Figure III-1. Conceptualized and actual prototype of the Robotic walker

2.2. Over-ground walking with omni-directional mobility

The motivation for using the omni-directional mobile platform is based on fundamental biomechanics principles. During normal walking, the pelvis moves in anterior-posterior (AP) and medio-lateral (ML) directions, and rotational (RT) path about vertical (VT) axis. These motions are important for energy efficient walking, and constraining these motions leads to abnormal muscle activation and gait pattern. Appropriate sensory inputs through proper feedback and active participation are essential to promote learning skills of the neurologically challenged patients. In addition, the over-ground walking with pelvic motion facilitation may provide correct afferent sensory input and increase attention of users.

As a novel approach shown in Figure III-2, the omni-directional mobile platform has been integrated into the robotic walker to support pelvic AP (V_{cy}), ML (V_{cx}), and RT (Ω) movements. This platform was developed using two sets of ASOC units consisting of two coaxial conventional wheels (see Figure III-2B). The ASOC was driven independently according to velocity commands at the central point [80]. It was reported that the omni-directional mobility can support at three DoFs without additional actuators for the pelvic motion support. Therefore, we believe the omni-directional platform can be effectively utilized in conjunction with the robotic device, as it does not require additional actuators for pelvic lateral and rotational movements.

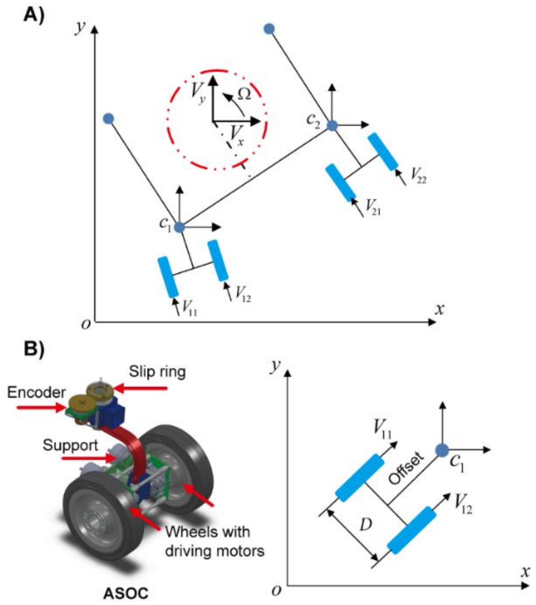


Figure III-2. A) Omni-directional mobility platform which can provide 3DoFs of pelvic motions (forward-backward, lateral, and rotational movements) B) Two sets of ASOC to achieve omni-directional mobility

For a basic concept of omni-directional mobility, the platform shown in Figure III-3 effectively has two degrees of freedom: forward speed and angular velocity. In the most basic case of a platform with two driven wheels at the front and two caster wheels at the back, it is simple to calculate the speed of each driven wheel for any given combination of forward and angular speed [81]:

Equation III-1

$$V_R = V + \omega r \text{ , } V_L = V - \omega r$$

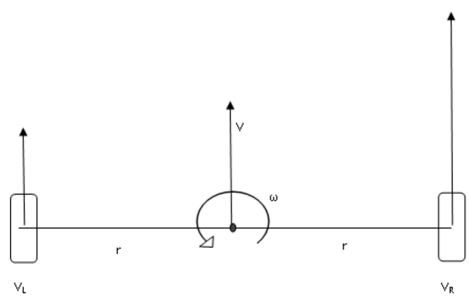


Figure III-3. Velocity relationship between center and ends

Consider a platform that is carried by active split offset castor units that move on a plane. The central point of the platform is defined by V_{cy}, V_{cx}, and its angular velocity, Ω (Figure III-4). Then V_{cy}, V_{cx}, and Ω are defined by the velocity of the center of each rods, V_{cy1}, V_{cx1}, V_{cy2} and V_{cx2} from Equation III-1 (see Eq. III-2). B corresponds to the length of the rod and ϕ is the angle between rods and x axis.

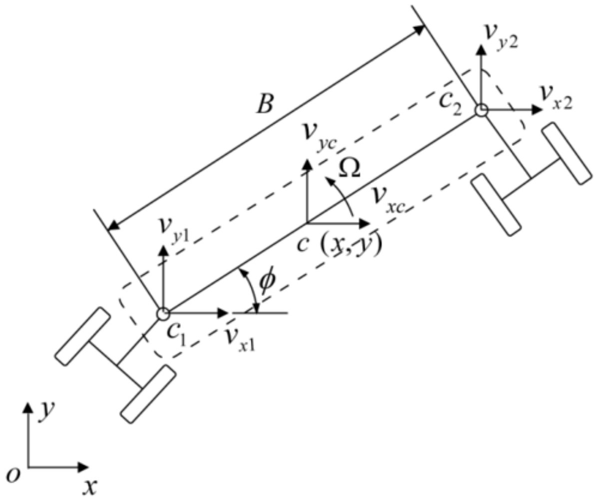


Figure III-4. The omni-directional mobility through ASOC

Equation III-2

$$\begin{bmatrix} V_{CX} \\ V_{CY} \\ \Omega \end{bmatrix} = \begin{bmatrix} \frac{1}{2} & 0 & \frac{1}{2} & 0 \\ 0 & \frac{1}{2} & 0 & \frac{1}{2} \\ -\frac{1}{B\cos\phi} & -\frac{1}{B\sin\phi} & \frac{1}{B\cos\phi} & \frac{1}{B\sin\phi} \end{bmatrix} \times \begin{bmatrix} V_{CX1} \\ V_{CY1} \\ V_{CX2} \\ V_{CY2} \end{bmatrix}$$

From Eq. III-2, each of the central points is defined by velocities of rod ends.

Equation III-3

$$\begin{bmatrix} V_{CX1} \\ V_{CY1} \\ \Omega \end{bmatrix} = \begin{bmatrix} \frac{1}{2} & 0 & \frac{1}{2} & 0 \\ 0 & \frac{1}{2} & 0 & \frac{1}{2} \\ \frac{1}{B\sin\phi} & -\frac{1}{B\cos\phi} & -\frac{1}{B\sin\phi} & \frac{1}{B\cos\phi} \end{bmatrix} \times \begin{bmatrix} V_{X1} \\ V_{Y1} \\ V_{X2} \\ V_{Y2} \end{bmatrix}$$

Equation III-4

$$\begin{bmatrix} V_{X1} \\ V_{Y1} \\ V_{X2} \\ V_{Y2} \end{bmatrix} = \begin{bmatrix} 1 & 0 & \left(\frac{B}{2}\right)\sin\phi \\ 0 & 1 & -\left(\frac{B}{2}\right)\cos\phi \\ 1 & 0 & -\left(\frac{B}{2}\right)\sin\phi \\ 0 & 1 & \left(\frac{B}{2}\right)\cos\phi \end{bmatrix} \times \begin{bmatrix} V_{CX1} \\ V_{CY1} \\ \Omega \end{bmatrix}$$

The position of the center of platform (V_{cx}, V_{cy}, and Ω) is then defined as the velocities of each rod end (V_{x1}, V_{y1}, V_{x2}, and V_{y2}). The velocity of end point will be controlled by the velocity of each wheel (see Eq. III-3 and Eq. III-4). As can be seen in the Figure III-5, the wheel velocities are V_{11}, V_{12}, and the joint velocities with respect to ground are V_f, and V_s. The two vectors u and $\dot{q}_w$ are defined as: $u = \begin{bmatrix} V_{11} & V_{12} \end{bmatrix}^T$ and $\dot{q}_w = \begin{bmatrix} V_f & V_s \end{bmatrix}^T$. The relation between $\dot{q}_w$ and u can be written as: $\dot{q}_w = J_w u = \begin{bmatrix} 1/2 & 1/2 \\ S/D & -S/D \end{bmatrix} u$, where J_w is the Jacobian matrix of the ASOC module in the moving coordinate frame X_w, and Y_w. The velocity of point C is defined as $\dot{q}$ and is given by the following:

$$\dot{q} = \begin{bmatrix} V_x & V_y \end{bmatrix}^T = R\dot{q}_w = \begin{bmatrix} \cos\alpha & -\sin\alpha \\ \sin\alpha & \cos\alpha \end{bmatrix} \dot{q}_w .$$

Therefore, J is defined as:

Equation III-5

$$J = \begin{bmatrix} \frac{1}{2}\cos\alpha_1 - \frac{S}{D}\sin\alpha_1 & \frac{1}{2}\cos\alpha_1 + \frac{S}{D}\sin\alpha_1 \\ \frac{1}{2}\sin\alpha_1 + \frac{S}{D}\cos\alpha_1 & \frac{1}{2}\sin\alpha_1 - \frac{S}{D}\cos\alpha_1 \end{bmatrix}$$

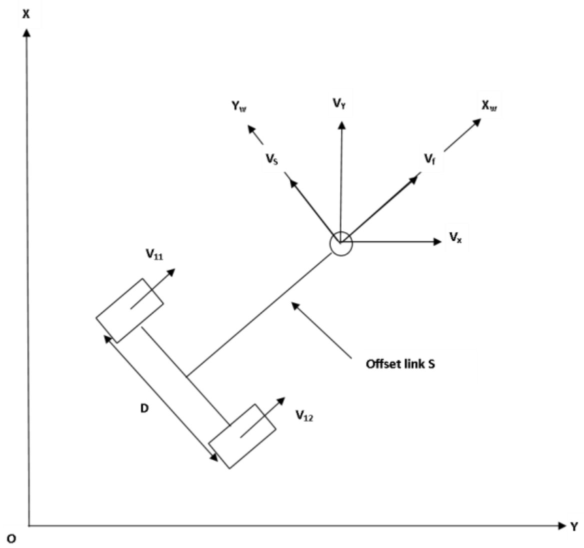


Figure III-5. An Active Split Offset Castor with its coordinate system

According to the equation III-5, the velocities of each rod end are defined below:

$$\begin{bmatrix} V_{X1} \\ V_{Y1} \end{bmatrix} = \begin{bmatrix} \frac{1}{2}\cos\alpha_1 - \frac{S}{D}\sin\alpha_1 & \frac{1}{2}\cos\alpha_1 + \frac{S}{D}\sin\alpha_1 \\ \frac{1}{2}\sin\alpha_1 + \frac{S}{D}\cos\alpha_1 & \frac{1}{2}\sin\alpha_1 - \frac{S}{D}\cos\alpha_1 \end{bmatrix} \times \begin{bmatrix} V_{11} \\ V_{12} \end{bmatrix}$$

$$\begin{bmatrix} V_{X2} \\ V_{Y2} \end{bmatrix} = \begin{bmatrix} \frac{1}{2}\cos\beta_1 - \frac{S}{D}\sin\beta_1 & \frac{1}{2}\cos\beta_1 + \frac{S}{D}\sin\beta_1 \\ \frac{1}{2}\sin\beta_1 + \frac{S}{D}\cos\beta_1 & \frac{1}{2}\sin\beta_1 - \frac{S}{D}\cos\beta_1 \end{bmatrix} \times \begin{bmatrix} V_{21} \\ V_{22} \end{bmatrix}$$

Therefore, the desired velocity of the platform (V_{cy}, $V_{cx,}$ and Ω) can be defined by the velocity of wheels, V_{11}, V_{12}, V_{21}, V_{22}, V_{31}, V_{32}, V_{41}, and $V_{42.}$

It had a number of advantages, including the use of a simple structure, high energy efficiency, and robust mobility even on uneven terrain. Thus it had the potential to simplify our mechanical design by eliminating additional actuators to facilitate pelvic motions. Furthermore, the omni-directional platform allows the user to move in any direction instead of traditional treadmill-based devices where only forward movement is allowed. Hence, a user can consistently move in any direction and any configuration by using the omni-directional platform.

2.3. Pelvic and Trunk Motion Support Brace

The pelvic and trunk motion support system is shown in Figure III-6. In this study, the pelvic support harness not only served as the physical interface between the robotic walker and the user, but also passively supported both pelvic tilt and obliquity. The pelvic pads were strapped around the waist of the subject, and the brace had an adjustable design to facilitate each patient's anatomy. The pelvic pads were mounted to the pelvic brace with a spherical flange bearing, allowing pelvic anterior-posterior tilt. The back side of the pelvic brace was connected to a swivel joint covered by fiberglass. Fiberglass was chosen due to its compliance characteristics that allowed superior-inferior movements of anterior-superior iliac spine (ASIS) in the frontal plane, thus accomplishing pelvic obliquity passively. In addition, the fiberglass was still stiff enough and capable of restricting any exaggerated vertical movement of ASIS usually present in stroke patients (i.e. hip-hiking) [82]. The trunk support brace was connected to a swivel joint with the fiberglass providing rotation along the frontal plane of the torso. As a result, trunk lateral bending was achieved and at the same time, the fiberglass allowed trunk flexion and extension in the sagittal plane to achieve trunk bending in AP direction.

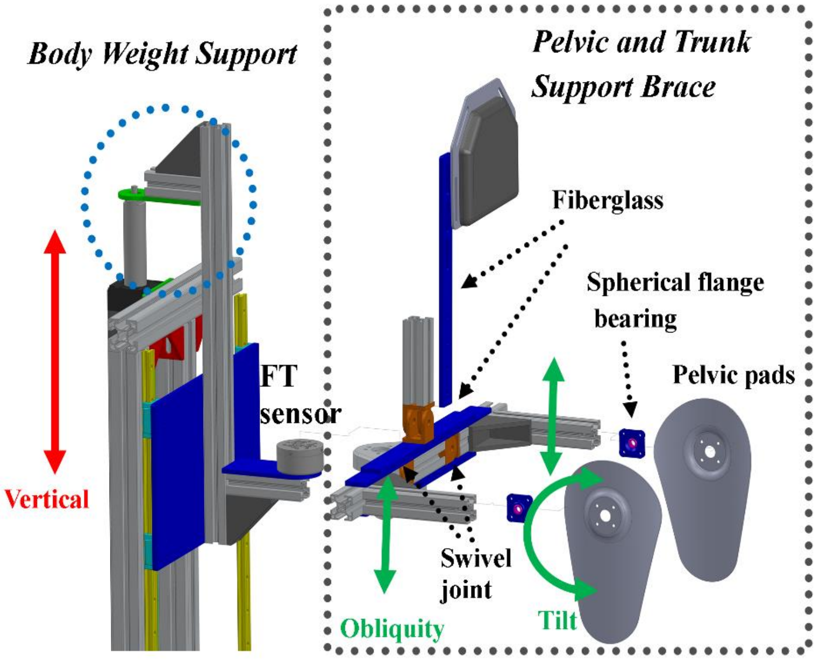


Figure III-6. The pelvic and trunk motion support system with BWS actuator

2.4. Body Weight Support Actuator

The use of an active BWS system which provided active unloading of the body mass of the subject to the desired percentage with unrestricted pelvic motion was proposed for effective BWS training (Figure III-7A). Therefore, the robotic walker allowed the pelvis and trunk to move vertically with pelvic AP, ML, and RT movements, as well as pelvic tilt and obliquity. The BWS actuator could provide all-in-one control through a PID controller, drive, and motor integrated into one compact component; the active body weight of the subject is maintained via the vertical axis of the force/torque (FT) sensor during dynamic walking (Figure III-7B). The BWS actuator is able to support vertical (VT) pelvic and trunk movements during walking.

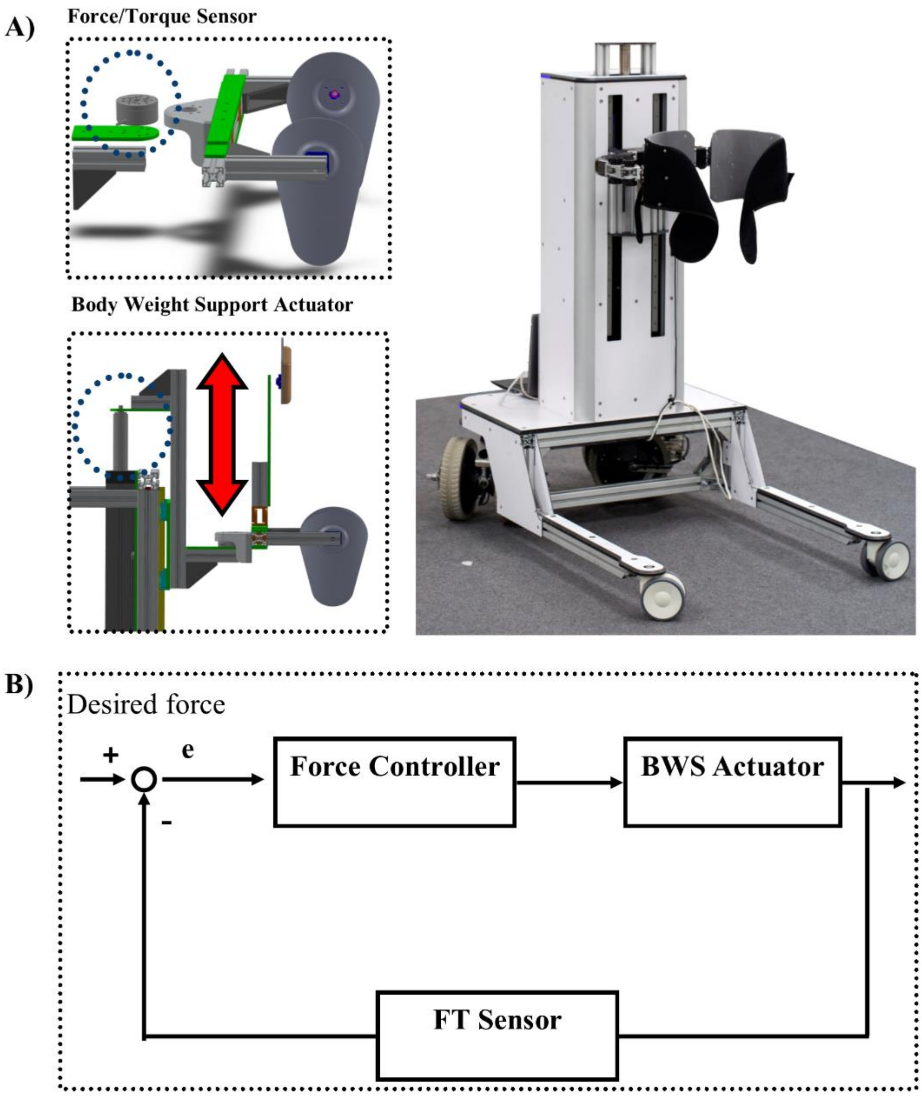


Figure III-7. PID control for the BWS actuator

In summary, the walker achieved six DoFs of pelvic motion by integrating an omni-directional platform, a body weight support unit, and a pelvic support brace (Table 3). The walker implemented four DoFs active pelvic support which corresponds to pelvic AL, ML, RT and VT movements through the omni-directional platform and the active BWS unit, while the remaining two DoFs, which were passive motions (pelvic tilt and obliquity), could be achieved through the pelvic support brace.

Table 3. DoFs and RoMs for Pelvic Motions in the Walker System

Pelvic DoFs			**Robot**	**RoMs**
Active	3DoFs	AP ML RT	Omni-directional Platform	No Limit
	1DoFs	VT	Body Weight Support Unit	No Limit
Passive	2DoFs	Tilt	Pelvic Motion Support Brace	No Limit
		Obliquity		-20 ° to 20 °

3. Control of the Robotic walker

3.1. Intuitive human-machine interface

For the walker control, the interface was critical in the controlling parameters relevant to the robotic assisted gait devices. The design of the robotic walker considered the user's gait characteristics to account for a continuous interaction between the robotic walker and the user. A key requirement for the interface was to adapt the intention for different levels of physical and mental functionality and provide a user-friendly experience to enable the subject to walk naturally.

The FT sensor was enclosed by a central section, which was affixed to the pelvic brace. This allowed for force transmission from the subject's pelvis to the pelvic brace, and finally detection from the FT sensor. The measured FT signal was used to drive the motion of the system. However, using FT signals directly to generate motion could result in instability due to fluctuation and noise in the signals. Instead, based on the interaction force detected, speeds in the AP, ML, and RT directions were generated with an adaptive admittance model (Figure III-8) [80, 83]. This method can achieve the most intuitive human physical interaction allowing the user to simply focus on walking without thinking about direction and speed control, resulting in a much reduced mental workload to drive the machine. The admittance control method consisted of a virtual mass M and damper parameter B to provide natural and intuitive interaction between the user and the device (Figure III-9).

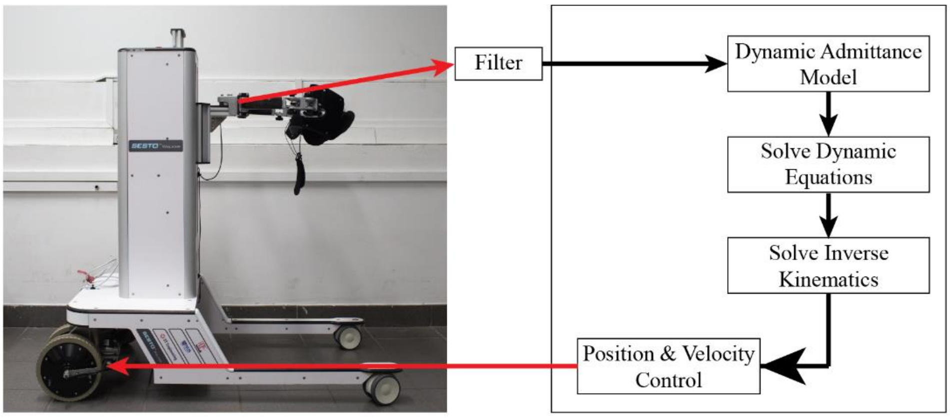


Figure III-8. Admittance-based model for walker control. Input comes from the force and torque 6 axis FT axis and the output generates the velocity of each of the four wheels.

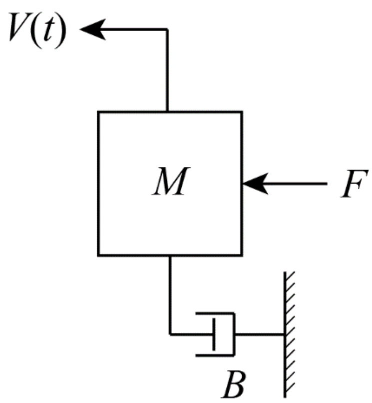


Figure III-9. Mass-damper admittance model.

With F as the user input force in the respective directions (forward-backward (anterior-posterior, (AP)), medio-lateral (ML) and rotational (RT) movement) and V as system response speed in the same direction, the transfer function of the system was defined as

Equation III-6

$$G(s) = \frac{V(s)}{F(s)} = \frac{1}{Ms + B}$$

where M is mass, and B is the damping parameter. The time response of the system for a step input was defined as

Equation III-7

$$V(t) = \frac{F}{B}(1 - e^{-\frac{t}{\tau}})$$

where τ is the time constant defined by $\tau = M / B$.

A linear 3 DoFs mass-damper model for the walker was defined by

Equation III-8

$$\begin{bmatrix} M_X & 0 & 0 \\ 0 & M_Y & 0 \\ 0 & 0 & J_Z \end{bmatrix} \cdot \begin{bmatrix} \dot{V}_X \\ \dot{V}_Y \\ \dot{\Omega} \end{bmatrix} + \begin{bmatrix} B_X & 0 & 0 \\ 0 & B_Y & 0 \\ 0 & 0 & B_Z \end{bmatrix} \cdot \begin{bmatrix} V_X \\ V_Y \\ \Omega \end{bmatrix} = \begin{bmatrix} F_X \\ F_Y \\ T_Z \end{bmatrix}$$

where $M_{X,Y}$, $B_{X,Y}$ and $F_{X,Y}$ are the mass, damping and force in lateral and forward directions, J_Z, B_Z and T_Z represent the moment of inertia, damping and torque in the vertical direction. This method also allows the combination of virtual force generated by the controller and the forces applied by the user, achieving shared control scheme [80].

3.2. Determination of the damping parameters

The mobility and performance of the walker is dependent on determining proper virtual mass and damper parameters of the admittance controller. The force requirement to reach a steady state velocity primarily depended on the damping parameters, thus the appropriate range of damping parameters was investigated through experiments with actual trials while the mass parameter was fixed at 2kg. In addition, the moment of inertia, J_z, and the damping B_z of rotational movement were fixed at J_z = 2 kg·m^2 and B_z = 20 N·s because these values showed the closest pelvic rotation patterns to normal gait. In this study, we focused on determining appropriate damping in AP and ML directions. Damping parameters were considered acceptable if the gait patterns with the walker is similar with that of normal walking.

A. *Testing protocols and analysis*

The center of mass (CoM) of the pelvic trajectory were obtained from five young and healthy subjects (average age: 27.25 years, average height: 172.25 cm, and average weight: 62.63 kg) and was used as a reference for normal gait. The study was conducted in a motion capture gait laboratory with eight high-speed infrared cameras (Vicon, Oxford, UK). Four optical retro reflective markers attached on the left and right anterior-superior iliac spine (ASIS), and the left and right posterior-superior iliac spine (PSIS). The AP, LT, and RT velocities of the pelvic CoM were then calculated.

The testing of the appropriate damping parameters was conducted on one subject (31 years old, height: 171.5 cm, and weight: 72 kg). The subject wore the harness and was strapped tightly to the walker. The subject was instructed to walk on a 10m walkway at a comfortable gait speed. The damping parameters varied from 80 to 160 Ns/m with increments of 20 Ns/m. The measured force and torque from the FT sensor were translated into velocities of the pelvic CoM through the admittance control model and compared to that of the reference velocities of healthy subject's normal gait.

B. *Testing results for range of damping*

Figure III-10 shows the velocities of forward-backward and lateral movements with various damping parameters. The velocity of forward movement showed a distinctive "W" shape for each gait cycle with mean velocity approximately 1.0 m/s (Figure III-10A). Overall, the mean forward velocity when using the walker was significantly reduced compared to that of normal walking. However, increasing the damping for AP motion resulted in a reduction of the AP velocity. At high AP damping (i.e. greater than 140 Ns/m), the subject experienced heaviness of the walker, which led to the reduced speed. At lower damping (i.e. less than 80 Ns/m), however, oscillatory motions were observed. Therefore, we determined that a damping range between 80 Ns/m to 140 Ns/m was appropriate as it provided a balance between stability (less oscillation) and weight (less resistance). Likewise, the lateral velocity did not alter when the lateral damping parameters were between 80 Ns/m to 140 Ns/m. The subject also experienced heaviness and oscillation in the lateral direction when the damping was higher than 140Ns/m and less than 80Ns/m respectively (Figure III-10B). This range of damping values showed most natural walking patterns without oscillation and heaviness of the system.

