ABSTRACT

A NOVEL APPROACH TO USER CONTROLLED AMBULATION OF LOWER EXTREMITY EXOSKELETONS USING ADMITTANCE CONTROL PARADIGM

by
Kiran Kartika Karunakaran

The robotic lower extremity exoskeletons address the ambulatory problems confronting individuals with paraplegia. Paraplegia due to spinal cord injury (SCI) can cause motor deficit to the lower extremities leading to inability to walk. Though wheelchairs provide mobility to the user, they do not provide support to all activities of everyday living to individuals with paraplegia.

Current research is addressing the issue of ambulation through the use of wearable exoskeletons that are pre-programmed. There are currently four exoskeletons in the U.S. market: Ekso, Rewalk, REX and Indego. All of the currently available exoskeletons have 2 active Degrees of Freedom (DOF) except for REX which has 5 active DOF. All of them have pre-programmed gait giving the user the ability to initiate a gait but not the ability to control the stride amplitude (height), stride frequency or stride length, and hence restricting users' ability to navigate across different surfaces and obstacles that are commonly encountered in the community. Most current exoskeletons do not have motors for abduction or adduction to provide users with the option for movement in coronal plane, hence restricting user's ability to effectively use the exoskeletons. These limitations of currently available pre-programmed exoskeleton models are sought to be overcome by an intuitive, real time user-controlled control mechanism employing admittance control by using hand-trajectory as a surrogate for foot

trajectory. Preliminary study included subjects controlling the trajectory of the foot in a virtual environment using their contralateral hand. The study proved that hands could produce trajectories similar to human foot trajectories when provided with haptic and visual feedback. A 10 DOF 1/2 scale biped robot was built to test the control paradigm. The robot has 5 DOF on each leg with 2 DOF at the hip to provide flexion/extension and abduction/adduction, 1 DOF at the knee to provide flexion and 2 DOF at the ankle to provide flexion/extension and inversion/eversion. The control mechanism translates the trajectory of each hand into the trajectory of the ipsilateral foot in real time, thus providing the user with the ability to control each leg in both sagittal and coronal planes using the admittance control paradigm. The efficiency of the control mechanism was evaluated in a study using healthy subjects controlling the robot on a treadmill. A trekking pole was attached to each foot of the biped. The subjects controlled the trajectory of the foot of the biped by applying small forces in the direction of the required movement to the trekking pole through a force sensor. The algorithm converted the forces to Cartesian position of the foot in real time using admittance control; the Cartesian position was converted to joint angles of the hip and knee using inverse kinematics. The kinematics, synchrony and smoothness of the trajectory produced by the biped robot was evaluated at different speeds, with and without obstacles, and compared with typical walking by human subjects on the treadmill. Further, the cognitive load required to control the biped on the treadmill was evaluated and the effect of speed and obstacles with cognitive load on the kinematics, synchrony and smoothness was analyzed.

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A NOVEL APPROACH TO USER CONTROLLED AMBULATION OF LOWER EXTREMITY EXOSKELETONS USING ADMITTANCE CONTROL PARADIGM

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TABLE OF CONTENTS

TABLE OF CONTENTS
(Continued)

TABLE OF CONTENTS
(Continued)

TABLE OF CONTENTS
(Continued)

TABLE OF CONTENTS
(Continued)

LIST OF TABLES

LIST OF FIGURES

LIST OF FIGURES
(Continued)

LIST OF FIGURES
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CHAPTER 1

INTRODUCTION

1.1 Objective

'*My disability exists not because I use a wheelchair, but because the broader environment isn't accessible'*- **Stella Young**

Individuals with paraplegia due to spinal cord injury (SCI) have motor and/or sensory deficits leading to an inability to walk, and, therefore rely on wheelchairs for mobility. Although wheelchairs provide mobility, they do not support all activities of everyday living. Current research addresses mobility using wearable lower extremity exoskeletons. Most of the currently available wearable exoskeletons have only two degrees of freedom (DOF) per leg in the sagittal plane with some of them providing passive control of the ankle in the sagittal plane. Also, all of the currently available exoskeletons are pre-programmed with the user having the ability to initiate gait patterns but having no control over the stride length, stride frequency or stride amplitude (height), hence restricting the user's ability to navigate irregular surfaces and obstacles. The objective of this dissertation is to address the current limitations by an intuitive, real time user- controlled control mechanism that uses hand-trajectory as a surrogate for foot trajectory.