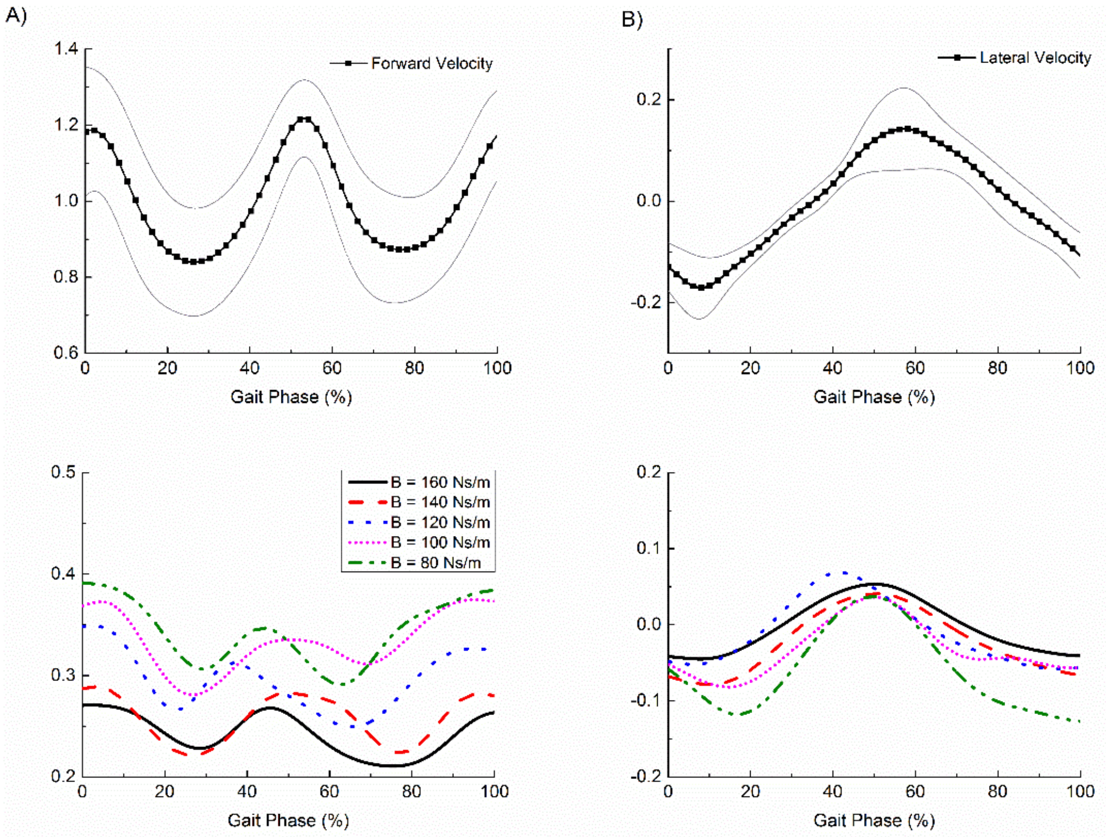


Figure III-10. Testing on appropriate damping ranges in forward-backward and lateral directions with various damping parameters from 80 Ns/m to 140 Ns/m

However, using a fixed parameter model could not take into account the user's need to accelerate and decelerate during normal gait. It should be noted that the more damping provided less mobility with increased stability, while less damping provided increased mobility with less stability. To address different requirements of the users during different phases of gait cycles, an admittance semi-empirical model with velocity-dependent damping was implemented (Eq. III-9) [80].

Equation III-9

$$b = b_m - \frac{b_m - b_o}{V_m}|V|$$

where b_m is the maximum damping, b_o is the minimum damping, V is current velocity of the walker, and V_m is the allowance of maximum speed. The velocity-dependent damping can increase stability during acceleration/ deceleration of normal gait while also providing high mobility at a steady gait.

4. System Implementation for the Robotic Walker

NI system was incorporated in order to control the omni-directional mobile platform for over-ground walking with the FT signal processing. The cRIO series was chosen due to the need for a real-time operating system as well as a system that could be flexibly customized to provide real-time control with field-programmable gate array (FPGA) and serial communication. Figure III-11 shows the connection of C-series module to the external hardware. The NI 9403 was used for digital input of motor controller, the two sets of NI 9239 were assigned for analog input signal from the FT sensor, and the NI 9401 module provided left and right encoder digital input and output. In addition, the RS232 serial port was adapted to control the BWS actuator. The system development and programming were done via LabVIEW™ 2013 on a host PC. Figure III-12 shows the block diagram detailing how the NI system is used to achieve the pelvic motion support (PMS) and BWS. In Figure III-12, it can be seen that the higher level control consists of the user interface on the host PC and the Real Time Operating System (RTOS) in the NI controller. The user interface receives the input parameter values set by the experimenter while the RTOS processes these values to calculate corresponding output values. All critical parameters such as FT signal with second order Butterworth filter with 0.5Hz cutoff frequency, left-right (L/R) encoder values, and motor driver command were obtained through the FPGA system with 1 kHz sampling rate. Additionally, the

AP, ML, and RT velocities of the walker and BWS system were achieved through admittance and PID controllers, respectively. The pelvic motion assistance and resistance functions were added to support the gait with damping based force-field control system. The BWS actuators support pelvic vertical movement during the gait, generating VT force according to certain percentages of body weight (0-50%). The FT signal, velocities of the walker, and encoder values were used for feedback to provide pelvic motion support while walking.

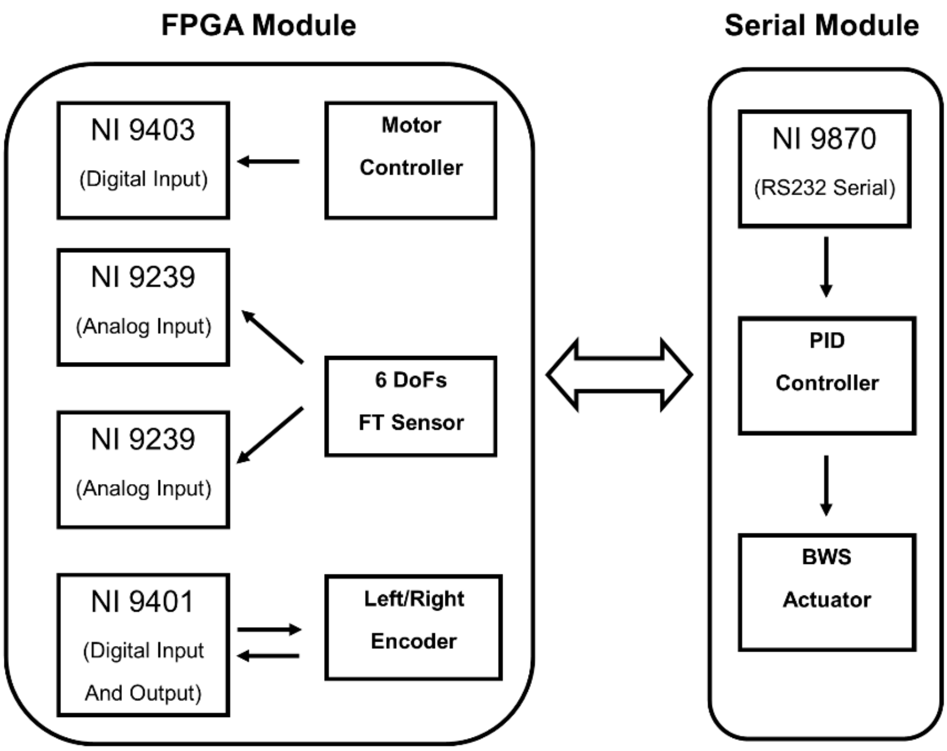


Figure III-11. Connection of NI Hardware to External Hardware

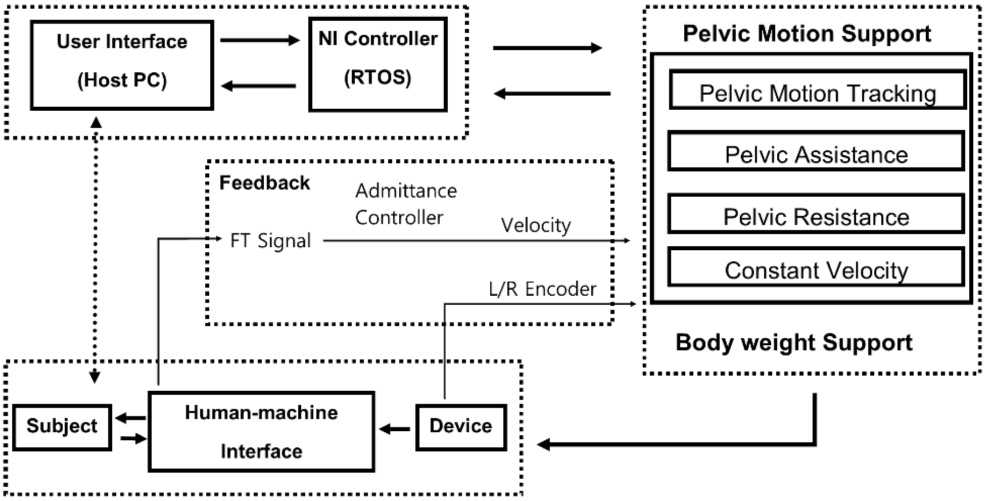


Figure III-12. Overall System to Control the Pelvic Motion and Body Weight Support of the Over-ground Gait Rehabilitation Device

5. Evaluation of the Feasibility of the Robotic Walker on Gait Dynamics

Preliminary gait experiments were performed to evaluate the feasibility of and to investigate the effects of the pelvic lateral and rotational facilitations with the robotic walker. The experiments were performed in three different conditions: (1) walking without walker, (2) walking with walker while pelvic lateral and rotational motions were allowed, and (3) walking with walker while pelvic lateral and rotational motion were restricted. We hypothesized that walking with no restriction would have minimal alteration to gait kinematic profiles and range of motions (RoMs) of lower limb, while walking with pelvic restrictions would significantly affect the gait dynamics.

5.1. Experimental protocol

Twelve healthy young subjects with age: 26.72 ±4.45 years, height: 1706.10 ± 93.05mm, and weight: 63.64 ±13.50 kg participated in this study. Subjects were only recruited if they had no prior clinical gait abnormalities or musculoskeletal and neurological disorders. A motion capture gait laboratory with 15 optical retroreflective markers and eight high-speed infrared cameras (Vicon, Oxford, UK) were used to obtain joint kinematics with sampling rate of 100Hz. Twelve surface wireless electromyography (EMG) electrodes (six electrodes on each limb) were attached on tibialis anterior (TA), gastrocnemius (GA), vastus medialis (VM), biceps femoris (BF), gluteus medius (GM), and adductor longus (AL) and simultaneously recorded with sampling frequency of 1000 Hz.

Experimental scenarios were comprised of walking without the walker (normal control, NC), no restriction (NR), and both restriction (BR) as mentioned above. BR condition was implemented by providing infinite lateral and rotational damping to the admittance controller. As walker performance primarily depends on selection of virtual mass and damper parameters, the admittance semi-empirical model with velocity dependent damping (Eq. III-9) was used with damping from 80 Ns/m to 140 Ns/m for forward and lateral movements. All subjects provided informed consent in accordance with Institutional Review Board standards and were instructed to walk naturally with their preferred speed on the 10m walkway in the gait laboratory. Three successful trials for each condition were collected for further analysis and interpretation.

5.2. Data analysis

The raw kinematic data was pre-processed using Nexus, software provided by the Vicon motion capture system. Each gait cycle was only considered successful if complete data was available over one whole stride. The records of gait phase events (i.e. heel strike and toe off time) and kinematic profiles of ankle, knee and hip flexion/extension angles were extracted from the software.

All kinematic profiles of the gait cycle were normalized from time in seconds to percentage of the gait cycle. RoM of kinematic profiles and gait performance parameters (step length, stride length, step width, and gait velocity) were obtained through a customized program in MATLAB (Mathworks, Natick, MA, USA). The stride and step length were normalized to the subject's leg length. The raw EMG signals were band-pass filtered with a pass-band between 2 Hz and 200 Hz to remove motion artifact and high frequency noise. After rectification, all EMG data was normalized by a maximum voluntary contraction value. Finally, a low pass filter with a cut-off frequency at 10 Hz was performed to produce a linear envelope representation of the signals [84, 85].

5.3. Statistics

All experimental data was analyzed using statistical software (SPSS Inc., Chicago, IL). A one-way ANOVA was performed for minimum, maximum values, RoMs of kinematic profiles, and gait performance to distinguish differences among experimental conditions. When significance was found through a one-way ANOVA, Tukey's post hoc test was processed to contrast differences among the experimental conditions.

5.4. Results of the preliminary experiments

A. Kinematics

Figure III-13 and Table 4 show the kinematic profiles, minimum, maximum values and RoMs of the ankle, knee, and hip joints. Overall, walking with NR did not alter ankle, knee, and hip joint angles over a gait cycle, while walking in BR showed significantly reduced ankle plantar flexion at pre-swing, knee flexion and hip flexion at mid-swing and terminal stance, respectively (Figure III-13 A), B), and C)).

Although statistical analysis found that ankle plantar flexion was significantly reduced in both the NR and BR conditions when compared to NC, this reduction was more significant in the BR condition (i.e. restricted pelvic motions) compared to NR (Figure III-13D). For the knee joint, a significantly reduced knee flexion angle was found only in the BR condition compared to normal walking (Figure III-13E). Maximum hip extension at terminal stance was significantly reduced at NR and BR, but there was no significant difference in maximum hip flexion at initial contact and terminal swing (Figure III-13D). All RoMs were significantly reduced for both NR and BR conditions but RoM reduction was significantly greater in BR than in NR (Figure III-13E, Table 4)

B. Gait descriptive parameters

The gait descriptive parameters such as normalized stride and step length, step width, and gait velocity are shown in Figure III-14 and Table 5. All gait parameters of NR and BR were significantly reduced compared to NC condition. However, substantial reductions in stride and step length were observed when the pelvic motions were restricted compared to NR.

C. Muscle Activation Patterns

Figure III-15 shows muscle activation profiles for all experimental conditions. From the profiles, minimal changes can be seen between NR condition and NC. The most obvious changes were observed in the TA, GA, and BF. In these three muscles, the EMG profiles show increased and prolonged activation across the gait cycle compared to both NR and NC. In fact, no alteration was observed in

NR compared to NC, showing the muscle activation in NR was not deviated from the variations of NC. This shows that the restriction of the lateral and rotational movements of the pelvic resulted in deviation of the muscle activation from normal gait. When these motions were allowed, however, even with slight changes in the kinematic profiles, muscle activation was largely similar to that of normal gait. There were no significant differences in VM, G_{Med}, and AL from normal and NC.

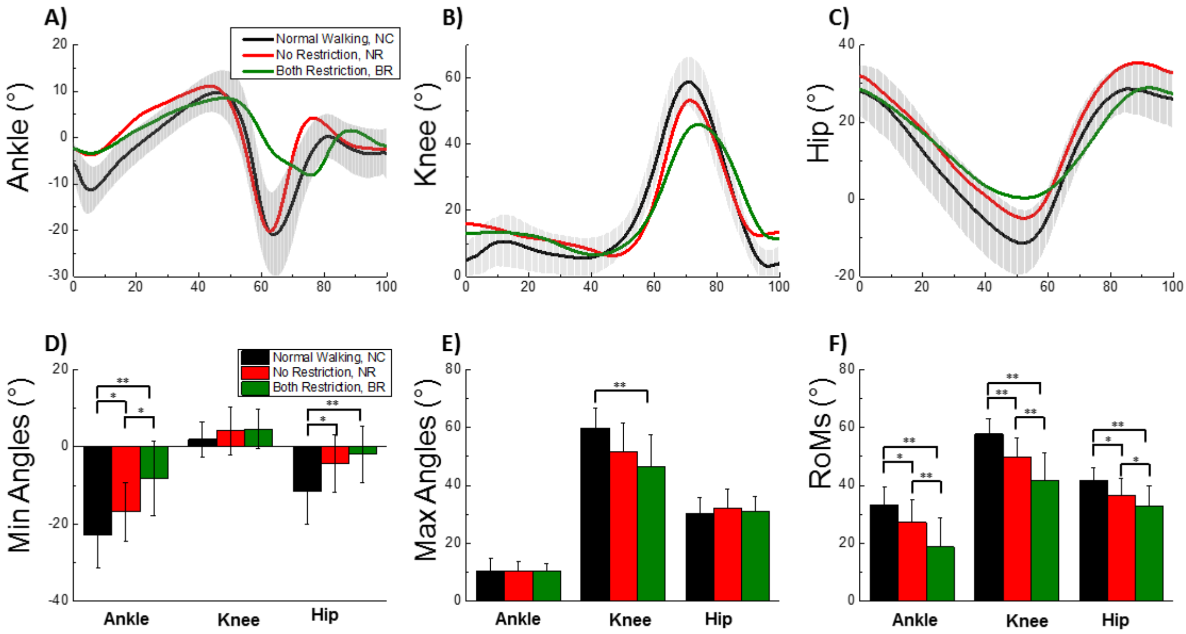


Figure III-13. Ankle, knee, and hip Joints kinematics (A), B), and C)), and their minimum, maximum, and range of motions (D), E), and F)) during each gait cycle. The black and gray line shows averaged kinematic profiles of normal walking and standard deviation. Red and green colors represent the kinematic results of NR and BR, respectively

Table 4. Kinematic parameters during walking with Robotic walker

	NC	NR	BR
Ankle dorsiflexion	10.36 ±4.21	10.41 ±3.35	10.45 ±2.64
Ankle Plantar flexion	22.79 ±8.53	16.84 ±7.59*	8.26 ±9.61**,ǂ
Ankle RoM	33.15 ±6.16	27.24 ±7.91*	18.71 ±9.87**,ǂǂ
Knee Flexion	59.76 ±7.16	54.28 ±8.10	46.49 ±11.04**
Knee RoM	57.79 ±5.30	49.76 ±6.65**	41.82 ±9.32**,ǂǂ
Hip Flexion	30.15 ±5.64	32.24 ±6.57	30.99 ±5.20
Hip Extension	11.55 ±8.41	4.41 ±7.48*	1.97 ±7.31**
Hip RoM	41.69 ±4.24	36.64 ±5.68*	32.97 ±6.70**,ǂ

* Statistical difference between NC and NR, LR, RR, and BR, P<0.05

** Statistical difference between NC and NR, LR, RR, and BR , P<0.01

ǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.05

ǂǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.01

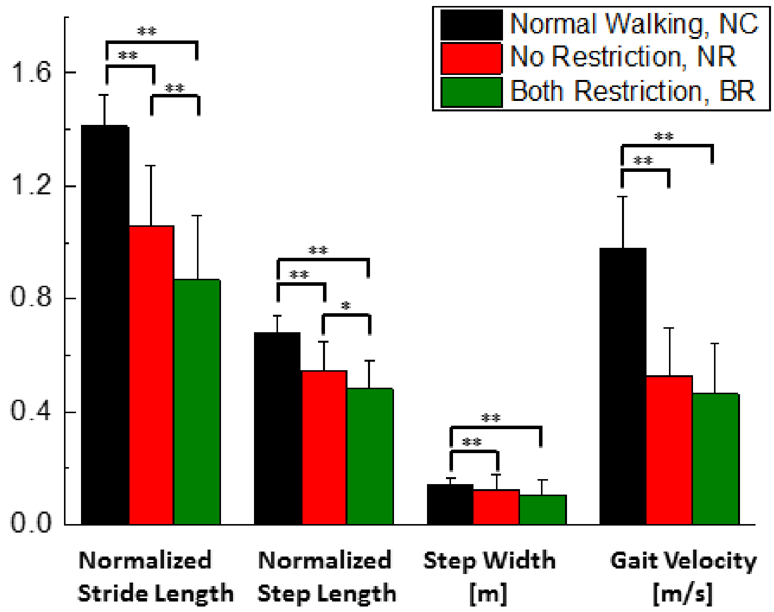


Figure III-14. Gait performance parameters such as normalized stride and step length, step width, and gait velocity.

Table 5. Gait performance parameters

	NC	NR	BR
Normalized Stride Length	1.410 ± 0.112	$1.055 \pm 0.214^{**}$	$0.866 \pm 0.227^{**,\ddagger\ddagger}$
Normalized Step Length	0.679 ± 0.063	$0.543 \pm 0.105^{**}$	$0.481 \pm 0.101^{**,\ddagger}$
Step Width (m)	0.143 ± 0.019	$0.122 \pm 0.056^{**}$	$0.106 \pm 0.052^{**}$
Gait Velocity (m/s)	0.977 ± 0.186	$0.524 \pm 0.175^{**}$	$0.463 \pm 0.179^{**}$

* Statistical difference between NC and NR, LR, RR, and BR, P<0.05

** Statistical difference between NC and NR, LR, RR, and BR , P<0.01

ǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.05

ǂǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.01

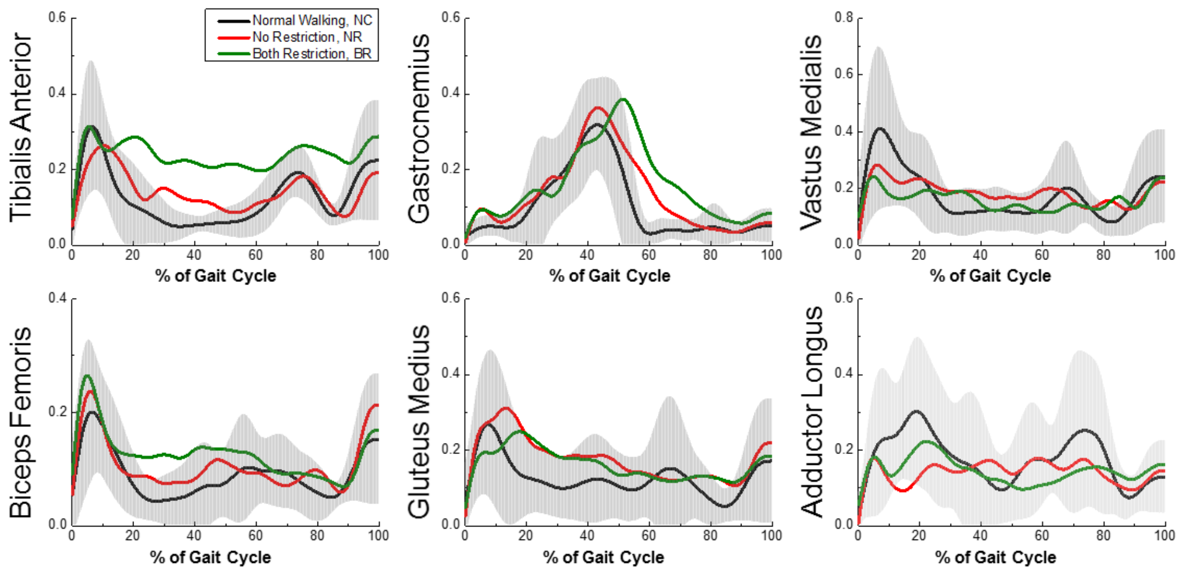


Figure III-15. Averaged surface EMG profiles for six major muscles in walking under three different conditions: NC (black line), NR (red line), and BR (green line). Gray line represents the standard deviation of NC.

6. Discussion

In this section, the integration of the robotic walker, and experimental evaluation of the developed device aimed at gait rehabilitation for the neurologically challenged patients is discussed.

6.1. Development of the robotic walker

The omni-directional mobility, human machine interface with pelvic brace and FT sensor were integrated into the platform. To avoid pelvic motion constraints and complexity of the actuation, we simplified the mechanical design for pelvic motion support by implementing omni-directional mobility, a BWS system, and a pelvic motion support brace.

A system with omni-directional mobility can move in any direction instantaneously without the need for complex actuators for maneuvering. The omni-directional mobility with admittance controller allow the user's pelvis to be actively controlled and can move in any direction through two pelvic translations (AP, and ML) and one DoF for pelvic RT. Additionally, the BWS system actively support the pelvic and trunk vertical movement in order to achieve a total of four active DoFs of pelvic motions. For passive pelvic motion support, the pelvic motion support brace allows for both pelvic tilt and obliquity. Therefore, a total six DoFs of pelvic motion are accomplished with the walker.

The basic function of the walker is to encourage natural walking during over-ground gait rehabilitation with pelvic motion support. The velocity-dependent damping model provides a slower response during gait acceleration with higher damping for stability, while providing lightness and smoothness during the steady state of walking period for mobility. When the user needs to stop or decelerate, the walker stops quickly as the damping increases, so that the user feels minimal dragging or inertial force.

6.2. Evaluation of the performances of the robotic walker

In this Chapter, the evaluation of the robotic walker on gait dynamics is investigated through the gait analysis. The results show feasibility and mobility of the walker and its potential use in gait rehabilitation for neurological disorders. The kinematic excursion of ankle, knee, and hip joint display no significant changes between NC and NR, while ankle plantar-flexion in mid-stance, knee flexion in mid-swing, and hip extension during mid-stance are significantly reduced under the BR condition. As a result, these reductions notably constrain RoMs of ankle, knee, and hip joint motions. For gait performance parameters, the normalized stride and step length, step width, and gait velocity are reduced in both NR and BR compared to NC, with a greater decrease in normalized stride and step length in the BR compared to NR condition. It shows that pelvic restriction severely affects the gait performances by interfering with normal walking. Despite a slight decrease in gait performance and RoMs of kinematics in the NR condition, the muscle activation patterns show clear indication of the biomechanical effects of the walker. It appears that profiles of the EMG burst amplitude when the pelvis is not restricted are not significantly altered from that of normal gait. However, prolonged and excessive activation are observed when pelvic motions are restricted, especially in the TA, GA, and the HS muscles (Figure III-15). In other words, the developed device can elicit normal muscle activation patterns in a natural and intuitive way without altering normal gait while walking. Thus, the results of this study complement other studies and prove the importance of facilitating pelvic lateral and rotational movements to elicit correct afferent input and thus restore defective gait patterns of stroke survivors after gait training.

6.3. Extra Capabilities of the Walker

The walker is shown to facilitate gait rehabilitation in a natural and intuitive way. However, several adjustments can be made to the admittance controller that expand on the capabilities of the walker and facilitate a wider rehabilitation regimen. Firstly, assistive or resistive functions could be added in the anterior-posterior direction as such functions have been shown to promote the error correction process that could accelerate motor learning during gait rehabilitation [86]. In addition, stroke survivors typically display larger pelvic LT and RT movements as a compensatory strategy to keep the body balanced during walking with significantly reduced AP velocity [70]. The walker can assist in such situations by providing either an assistive or a resistive lateral force to correct excessive pelvic lateral movements when the patient might have deviated from a desired range. Lastly, forces are applied to perturb patients at the pelvis, and can assist in balance training where patients learn to adapt to such perturbations while still providing additional safety as opposed to more conventional and primitive rehabilitation techniques. Finally yet importantly, the BWS function is provided to reduce burden on patients and therapists. The biomechanical effects of pelvic motion restriction, BWS, and resistance training are discussed in the Chapter IV, V, and VI, respectively.

CHAPTER IV. RESTRICTION OF PELVIC LATERAL AND ROTATIONAL MOTIONS ALTERS LOWER LIMB KINEMATICS AND MUSCLE ACTIVATION PATTERN DURING OVER-GROUND WALKING

1. Introduction

The robotic walker for successful robotic gait rehabilitation was developed, and its feasibility on gait dynamics was investigated in Chapter III. As highlighted in Chapter II and III, the pelvis is highly associated with gait performance of stroke survivor, and fixation of pelvis can lead to abnormal gait and muscle activation pattern. Therefore, the scope of this Chapter is to investigate detailed insight of the restriction of pelvic lateral and rotational movements during walking in healthy young subjects as an extended evaluation from Chapter III.

Pelvis plays a central role in human locomotion. Six gait determinants (GDs), comprised of pelvic rotation (RT), obliquity, knee flexion, foot and knee mechanism, and lateral displacement (LD) of the pelvis, have been defined as the primary functions of gait which minimize vertical and LD of the center of mass (CoM) [67, 69]. The pelvic motions, which accounts for three of the six GDs, control the whole body balance, transmit force between lower and upper limbs, and increase energy efficiency of gait [87-89]. Pelvic LD and RT, defined as side-to-side movement of the pelvis and rotation of the pelvis about a vertical axis respectively [90], are especially important for manipulating the vertical displacement of CoM, step and stride length, and horizontal balance during normal gait [70, 88, 91, 92].

Recently, the importance of facilitating pelvic LD and RT motions has been emphasized in the area of gait rehabilitation, in order to give a natural and aesthetic gait pattern after gait training for neurologically challenged subjects. With the prevalent use of robotic devices for gait rehabilitation, however, these movements are often limited by such robotic devices [93, 94]. Any restriction on pelvis leads to alterations in the gait kinematics and severely limit frontal and transverse rotations [11]. Previous studies have reported that the neurologically challenged patients have shown abnormal pelvic motions with increased pelvic LD and RT movements due to compensation for weakened and spastic muscles [70, 72-74, 95]. As correct afferent sensory inputs, which carries nerve impulses from receptors toward the central nervous system are the most critical factor for successful gait rehabilitation [11], such pelvic restrictions caused by robotic devices can diminish the quality of gait training. To tackle this issue, Hidler [94] suggested that there is a need for robotic-gait training devices to include additional degrees of freedom (DoFs) for pelvic motion in order to facilitate more normative muscle activation patterns, after investigating abnormal electromyography (EMG) patterns caused by pelvic restriction [94]. However, the addition of extra DoFs for pelvic motion to the robotic devices without understanding their contributions to human gait makes the mechanical structure of robotic devices more complicated. Consequently, it might cause other abnormal gait patterns due to the effects of compensatory movements. Moreover, a certain level of pelvic LD and RT restriction may be beneficial for the early stages of gait intervention to increase lateral balance during walking, as neurologically challenged patients have larger pelvic LD and RT movements compared to healthy individuals [93].

In this aspect, it is imperative to investigate influences of pelvic motions on human gait to verify the impact of these motions on actual gait patterns and to provide a better robotic gait rehabilitation for the neurologically challenged patients. However, to the best of our knowledge, few studies on pelvic restriction to lower limb dynamics are available in the literature. Within this limited information, only one study conducted by Veneman and his co-workers [93], which investigated the effects of pelvic fixation in the horizontal plane during walking on a treadmill was found. This study reported that this fixation can affect gait kinematic patterns by reducing step width and sagittal plane trunk rotations, and by increasing step length and coronal plane trunk motion [93]. However, the findings of this study [93] were limited to a kinematic point of view, without taking a mechanical understanding into account by observing muscle activation patterns. In order to have a better understanding of pelvic motion restriction, investigating muscle activation patterns with the kinematic and gait descriptive parameters is crucial. Additionally, the mechanisms involved between treadmill and over-ground walking are different, resulting in altered walking patterns [36, 96]. In the perspectives of the robustness of the experimental framework, it is necessary to investigate the respective effects of pelvic LD and RT restrictions on human gait dynamics during over-ground walking. Therefore, this study aimed to examine the biomechanical effects of pelvic LD and RT restrictions on lower limb dynamics including gait descriptive parameters, kinematics, and muscle activation during over-ground walking. The underlying hypotheses of this study are 1) walking without pelvic restriction could be the most natural pattern, and 2) pelvic restriction would alter gait dynamics.

2. Methods and Materials

2.1. Pelvic motion restriction with a novel robotic walker for over-ground gait rehabilitation

A novel robotic walker for over-ground gait rehabilitation has been developed and described in Chapter III. As shown in Fig. VI-1A, the walker was designed to facilitate 6 DoFs of pelvic motions. The omni-directional mobile platform has been designed to support 3 DoFs pelvic motions including pelvic forward-backward (V_{cy}), lateral (V_{cx}), and rotational (Ω) movements (Figure IV-1B, i)). The body weight support unit was proposed to support vertical displacement of the pelvis (Figure IV-1B, ii)), while pelvic tilt and obliquity were passively supported by a pelvic motion support brace. Control of the walker is achieved by detecting a force/torque (FT) signal from pelvis which is tightly enclosed by the pelvic motion support brace (Figure IV-1B, iii)). Based on the interaction force detected, speeds in the forward, lateral, and rotation directions are generated with an adaptive admittance model which is comprised of a virtual mass and damper parameters [80]. With this system, users can simply focus on walking, without thinking of directions and speed control, thereby their mental workload can be considerably reduced. The previous study in Chapter III [83] showed that walking with the robotic walker did not alter the three dimensional trajectories of human CoM, by providing realistic gait patterns. Pelvic restriction was accomplished by assigning infinite virtual damping parameters for the lateral and rotational DoFs. In other words, the infinite damping of lateral and rotational movements were designated to restrict the corresponding pelvic motions, and only the forward-backward movement was allowed when all the pelvic motions were restricted.

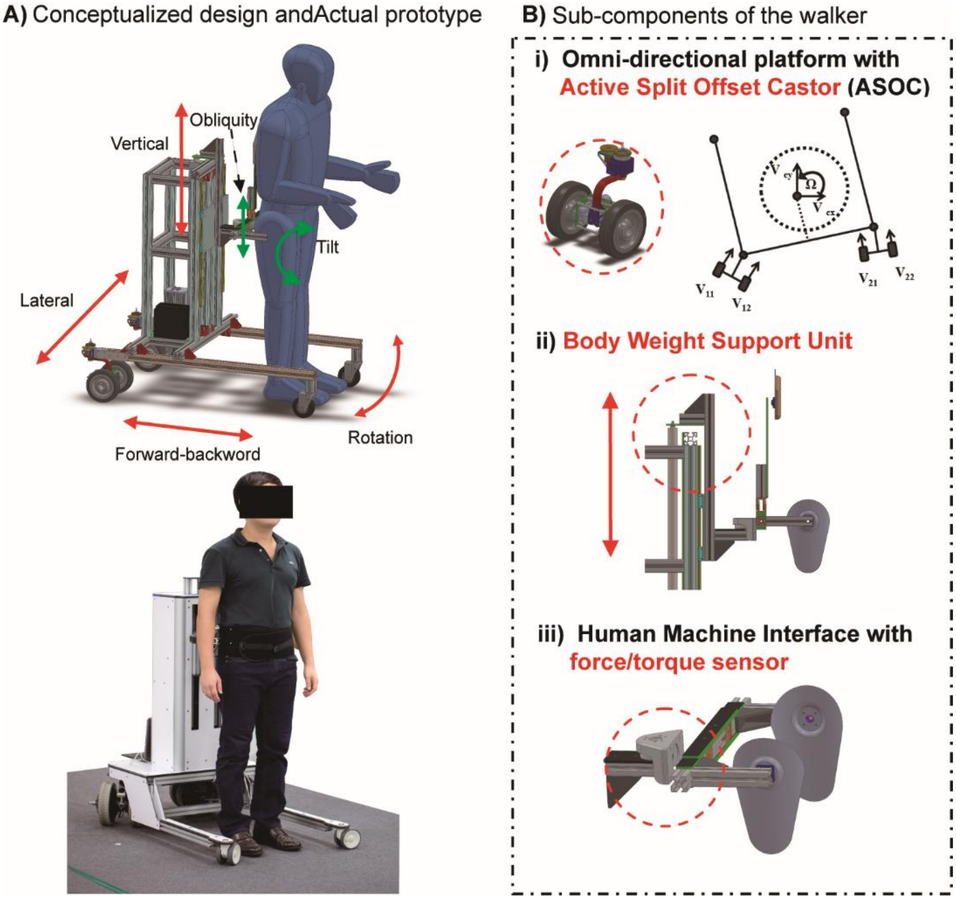


Figure IV-1. A) The conceptualized design and actual prototype of the novel robotic walker for pelvic motion support. B) The system consists of omni-directional mobile platform with ASOC, pelvic and trunk motion support brace unit with active BWS actuator, human-machine interface with FT sensor.

2.2. Participants and experimental design

12 healthy subjects (age: 26.72 ±4.45 years, height: 1706.10 ±93.05 mm, and weight: 63.64 ±13.50 kg) participated in this study. We excluded any subjects with any gait abnormalities or musculoskeletal and neurological disorders. All subjects were instructed to walk along the 30m corridor for 10 minutes prior to the actual trials, to be acclimatized with the walker. The maximum voluntary contractions of the muscles targeted in this study were measured prior to the actual experiment. The actual trials were performed in five different conditions: 1) walking without the walker (normal control, NC); 2) walking with walker, while pelvic lateral and rotational motions are allowed (no restriction, NR); 3)

walking with the walker with only pelvic lateral motion restriction (lateral restriction, LR); 4) walking with the walker with only pelvic rotation restriction (rotation restriction, RR); and 5) walking with the walker, with both pelvic lateral and rotation motion restricted (both restriction, BR). All subjects were instructed to walk naturally with their preferred speed on a 10m walk way. Subjects were asked to repeat the trial until 3 successful strides were achieved. All subjects gave informed consent in accordance with Institutional Review Board standards.

2.3. Data collection and analysis

Data was collected with 8 high speed optical cameras (Vicon, Oxford, UK), and a lightweight wireless EMG (Delsys, Boston, Mass., USA). All instruments were time synchronized. 15 retro reflective markers were attached to the subjects' body landmarks according to the Plug-in-gait marker set. 12 surface wireless EMG electrodes were attached to six major muscle groups – tibialis anterior (TA), gastrocnemius (GA), vastus medialis (VM), biceps femoris (BF), gluteus medius (GM), and adductor longus (AL) with a sampling frequency of 1000 Hz.

A. Joint kinematics and gait descriptive parameters

The raw kinematic data was low-pass filtered via zero-lag 4th-order Butterworth filter with cut-off frequency of 6 Hz to remove motion artifacts and high random noise [97]. The lower limb kinematics such as ankle, knee, and hip joints angles were obtained via motion analysis software (Nexus, Vicon, UK). The RoMs of each joint were calculated based on the joint angles. The gait descriptors including the stride length, step length, step width, gait velocity, and percentage of stance phase were measured based on 3 dimensional coordinates of the heel and toe markers through a customized program in MATLAB (MathWorks, Natick, MA). The stride length was defined as the distance between the positions of heel marker from first initial contact to ensuing initial contact. The step length was defined as the maximum distance between left and right heel markers in the forward direction. The step width was defined as the lateral distance between the positions of ankle markers, while the gait velocity was estimated with a migrated distance of the pelvic markers divided by stride time. The stride length and step length were normalized against the subjects' leg length.

B. EMG activation duration-intensity

The raw EMG signals were first band-pass filtered between 2 and 200 Hz to remove motion artifact and high frequency noise. After rectification of the band pass filtered EMG, all EMG data was normalized against the respective maximum voluntary contraction (MVC) value which was measured prior to the gait experiment. The low pass filter with a cut-off frequency of 10 Hz was used to produce a linear envelope representation. The enveloped EMG data was then used to quantify the duration-intensity of muscle activity. Amplitudes of enveloped EMG were then classified into 5 groups according to the relative intensity of the selected group over MVC [98, 99], i.e. 10%, 20%, 30%, 40%, and 50% intensities of muscle activation. Activation durations of each classified EMG were calculated during the earlier 5 categorized groups of the muscle intensity, for example, TA muscle duration-intensity was defined as: TA10, TA20, TA30, TA40, and TA50 respectively. The same procedure was repeated for the remaining muscles data; GA, VM, BF, GM, and AL.

2.4. Statistical Analysis

Statistical analysis (SPSS Inc., Chicago, IL) was conducted in order to identify the gait changes according to the five conditions mentioned above. One way ANOVA was performed to distinguish the gait kinematic parameters as well as the differences in muscle activation duration-intensity parameters among the experimental conditions. For the cases exhibiting significant differences at ANOVA, Tukey's post hoc test was performed subsequently. All significance levels were set at $p < 0.05$.

3. Results

3.1. Gait descriptive parameters

Table 6 shows the gait descriptive parameters according to the experimental conditions mentioned. The normalized stride and step length, and gait velocity showed significantly lower values for NR, LR, RR, and BR, as compared to the NC ($p < 0.001$). The NR showed significantly longer stride and step length as compared to LR and BR ($p < 0.05$). However, there was no significant difference in step width among the conditions. The percentage of stance phase was prolonged in the condition of LR, RR ($p < 0.001$), and BR ($p < 0.05$) compared with NC.

Table 6. The gait performance parameters during pelvic restriction walking.

	NC	NR	LR	RR	BR
Norm. Stride Length	1.410 ±0.112	1.055 ±0.214**	0.806 ±0.263**,ǂ	0.951 ±0.215**	0.866 ±0.227**,ǂǂ
Norm. Step Length	0.679 ±0.063	0.543 ±0.105	0.470 ±0.243ǂ	0.507 ±0.108*	0.481 ±0.101**
Step Width (m)	0.143 ±0.019	0.122 ±0.056	0.146 ±0.055	0.130 ±0.057	0.106 ±0.052
Gait Velocity (m/s)	0.977 ±0.186	0.524 ±0.175**	0.445 ±0.172**	0.487 ±0.168**	0.463 ±0.179**
% of Stance Phase	61.34 ±1.68	64.11 ±5.17	70.70 ±7.78**,ǂǂ	66.09 ±6.08*	67.94 ±8.23**

* Statistical difference between NC and NR, LR, RR, and BR, P<0.05
** Statistical difference between NC and NR, LR, RR, and BR , P<0.01
ǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.05
ǂǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.01

3.2. Kinematic profiles and range of motions (RoMs)

Figure IV-2 and Table 7 show kinematic and RoMs of each joint are shown. There were significant reductions in ankle plantarflexion at terminal stance, knee flexion at mid-swing, and hip extension at mid-stance, in LR, RR, and BR conditions as compared to NC ($p < 0.001$), (Table 7). The ankle, knee, and hip joint angular profiles of NR condition resembled those of NC, and showed no significant deviation from normal walking. The RoMs of ankle, knee, and hip joints were reduced for all conditions as compared to NC. However, significant reduction of ankle RoM in LR and BR ($p < 0.05$), knee RoM in LR, RR, and BR ($p < 0.05$), and hip RoM in LR and BR ($p < 0.05$) were found as compared to the NR. These reductions showed that a fixation of the pelvic LD and RT movements constrained the RoMs of ankle, knee, and hip joints by reducing and altering their excursion during gait.

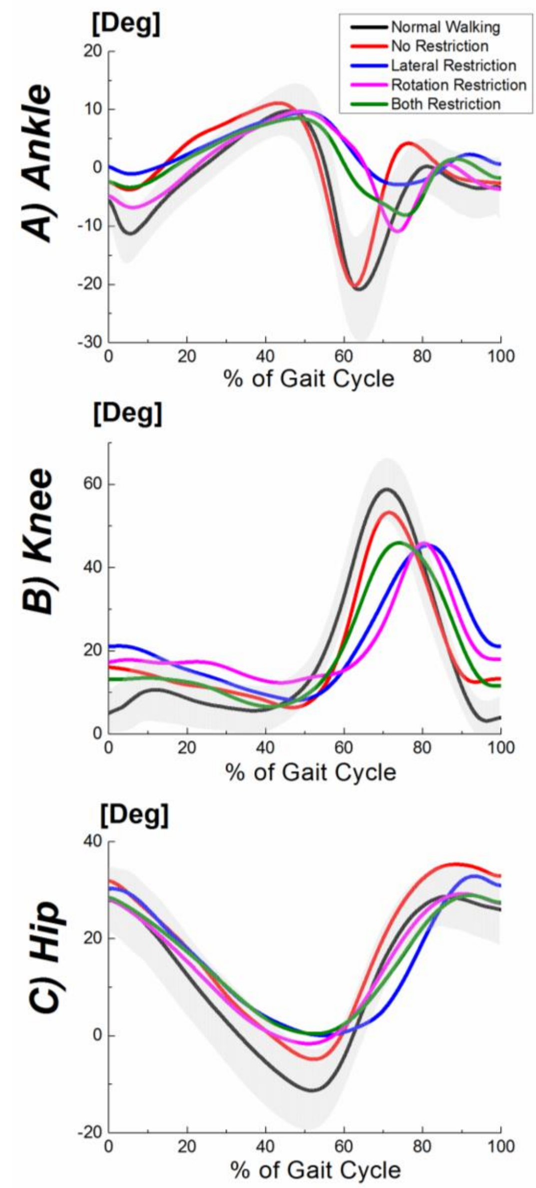


Figure IV-2. Comparison of A) Ankle, B) knee, and C) hip joint kinematics profiles during the gait under the condition of NC (black), NR (red), LR (blue), RR (pink), and BR (green).

Table 7. Range of motions of ankle, knee, and hip joints during walking.

	NC	NR	LR	RR	BR
Ankle Dorsiflextion	10.36 ±4.21	10.40 ±3.35	11.55 ±3.59^{**,ǂ}	9.63 ±3.68**	10.45 ±2.64^{**,ǂ}
Ankle Plantarflexion	22.79 ±8.53	16.84 ±7.59	4.36 ±10.88	11.63 ±9.69	8.26 ±9.61
Ankle RoM	33.15 ±6.16	27.24 ±7.91	15.91 ±11.36^{**,ǂ}	21.26 ±9.62**	18.71 ±9.87^{**,ǂ}
Knee Flexion	59.76 ±7.16	54.28 ±8.10	47.92 ±12.02**	46.25 ±12.65^{**,ǂ}	46.49 ±11.04^{**,ǂ}
Knee Extension	1.97 ±4.52	4.22 ±6.22	5.84 ±6.18*	4.09 ±5.50	4.67 ±5.10
Knee RoM	57.79 ±5.30	49.76 ±6.64*	42.08 ±9.11^{**,ǂ}	42.36 ±10.80^{**,ǂ}	41.82 ±9.32^{**,ǂ}
Hip Flexion	30.15 ±5.64	32.24 ±6.57	30.97 ±4.97	30.50 ±4.57	30.99 ±5.20
Hip Extension	11.55 ±8.41	4.41 ±7.48*	1.04 ±6.97**	2.95 ±7.89**	1.97 ±7.31**
Hip RoM	41.69 ±4.24	36.64 ±5.68*	32.01 ±5.94^{**,ǂ}	33.45 ±6.02**	32.97 ±6.70^{**,ǂ}

* Statistical difference between NC and NR, LR, RR, and BR, P<0.05
** Statistical difference between NC and NR, LR, RR, and BR , P<0.01
ǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.05
ǂǂ Statistical difference between NR and NC, LR, RR, and BR, P<0.01

3.3. Duration-intensity of EMG activation

Muscle activation profiles and activation duration-intensity are shown in Figure IV-3 and 4. The black line shows the average muscle activation profile, and gray line shows standard deviation of the NC. No significant alteration between NR and NC was detected. However, the TA muscle in LR and BR conditions showed remarkable over-activation from the mid-stance to terminal swing. The profile of the GA muscle showed prolonged activation patterns in the mid-stance, exhibiting an enlarged stance phase (Table 6.) in the LR and BR conditions. In addition, the BF muscle in the RR condition showed significantly higher activation during the stance phase. These results can be statistically quantified through EMG duration-intensity graphs as shown in Figure IV-4. The EMG duration-intensity for TA and BF during stance phase showed significantly higher value in LR and BR, compared to NC ($p < 0.05$). TA10, TA20, TA30, TA40 were significantly increased in LR and BR conditions, while BF10 was increased in RR condition compared with NC in the stance phase. For the swing phase, TA20 and TA30 showed significantly larger activation duration in LR compared to NC and NR conditions. Finally, GA10 and GA20 in LR and BR conditions were significantly increased compared to NC ($p < 0.01$) and NR ($p < 0.05$). No statistical difference was found in VM, GM, and AL muscles.

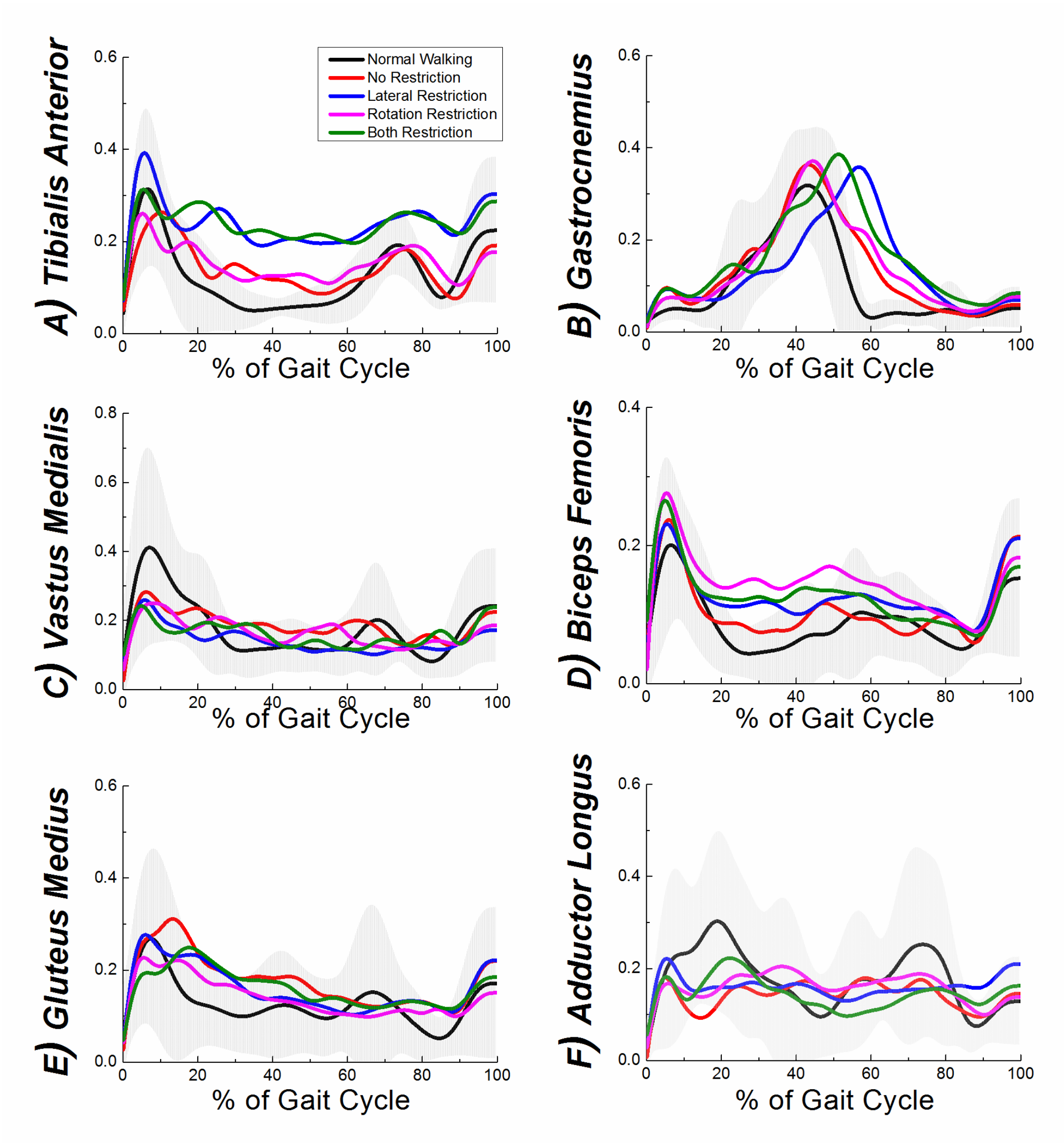


Figure IV-3. Averaged and enveloped surface EMG profiles for 6 major muscles; A) TA, B) GA, C) VM, D) BF, E) GM, and F) AL under walking conditions. The black and gray lines show the NC and its standard deviation. Red, blue, pink, and green lines show the NR, LR, RR, and BR, respectively.