The control mechanism translates walking-like movements produced by the hand to kinematics of gait of the exoskeleton in real time. This approach was tested on a 10 DOF, 1/2 scale robot. The robot has 5 DOF on each leg with 2 DOF at the hip to provide flexion/extension and abduction/adduction, 1 DOF at the knee to provide flexion and 2 DOF at the ankle to provide flexion/extension and inversion/eversion. The control mechanism employs admittance control to translate the trajectory of each hand into the trajectory of the ipsilateral foot in real time, thus providing the user with ability to control each leg in both sagittal and coronal planes.

Hypothesis: The trajectory produced by the hands while replicating the human gait pattern can provide an intuitive control mechanism required to control an exoskeleton and also provide the user with the ability to perform the normal gait cycle and navigate obstacles.

Preliminary Study: This study included mapping hand trajectories to foot trajectories in a virtual environment to validate the need for the use of haptic feedback in an effective control mechanism for the exoskeleton.

Specific Aim 1: To design a real time control mechanism for a 10 DOF 1/2 scale exoskeleton prototype.

Sub Aim 1: To build a 3 DOF leg for achieving swing cycle with joints at the hip, knee and ankle in the sagittal plane that will be controlled by the hand trajectories using admittance control.

The aim is to design a prototype for an exoskeleton for the lower extremities which would conform to the anthropometry of human lower extremities and provide users with the ability to control the amplitude (height of the stride) and stride length of the gait in real time in the sagittal plane. The prototype would be a ½ scale biped that would have actuators at the hip, knee and ankle to provide 3 DOF in the sagittal plane.

Sub Aim 2: To add 2 more DOF to the leg, one at the hip and ankle to provide control in coronal plane.

Humans navigate obstacles by abduction/adduction of the hip and ankle in the coronal plane in combination with flexion and extension of hip, knee and ankle in sagittal plane. The current exoskeletons do not have motors for abduction or adduction to provide users with the option for movement in coronal plane, hence restricting user's ability to effectively use the exoskeletons in everyday activities. The aim is to design a prototype to provide users with the ability to control the leg in coronal plane in addition to sagittal plane.

Sub Aim 3: To develop an algorithm to control the gait cycle of both legs with transition from stance phase to swing phase and vice versa.

Typical human gait cycle consists of approximately 40% swing phase and 60% stance phase. Gait comprises of one leg in stance phase and the other leg in swing phase or both legs in stance. The leg in stance phase has the hip and ankle joint contributing to the gait in accordance with that of the swing leg. The aim is to develop two control paradigms to control the gait cycle: Hybrid Control and Complete Control. The Hybrid Control algorithm involves mapping one hand trajectory to its ipsilateral foot trajectory of the swing leg while the algorithm moves the stance leg to the required joint angles to complete the gait step. The Complete Control algorithm allows the user's hands to control both swing and stance legs.

Specific Aim 2: To compare the gait of the biped with human gait patterns in terms of trajectory and obstacle navigation.

The aim is to show that the kinematics generated using the hand resemble normal human gait kinematics and to identify if the user can navigate obstacles by controlling the gait using hands.

Sub Aim 1: To compare gait patterns of human gait in varying speeds with that of biped gait on a treadmill.

The aim of this study is to evaluate the gait of the biped to that of normal human gait. The aim is to compare and prove that the gait of the robot can be controlled by the user under constant speed conditions and as well as under varying speed conditions.

Sub Aim 2: Compare the gait patterns of exoskeleton and healthy subjects in the presence of obstacles.

The goal is to investigate the similarities in obstacle navigation of hand controlled bipeds and humans with normal functional lower extremities. Current exoskeletons are pre-programmed to gait patterns in sagittal plane and do not provide the option to change the stride length or stride amplitude (height of the stride). This aim evaluates this hand controlled biped robot's foot trajectories with human gait trajectories in sagittal plane in the presence of obstacles.

Specific Aim 3: To evaluate and compare the influence of cognitive load on gait of the biped with that of human gait pattern in terms of trajectory and obstacle navigation.

Central pattern generators in spinal cord are believed to produce cyclic gait patterns leading, therefore to lower cognitive load during walking. The objective is to evaluate the effect of cognitive task on the kinematics of the biped walking and also to evaluate the cognitive load of controlling the gait of the biped with hands.

Sub Aim 1: To compare gait patterns of human gait and of biped gait on a treadmill at varying speeds while also performing a cognitive task.

The goal is to evaluate the cognitive load required to perform the task of controlling the biped using hands as compared to normal human walking at constant speeds and varying speeds. Also, evaluate the effect of the cognitive task on the kinematics of the biped gait.

Sub Aim 2: To compare the gait patterns of exoskeleton and healthy subjects in the presence of obstacles at varying speeds while also performing a cognitive task.

The objective is to investigate the cognitive load required to control the biped using hands as compared to normal human walking while navigating obstacles at constant speeds and varying speeds.