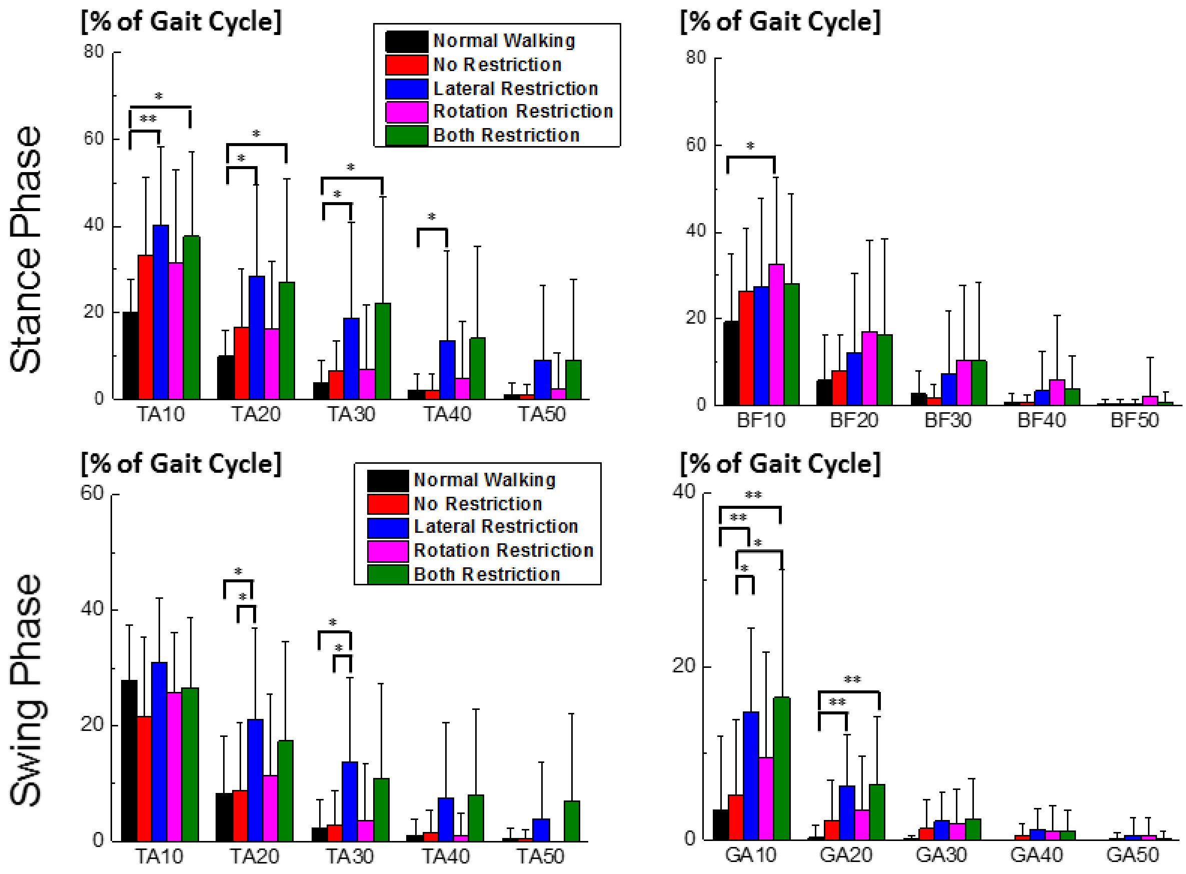


Figure IV-4. The EMG duration-intensity results according to the experiment conditions. Black bar shows the condition for NC, red for NR, blue for LR, pink for RR, and green for BR.

4. Discussion and Conclusion

From the perspective of gait kinematics and muscle activation, we confirmed significant reductions in gait performances and RoMs of kinematic profiles as well as increases in muscle activation patterns when the pelvic LD and RT were restricted during over-ground walking.

The gait descriptive parameters, especially the normalized stride and step length, showed significant reductions in LR and BR conditions compared with NR. It is important to note that the stride and step length were severely affected by restricting LD rather than RT. However, these results differ from the outcomes on the pelvic fixation during treadmill walking as reported by Venenam et al [93]. The previous study reported that a pelvic horizontal restriction resulted in a longer step length, and claimed that pelvic fixation could be helpful in increasing step length in actual clinical trials. The difference between the results of our study and the previous study could be attributed to variations in experimental designs. It might be due to either different walking mechanisms between walking on treadmill and over-ground or different methods for restricting pelvic motions during experimental trials. Specifically, walking on a treadmill can be different from over-ground walking, showing an increase in cadence and decreases in stride length and joint excursion on a treadmill [37]. In addition, the pelvic fixation was implemented using a waist girdle connected to a frame of the treadmill in the previous study, while the pelvic restriction was accomplished by assigning infinite virtual damping parameters for LD and RT movements in this study. Despite the inconsistency between these two studies, our results suggest that pelvic restriction significantly affected the gait descriptive parameters by not only reducing the normalized stride and step length, and gait velocity, but also increasing percentage of the stance phase.

In terms of lower limb kinematics, the pelvic restrictions caused limited ankle plantarflexion at the terminal stance, knee flexion at mid-swing, and hip extension at mid-stance, contributing to the reduction of RoMs in all lower limb joints. These reductions in RoMs and altered kinematic profiles might have caused other abnormal gait patterns in the subjects due to compensatory strategies [66]. Furthermore, since the ankle RoM is a key factor in gait efficiency for cerebral palsy patients [75], the reduced RoM in the ankle joint due to pelvic restriction might hinder the efficiency and performance of the actual clinical trials.

We identified the muscle activation patterns in terms of both duration and intensity of major muscles in lower limbs. The muscle activation patterns provided mechanistic causes of kinematic patterns and showed clear indication of the biomechanical effects of the walker with and without pelvic motion facilitation. It appeared that the EMG amplitude profiles of NR were not significantly altered from those of NC. On the other hand, a significant increase in EMG activation duration-intensity at the TA muscles was found in LR and BR compared to that of NC in both the stance and swing phases. However, there was no significant difference between the NR and RR. From the results described above, the pelvic lateral restriction may cause body load concentration towards the stance limb for weight acceptance. Consequently, the applied body load may have caused an increased muscle activation intensity and duration at the TA muscle during the single limb stance period as the subjects were trying to maintain sagittal plane balance. In addition, the GA muscle duration-intensity was significantly increased with prolonged EMG profiles in LR and BR conditions, especially in the swing phase. The prolonged GA activation is an indication of a raised stance period (Table 7) to stabilize the gait patterns caused by LR and BR walking. Furthermore, the pelvic rotational restriction caused an increased activation duration in the BF muscle (BF10). This might have been attributed to a compensation for the increased knee flexion during mid-stance (Figure IV-2), consequently the BF muscle was over-activated to maintain the flexed knee motion and to achieve locomotion against the pelvic rotational restriction.

In conclusion, gait with pelvic motion facilitation can elicit normal muscle activation patterns in a natural manner without altering normal gait dynamics. On the other hand, gait with pelvic restriction severely affected gait dynamics by reducing the gait performances with significant decreases in the RoMs of the ankle, knee, and hip angles, and increases in lower limb muscles activation duration-intensity. These alterations will hinder the subjects from learning natural gait patterns and having correct afferent sensory input and sensory feedback which are the most critical factors for a successful gait training. Therefore, the efficacy or functional outcome after gait rehabilitation can be significantly reduced if the pelvic lateral and rotational motions are restricted. Additionally, the pelvic restriction may result in a higher metabolic cost for gait training as it requires increased and prolonged muscle activities [89]. These findings can serve as a cornerstone of the further development of robotic gait rehabilitation by providing clear evidence of the pelvic LD and RT restriction on the lower limb gait dynamics.

With a growing interest in pelvic motion support for gait rehabilitation, the effects of pelvic motion restriction on the lower limb during over-ground walking in terms of gait kinematics, gait descriptive parameters, and muscle activation were investigated in this study. However, there were a few design parameters that could be improved by incorporating the role of trunk motion against the pelvic restrictions as a future scope of this work. It should be noted that the main contribution of this study is to examine the biomechanical effects of the pelvic LD and RT restrictions on the lower limb, and we believe that this study would help the scientific community to develop a better understanding on the pelvic restriction during over-ground walking. While the results of this study showed the necessity of pelvic motion facilitation in gait rehabilitation, an experiment with healthy young subjects may not be enough to provide clinical influence for neurologically challenged patients. As a future scope of research, we are looking to address this issue by conducting preliminary experiments with neurologically inhibited patients. More sophisticated robotic gait rehabilitation will be possible when therapists are aware of the results of this study and its clinical verification.

CHAPTER V. BIOMECHANICAL EFFECTS OF BODY WEIGHT SUPPORT WITH A NOVEL ROBOTIC WALKER FOR OVER-GROUND GAIT REHABILITATION

1. Introduction

Impairments in the musculoskeletal system after neurological disorders may hinder individuals with gait abnormalities from performing regular activities of daily living. This abnormality results in gait inefficiency, mainly due to the loss of strength and balance, and coordination of the limbs [12, 79, 100]. Gait rehabilitation aims at restoring basic locomotive function and can be achieved through an appropriately designed regime that includes over-ground walking training, body weight support (BWS), and strength training [101, 102]. Especially in clinical settings, BWS gait training promotes better functional outcomes for neurologically challenged patients, by reducing gravitational force acting on the body, and by increasing stability, comfort, and the patients' balance with external assistance [11, 103, 104]. In a study investigating the effects of BWS treadmill training on locomotion with 100 stroke survivors, Barbeau and his colleague concluded that gait training with BWS significantly increased all clinical outcomes in terms of walking speed, endurance, balance, and motor recovery, proving that BWS is an effective training method for improving gait and postural abilities [105]. Similarly, Peurala et al. studied the effectiveness of BWS gait training, and concluded that gait descriptors were largely improved in the group using BWS after intensive rehabilitation training [106]. Furthermore, promising results in gait parameters after BWS training have been reported in patients with neurological disorders such as spinal cord injuries [107-109], cerebral palsy [101, 110-114], and Parkinson's disease [18, 115, 116].

Although an improvement in gait functionality has been clearly documented, the biomechanical effects of BWS still remain contradictory in both treadmill and over-ground based rehabilitation devices. Threlkeld et al. reported decreased cadence, stance phase and double limb support (DLS) time with increased step length at 50% and 70% levels of BWS as compared to 0% BWS, while Lewek reported that use of BWS did not alter how stance time and step length were manipulated during treadmill walking [103, 104]. In addition, Van Hedel et al. reported abnormal over-activated muscle profile changes at rectus femoris, lateral and medial hamstring, and vastus medialis during mid-stance with increased BWS level[117], while Burnfield et al. found reduced EMG mean activation at the gluteus medius and vastus lateralis, and decreased duration at soleus in an BWS environment without differences in flexor muscles [118]. An explanation for the inconsistencies in biomechanical effects of BWS is that there is a strong reliance on the devices used for the experiments, implying that the role of stance phase stability is shared with the mechanical structure of the devices. Another possible reason for the inconsistency observed may be the use of an overhead harness in the treadmill-based BWS device which causes a different gait pattern from an actual over-ground gait. It has been reported that walking on a treadmill can lead to greater cadence, forward tilted trunk motion, and an increased vertical acceleration [119], which potentially affects gait functional outcomes as a result of different sensorimotor feedback and proprioceptive input compared to over-ground walking, especially for older patients and stroke survivors [37, 96, 120]. Moreover, an overhead harness BWS system can often cause restriction to pelvic motion in the horizontal plane, such as pelvic rotation and lateral displacement. These can severely deviate the gait kinematics and temporospatial parameters from normative patterns [11, 93]. Celestino et al. reported that BWS with over-ground walking can significantly increase the gait functional outcomes compared to BWS with treadmill walking, showing that walking over the ground promoted gait patterns of Cerebral Palsy patients more similar to their typically developing peers, while greater instability was observed during treadmill walking compared to over-ground walking [110].

Therefore, the use of a BWS system allowing pelvic motion facilitation and over-ground walking has two significant advantages: it provides a better clinical application and provides an understanding of the biomechanical effects of BWS to guide an intervention during gait rehabilitation. The effects of BWS during self-controlled over-ground gait has been studied with regards to walking speed in both healthy elderly populations and individuals with chronic stroke, using a robotic gait device called KineAssist [102]. Self-selected gait speed decreased with increased levels of BWS for the healthy subjects, whereas gait speed was increased by 18% for the post-stroke subjects with increased BWS level compared to the 0% BWS condition. These findings imply that if BWS were used at appropriate levels, it would provide an objective and reliable gait rehabilitation tool for physical therapists in a clinical setting. However, both research projects [102, 110] investigated kinematic or temporospatial gait parameters without looking at the patterns of muscle activity.

To fill the knowledge gaps in the previous studies, a novel robotic over-ground walking system (Robotic Walker) with pelvic motion support has been developed (Figure V-1). This robotic walker allows for active pelvic anterior-posterior (AP), medio-lateral (ML), rotational (RT), and vertical (VT) movements, and for passive pelvic tilt and obliquity without altering normal gait patterns. The details of the walker are described in the next section. Therefore, the aim of this study is to comprehensively investigate the biomechanical effects of BWS in terms of gait kinematics, temporospatial parameters, and muscle activation in healthy subjects using the Robotic Walker.

One important criterion for effective and successful BWS training is the minimization of step-to-step transition (SST) cost in single limb and double limb support periods, to which two thirds of the metabolic cost is attributable [121, 122]. Another critical factor is that the load applied to the lower limb to support body weight during the stance phase and to move the limb in the swing phase should be reduced without altering normal muscle activation patterns [48]. In this regard, we hypothesized that an increased level of BWS will 1) reduce stance phase temporal parameters including single limb support and double limb support time, 2) reduce muscle activation time and amplitude without altering normative patterns, and 3) increase the lateral stability which can be shown by reduced amplitudes of hip ab/adductor muscle.

2. Methods

2.1. The robotic walker with BWS system

A BWS module is implemented on top of the robotic walker (Figure V-1A) [83]. This walker is capable of supporting a patient's pelvic motion through six DoFs for natural gait patterns. The use of a BWS system which provides active unloading of the body mass of the subject to the desired percentage with unrestricted pelvic motion is proposed for effective BWS training (Figure V-1B). Therefore, the robotic walker allows the pelvis and trunk to move vertically with pelvic AP, ML, and RT movements, as well as pelvic tilt and obliquity. The BWS actuator provides all-in-one control through a proportional-integral-derivative (PID) controller, drive, and motor integrated into one compact component; the active body weight of the subject is maintained via the vertical axis of the force/torque (FT) sensor during dynamic walking.

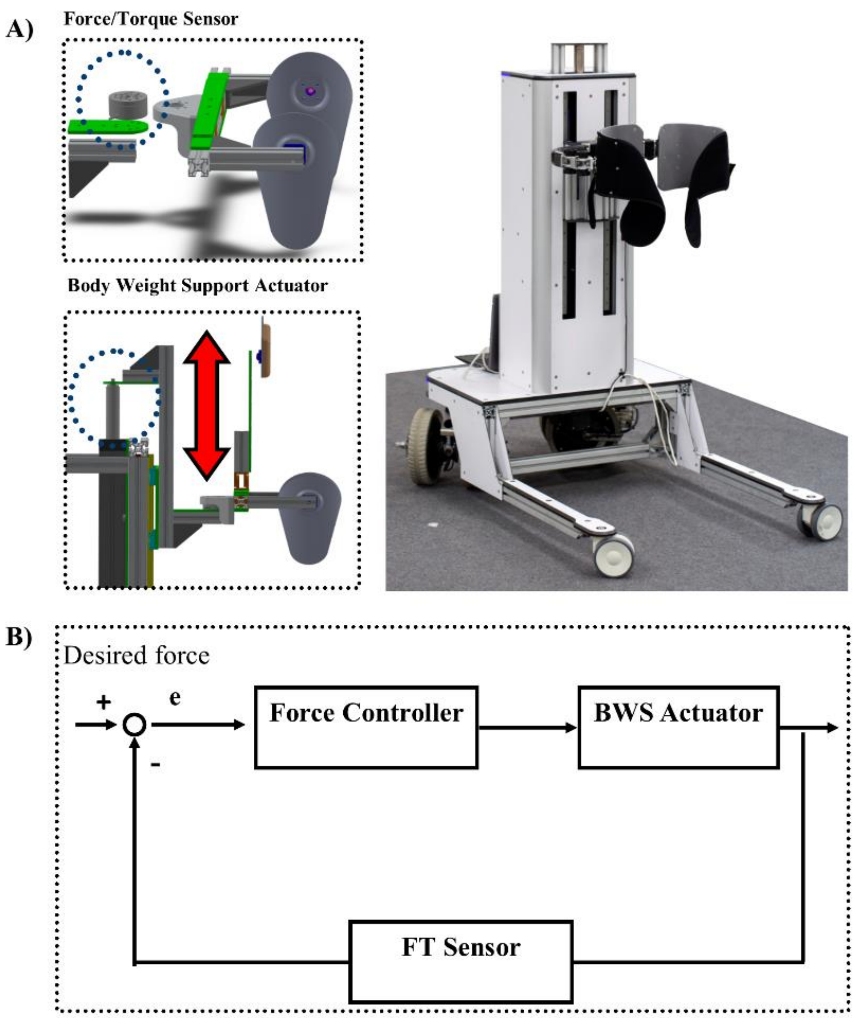


Figure V-1. Robotic Walker and Control of the BWS unit

2.2. Subjects and experimental protocol

Ten healthy young subjects (age: 25.1 ± 4.4 years old, weight: 62.1 ± 9.1 kg, height; 168 ± 5.0 cm) with no known history of gait disorders, lower extremity injuries, and neurological disease participated in this study. All subjects signed a consent form which was approved by the Institutional Review Board of the National University of Singapore.

Eight high speed motion capture cameras (Vicon, Oxford, UK) and a sixteen channel wireless bi-polar electromyography (EMG) were used to obtain gait kinematic data, temporospatial parameters, and muscle activation data. Fifteen retroreflective optical markers were placed on the subject's pelvis and lower limb (sacrum, left and right anterior superior iliac spine, thigh, knee, tibia, ankle, heel, and toe). The precise 3D positions of the markers were recorded with a sampling frequency of 100 Hz on Vicon Nexus software (Nexus 1.8.3, Vicon, Oxford, UK). Nine EMG (Delsys, MA, USA) sensors were attached to the tibialis anterior (TA), gastrocnemius (GA), soleus (SOL), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), gluteus maximus (G_{Max}), gluteus medius (G_{Med}) and adductor longus (AL) according to the SENIAM protocol to quantify muscle activity with sampling frequency at 1000 Hz [123]. All of the experimental instruments were time synchronized. With all of the instrumentation in place, the subjects were first instructed to walk around with the Robotic Walker for 5 to 10 minutes to become acclimatized. In the actual experiment, the subjects were instructed to walk on a 10m distance walkway with incremental amount of BWS forces from 0% to 40% of body weight at 10% intervals. The experimental conditions consisted of the 0%, 10%, 20%, 30%, and 40% BWS level (Figure V-2). All subjects were asked to walk through more than three successful trials with different experimental conditions.

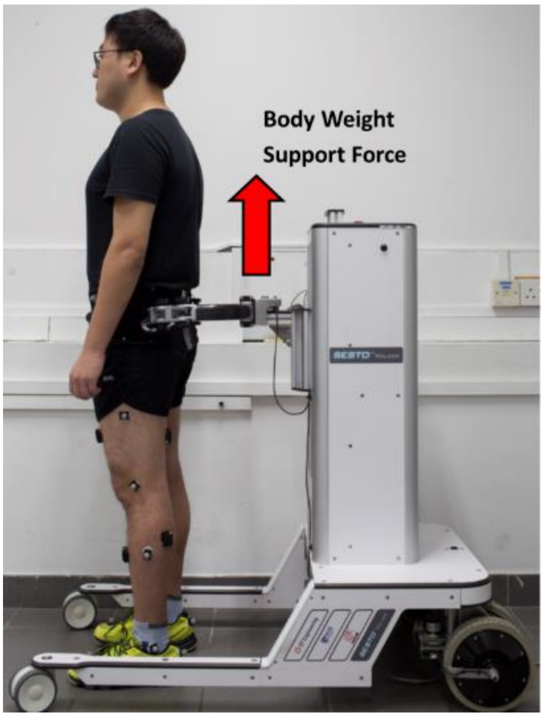


Figure V-2. Provision of body weight support force

2.3. Data analysis and statistics

A. Kinematic parameters and temporospatial gait parameters

3D trajectories of 15 optical markers were low-pass filtered through a zero-lag 4th order Butterworth filter with a cut-off frequency of 6 Hz [97]. During each trial, heel strike (HS) and toe-off (TO) events were determined by using the vertical trajectory and velocity of the foot [124]; two strides in the middle of the walkway were used to calculate temporospatial gait parameters for each trial. Ankle, knee, and hip joint angles in the sagittal plane and their minimum, maximum, and range of motions (RoMs) were calculated from customized software based on the positions of each marker, and the 3D trajectories of the temporospatial gait parameters were further analyzed using MATLAB. The spatial parameters (stride length, step length, step width) were calculated based on the distance of markers on the left and right foot, and the temporal parameters (gait velocity, stride time, stance time, swing time, single limb support time (SLS), double limb support (DLS), and percentage of stance phase) were determined based on HS and TO time. The stride and step length were normalized over the subjects' leg length.

B. EMG Analysis

EMG signals were firstly band-pass filtered with a zero-lag 4th order Butterworth filter with a cut-off frequency between 2 and 200 Hz. The band-passed signals were rectified, and then normalized by a maximum voluntary contraction value which was collected from overall experimental trials to obtain relative amplitude of the signal. The rectified and normalized EMG signals were then low-pass filtered at 10 Hz for a linear envelope, which represented the instantaneous amplitude of the signal [99]. The enveloped signals were then used to calculate activation amplitude and duration for quantitative analysis. The mean EMG amplitude was determined in the overall gait cycle, stance, and swing phase. The temporal signals were considered activated if the relative amplitude exceeded 20%, otherwise the signals were defined as inactive. Finally, the EMG activation durations were calculated in the overall gait cycle, stance, and swing phase.

C. Statistics

One way ANOVA was used to test significance among the conditions for the joint kinematics, temporospatial gait parameters, and EMG mean amplitude and duration with Tukey's post hoc analysis to contrast differences among the experimental conditions.

3. Results

3.1. Provision of BWS force with the robotic walker

Figure V-3 shows the amount of vertical force acting on the FT sensor with increasing levels of BWS (i.e. 0%, 10%, 20%, 30%, and 40% of BW). As expected, with increasing BWS, subjects exerted a significantly greater unloading force for vertical movement ($p < 0.001$).

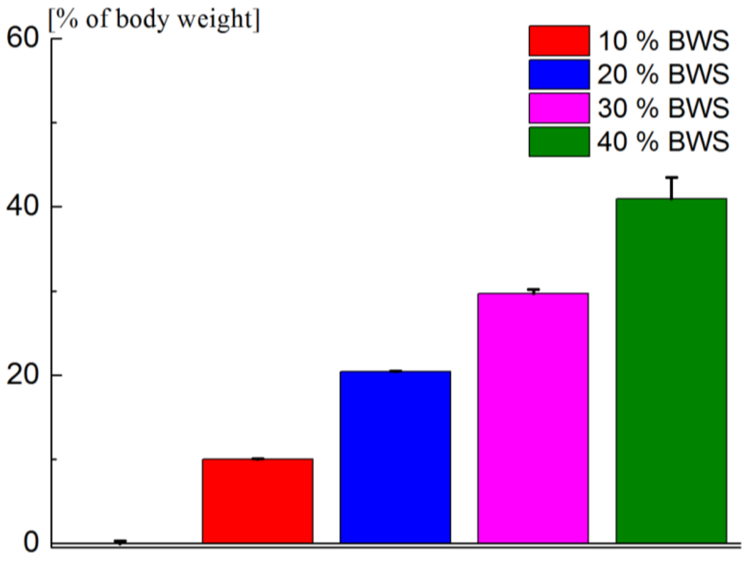


Figure V-3. Amount of vertical force exerted on FT sensor with increasing level of BWS

3.2. Gait kinematics

Table 8 shows the minimum, maximum, and RoMs of ankle, knee, and hip joint angles. No significant differences were found in ankle joint readings with increasing BWS level. However, maximum knee flexion and RoM were significantly reduced at 40% BWS compared to 0% BWS ($p < 0.001$). Similarly, the maximum hip extensions were reduced at 20%, 30%, and 40% BWS ($p < 0.001$) and hip RoMs were also diminished at 30% and 40% BWS in comparison with 0% BWS ($p < 0.001$) (Table 8).

Table 8. Minimum, maximum, and range of motions of ankle, knee, and hip joint angles

	Ankle			Knee		Hip		
Conditions	**Plantar-flexion**	**Dorsi-flexion**	**RoM**	**Flexion**	**RoM**	**Flexion**	**Extension**	**RoM**
0% of BWS	-11.41±9.06	10.37±4.95	21.78±9.52	52.28±8.33	52.38±6.91	27.32±8.16	-7.15±7.37	34.47±4.76
10% of BWS	-15.13±9.89	10.11±5.17	25.24±10.28	49.98±9.06	49.62±7.51	29.81±9.03	-3.46±7.35	33.27±5.76
20% of BWS	-13.71±10.01	8.37±4.67	22.08±9.16	48.06±8.69	48.90±8.13	30.47±7.87	0.19±6.62**	30.28±6.03
30% of BWS	-15.02±10.05	8.63±4.73	23.66±9.82	47.80±8.86	48.32±8.01	29.17±6.65	1.19±6.78**	27.98±5.44**
40% of BWS	-14.02±10.04	8.62±4.71	22.64±8.98	46.33±9.41**	47.27±8.78**	29.41±7.35	2.43±8.31**	26.98±6.90**

* Statistical difference from 0% BWS, $p < 0.05$
** Statistical difference from 0% BWS, $p < 0.001$

3.3. Temporospatial Gait Parameters

Temporospatial gait parameters for the different conditions in accordance with BWS level are shown in Table 9. Gait spatial parameters such as normalized stride and step length, and step width were not influenced, while temporal parameters were significantly altered by the level of BWS.

Specifically, as BWS level increased, a significant increase in gait velocity at 10% and 40% BWS and shortened stride and stance time at 10%, 20%, 30%, and 40% BWS were observed compared to 0% BWS ($p < 0.001$). In addition, swing time was longer at 40% BWS; SLS time, DLS time and percentage of stance phase at 10% to 40% of BWS were significantly shortened compared to 0% BWS ($p < 0.001$).

Table 9. Temporospatial parameters with increasing level of BWS

Conditions	0% of BWS	10% of BWS	20% of BWS	30% of BWS	40% of BWS
	Mean ± SDT	Mean ± STD	Mean ±STD	Mean ± STD	Mean ± STD
Normalized Stride Length	0.97 ± 0.14	0.98 ± 0.17	0.96 ± 0.15	0.98 ± 0.14	0.98 ± 0.14
Normalized Step Length	0.51 ± 0.06	0.51 ± 0.08	0.49 ± 0.08	0.50 ± 0.06	0.50 ± 0.07
Step Width(m)	0.14 ± 0.03	0.14 ± 0.05	0.14 ± 0.04	0.15 ± 0.04	0.15 ± 0.05
Velocity (m/s)	0.366 ± 0.07	0.413 ± 0.14*	0.395 ± 0.06	0.406 ± 0.06	0.413 ± 0.06*
Stride Time (s)	2.31 ± 0.24	2.06 ± 0.28**	2.04 ± 0.17**	2.05 ± 0.25**	2.00 ± 0.16**
Stance Time(s)	1.61 ± 0.21	1.34 ± 0.31**	1.29 ± 0.19**	1.29 ± 0.21**	1.21 ± 0.15**
Swing Time (s)	0.69 ± 0.12	0.72 ± 0.21	0.75 ± 0.12	0.77 ± 0.15	0.80 ± 0.16**
SLS Time	1.15 ± 0.19	0.99 ± 0.31*	0.97 ± 0.26*	0.99 ± 0.16*	0.91 ± 0.43**
DLS Time	0.47 ± 0.16	0.35 ± 0.15*	0.33 ± 0.21*	0.30 ± 0.17**	0.32 ± 0.32*
% of Stance Phase (%)	69.83 ± 4.46	64.82 ± 0.69**	63.19 ± 5.66**	62.61 ± 5.82**	60.30 ± 6.79**

* Statistical difference from 0% BWS, $p < 0.05$
** Statistical difference from 0% BWS, $p < 0.001$

3.4. EMG parameters

Figure V-4 depicts the enveloped EMG profiles in the lower extremity muscles, recorded at self-selected speed for designated BWS conditions. Clear indication of linearly decreased muscle activation patterns can be observed in most of the major muscles, especially for ankle dorsi/plantar flexion, and hip flexion/extension. For comprehensive analysis of the influence of varying amounts of BWS unloading on EMG activity, the changes in both amplitude and duration were computed in this study.

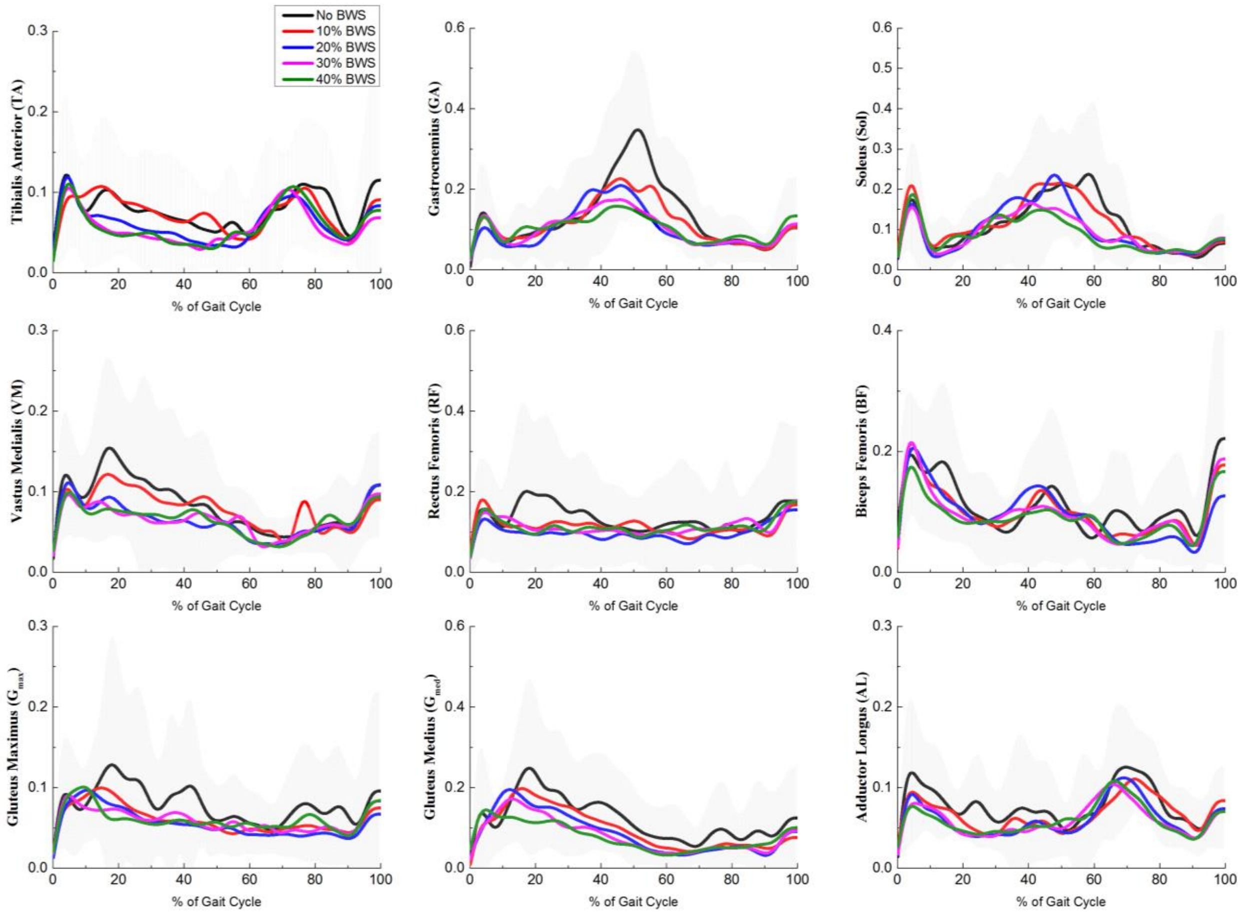


Figure V-4. Enveloped EMG profiles from 9 major muscles during walking with BWS. The black line and gray line show averaged EMG profiles and its' standard deviation in 0% BWS. Red, blue, pink, and green line shows the enveloped EMG profiles in 10%, 20%, 30%, and 40% BWS, respectively.

A. Mean amplitude of EMGs

For the quantitative measurements of amplitude domain, the mean EMG activation in overall gait cycle (Figure V-5), stance and swing phase (Table 10) are indicated. In overall gait cycle, TA mean amplitude at 30% and 40% BWS, and GA at 20%, 30%, and 40% were significantly reduced compared to 0% BWS ($p < 0.05$). In addition, the G_{Med} (30% BWS ($p < 0.05$), and 40% BWS ($p < 0.001$)) and AL (20%, 30%, and 40% BWS ($p < 0.05$)), contributing to hip ab/adduction, were significantly reduced. The mean EMG amplitude in the stance phase showed similar patterns to that in the overall gait cycle, but AL muscle amplitude in the swing phase was significantly reduced at 30% and 40% ($p < 0.05$) compared to 0% BWS. Despite significant reductions found in ankle dorsi/plantar flexor and hip flexor/extensor muscle activation, no significant differences were observed in Sol, VM, RF, BF, and G_{Max} muscles.

B. Activation duration of EMGs

For the quantitative measurements of the temporal domain, the activation duration of EMG in the stance and swing phase was calculated (Figure V-6 and Table 11). During the stance phase, for ankle joint plantarflexor and dorsiflexor, duration of TA at 30% and 40% of BWS ($p < 0.001$), and GA at 20%, 30%, and 40% of BWS ($p < 0.001$) were significantly shortened as compared to 0% BWS. For knee extensor and hip flexor, the duration of VM at 40%, and RF at 20% was significantly shortened as compared to 0% of BWS. Likewise, hip joint flexor/extensor and ad/abductor, G_{Max} in all conditions ($p < 0.001$), G_{Med} in 30% and 40% ($p < 0.05$), and AL in all conditions ($p < 0.001$) were significantly reduced compared with 0% of BWS. In addition, during the swing phase, slight but significant changes were found at GA at 20%, 30%, and 40% BWS, Sol in 40% BWS, BF in 30% BWS, and G_{Med} in 30% of BWS (Table 11).

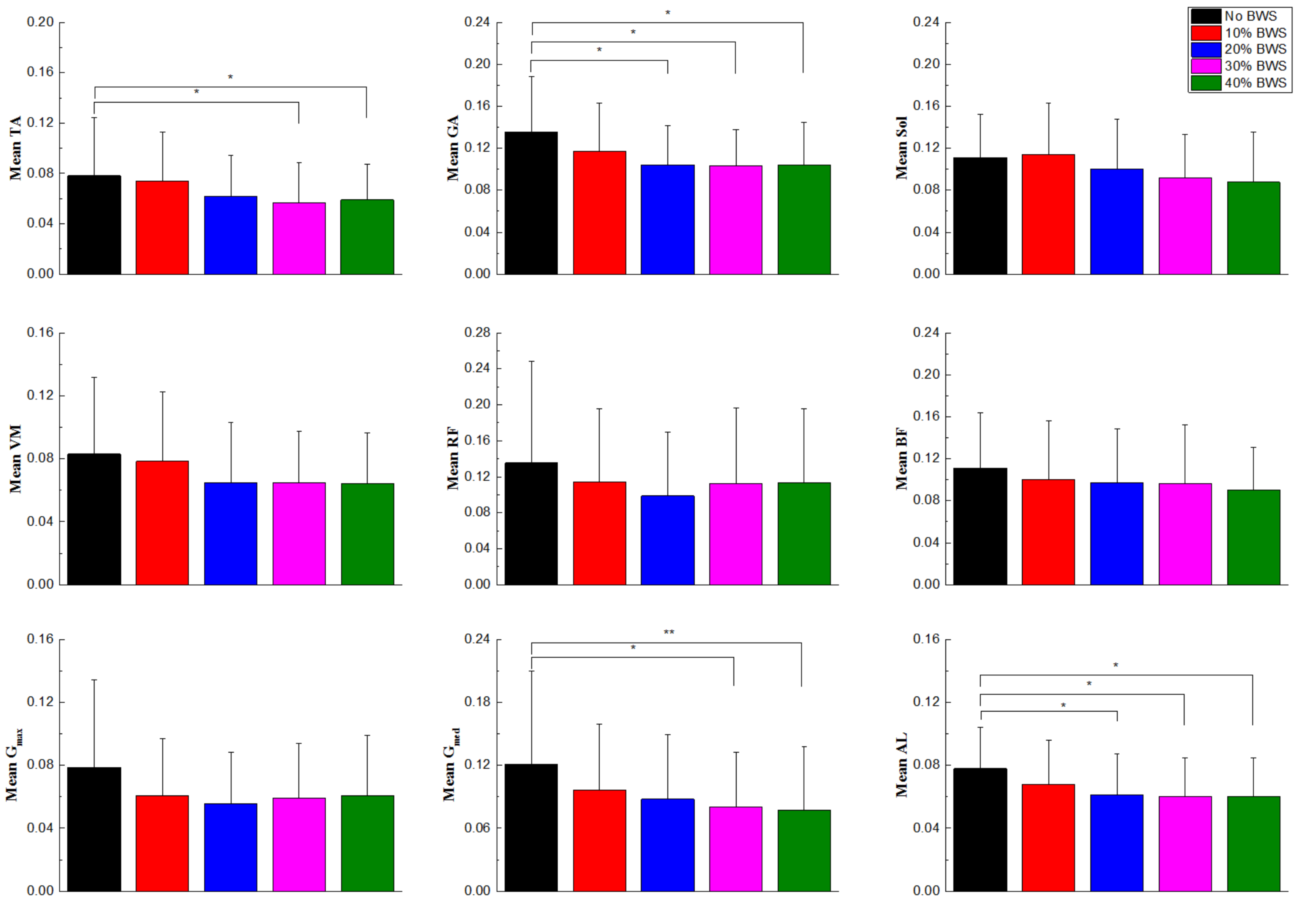


Figure V-5. Averaged EMG amplitude from 9 major muscles during walking with BWS. The black bar shows averaged EMG amplitude in overall gait cycle at 0% BWS. Red, blue, pink, and green bars show the EMG amplitudes in 10%, 20%, 30%, and 40% BWS, respectively.

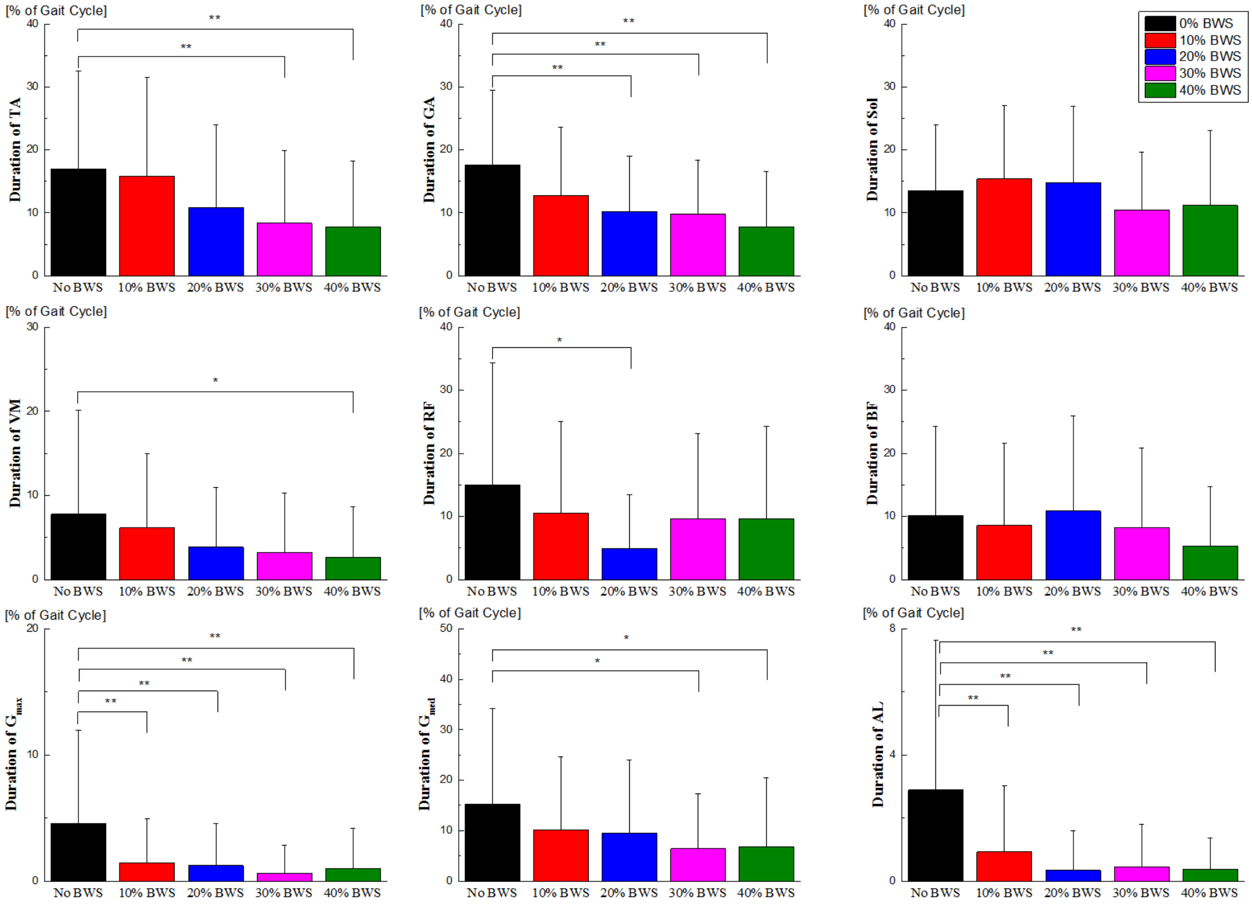


Figure V-6. Averaged EMG activation duration from 9 major muscles during walking with BWS. The black bar shows averaged EMG activation duration in stance phase at 0% BWS. Red, blue, pink, and green bars show the EMG amplitudes in 10%, 20%, 30%, and 40% BWS, respectively.

Table 10. Averaged EMG amplitude of 9 major muscles in stance and swing phase

Conditions		0% of BWS	10% of BWS	20% of BWS	30% of BWS	40% of BWS
		Mean ± STD	Mean ± STD	Mean ± STD	Mean ± STD	Mean ± STD
Stance Phase	**TA**	7.34 ± 4.72	7.33 ± 4.43	5.70 ± 3.69	5.22 ± 3.62	5.11 ± 3.38*
	GA	16.41 ± 6.00	13.91 ± 6.09	12.15 ± 4.06**	12.01 ± 3.81**	11.60 ± 4.69**
	Sol	13.75 ± 6.09	14.28 ± 6.52	12.44 ± 6.17	10.99 ± 5.38	10.65 ± .02
	VM	9.54 ± 6.29	8.84 ± 5.21	6.85 ± 4.46	6.86 ± 3.84	6.81 ± 4.02
	RF	13.86 ± 11.60	11.87 ± 8.60	9.48 ± 6.53	10.82 ± 8.10	10.90 ± 7.74
	BF	11.49 ± 6.41	11.03 ± 6.58	11.67 ± 7.22	10.48 ± 6.92	9.81 ± 5.08
	G_{Max}	8.38 ± 5.58	6.53 ± 3.54	6.17 ± 3.29	6.30 ± 3.30	6.26 ± 3.46
	G_{Med}	14.13 ± 10.02	11.82 ± 7.62	10.80 ± 7.24	9.57 ± 6.29*	8.70 ± 6.61**
	AL	7.40 ± 2.76	6.04 ± 2.68	5.56 ± 2.49*	5.80 ± 2.69*	5.74 ± 2.63*
Swing Phase	**TA**	8.72 ± 6.06	7.52 ± 4.65	7.09 ± 3.52	6.58 ± 3.35	7.38 ± 3.89
	GA	8.31 ± 5.42	7.64 ± 4.53	7.19 ± 4.73	7.23 ± 4.03	8.14 ± 5.01
	Sol	6.22 ± 3.46	6.15 ± 3.62	5.45 ± 2.80	5.79 ± 3.04	5.31 ± 3.23
	VM	6.03 ± 3.06	5.98 ± 4.57	5.79 ± 4.56	5.77 ± 3.56	5.74 ± 3.61
	RF	12.92 ± 11.49	10.69 ± 7.61	10.64 ± 8.73*	12.13 ± 9.74	12.14 ± 9.84
	BF	10.37 ± 6.49	8.15 ± 6.07	5.99 ± 5.04	8.14 ± 6.29	7.55 ± 5.27
	G_{Max}	6.93 ± 5.86	5.21 ± 3.99	4.46 ± 3.64	5.21 ± 4.26	5.69 ± 5.35
	G_{Med}	8.41 ± 8.28	5.51 ± 5.64	4.99 ± 6.56	5.22 ± 4.04	5.86 ± 6.43
	AL	8.51 ± 3.09	8.15 ± 4.28	7.16 ± 3.35	6.39 ± 2.76*	6.57 ± 3.21*

* Statistical difference from 0% BWS, $p < 0.05$

** Statistical difference from 0% BWS, $p < 0.001$

Table 11. Averaged EMG activation duration of 9 major muscles in stance and swing phase

Conditions		0% of BWS	10% of BWS	20% of BWS	30% of BWS	40% of BWS
		Mean ± STD	Mean ± STD	Mean ± STD	Mean ± STD	Mean ± STD
Stance	TA	17.02 ± 15.51	15.86 ± 15.63	10.85 ± 13.14	8.34 ± 11.60**	7.78 ± 10.43**
	GA	17.65 ± 11.89	12.73 ± 10.85	10.17 ± 8.78**	9.86 ± 8.53**	7.78 ± 8.82**
	Sol	13.50 ± 10.53	15.39 ± 11.72	14.80 ± 12.17	10.51 ± 9.16	11.16 ± 11.92
	VM	7.78 ± 12.33	6.16 ± 8.86	3.84 ± 7.15	3.20 ± 7.10	2.66 ± 5.96*
	RF	15.02 ± 19.30	10.51 ± 14.54	4.96 ± 8.54*	9.69 ± 13.46	9.70 ± 14.56
	BF	10.15 ± 14.15	8.60 ± 12.97	10.90 ± 15.03	8.28 ± 12.57	5.34 ± 9.44
	G_{Max}	4.56 ± 7.39	1.47 ± 3.51**	1.24 ± 3.33**	0.64 ± 2.22**	0.99 ± 3.22**
	G_{Med}	15.21 ± 19.05	10.21 ± 14.43	9.51 ± 14.49	6.41 ± 10.84*	6.81 ± 13.67*
	AL	2.89 ± 4.76	0.93 ± 2.09**	0.34 ± 1.25**	0.46 ± 1.34**	0.36 ± 1.01**
Swing	TA	11.37 ± 9.68	8.70 ± 7.95	8.61 ± 8.35	7.38 ± 7.57	10.46 ± 9.06
	GA	5.14 ± 6.65	2.73 ± 4.25	1.52 ± 3.76**	1.49 ± 3.2**1	2.16 ± 4.49*
	Sol	3.00 ± 4.20	2.28 ± 4.05	1.19 ± 2.41	1.40 ± 3.15	0.66 ± 1.70**
	VM	0.91 ± 3.20	1.51 ± 3.88	1.25 ± 3.94	0.84 ± 2.82	0.95 ± 2.72
	RF	8.58 ± 11.73	4.40 ± 7.67	4.40 ± 8.68	7.29 ± 11.64	8.46 ± 11.70
	BF	5.14 ± 6.16	3.13 ± 5.31	1.80 ± 3.88**	3.04 ± 5.65	2.61 ± 4.27
	G_{Max}	1.75 ± 4.20	0.37 ± 1.34	0.23 ± 1.13	0.70 ± 2.54	1.18 ± 3.57
	G_{Med}	4.64 ± 9.26	2.08 ± 6.75	1.45 ± 4.94	0.45 ± 1.55**	1.89 ± 5.37
	AL	1.73 ± 3.44	2.02 ± 3.66	1.51 ± 2.96	0.78 ± 1.47	1.35 ± 2.41

* Statistical difference from 0% BWS, $p < 0.05$

** Statistical difference from 0% BWS, $p < 0.001$

4. Discussion

This study examined the unique effects of BWS with the robotic walker in terms of joint kinematics, temporospatial gait parameters, and EMG mean amplitude and activation duration. The robotic walker can achieve normal gait patterns without altering any joint kinematics and EMG activation through pelvic motion facilitation during walking [83]. It is important to note that the achievement of this natural gait is the key addition of this study.

4.1. Kinematics and gait parameters

The mean peak angles and RoM showed a significant inverse relation with BWS levels. In particular, decreased peak flexion and RoM of the knee at 40% BWS, and the maximum extension and RoM of the hip at 20%, 30%, and 40% BWS level were observed. In contrast, the ankle joint kinematics were not significantly influenced by increased BWS levels. These results both support and contradict previous research. Fischer and his colleague reported a significant reduction in maximum knee flexion at 30% BWS and maximum hip flexion at 15% and 30% BWS [125]. These results are similar to ours showing a decrease in peak knee flexion at mid-swing. However, for hip joint kinematics the previous study showed a decrease in hip flexion while our results showed increased maximum hip extension with increased BWS level. It is important to note that the gait kinematic patterns are highly dependent on the devices used for the experiments, thus the kinematic patterns can be different from the previous study [125].

With increasing amounts of bodyweight unloading, the product of spatial gait variables such as step and stride length, and step width appears to be less significantly affected while temporal parameters such as velocity, stride time, stance time, swing time, SLS time, DLS time and percentage of stance phase were largely influenced with increased BWS levels. Contrary to previous assessment of over-ground walking with BWS using healthy individuals [102], the participants of our study walked faster in the 10% and 40% BWS conditions than in the no bodyweight unloading condition. Increase in gait speed at these BWS conditions was accompanied by shorter stride and stance time. This contradiction may have arisen from the different mechanical structure and control strategies of the devices used for the experiments. The reduction in the absolute duration of the stride over the gait cycle observed in this study is mainly attributable to a decrease in the stance duration, as the duration of the swing phase does not change much with varying amounts of bodyweight unloading, except at 40% BWS level. Such systematic decline in the proportion of the stance phase in the gait cycle at all BWS conditions indicates improved stability of the subjects' dynamic gait with increased BWS, as, during unsteady walking, patients generally remain with at least one foot in contact with the ground by spending more time in stance phase. In addition, especially during DLS, additional effort should be exerted by pushing off at the ankle or powering the hip to maintain a steady walking speed or to step back and forth [121]. Thus, the significantly shortened stance, SLS, and DLS time may be explained by the reduced time required for SST and increased step frequency rather than step or stride length in the BWS conditions. These results confirmed our first hypothesis that the BWS unloading will shorten the stance phase including single limb support and double limb support time.

4.2. EMG amplitude and duration

The EMG data collected in this study suggests a specific mechanistic cause that underlies the observations in the kinematic and temporal gait changes, in line with the increase in BWS level. In agreement with our second hypothesis, the results of our study showed that the intensity of muscle activation at ankle, knee, and hip joints in the sagittal plane and that at hip joint in the frontal plane were significantly reduced with increasing BWS levels. The systematic decline in EMG activity amplitude with increasing levels of BWS was expected in neurologically non-impaired individuals, due to a reduction in antigravity muscle activation, which has a greater influence in decreased mechanical loading conditions [126].

At the ankle joint, interestingly, the use of BWS slightly but significantly reduced the ankle dorsiflexor in the overall gait cycle and stance phase. It can be explained that the use of 30% or 40% of BWS may reduce ankle dorsiflexor load during mid-stance for weight acceptance rather than helping to elevate the foot during the swing phase (Figure V-5 and Table 10). In contradiction to Lewek’s (2011) study reporting unchanged plantarflexor muscle activity with BWS, the amplitude and duration in the GA muscle was linearly and significantly reduced at the 20%, 30%, and 40% BWS conditions without altering the normative EMG pattern. As a critical component for body propulsion, the reduced GA muscle activity with increased gait velocity and decreased DLS time in high BWS conditions, could be explained in that the SST can be achieved with relatively little active muscle powering to gain energy efficiency in dynamic walking [121].

Slightly reduced EMG amplitudes were found in the knee flexor and extensor such as VM, RF and BF, but no significant differences were observed in higher BWS levels compared to 0% BWS (Figure V-5 and Table 10). Although the effects of BWS on the knee joint were minor, the activation duration of VM at 40% and RF at 20% were significantly reduced as compared to 0% BWS (Figure V-6).