1.2 Background and Significance

Spinal cord injury (SCI) is usually a result of fracture or dislocation of vertebrae caused by sudden traumatic blow to the spine (http://www.ninds.nih.gov/disorders/sci/sci.htm). As this spinal cord is a transmission line that carries neuronal signals, SCI affects communication between the brain and the extremities. SCI can be partial or complete depending on the severity of the injury. In a complete spinal cord injury, the cord is severed and no signals are transmitted beyond the point of injury, resulting in a complete loss of motor and/or sensory functions on the regions below the injury. In partial spinal cord injury, some of the signals are transmitted; hence the individual with injury may retain some motor and/or sensory functions below the point of injury (http://www.ninds.nih.gov/disorders/sci/sci.htm).

Motor and/or sensory function that is lost is determined by the level of the injury/lesion on the spinal cord. There are, therefore, two main classifications of SCI based on the level of the injury/lesion on the spinal cord viz. paraplegia and tetraplegia. Tetraplegia (also known as quadriplegia) is caused by injuries or lesions to the cervical segments of spinal cord, resulting in complete or incomplete paralysis to both upper and lower extremities (http://www.spinalinjury101.org/details). Paraplegia is caused by injuries or lesions to the thoracic, lumbar or sacral regions (i.e. T1 or below Figure 1.1 a) of the spinal cord, resulting in complete or partial paralysis of the lower extremities (http://www.spinalinjury101.org/details).

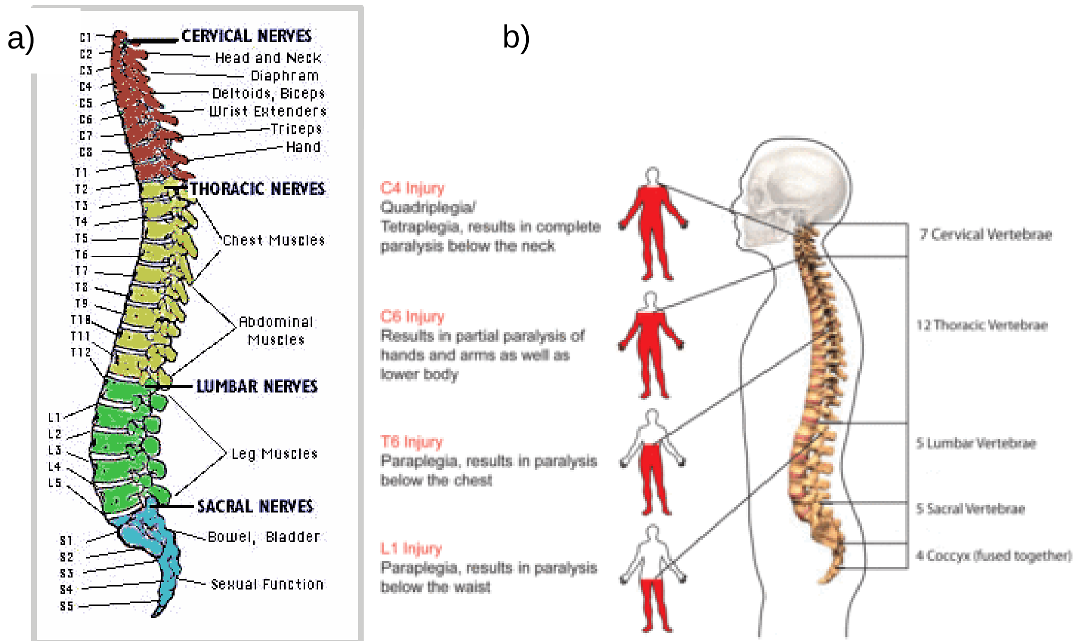


Figure1.1 a) Diagram showing the relationship between vertebrae and function. b) Diagram showing the relationship between vertebrae and level of injury and kind of injury.
Sources: http://www.atlantainjurylawyer.com/spinal-cord-injury.html, http://www.bel13vefoundation.org/spinal-cord-injury.

Injury or lesion to thoracic nerves T1 –T5 affects mid back and abdominal muscles and results in the inability of the individual to use his/her trunk or legs. Injury or lesion to thoracic nerves T6 –T12 affects functions below the abdominal or back muscles, and the individual will have normal upper body movement with the ability to control and balance the trunk but will have no voluntary control of lower extremities. Injuries to lumbar and sacral regions also cause some loss of function to lower extremities especially legs and hips (http://www.spinalinjury101.org/details) as shown in Figure 1.1 b. Paraplegia and tetraplegia can further be classified as complete or incomplete; individuals with sensory and motor impairment to their lower extremities due to complete injuries or lesions to thoracic levels or below are referred to as individuals with complete paraplegia, and individuals with sensory and motor impairment to their upper and lower extremities due to complete injuries or lesions to cervical levels or below are referred to as individuals with complete tetraplegia.