For the hip joint flexor and extensor, there was a significant reduction in G_{Max} duration with increases in the BWS level, due to the reduced maximum hip extension at high BWS levels. Furthermore, remarkable muscle activation changes were found in G_{Med} and AL, which are hip ab/adductors. Both amplitude and duration of G_{Med} and AL were significantly reduced with increased BWS level. During gait, a strong hip adduction torque is required in the loading response period, followed by the rapid transfer of BW onto the limb, and these demands continue throughout the stance period [127]. The G_{Med} muscle decelerates the rapid drop of the pelvis over the loading response period and maintains lateral stability during dynamic walking. Thus, the reduced G_{Med} muscle activation may be attributed to the reduced lateral momentum created in the pelvis and trunk induced by the unloading forces, following the smaller effort required for balancing the body in the horizontal plane. Furthermore, the hip flexion in pre-swing is initiated by both AL and RF muscles, whilst the AL muscle is further activated in pre-swing to restrain the abducting torque at the hip generated by BW falling towards the other limb [127]. The significantly reduced muscle load in AL implies an abductor torque that must be restrained to preserve weight-bearing balance with relatively little exertion. The reduced

G_{Med} and AL prove our third hypothesis that BWS systems should be able to increase lateral stability by reducing hip ab/adductor muscle amplitudes.

4.3. Clinical implications

Firstly, for effective and successful gait rehabilitation, the BWS levels selected must satisfy the normal sensorimotor input and extract the proportionately scaled motor responses required for normative gait [48]. However, the over-head harness BWS scheme often restricts pelvic lateral and rotational movement, resulting in abnormal gait patterns such as reduced step width and trunk rotation, and increasing step length with significantly altered muscle activation [93, 94]. The restriction of the pelvis would finally affect gait functional outcomes after gait rehabilitation. The robotic walker, which can facilitate 6 DoFs of pelvic motion combined with BWS ability, may provide neurologically challenged patients with afferent sensory feedback with linearly decreased muscle activation without altering normative EMG excursion during BWS training (Figure V-4).

Secondly, a previous study pinpointed that pelvic lateral displacement in patients with acute hemiparetic stroke was significantly increased to keep the body balanced from dysfunction of voluntary joint movements [70]. Such reduced G_{Med} and AL activation as the BWS level increased are of great importance in clinical trials for keeping patients' body laterally balanced, and this has a significant implication in increasing energy efficiency for maintaining lateral stability of neurological patients.

Last but not least, our findings show that with increasing BWS level, healthy individuals' DLS time is shortened and muscle activation decreased. Previous research has suggested that SST work exacts a proportional metabolic cost of walking [128, 129], and that this is calculated from the time-integral of external mechanical power from each leg during DLS when both feet are in contact with the ground [128]. In addition, the decreased intensity and duration of ankle plantar flexor and hip muscles in this study provide a clear example of reduced SST work and increased energy efficiency by minimizing SST cost [121]. Especially, despite the apparent decreases in the level of EMG activity at the ankle joint, there was no notable change in kinematic variables at the ankle. This finding is important for the clinical application suggesting that with increased BWS levels less ankle muscle strength is required to perform the same motions. The total metabolic cost for ankle movements during dynamic walking is decreased while providing the same mechanical work output, proving relatively high muscle efficiency at the ankle joint as BWS levels increase. Therefore, it is expected to increase effectiveness of gait training by lowering metabolic cost and increasing patients' mobility and stability.

5. Conclusion

The aim of this study has been to sufficiently investigate lower limb kinematics, temporospatial gait parameters, and EMG activity to address the question of how the gait variables adapt to reduced gravity over the gait cycle. Our unique Robotic Walker, which allows pelvic movements, successfully reduced gravitational force and loading during gait. The findings of this study demonstrate the linearly decreased intensity and duration of muscle activation without altering normal pattern, high muscle efficiency for weight bearing and propulsion in the sagittal plane and for lateral balance. The findings of this study help guide the rehabilitation strategies and the future design of assistive robotic devices by highlighting the effectiveness of BWS gait training aimed at lowering metabolic costs and increasing the stability of the patient. Although we only observed healthy individuals, our findings shed some light on determining the possible load-related sensory mechanisms that affect locomotor output.

People with post-stroke hemiparesis or other neurological disorders would respond differently than non-impaired healthy subjects with increased amounts of BW unloading. Therefore, further studies with neurologically challenged patients suffering from any gait impairments will be useful in assessing the effectiveness of BWS gait training on ground-level. Thus, the use of a BWS system during over-ground walking will be recommended as a useful intervention strategy for gait rehabilitation.

CHAPTER VI. RESISTANCE TRAINING USING A NOVEL OVER-GROUND GAIT WALKER: A PRELIMINARY STUDY ON HEALTHY SUBJECTS

1. Introduction

Gait is the most basic form of human locomotion and comprises an intricate network between the neurophysiological network and the musculoskeletal system. The loss of control which involves a constant communication between the efferent signals from the central command and afferent signals from sensory feedback leads to the loss of a basic ability to walk [6]. Gait rehabilitation is aimed at restoring this basic locomotive function, and can be achieved through a well-planned regime that includes balance training, weight bearing exercises, and strength training [14].

Muscle weakness, which is the most common symptom of gait abnormality, can restrain patients from being able to meet the demands of walking and can occur due to age [6] and/or due to disuse of muscular atrophy caused by neurological disorders such as a stroke [9, 130], Parkinson's disease [131] and even spinal cord and traumatic brain injury [132]. While other factors such as spasticity or muscular contracture generally play a part in gait abnormalities, the contribution of muscle weakness differs among patients due to their ability to compensate it by altering timings of certain gait events or performing exaggerated motions of other joints [133]. Furthermore, such gait abnormalities usually do not require medical or surgical intervention in where strengthening exercises usually suffice. To improve muscle strength and power, strength training has been recommended and widely adopted in gait training with positive results [50]. This strength training is a part of gait rehabilitation, which complements other training scopes such as balance control and weight bearing.

It has been shown that strength training can improve neural adaptations such as motor unit activation and synchronization, thus leading to higher muscular strength and better control [51].

Conventional strength training takes the form of generic lower limb exercises such as hip or knee flexion using weights and complex lower limb exercises [134-138]. Although this type of graded strength training may improve the ability to generate force, it cannot be transferred into improvement in gait functionality [139]. A recent review highlighted that many studies, which conducted strength training on neurological patients, did not show positive gait outcomes due to the lack of task-specificity, which is the repetitive practice of a task that is specific to the intended outcome [52, 140]. In this regard, incorporated task specific strength training has been emphasized, and practices in a variety of walking tasks have been targeted to improve weight-bearing, aerobic, functional strengthening and balance for the neurologically challenged patients [138, 141-143].

In the most traditional form, several studies have used weights attached to subject's waist which were connected to a pulley system while walking on a treadmill, and have showed that such a method could serve as an easier and cheaper alternative to providing resistance, since all subjects had an increase in metabolic cost [144]. Blanchette and Bouyer provided resistance to the ankle joint during treadmill walking, and EMG analysis revealed an increase in hamstring activity and this increase was kept even upon the removal of the resistance [145]. Using more advanced robotics, Lam et al. was able to modify the Lokomat's control system by applying a resistance to the hip and knee joint during the swing phase [146]. Results from their study showed that knee flexor activities were increased although knee flexion was reduced during swing, yet, great variability was found among the subjects.

It is clear that manipulation of the various resistance variables such as number of sets, position of the force applied, and intensity or load can stimulate the muscles in very different ways. However, it was recently emphasized that in order to improve gait outcomes, strength training has to focus on three power events in a gait cycle: 1) ankle plantarflexion push off in the late stance, 2) hip extension in the early stance and 3) hip flexion in the terminal stance [52]. Correspondingly, the main interest of this study will be on the task-specific resistance training applied at the human Center of Mass (CoM) to improve muscle strength. This form of exercise has been proven to be effective in strengthening muscles and improving overall physical capacity. However, the effects of this exercise on the three-power gait events mentioned above has not been investigated. For the task-specific gait training, a novel robotic walker for over-ground gait training which can provide resistive force at the CoM (pelvis) was recently developed [147]. Therefore, the aims of this study were to investigate the effects of a resistance force applied to the CoM on gait dynamics in terms of gait kinematics and electromyography (EMG), and to investigate if this type of exercise can satisfy the task-specific and effective strength training for gait. To do so, three objectives needed to be achieved. Firstly, the gait training device should be capable of providing resistance while over-ground walking. Secondly, the biomechanical gait changes (i.e. kinematics and electromyographic) should be analyzed according to the applied resistance. Finally, these biomechanical changes should be valid enough to determine if such a resistance training can be effective and task-specific to gait.

2. Methods

2.1. Provision of resistance force with the robotic walker

Resistance was provided by means of an offset from the measured force from the FT sensor (Figure VI-1). The force after the offset was then used as an input for the mass-damper admittance model (Figure VI-2). As a result, subjects needed more force to move forward with the walker creating a dragging or pulling effect on subjects.

Fig.VI-1 The robotic walker and anterior force applied with increasing resistance

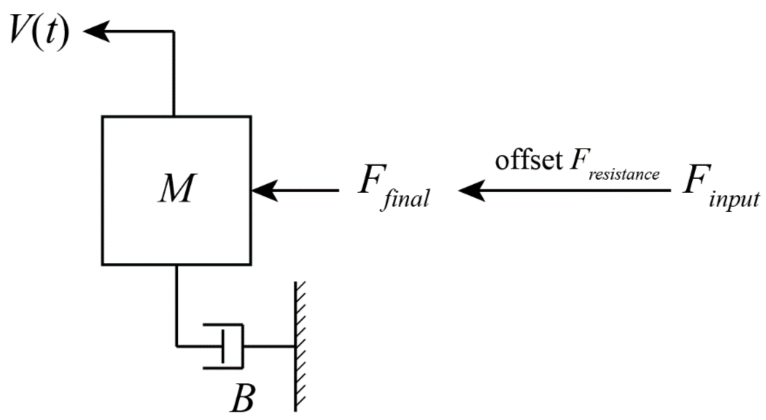


Figure VI-2.Provision of the resistance force using mass-damper admittance controller with force off-set

2.2. Experimental protocol

For this study, 10 young and healthy subjects (24.2 $\pm$3.3 years old; 7 males and 3 female) without any history of neurological disease and musculoskeletal injuries which affect their gait were recruited. To obtain gait kinematic data, a 3D motion capture system was used (Vicon, Oxford, UK). 15 reflective markers were placed on the subjects' pelvis and lower limb [148]. The precise 3D location of each marker was determined by 8 infrared cameras (100 Hz sampling frequency) and recorded using software (Nexus 1.8.3, Vicon, Oxford, UK). Wireless bi-polar electromyograph (EMG) sensors (Delsys, MA, USA) were used to quantify muscle activity (1000 Hz sampling frequency). EMG data were obtained from nine muscles – tibialis anterior (TA), medial gastrocnemius (MG), soleus (SOL), vastus medialis (VM), rectus femoris (RF), bicep femoris (BF), gluteus maximus (GMax), gluteus medius (GMed) and adductor longus (AL). The sensors were positioned in accordance to the SENIAM protocol [149]. Since this study was conducted on healthy subjects, gait symmetry was assumed and EMG sensors were placed only on the left leg.

The subjects were first instructed to walk along the platform at a self-selected comfortable speed. Gait kinematics and EMG data were collected to check if there were any gait abnormalities for all subjects. The subjects were then strapped onto the walker and allowed to acclimatize to the walker for 10 minutes. Then resistance was given using the walker with five intervals, proportional to each subjects' body weight (BW) – 0% (R0), 2.5% (R2.5), 5% (R5), 7.5% (R7.5), and 10% (R10). For each trial, subjects walked along a 10m walkway. Two strides in the middle of the walkway for each trial were used for analysis. Heel-strike (HS) and toe-off (TO) events were determined by the algorithm designed by O'Connor et al. [150]. Each trial was subsequently divided into individual strides, then each stride was defined as the HS of the left foot to the subsequent left HS. Each stride was then further divided into stance phase and swing phase.

2.3. Data analysis

Raw data was analyzed using a custom program written in MATLAB (Mathwork, MA, USA). The marker trajectories and kinematic data of each stride were filtered by a zero-lag 4th order Butterworth low-pass filter (6 Hz cutoff frequency). The EMG signals were filtered and rectified using a zero-lag 4th order Butterworth band-pass filter (2 - 400 Hz cut-off frequency). The rectified EMG was subsequently normalized to each subject's maximum EMG value over all trials, and linearly enveloped using a zero-lag 4th order Butterworth low-pass filter (10 Hz cut-off frequency) [85]. Finally, the mean amplitude was calculated for both each subject and each stride. The power spectrum was obtained using a Fast Fourier Transfom of the band-passed signal. The mean frequency (MNF) of the EMG signal was also calculated from the power spectrum [151]. In this study, the resistance by the walker was applied to the subject's CoM. However, the effects of the resistance was insignificant on the swing phase, thus, only the stance phase was considered for this study.

2.4. Statistical analysis

A one-way ANOVA (Analysis of Variance) was used to test the significant differences of gait kinematic and EMG mean amplitude and frequency according to increased level of the resistance. Least significant difference (LSD) post hoc analysis was then conducted to determine where the differences occurred between groups. All significance levels were set at $p < 0.05$.

3. Results

When resistance was applied, all subjects had a resultant increase in anterior force that they exerted on the walker (Table 12).

3.1. Kinematic parameters

Figure VI-3 and Table 12 respectively depict gait kinematic profiles and parameters, respectively. During the initial contact (IC), the higher resistance increased the angles in all three joints (i.e. more flexed) (ankle: $p = 0.0353$; knee and hip $p < 0.001$). Ankle maximum dorsiflexion increased with resistance ($p = 0.0139$), but no change in maximum plantarflexion was observed. For the knee, maximum knee extension was reduced ($p = 0.0002$) but no changes were observed in maximum knee flexion. Higher resistance increased the maximum flexion at the hip joint ($p = 0.0028$), but the maximum hip extension exhibited no changes. While higher resistance increased the range of motion (RoM) at hip joint ($p < 0.0001$), it reduced the RoM at the knee joints ($p < 0.0209$).

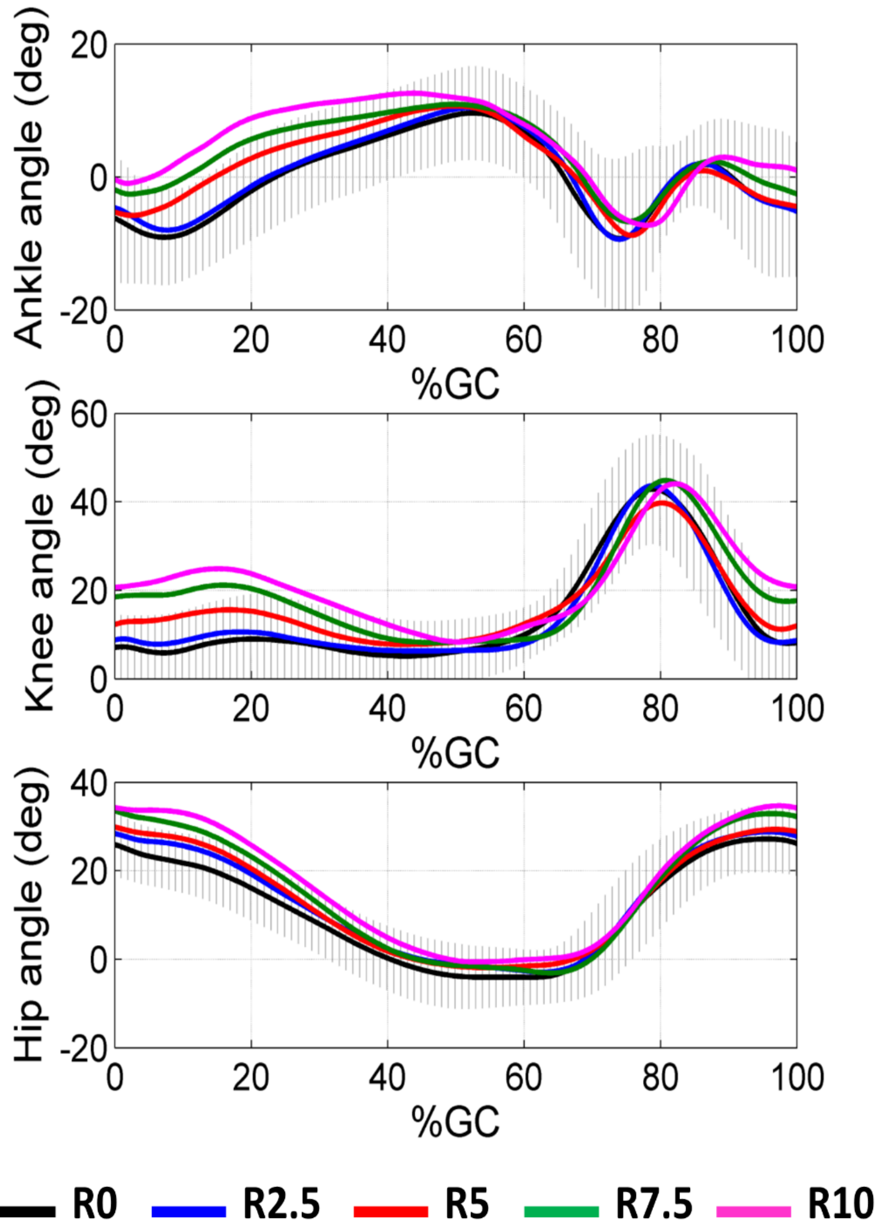


Figure VI-3. Ankle, knee, and hip flexion and extension angles. The black line and gray line show joint angles without resistance applied (R0) and its' standard deviation. Blue, red, green, and pink lines show the joint angles in 2.5%, 5%, 7.5%, and 10% BWS, respectively.

Table 12 Summary of kinematic parameters with increasing resistance

Parameters (deg)	0%BW	2.5%BW	5%BW	7.5%BW	10%BW
Force Applied (% BW)	8.4 ±2.1	11.1 ±1.5	13.5 ±2.1	15.8 ±3.1	17.1 ±3.8
Ankle IC	-6.2 ±9.6	-4.6 ±8.8	-5.3 ±11.0	-1.9 ±9.2*	-0.4 ±8.2**
Knee IC	7.1 ±7.5	8.8 ±6.3	12.3 ±7.5	18.5 ±9.7*	20.7 ±8.8**
Hip IC	25.9 ±7.1	28.4 ±7.8	29.9 ±9.6*	33.5 ±11.8**	34.3 ±12.7**
Ankle D-Flexion	11.1 ±6.9	11.7 ±6.1	12.8 ±6.4	14 ±5.3*	15.4 ±5.6*
Ankle P-Flexion	-16.7 ±8.3	-16.4 ±9.8	-18.7 ±12.2	-17.6 ±13.0	-17.5 ±12.8
Knee Flexion	50.3 ±8.6	48.7 ±7.4	47.5 ±8.4	49.7 ±10.6	51.1 ±9.0
Knee Extension	0.6 ±5.2	1.5 ±5.0	4.0 ±5.8*	4.3 ±4.8*	5.2 ±5.2**
Hip Flexion	28.4 ±7.3	30.5 ±8.5	31.9 ±9.9	34.9 ±11.7*	36.9 ±13.4**
Hip Extension	-5.9 ±6.3	-4.0 ±7.6	-4.8 ±8.0	-4.1 ±8.8	-3.3 ±9.0
Ankle ROM	27.8 ±8.3	28.1 ±11.7	31.5 ±13.1	31.6 ±13.3	32.9 ±13.7
Knee ROM	49.7 ±7.9	47.2 ±7.5	43.5 ±7.9*	45.4 ±10.1*	45.8 ±8.2*
Hip ROM	34.4 ±3.6	34.5 ±4.2	36.7 ±6.5	39.0 ±7.2*	40.2 ±9.0**

* Statistical difference from 0% BW, $p < 0.05$

** Statistical difference from 0% BW, $p < 0.001$

3.2. Electromyographic parameters

The provision with resistance resulted in some changes in muscle activation patterns as seen by the variation in the amplitude profiles of each muscle (Figure VI-4A). The mean amplitudes of the TA (p < 0.0001), VM (p < 0.0001), RF (p = 0.0066), GMax (p < 0.0001) and AL (p < 0.0001) all increased with higher resistance in the stance phase. The applied resistance force did not significantly alter the activations of the MG, SOL, BF and GMed muscles. Mean frequencies of the VM (p = 0.0157), GMax (0.0057) and AL (p = 0.0122) all decreased with higher resistance (Table 13). Mean frequency of the RF also decreased but was not statistically significant ($p = 0.0708$). Resistance had no effect on the mean frequencies of the TA, MG, SOL, BF and GMed.

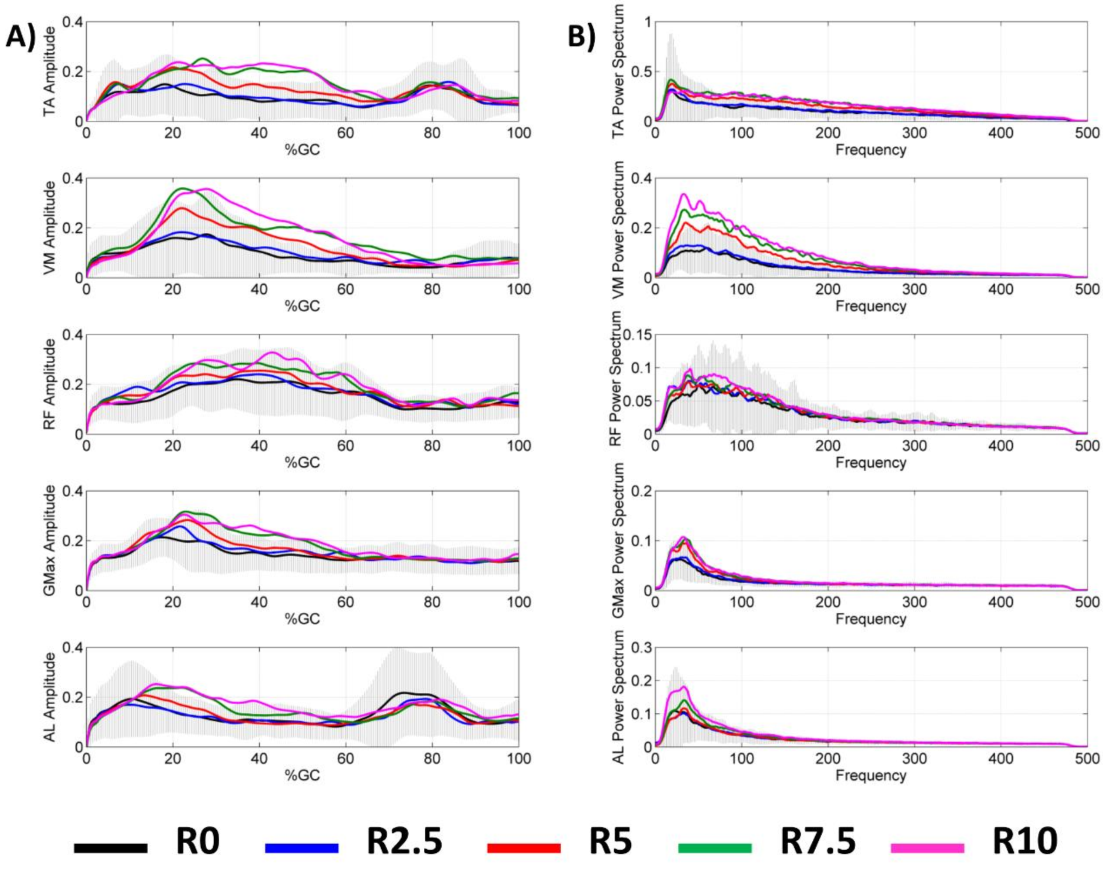


Figure VI-4. Enveloped EMG profiles from 5 muscles during walking with the various resistance forces. The black line and gray line show averaged EMG profiles and its' standard deviation in 0% of resistance. Blue, red, green, and pink lines show the enveloped EMG profiles in 2.5%, 5%, 7.5%, and 10% resistance, respectively.

Table 13 Summary of mean normalised EMGs of all 9 muscles during the stance phase with increasing resistance. Mean amplitude shown is normalised to each subject's maximum value among all the trials.

	Muscle	0%BW	2.5%BW	5%BW	7.5%BW	10%BW
Mean Amplitude	TA	0.08 ±0.03	0.09 ±0.03	0.12 ±0.04*	0.14 ±0.05**	0.14 ±0.07**
	MG	0.13 ±0.04	0.12 ±0.04	0.13 ±0.05	0.13 ±0.06	0.14 ±0.06
	SOL	0.14 ±0.06	0.12 ±0.04	0.13 ±0.07	0.13 ±0.07	0.13 ±0.07
	VM	0.1 ±0.07	0.11 ±0.07	0.14 ±0.09*	0.18 ±0.11**	0.18 ±0.08**
	RF	0.15 ±0.06	0.17 ±0.06	0.18 ±0.05	0.2 ±0.07*	0.2 ±0.07**
	BF	0.2 ±0.08	0.17 ±0.06	0.18 ±0.07	0.19 ±0.07	0.21 ±0.07
	G_{Max}	0.13 ±0.05	0.15 ±0.06	0.16 ±0.05*	0.18 ±0.06**	0.19 ±0.07**
	G_{Med}	0.21 ±0.08	0.17 ±0.04	0.17 ±0.05	0.2 ±0.06	0.21 ±0.08
	AL	0.12 ±0.06	0.11 ±0.05	0.12 ±0.06	0.14 ±0.07*	0.16 ±0.09**
Mean Frequency	TA	173.9 ±38.2	169.3 ±39.4	171.8 ±38.8	171.1 ±38.8	178.8 ±35.9
	MG	177.1 ±38.2	174.2 ±39.9	179.3 ±35	176.2 ±40.2	180.4 ±38.9
	SOL	157.2 ±28	146.7 ±29.8	152.6 ±25.3	152.7 ±24.1	155.8 ±26.6
	VM	153.2 ±23.4	147 ±21.5	142.4 ±22.9*	137.7 ±20.4*	138.8 ±21.1*
	RF	178.2 ±29	165.5 ±19.4	166 ±16.4	165.9 ±19.6	159.8 ±21.4*
	BF	162.3 ±19.9	161.1 ±22.3	159 ±20.2	158.8 ±26.2	157.7 ±21.8
	G_{Max}	181.7 ±21	174.4 ±15.3	165.7 ±24.3*	162.1 ±26*	161.7 ±28.6*
	G_{Med}	143.9 ±28.6	152.4 ±30.4	149.7 ±31.1	143.8 ±27.4	146.7 ±32.4
	AL	157.1 ±30.4	158.3 ±24.5	154.8 ±25.1	145.9 ±25.9	139.9 ±31.2*

* Statistical difference from 0% BW, $p < 0.05$

** Statistical difference from 0% BW, $p < 0.001$

4. Discussion

This study has not only proven that the walker is capable of providing resistance in the posterior direction during over-ground gait walking, but also confirmed that interacting force between human and the walker can be proportionally increased as resistance level is increased.

The provision of resistance affected on the subjects' gait kinematics. During the initial contact, all three joints were in greater flexion with increasing resistance. The maximum flexion of the ankle and hip increased while the maximum extension of the knee was reduced (i.e. more flexed). Thus, overall, the provision of resistance with the walker resulted in greater flexion on the ankle, knee and hip joint in the sagittal plane. The kinematic variations in this study had different results from those of other studies. Blanchette and Bouyer reported a reduction in peak knee flexion and ankle plantarflexion during swing when resistance was applied using elastic tubing at the ankle joint [145]. While Lam et al. also confirmed reduced knee extension in stance, reduced knee flexion was also observed when resistance was applied using the Lokomat [143]. However, it should be noted that the method of providing resistance adapted for this study greatly differs from that of other robotic devices. Devices such as the Lokomat directly resist lower limb joints via an exoskeleton, and resistance was largely present during the swing phase. In this study, a constant resistance was imposed via the pelvic support of the walker, thus resistance was more prominent during the stance phase. In the most primitive form, this can be analogous to sled or tire dragging exercise, commonly done by athletics.

Consequently, the imposed resistance on gait dynamics also affected the muscle activation patterns. At the ankle joint, the pre-tibial muscles usually work to pull the tibia over the foot (i.e. heel rocker) thus requiring their activation up until mid-stance [133]. When resistance was imposed however, the progression of the tibia was restricted and thus increased and prolonged activation of the TA muscle was required. This interaction between the body vector and the resistance force meant that the line of ground reaction force (GRF) was tilted in the posterior direction during the stance phase. Thus with higher resistance, the moment arm of GRF from the knee joint was increased, causing higher knee flexion torque. So both the knee extensor and hip flexor muscles (VM and RF) were increased accordingly. Finally, the resistive force acting on the CoM caused strong flexion moment at the hip joint during the stance period, thus resulting in increased hip extensor (GMax) activation. Therefore, it is expected that the increased muscle activation caused by resistance will promote muscle strength in ankle dorsiflexor (TA), knee extensor (VM), and hip flexor and extensor (RF, GMax, and AL) by promoting neural adaptations and motor neuron excitability, and by decreasing presynaptic inhibition [152].

For task specific resistive gait training, three components of gait cycle must be trained – plantar flexion and hip flexion in terminal stance and hip extension in early stance [52]. On closer inspection of this data, it was found that an increase in activation in hip flexion during terminal stance ($p = 0.0038$) and hip extension during early stance ($p = 0.0213$) were observed. The early stance was defined as the first double limb support phase (i.e. left HS to right TO) and terminal stance was the second double limb support phase (i.e. right HS to left TO). Plantar flexion during terminal stance increased without statistical significance. This could have been a result of the resistance being applied at the subject CoM, which means that greater force required for propulsion was supplied largely by the hip and lesser by the end effector muscles. It is expected that with the prolonged resistive training, the increased activity that patients' muscles generate would increase their muscular strength and thus would show better gait performances through improved propulsion by hip joint. These findings confirms that the resistance provided at the CoM is a task-specific method of gait rehabilitation by increasing muscle activation in at least 2 power events of the gait cycle.

From the analysis of the power spectrum, we found that the mean frequencies of the VM, RF, GMax and AL muscles were reduced with increased resistance (Table 13). This suggests that despite having a larger amount of motor units activation (i.e. increased amplitude), these motor units were firing at a lower rate (i.e. decreased frequency). The reduced MNF could have been attributed to the fatigue caused by the resistance training. The fatigue effect can be observed with decreased fast twitch (higher frequency) while slow twitch (lower frequency) retained [151]. However, the effects of fatigue would be minor as the level of resistances were well within each subject's physical capabilities and sufficient rest was provided between each trial. In this study, the profile of the frequency domain showed that overall power increased across the entire frequency spectrum but greater increase was observed at the lower frequencies (Figure VI-4B). This could imply a change in muscle fiber recruitment where slow twitch fibers (Type I) were dominantly recruited as resistance was increased, while fast twitch fibers (Type II) were relatively less involved [153]. It has been well documented that the high participation of slow twitch muscle will be more efficient at a low intensive aerobic exercise for patients undergoing gait rehabilitation [154]. Such growth will improve overall strength and endurance, and thus may improve the mobility and activities of daily life.

Although the biomechanical effects and task-specificity of the resistance training with the walker were proven in this study, it may be impracticable in treating patients with certain gait disorders such as acute stroke patients who are severely affected in their mobility. The fact that only young and healthy subjects were recruited will limit this study to a preliminary one only, and the effectiveness of resistance training can only be understood truly when it is brought into a clinical setting with a long-term study. Despite differences in gait dynamics, it is expected that patients would experience similar biomechanical changes to those of healthy subjects and benefit from the increase in muscular strength.

5. Conclusion

This study tested the effects of a resistance force applied at the CoM on kinematic and muscle activation patterns during over-ground walking. We found that the provision of the resistance significantly affects subjects' gait kinematics by increasing flexion angles. In addition, we also found that as the level of resistance increased, the amount of motor unit activations were increased with lower firing rates at knee flexors and hip flexor and extensor. Thereby, we conclude that this type of resistance training can improve the muscular strength and endurance in a task-specific manner. This study will serve as a cornerstone of understanding the biomechanical effects of the resistance training and will be expanded to the long-term study with actual neurologically challenged patients in the future.

CHAPTER VII. CONCLUSION AND FUTURE WORKS

The study in Chapter II examined the mechanism differences that altered gait performance between normal and stroke patients. Consequently, it was shown that while the primary joints excursion including ankle, knee, and hip RoM are the main contributors of gait performances of the control group, the pelvic tilt, and pelvic lateral displacement and rotation also play an important role to ensure gait velocity, step and stride length for stroke group. In addition, given the fact that the pelvic motions are excessively involved in gait performances of the stroke survivors, the need to support pelvic motions during or after gait rehabilitation was emphasized in this study.

For the study in Chapter III, a novel robotic walker for pelvic motion support was built and the biomechanical effects of the walker were tested in kinematic and muscle activation perspectives. The robotic walker can support pelvic lateral and rotational movements without complex actuators in an effective manner. The findings of this study conclude that gait closely resembled free over-ground walking with minimal alteration of the normal gait dynamics. It is further expected that the walker can provide satisfactory functional outcomes by providing more aesthetic gait patterns with proper sensory input and feedback to neurologically challenged patients.

The study in Chapter IV investigated the biomechanical effects of pelvic motion restriction during gait. The gait with pelvic motion facilitation can elicit normal muscle activation patterns in a natural manner without altering normal gait dynamics. On the other hand, gait with pelvic restriction severely affected gait dynamics, indicating the necessity of pelvic motion facilitation in gait rehabilitation.

The study in Chapter V investigated lower limb kinematics, temporospatial gait parameters, and EMG activity to address the question of how the gait variables adapt to reduced gravity over the gait cycle. The walker successfully reduced gravitational force and loading during gait. The linearly reduced muscles' activation amplitude and duration with the BWS unit implemented into the robotic walker can demonstrate an important indication of reduced step-to-step transition (SST) cost and energy expenditure, and increased lateral body balance with greater stabilization during gait. These findings provide a better understanding of the biomechanical effects of BWS during gait, which will help guide the design of rehabilitation strategies.

The study in Chapter VI has shown the effects of task-specific gait resistance training using the walker. The resistance applied at CoM significantly affected gait kinematics by increasing flexion angle, and increased muscle activation at knee flexors and hip flexor and extensor with lower firing rates. We conclude that this type of resistance training can improve the muscular strength and endurance in a task-specific manner.

In conclusion, the present dissertation underscores the importance of pelvic motion support during gait rehabilitation, provides a design description of the novel robotic over-ground walker, and demonstrates its biomechanical effects for gait training. The findings of this study will help guide the gait rehabilitation protocol with robotic gait rehabilitation devices and the future design of assistive robotic devices.

It should be noted that the gait experiments on healthy young subjects were performed with the purpose of evaluating various functions of the walker.

Although gait analysis with the walker showed the necessity of pelvic motion facilitation, BWS, and task-specific resistance training, experiments with healthy young subjects may not be applicable to the neurologically challenged patient. Hence, preliminary experiments with stroke patients will be conducted to study the effects of the walker for any actual clinical application.

BIOGRAPHY

[1] https://en.wikipedia.org/wiki/Neurological_disorder.

[2] http://www.who.int/features/qa/55/en/.

[3] D. Mozaffarian, E. J. Benjamin, A. S. Go, D. K. Arnett, M. J. Blaha, M. Cushman, *et al.*, "Heart Disease and Stroke Statistics—2015 Update A Report From the American Heart Association," *Circulation,* vol. 131, pp. e29-e322, 2015.

[4] B. A. Knarr, D. S. Reisman, S. A. Binder-Macleod, and J. S. Higginson, "Understanding compensatory strategies for muscle weakness during gait by simulating activation deficits seen post-stroke," *Gait & posture,* vol. 38, pp. 270-275, 2013.

[5] M. Pietrusinski, I. Cajigas, M. Goldsmith, P. Bonato, and C. Mavroidis, "Robotically generated force fields for stroke patient pelvic obliquity gait rehabilitation," in *Robotics and Automation (ICRA), 2010 IEEE International Conference on*, 2010, pp. 569-575.

[6] D. A. Winter, *Biomechanics and motor control of human gait: normal, elderly and pathological*, 1991.

[7] K. G. Pearson, "Generating the walking gait: role of sensory feedback," *Prog Brain Res,* vol. 143, pp. 123-9, 2004.

[8] P. A. Goldie, T. A. Matyas, and O. M. Evans, "Deficit and change in gait velocity during rehabilitation after stroke," *Archives of Physical Medicine and Rehabilitation,* vol. 77, pp. 1074-1082, 10// 1996.

[9] S. J. Olney and C. Richards, "Hemiparetic gait following stroke. Part I: Characteristics," *Gait & Posture,* vol. 4, pp. 136-148, 4// 1996.

[10] P.-H. Chen, R.-L. Wang, D.-J. Liou, and J.-S. Shaw, "Gait Disorders in Parkinson's Disease: Assessment and Management," *International Journal of Gerontology,* vol. 7, pp. 189-193, 12// 2013.

[11] A. Pennycott, D. Wyss, H. Vallery, V. Klamroth-Marganska, and R. Riener, "Towards more effective robotic gait training for stroke rehabilitation: a review," *J Neuroeng Rehabil,* vol. 9, p. 65, 2012.

[12] J. M. Potter, A. L. Evans, and G. Duncan, "Gait speed and activities of daily living function in geriatric patients," *Archives of physical medicine and rehabilitation,* vol. 76, pp. 997-999, 1995.

[13] C. M. Mumma, "Perceived Losses Following Stroke," *Rehabilitation Nursing,* vol. 11, pp. 19-24, 1986.

[14] R. Dickstein, "Rehabilitation of gait speed after stroke: a critical review of intervention approaches," *Neurorehabilitation and Neural Repair,* 2008.

[15] E. Dietrichs, "[Brain plasticity after stroke--implications for post-stroke rehabilitation]," *Tidsskrift for den Norske laegeforening: tidsskrift for praktisk medicin, ny raekke,* vol. 127, pp. 1228-1231, 2007.

[16] B. B. Johansson, "Brain plasticity and stroke rehabilitation The Willis lecture," *Stroke,* vol. 31, pp. 223-230, 2000.

[17] I. H. Robertson and J. M. Murre, "Rehabilitation of brain damage: brain plasticity and principles of guided recovery," *Psychological bulletin,* vol. 125, p. 544, 1999.

[18] A. Picelli, C. Melotti, F. Origano, A. Waldner, A. Fiaschi, V. Santilli, *et al.*, "Robot-Assisted Gait Training in Patients With Parkinson Disease A Randomized Controlled Trial," *Neurorehabilitation and neural repair,* vol. 26, pp. 353-361, 2012.

[19] S. Hesse, C. Bertelt, A. Schaffrin, M. Malezic, and K.-H. Mauritz, "Restoration of gait in nonambulatory hemiparetic patients by treadmill training with partial body-weight support," *Archives of physical medicine and rehabilitation,* vol. 75, pp. 1087-1093, 1994.

[20] B. H. Dobkin, "An overview of treadmill locomotor training with partial body weight support: a neurophysiologically sound approach whose time has come for randomized clinical trials," *Neurorehabilitation and neural repair,* vol. 13, pp. 157-165, 1999.

[21] M. Visintin, H. Barbeau, N. Korner-Bitensky, and N. E. Mayo, "A new approach to retrain gait in stroke patients through body weight support and treadmill stimulation," *Stroke,* vol. 29, pp. 1122-1128, 1998.

[22] 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.

[23] 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.

[24] I. Díaz, J. J. Gil, and E. Sánchez, "Lower-limb robotic rehabilitation: literature review and challenges," *Journal of Robotics,* vol. 2011, 2011.

[25] 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.

[26] K. P. Westlake and C. Patten, "Pilot study of Lokomat versus manual-assisted treadmill training for locomotor recovery post-stroke," *Journal of neuroengineering and rehabilitation,* vol. 6, p. 18, 2009.

[27] R. Riener, L. Lunenburger, S. Jezernik, M. Anderschitz, G. Colombo, and V. Dietz, "Patient-cooperative strategies for robot-aided treadmill training: first experimental results," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 13, pp. 380-394, 2005.

[28] J. F. Veneman, R. Kruidhof, E. E. Hekman, R. Ekkelenkamp, E. H. Van Asseldonk, and H. Van Der Kooij, "Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 15, pp. 379-386, 2007.

[29] N. Tufekciler, E. van Asseldonk, and H. van der Kooij, "Velocity-dependent reference trajectory generation for the LOPES gait training robot," in *Rehabilitation Robotics (ICORR), 2011 IEEE International Conference on*, 2011, pp. 1-5.

[30] H. Vallery, E. H. van Asseldonk, M. Buss, and H. van der Kooij, "Reference trajectory generation for rehabilitation robots: complementary limb motion estimation," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 17, pp. 23-30, 2009.

[31] E. H. Van Asseldonk, J. F. Veneman, R. Ekkelenkamp, J. H. Buurke, F. C. 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," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 16, pp. 360-370, 2008.

[32] S. K. Banala, S. H. Kim, S. K. Agrawal, and J. P. Scholz, "Robot assisted gait training with active leg exoskeleton (ALEX)," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 17, pp. 2-8, 2009.

[33] K. N. Winfree, P. Stegall, and S. K. Agrawal, "Design of a minimally constraining, passively supported gait training exoskeleton: ALEX II," in *Rehabilitation Robotics (ICORR), 2011 IEEE International Conference on*, 2011, pp. 1-6.

[34] 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," *Mechatronics, IEEE/ASME Transactions on,* vol. 15, pp. 216-225, 2010.

[35] S. Freivogel, J. Mehrholz, T. Husak-Sotomayor, and D. Schmalohr, "Gait training with the newly developed'LokoHelp'-system is feasible for non-ambulatory patients after stroke, spinal cord and brain injury. A feasibility study," *Brain Injury,* vol. 22, pp. 625-632, 2008.

[36] N. Chockalingam, F. Chatterley, A. C. Healy, A. Greenhalgh, and H. R. Branthwaite, "Comparison of pelvic complex kinematics during treadmill and overground walking," *Archives of physical medicine and rehabilitation,* vol. 93, pp. 2302-2308, 2012.

[37] J. R. Watt, J. R. Franz, K. Jackson, J. Dicharry, P. O. Riley, and D. C. Kerrigan, "A three-dimensional kinematic and kinetic comparison of overground and treadmill walking in healthy elderly subjects," *Clinical biomechanics,* vol. 25, pp. 444-449, 2010.

[38] L. L. Cai, A. J. Fong, C. K. Otoshi, Y. Liang, J. W. Burdick, R. R. Roy*, et al.*, "Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning," *The Journal of neuroscience,* vol. 26, pp. 10564-10568, 2006.

[39] J. W. Krakauer, "Motor learning: its relevance to stroke recovery and neurorehabilitation," *Current opinion in neurology,* vol. 19, pp. 84-90, 2006.

[40] J. F. Veneman, J. Menger, E. H. van Asseldonk, F. C. van der Helm, and H. van der Kooij, "Fixating the pelvis in the horizontal plane affects gait characteristics," *Gait & posture,* vol. 28, pp. 157-163, 2008.

[41] 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.

[42] J. Patton, D. A. Brown, M. Peshkin, J. J. Santos-Munné, A. Makhlin, E. Lewis*, et al.*, "KineAssist: design and development of a robotic overground gait and balance therapy device," *Topics in stroke rehabilitation,* vol. 15, pp. 131-139, 2008.

[43] Y. Allemand, Y. Stauffer, R. Clavel, and R. Brodard, "Design of a new lower extremity orthosis for overground gait training with the WalkTrainer," in *Rehabilitation Robotics, 2009. ICORR 2009. IEEE International Conference on*, 2009, pp. 550-555.

[44] Z. Guo, H. Yu, and Y. H. Yin, "Developing a Mobile Lower Limb Robotic Exoskeleton for Gait Rehabilitation," *Journal of Medical Devices*, vol. 8, p. 044503, 2014.

[45] P. Wang, K. Low, and A. McGregor, "A subject-based motion generation model with adjustable walking pattern for a gait robotic trainer: NaTUre-gaits," in *Intelligent Robots and Systems (IROS), 2011 IEEE/RSJ International Conference on*, 2011, pp. 1743-1748.

[46] T. Watanabe, T. Tono, Y. Nakashima, K. Kawamura, J. Inoue, Y. Kijima, *et al.*, "Analysis of interaction between therapist and hemiplegic patient for control of lateral pelvic motion during robotic gait training," in *Robotics and Automation (ICRA), 2013 IEEE International Conference on*, 2013, pp. 2663-2668.

[47] M. Pietrusinski, I. Cajigas, G. Severini, P. Bonato, and C. Mavroidis, "Robotic gait rehabilitation trainer," *Mechatronics, IEEE/ASME Transactions on*, vol. 19, pp. 490-499, 2014.

[48] S. R. Goldberg and S. J. Stanhope, "Sensitivity of joint moments to changes in walking speed and body-weight-support are interdependent and vary across joints," *Journal of biomechanics*, vol. 46, pp. 1176-1183, 2013.

[49] S. K. Banala, S. H. Kim, S. K. Agrawal, and J. P. Scholz, "Robot assisted gait training with active leg exoskeleton (ALEX)," *IEEE Trans Neural Syst Rehabil Eng*, vol. 17, pp. 2-8, Feb 2009.

[50] A. O. Brady and C. R. Straight, "Muscle capacity and physical function in older women: What are the impacts of resistance training?," *Journal of Sport and Health Science*, vol. 3, pp. 179-188, 2014.

[51] W. R. Frontera, V. A. Hughes, L. S. Krivickas, S.-K. Kim, M. Foldvari, and R. Roubenoff, "Strength training in older women: Early and late changes in whole muscle and single cells," *Muscle & Nerve*, vol. 28, pp. 601-608, 2003.

[52] G. Williams, M. Kahn, and A. Randall, "Strength training for walking in neurologic rehabilitation is not task specific: a focused review," *American Journal of Physical Medicine & Rehabilitation*, vol. 93, pp. 511-22, 2014.

[53] A. Lamontagne, F. Malouin, C. Richards, and F. Dumas, "Mechanisms of disturbed motor control in ankle weakness during gait after stroke," *Gait & posture*, vol. 15, pp. 244-255, 2002.

[54] S. G. Chung, E. van Rey, Z. Bai, E. J. Roth, and L.-Q. Zhang, "Biomechanic changes in passive properties of hemiplegic ankles with spastic hypertonia," *Archives of physical medicine and rehabilitation*, vol. 85, pp. 1638-1646, 2004.

[55] A. Cooper, I. Musa, R. Van Deursen, and C. Wiles, "Electromyography characterization of stretch responses in hemiparetic stroke patients and their relationship with the Modified Ashworth Scale," *Clinical rehabilitation*, vol. 19, pp. 760-766, 2005.

[56] J. Fleuren, G. Snoek, G. Voerman, and H. Hermens, "Muscle activation patterns of knee flexors and extensors during passive and active movement of the spastic lower limb in chronic stroke patients," *Journal of Electromyography and Kinesiology*, vol. 19, pp. e301-e310, 2009.

[57] D. L. Damiano, E. Laws, D. V. Carmines, and M. F. Abel, "Relationship of spasticity to knee angular velocity and motion during gait in cerebral palsy," *Gait & posture*, vol. 23, pp. 1-8, 2006.

[58] C. Bleyenheuft, Y. Bleyenheuft, P. Hanson, and T. Deltombe, "Treatment of genu recurvatum in hemiparetic adult patients: a systematic literature review," *Annals of physical and rehabilitation medicine,* vol. 53, pp. 189-199, 2010.

[59] J. L. Hicks, S. L. Delp, and M. H. Schwartz, "Can biomechanical variables predict improvement in crouch gait?," *Gait & posture,* vol. 34, pp. 197-201, 2011.

[60] J. Boudarham, R. Zory, F. Genet, G. Vigné, D. Bensmail, N. Roche*, et al.*, "Effects of a knee–ankle–foot orthosis on gait biomechanical characteristics of paretic and non-paretic limbs in hemiplegic patients with genu recurvatum," *Clinical Biomechanics,* vol. 28, pp. 73-78, 2013.

[61] I. Campanini, A. Merlo, and B. Damiano, "A method to differentiate the causes of stiff-knee gait in stroke patients," *Gait & posture,* vol. 38, pp. 165-169, 2013.

[62] A. S. Arnold, F. C. Anderson, M. G. Pandy, and S. L. Delp, "Muscular contributions to hip and knee extension during the single limb stance phase of normal gait: a framework for investigating the causes of crouch gait," *Journal of biomechanics,* vol. 38, pp. 2181-2189, 2005.

[63] A. S. Arnold, M. Q. Liu, M. H. Schwartz, S. Ounpuu, and S. L. Delp, "The role of estimating muscle-tendon lengths and velocities of the hamstrings in the evaluation and treatment of crouch gait," *Gait & posture,* vol. 23, pp. 273-281, 2006.

[64] M. D. Lewek, T. G. Hornby, Y. Y. Dhaher, and B. D. Schmit, "Prolonged quadriceps activity following imposed hip extension: a neurophysiological mechanism for stiff-knee gait?," *Journal of neurophysiology,* vol. 98, pp. 3153-3162, 2007.

[65] M. A. Gatti, O. Freixes, S. A. Fernández, M. E. Rivas, M. Crespo, S. V. Waldman*, et al.*, "Effects of ankle foot orthosis in stiff knee gait in adults with hemiplegia," *Journal of biomechanics,* vol. 45, pp. 2658-2661, 2012.

[66] S. Schmid, K. Schweizer, J. Romkes, S. Lorenzetti, and R. Brunner, "Secondary gait deviations in patients with and without neurological involvement: A systematic review," *Gait & posture,* vol. 37, pp. 480-493, 2013.

[67] J. B. Saunders, V. T. Inman, and H. D. Eberhart, "The major determinants in normal and pathological gait," *J Bone Joint Surg Am,* vol. 35-A, pp. 543-58, Jul 1953.

[68] U. Della Croce, P. O. Riley, J. L. Lelas, and D. C. Kerrigan, "A refined view of the determinants of gait," *Gait & posture,* vol. 14, pp. 79-84, 2001.

[69] D. C. Kerrigan, P. O. Riley, J. L. Lelas, and U. D. Croce, "Quantification of pelvic rotation as a determinant of gait," *Archives of physical medicine and rehabilitation,* vol. 82, pp. 217-220, 2001.

[70] K. J. Dodd and M. E. Morris, "Lateral pelvic displacement during gait: abnormalities after stroke and changes during the first month of rehabilitation," *Archives of physical medicine and rehabilitation,* vol. 84, pp. 1200-1205, 2003.

[71] E. De Bujanda, S. Nadeau, and D. Bourbonnais, "Pelvic and shoulder movements in the frontal plane during treadmill walking in adults with

stroke," *Journal of stroke and cerebrovascular diseases,* vol. 13, pp. 58-69, 2004.

[72] M. Gaston, E. Rutz, T. Dreher, and R. Brunner, "Transverse plane rotation of the foot and transverse hip and pelvic kinematics in diplegic cerebral palsy," *Gait & posture,* vol. 34, pp. 218-221, 2011.

[73] J. Salazar-Torres, B. McDowell, C. Kerr, and A. Cosgrove, "Pelvic kinematics and their relationship to gait type in hemiplegic cerebral palsy," *Gait & posture,* vol. 33, pp. 620-624, 2011.

[74] B. K. Krautwurst, S. I. Wolf, D. W. Heitzmann, S. Gantz, F. Braatz, and T. Dreher, "The influence of hip abductor weakness on frontal plane motion of the trunk and pelvis in patients with cerebral palsy," *Research in developmental disabilities,* vol. 34, pp. 1198-1203, 2013.

[75] C. M. Kim and J. J. Eng, "Magnitude and pattern of 3D kinematic and kinetic gait profiles in persons with stroke: relationship to walking speed," *Gait & posture,* vol. 20, pp. 140-146, 2004.

[76] T. H. Cruz, M. D. Lewek, and Y. Y. Dhaher, "Biomechanical impairments and gait adaptations post-stroke: multi-factorial associations," *Journal of biomechanics,* vol. 42, pp. 1673-1677, 2009.

[77] S. R. Goldberg, S. Ounpuu, A. S. Arnold, J. R. Gage, and S. L. Delp, "Kinematic and kinetic factors that correlate with improved knee flexion following treatment for stiff-knee gait," *Journal of biomechanics,* vol. 39, pp. 689-698, 2006.

[78] R. Gross, F. Leboeuf, O. Rémy-Néris, and B. Perrouin-Verbe, "Unstable gait due to spasticity of the rectus femoris: gait analysis and motor nerve block," *Annals of physical and rehabilitation medicine,* vol. 55, pp. 609-622, 2012.

[79] A.-L. Hsu, P.-F. Tang, and M.-H. Jan, "Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke," *Archives of physical medicine and rehabilitation,* vol. 84, pp. 1185-1193, 2003.

[80] H. Yu, M. Spenko, and S. Dubowsky, "An adaptive shared control system for an intelligent mobility aid for the elderly," *Autonomous Robots,* vol. 15, pp. 53-66, 2003.

[81] H. Y. Yu, M. Spenko, and S. Dubowsky, "Omni-directional mobility using active split offset castors," *Journal of Mechanical Design,* vol. 126, pp. 822-829, Sep 2004.

[82] B. Kopp and M. Wolff, "Brain mechanisms of selective learning: Event-related potentials provide evidence for error-driven learning in humans," *Biological Psychology,* vol. 51, pp. 223-246, 2000.

[83] K.-R. Mun, H. Yu, C. Zhu, and M. S. Cruz, "Design of a novel robotic over-ground walking device for gait rehabilitation," in *Advanced Motion Control (AMC), 2014 IEEE 13th International Workshop on*, 2014, pp. 458-463.

[84] C. A. Byrne, D. T. O'Keeffe, A. E. Donnelly, and G. M. Lyons, "Effect of walking speed changes on tibialis anterior EMG during healthy gait for FES envelope design in drop foot correction," *Journal of Electromyography and Kinesiology,* vol. 17, pp. 605-616, 10// 2007.

[85] G. Bovi, M. Rabuffetti, P. Mazzoleni, and M. Ferrarin, "A multiple-task gait analysis approach: kinematic, kinetic and EMG reference data for healthy young and adult subjects," *Gait & posture,* vol. 33, pp. 6-13, 2011.

[86] M. Wu, J. M. Landry, J. Kim, B. D. Schmit, S.-C. Yen, and J. MacDonald, "Robotic resistance/assistance training improves locomotor function in individuals poststroke: a randomized controlled study," *Archives of physical medicine and rehabilitation,* vol. 95, pp. 799-806, 2014.

[87] A. D. Kuo, "The six determinants of gait and the inverted pendulum analogy: A dynamic walking perspective," *Human movement science,* vol. 26, pp. 617-656, 2007.

[88] B. W. Liang, W. H. Wu, O. G. Meijer, J. H. Lin, G. R. Lv, X. C. Lin*, et al.*, "Pelvic step: the contribution of horizontal pelvis rotation to step length in young healthy adults walking on a treadmill," *Gait Posture,* vol. 39, pp. 105-10, Jan 2014.

[89] K. E. Gordon, D. P. Ferris, and A. D. Kuo, "Metabolic and mechanical energy costs of reducing vertical center of mass movement during gait," *Archives of physical medicine and rehabilitation,* vol. 90, pp. 136-144, 2009.

[90] R. Baker, "Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms," *Gait & posture,* vol. 13, pp. 1-6, 2001.

[91] Y. Huang, O. G. Meijer, J. Lin, S. M. Bruijn, W. Wu, X. Lin*, et al.*, "The effects of stride length and stride frequency on trunk coordination in human walking," *Gait & posture,* vol. 31, pp. 444-449, 2010.

[92] K. Dodd, T. Wrigley, P. Goldie, M. Morris, and C. Grant, "Quantifying lateral pelvic displacement during walking," *Clinical Biomechanics,* vol. 13, pp. 371-373, 1998.

[93] J. F. Veneman, J. Menger, E. H. van Asseldonk, F. C. van der Helm, and H. van der Kooij, "Fixating the pelvis in the horizontal plane affects gait characteristics," *Gait Posture,* vol. 28, pp. 157-63, Jul 2008.

[94] J. M. Hidler and A. E. Wall, "Alterations in muscle activation patterns during robotic-assisted walking," *Clinical Biomechanics,* vol. 20, pp. 184-193, 2005.

[95] H. S. Linley, E. A. Sled, E. G. Culham, and K. J. Deluzio, "A biomechanical analysis of trunk and pelvis motion during gait in subjects with knee osteoarthritis compared to control subjects," *Clinical Biomechanics,* vol. 25, pp. 1003-1010, 2010.

[96] K. Parvataneni, L. Ploeg, S. J. Olney, and B. Brouwer, "Kinematic, kinetic and metabolic parameters of treadmill versus overground walking in healthy older adults," *Clinical biomechanics,* vol. 24, pp. 95-100, 2009.

[97] E. Kristianslund, T. Krosshaug, and A. J. van den Bogert, "Effect of low pass filtering on joint moments from inverse dynamics: implications for injury prevention," *Journal of biomechanics,* vol. 45, pp. 666-671, 2012.

[98] D. M. Bojanic, B. D. Petrovacki-Balj, N. D. Jorgovanovic, and V. R. Ilic, "Quantification of dynamic EMG patterns during gait in children with cerebral palsy," *Journal of neuroscience methods,* vol. 198, pp. 325-331, 2011.

[99] K. Lyons, J. Perry, J. K. Gronley, L. Barnes, and D. Antonelli, "Timing and relative intensity of hip extensor and abductor muscle action during level and stair ambulation An EMG study," *Physical Therapy,* vol. 63, pp. 1597-1605, 1983.

[100] P. A. Goldie, T. A. Matyas, and O. M. Evans, "Gait after stroke: initial deficit and changes in temporal patterns for each gait phase," *Archives of physical medicine and rehabilitation,* vol. 82, pp. 1057-1065, 2001.

[101] S. Hesse, C. Werner, A. Bardeleben, and H. Barbeau, "Body weight-supported treadmill training after stroke," *Current atherosclerosis reports,* vol. 3, pp. 287-294, 2001.

[102] J. K. Burgess, G. C. Weibel, and D. A. Brown, "Overground walking speed changes when subjected to body weight support conditions for nonimpaired and post stroke individuals," 2010.

[103] M. D. Lewek, "The influence of body weight support on ankle mechanics during treadmill walking," *Journal of biomechanics,* vol. 44, pp. 128-133, 2011.

[104] A. J. Threlkeld, L. D. Cooper, B. P. Monger, A. N. Craven, and H. G. Haupt, "Temporospatial and kinematic gait alterations during treadmill walking with body weight suspension," *Gait & posture,* vol. 17, pp. 235-245, 2003.

[105] H. Barbeau and M. Visintin, "Optimal outcomes obtained with body-weight support combined with treadmill training in stroke subjects," *Archives of physical medicine and rehabilitation,* vol. 84, pp. 1458-1465, 2003.

[106] S. H. Peurala, I. M. Tarkka, K. Pitkänen, and J. Sivenius, "The effectiveness of body weight-supported gait training and floor walking in patients with chronic stroke," *Archives of physical medicine and rehabilitation,* vol. 86, pp. 1557-1564, 2005.

[107] T. Senthilvelkumar, H. Magimairaj, J. Fletcher, G. Tharion, and J. George, "Comparison of body weight-supported treadmill training versus body weight-supported overground training in people with incomplete tetraplegia: a pilot randomized trial," *Clinical rehabilitation,* vol. 29, pp. 42-49, 2015.

[108] J. Kressler, M. S. Nash, P. A. Burns, and E. C. Field-Fote, "Metabolic responses to 4 different body weight-supported locomotor training approaches in persons with incomplete spinal cord injury," *Archives of physical medicine and rehabilitation,* vol. 94, pp. 1436-1442, 2013.

[109] 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.

[110] M. L. Celestino, G. L. Gama, and A. M. Barela, "Gait characteristics of children with cerebral palsy as they walk with body weight unloading on a treadmill and over the ground," *Research in developmental disabilities,* vol. 35, pp. 3624-3631, 2014.

[111] C. O. Sousa, J. A. Barela, C. L. Prado-Medeiros, T. F. Salvini, and A. M. Barela, "Gait training with partial body weight support during overground walking for individuals with chronic stroke: a pilot study," *Journal of neuroengineering and rehabilitation,* vol. 8, p. 48, 2011.

[112] M. K. Aaslund, J. L. Helbostad, and R. Moe-Nilssen, "Walking during body-weight-supported treadmill training and acute responses to varying walking speed and body-weight support in ambulatory patients post-stroke," *Physiotherapy theory and practice,* vol. 29, pp. 278-289, 2013.

[113] A. S. Delussu, G. Morone, M. Iosa, M. Bragoni, M. Traballesi, and S. Paolucci, "Physiological responses and energy cost of walking on the Gait

Trainer with and without body weight support in subacute stroke patients," *Journal of neuroengineering and rehabilitation*, vol. 11, p. 54, 2014.

[114] M. J. Kurz, W. Stuberg, and S. L. DeJong, "Body weight supported treadmill training improves the regularity of the stepping kinematics in children with cerebral palsy," *Developmental neurorehabilitation,* vol. 14, pp. 87-93, 2011.

[115] M. H. Rose, A. Løkkegaard, S. Sonne-Holm, and B. R. Jensen, "Improved clinical status, quality of life, and walking capacity in Parkinson's disease after body weight-supported high-intensity locomotor training," *Archives of physical medicine and rehabilitation,* vol. 94, pp. 687-692, 2013.

[116] I. Miyai, Y. Fujimoto, Y. Ueda, H. Yamamoto, S. Nozaki, T. Saito, *et al.*, "Treadmill training with body weight support: its effect on Parkinson's disease," *Archives of physical medicine and rehabilitation,* vol. 81, pp. 849-852, 2000.

[117] H. Van Hedel, L. Tomatis, and R. Müller, "Modulation of leg muscle activity and gait kinematics by walking speed and bodyweight unloading," *Gait & posture,* vol. 24, pp. 35-45, 2006.

[118] J. M. Burnfield, S. L. Irons, T. W. Buster, A. P. Taylor, G. A. Hildner, and Y. Shu, "Comparative analysis of speed's impact on muscle demands during partial body weight support motor-assisted elliptical training," *Gait & posture,* vol. 39, pp. 314-320, 2014.

[119] M. K. Aaslund and R. Moe-Nilssen, "Treadmill walking with body weight support: Effect of treadmill, harness and body weight support systems," *Gait & posture,* vol. 28, pp. 303-308, 2008.

[120] 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.

[121] A. D. Kuo and J. M. Donelan, "Dynamic principles of gait and their clinical implications," *Physical therapy,* vol. 90, pp. 157-174, 2010.

[122] J. M. Donelan, R. Kram, and A. D. Kuo, "Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking," *Journal of Experimental Biology,* vol. 205, pp. 3717-3727, 2002.

[123] H. J. Hermens, B. Freriks, R. Merletti, D. Stegeman, J. Blok, G. Rau, *et al.*, "European recommendations for surface electromyography," *Roessingh Research and Development,* vol. 8, pp. 13-54, 1999.

[124] C. M. O'Connor, S. K. Thorpe, M. J. O'Malley, and C. L. Vaughan, "Automatic detection of gait events using kinematic data," *Gait & posture,* vol. 25, pp. 469-474, 2007.

[125] A. G. Fischer and A. Wolf, "Assessment of the Effects of Body Weight Unloading on Overground Gait Biomechanical Parameters," *Clinical Biomechanics,* 2015.

[126] Y. Ivanenko, R. Grasso, V. Macellari, and F. Lacquaniti, "Control of foot trajectory in human locomotion: role of ground contact forces in simulated reduced gravity," *Journal of neurophysiology,* vol. 87, pp. 3070-3089, 2002.

[127] J. Perry and J. R. Davids, "Gait analysis: normal and pathological function," *Journal of Pediatric Orthopaedics,* vol. 12, p. 815, 1992.

[128] J. M. Donelan, R. Kram, and A. D. Kuo, "Simultaneous positive and negative external mechanical work in human walking," *Journal of biomechanics,* vol. 35, pp. 117-124, 2002.

[129] A. D. Kuo, J. M. Donelan, and A. Ruina, "Energetic consequences of walking like an inverted pendulum: step-to-step transitions," *Exercise and sport sciences reviews,* vol. 33, pp. 88-97, 2005.

[130] D. H. Sutherland and J. R. Davids, "Common gait abnormalities of the knee in cerebral palsy," *Clinical Orthopaedics and Related Research,* vol. 288, pp. 139-147, 1993.

[131] P. H. Chen, R. L. Wang, D. J. Liou, and J. S. Shaw, "Gait Disorders in Parkinson's Disease: Assessment and Management," *International Journal of Gerontology,* vol. 7, pp. 189-193, 2013.

[132] 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.

[133] J. Perry, T. Slac, and J. R. Davids, "Gait Analysis: Normal and Pathological Function," *Journal of Pediatric Orthopaedics,* vol. 12, p. 815, 1992.

[134] R. Labruyere and H. J. van Hedel, "Strength training versus robot-assisted gait training after incomplete spinal cord injury: a randomized pilot study in patients depending on walking assistance," *Journal of NeuroEngineering and Rehabilitation,* vol. 11, p. 4, 2014.

[135] R. Topp, A. Mikesky, J. Wigglesworth, W. Holt, and J. E. Edwards, "The Effect of a 12-week Dynamic Resistance Strength Training Program on Gait Velocity and Balance of Older Adults," *The Gerontologist,* vol. 33, pp. 501-506, August 1, 1993 1993.

[136] U. B. Flansbjer, M. Miller, D. Downham, and J. Lexell, "Progressive resistance training after stroke: effects on muscle strength, muscle tone, gait performance and perceived participation," *J Rehabil Med,* vol. 40, pp. 42-8, Jan 2008.

[137] T. A. Scandalis, A. Bosak, J. C. Berliner, L. L. Helman, and M. R. Wells, "Resistance Training and Gait Function in Patients with Parkinson's Disease," *American Journal of Physical Medicine & Rehabilitation,* vol. 80, pp. 38-43, 2001.

[138] C. J. Hass, T. A. Buckley, C. Pitsikoulis, and E. J. Barthelemy, "Progressive resistance training improves gait initiation in individuals with Parkinson's disease," *Gait & posture,* vol. 35, pp. 669-673, 2012.

[139] J. J. Eng and P.-F. Tang, "Gait training strategies to optimize walking ability in people with stroke: a synthesis of the evidence," 2007.

[140] K. J. Sullivan, D. A. Brown, T. Klassen, S. Mulroy, T. Ge, S. P. Azen, *et al.*, "Effects of task-specific locomotor and strength training in adults who were ambulatory after stroke: results of the STEPS randomized clinical trial," *Physical therapy,* vol. 87, pp. 1580-1602, 2007.

[141] G. Williams, M. Kahn, and A. Randall, "Strength training for walking in neurologic rehabilitation is not task specific: a focused review," *American Journal of Physical Medicine & Rehabilitation,* vol. 93, pp. 511-522, 2014.

[142] J. L. Emken and D. J. Reinkensmeyer, "Robot-enhanced motor learning: accelerating internal model formation during locomotion by transient

dynamic amplification," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 13, pp. 33-39, 2005.
[143] T. Lam, M. Anderschitz, and V. Dietz, "Contribution of feedback and feedforward strategies to locomotor adaptations," *Journal of Neurophysiology,* vol. 95, pp. 766-773, 2006.
[144] Y.-H. Chang and R. Kram, "Metabolic cost of generating horizontal forces during human running," *Journal of Applied Physiology,* vol. 86, pp. 1657-1662, 1999.
[145] A. Blanchette and L. J. Bouyer, "Timing-specific transfer of adapted muscle activity after walking in an elastic force field," *Journal of neurophysiology,* vol. 102, pp. 568-577, 2009.
[146] T. Lam, M. Wirz, L. Lünenburger, and V. Dietz, "Swing phase resistance enhances flexor muscle activity during treadmill locomotion in incomplete spinal cord injury," *Neurorehabilitation and Neural Repair,* vol. 22, pp. 438-446, 2008.
[147] K. R. Mun, H. Yu, C. Zhu, and M. S. Cruz, "Design of a novel robotic over-ground walking device for gait rehabilitation," in *Advanced Motion Control (AMC), 2014 IEEE 13th International Workshop on*, 2014, pp. 458-463.
[148] M. Spanjaard, N. D. Reeves, J. H. van Dieën, V. Baltzopoulos, and C. N. Maganaris, *Gastrocnemius muscle fascicle behavior during stair negotiation in humans* vol. 102, 2007.
[149] H. J. Hermens, B. Freriks, R. Merletti, D. Stegeman, J. Blok, G. Rau, *et al.*, "European recommendations for surface electromyography," *Roessingh Research and Development, Enschede,* 1999.
[150] C. M. O'Connor, S. K. Thorpe, M. J. O'Malley, and C. L. Vaughan, "Automatic detection of gait events using kinematic data," *Gait Posture,* vol. 25, pp. 469-74, Mar 2007.
[151] A. Phinyomark, C. Limsakul, H. Hu, P. Phukpattaranont, and S. Thongpanja, *The usefulness of mean and median frequencies in electromyography analysis*: INTECH Open Access Publisher, 2012.
[152] P. Aagaard, E. B. Simonsen, J. L. Andersen, P. Magnusson, and P. Dyhre-Poulsen, "Neural adaptation to resistance training: changes in evoked V-wave and H-reflex responses," *Journal of Applied Physiology,* vol. 92, pp. 2309-2318, 2002.
[153] W. Scott, J. Stevens, and S. A. Binder–Macleod, "Human skeletal muscle fiber type classifications," *Physical therapy,* vol. 81, pp. 1810-1816, 2001.
[154] J. O. Holloszy and F. W. Booth, "Biochemical adaptations to endurance exercise in muscle," *Annual review of physiology,* vol. 38, pp. 273-291, 1976.

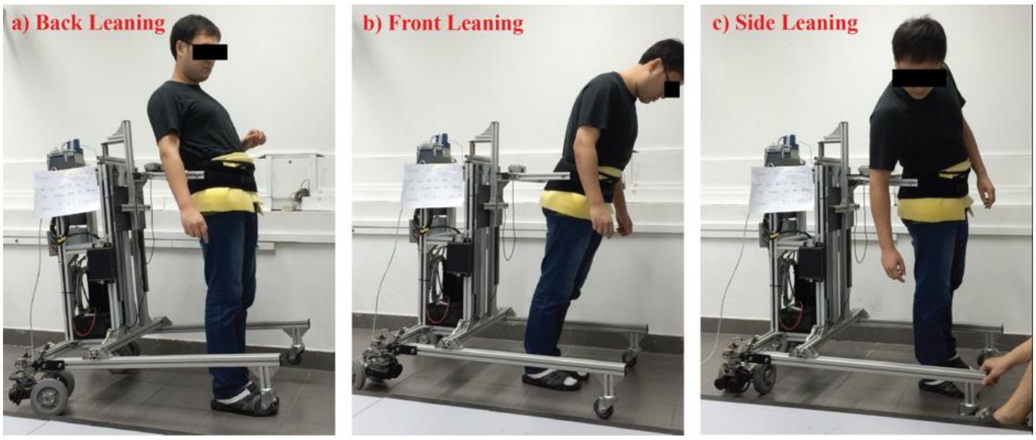
APPENDICES

Appendix A: Safety test of the robotic walker

1. Static stability of the Robotic Walker

Safety is of utmost importance in robotic gait rehabilitation for neurologically challenged patients. For example, falling is the most serious problem for the user and fear of falling can increase anxiety in patients as well as decrease the effectiveness of the intervention. To look into the static stability of the robotic walker, we have empirically tested force and torque required.

Static stability tests were conducted to determine the force and torque required for a subject to tip the walker when the robotic walker was immobile. The subject (75 kg) leaned towards the front, back and sides (left and right) (Figure A). For the front leaning test, subject stopped when the walker was being dragged forward. For the back and side leaning tests, subject stopped when two of the wheels were lifted off the ground. The subject conducted three repeats and the results are summarized in Table A and an example of a repeat is shown in Figure B. Typical force and torque values of normal walking are presented in Table B, and are significantly lower than those recorded in Table A**Error! Reference source not found.**. The forward and backward forces for front-back leaning are about 5 times bigger than the normal walking, and the force for side leaning requires 7-8 times higher than the normal walking. The torque required for falling also showed significantly higher than the normal walking (See Table A and B).



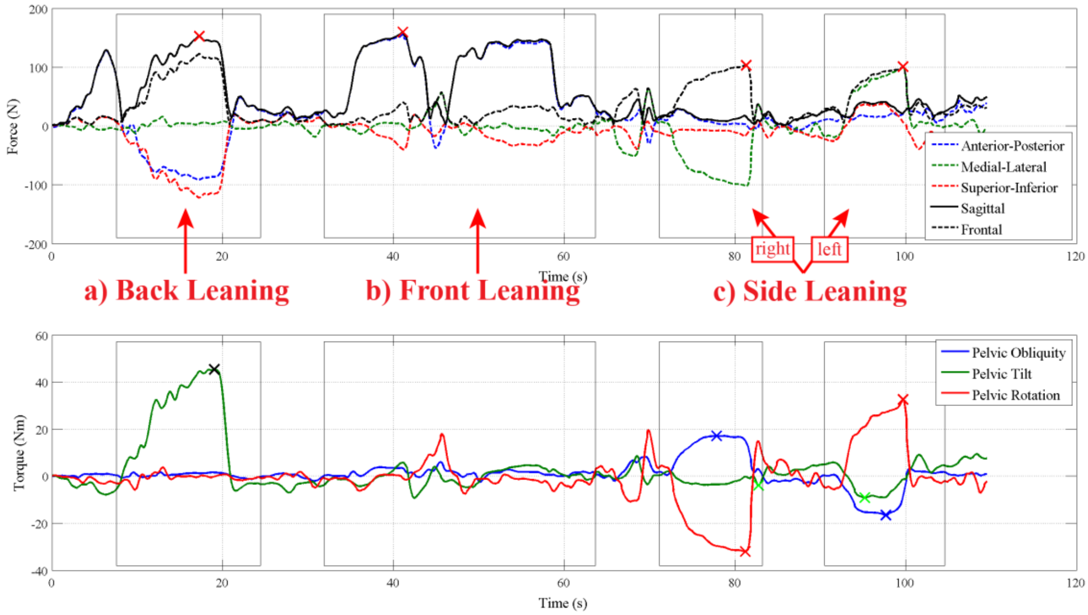
A. Static stability test of the robotic Walker

Table A. Summary of stability test. All values are absolute values. Side includes the mean of both left and right.

	Plane	Back (n = 3)	Front (n = 3)	Side (n = 6)
Force (N)	-	135.9 ±14.5	141.2 ±18.4	98.2 ±10.4
Torque (Nm)	X(Pelvic Obliquity)	-	-	17.4 ±3.4
	Y(Pelvic Tilt)	38 ±6.6	-	10.7 ±6.5
	Z(Pelvic Rotation)	-	-	30.8 ±2.5

Table B. Force and Torque required for normal walking.

	Normal Walking
Forward Force	25-35 (N)
Lateral Force	12-15 (N)
Tx (Pelvic Obliquity)	2.7 (N·m)
Ty (Pelvic Tilt)	2 (N·m)
Tz (Pelvic Rotation)	6 (N·m)



B. Example of result of static stability test

2. Fall prevention Methods

This data presented above proves the stability of the robotic walker. Although the robotic walker is safe in the static situation, it also needs to achieve dynamic stability and to provide fall prevention during gait training. To prevent the occurrence of falls during the experiments, we implemented 3 types of fall prevention functions into the walker as follows: (1) set the range of velocity in three-dimension of forward-backward, lateral, and pelvic rotational velocity, (2) a safety button with remote controller was provided, and (3) emergency switch was used to allow the walker to stop moving instantly in view of any possible dangerous situation.

For the function 1), the walker will immediately stop moving when the Walker has been reached to a threshold speed. The speed limit is currently set at 1.0 m/s forward velocity and 0.4 m/s lateral velocity. The experimenter can adjust these values based on the physical capabilities (i.e. natural walking speed) of subjects. For the function 2) and 3), a minimum of two experimenters will be at the side of the patient to provide support when required at all times. In any case of falling events, one will control the remote controller and the other will regulate the emergency stop switch, then the subject will be attended to by the experimenters. Therefore, this device allows the users for their own control that could increase motivation of the subjects and provide correct sensory input during the gait without anxiety of falling. This could potentially promote functional outcomes after gait rehabilitation.

Journal Publications and Conference Proceedings

1. **<u>K.R. Mun</u>**, Z. Guo, and H. Yu, "Restriction of pelvic lateral and rotational motions alters lower limb kinematics and muscle activation pattern during over-ground walking", Gait & Posture, submitted, 2015.

2. **<u>K.R. Mun</u>**, S.B. Lim, Z. Guo, and H. Yu, "Biomechanical effects of body weight support with a novel robotic walker for over-ground rehabilitation", Journal of Biomechanics, submitted, 2015.

3. **<u>K.R. Mun</u>**, B.S. Yew, Z. Guo, and H. Yu, "Resistance training using a novel robotic walker for over-ground gait rehabilitation: a preliminary study on healthy subjects", Journal of Human Movement Science, submitted, 2015.

4. **<u>K.R. Mun</u>**, Z. Guo, and H. Yu, "Development and Evaluation of a Novel Over-ground Robotic Walker for Pelvic Motion Support", 14th IEEE/RAS-EMBS International Conference on Rehabilitation Robotics (ICORR), Singapore, 2015

5. **<u>K.R. Mun</u>**, H. Yu, and C. Zhu, "Design of a Novel Robotic Over-Ground Walking Device for Gait Rehabilitation", The 13th International Workshop on Advanced Motion Control (AMC), Japan, 2014

Conference Presentation

1. **K.R. Mun**, Z. Guo, and H. Yu, "Development and Evaluation of a Novel Over-ground Robotic Walker for Pelvic Motion Support", 14th IEEE/RAS-EMBS International Conference on Rehabilitation Robotics (ICORR), Singapore, 2015

2. **K.R. Mun**, H. Yu, and C. Zhu, "Design of a Novel Robotic Over-Ground Walking Device for Gait Rehabilitation", The 13th International Workshop on Advanced Motion Control (AMC), Japan, 2014

3. **K.R. Mun** and H. Yu," A Novel Robotic Walker for Over-ground Gait Rehabilitation", World Congress of Gerontology and Geriatrics in Korea, IAGG, 2013

4. **K.R. Mun** and H. Yu, "A novel robotic walker for over-ground gait rehabilitation", Asia-korean conference in Singapore, 2013