One visible effect of paraplegia is the inability of individuals to walk, and hence their need to rely on wheelchairs for mobility. Inability to walk contributes to a number of medical complications including pressure ulcers (decubitus), thrombosis, fractures, cardiovascular conditioning, pulmonary embolism, decreased muscle mass, diabetes and obesity (http://www.spinalinjury101.org/details). Secondary medical complications play an important role in the continuing care for people with SCI as they increase the lifetime cost of care (McKinley, 1999). Allowing these individuals to walk and use their lower extremities would help in reducing the secondary complications and hence reduce the cost of care. Pressure ulcers and fractures in the lower extremity are the most frequent secondary complications (McKinley, 1999). A pressure ulcer is caused by reduction in capillary blood flow due to prolonged wheel chair use or staying in bed. Pressure ulcers are the most frequent secondary complication and the likelihood of contracting a pressure ulcer increases with years with the injury (McKinley, 1999). Also, the level of injury has no significant effect on the likelihood of individuals developing pressure ulcers, but the severity of the ulcer is higher in people with complete paraplegia and complete tetraplegia (McKinley, 1999). Sublesional osteoporosis due to inactivity is observed in individuals with SCI due to reduced muscle activity and mechanical loading, resulting in bone loss and muscle atrophy. The fracture rate especially in long bones in the lower extremities was observed to be high in people with SCI (McKinley, 1999). It was observed that bone loss does not plateau but continues through the years, hence leading to increase in number of fractures through the life time (Giangregorio, 2006, Frotzler, 2015). In addition to bone loss, alterations to bone area and geometry were also reported. Muscle atrophy was observed between 6 to 24 weeks post injury by about 16% (Giangregorio, 2006). This reduction in muscle also leads to decreased metabolic rate and increased storage as the energy intake is not adequately adjusted to energy expenditure leading to obesity (Giangregorio, 2006).

Secondary complications are important to address as they can impair the individual's functional ability, interfere with employment and educational pursuit. It could also lead to lost work days, necessity for increased attendant or skilled care (McKinley, 1999).

Studies have shown that there is increased muscle mass and muscle area in individuals with acute SCI with treadmill training with body weight support but very little change was observed with just standing (McKinley, 1999). Hence, walking with exoskeletons could aid in mitigating the above mentioned secondary effects.

There are currently 273,000 persons living in United States with Spinal cord injury (SCI) and there are approximately 12,000 new cases added every year. Also, more than half the injuries occurred among young adults between the ages of 16 and 30 and their average life expectancy is 45 years for paraplegics and 40 years for low tetraplegics and 36 years for high tetraplegics. This statistics emphasizes the need to address the various difficulties faced by the individuals with paraplegia as the population of SCI is a growing population (http://www.sci-info-pages.com/facts.html). Estimated lifetime costs often exceed $1, 00, 000 and average yearly costs for rehospitalization, emergency room, physician visit costs exceed $5000 (McKinley, 1999).

The long term goal of research in SCI would be to cure paralysis by axonal growth/regeneration (Anderson, 2004). Until the last several decades, SCI was considered irreversible, which was proven wrong by the research advancement in axonal growth/regeneration. Researchers have shown significant advancement in regenerating neurons in rodents to enhance functions such as bladder control and respiratory function (Anderson, 2004). In spite of such great progress, there are still many unknowns with regard to successful regeneration of neurons and axons in chronic

SCI. The research on axonal growth/regeneration in humans is decades away. Hence a more realistic approach at this juncture would be to improve the quality of life of people with SCI.

Table 1.1 Average Yearly Expenses and Estimated Lifetime Costs in SCI

Severity Of Injury	Average Yearly Expenses		Estimated Lifetime Cost By Age At Injury	
	First Year	Each Subsequent Year	25 years old	50 years old
High Tetraplegia (c1-c4)	$1,044,197	$181,328	$4,633,137	$2,546,2954
Low Tetraplegia(c5-c8)	$754,524	$111,237	$3,385,259	$2,082,237
Paraplegia	$508,904	$67,415	$3,265,84	$1,486,835
Incomplete Motor Functional at any Level	$ 340,787	$41,393	$1,547,858	$1,092,521

Source: https://www.nscisc.uab.edu

Anderson et al. studied the priorities of the individuals with paraplegia and tetraplegia in reference to improving the quality of life. They reported that individuals with paraplegia and tetraplegia considered walking as one among the top three priorities. Also, the time after injury did not influence the preference of walking; implying that walking remained a top priority for individuals with paraplegia and tetraplegia even after years of injury. Their study also noted that individuals with SCI believed exercise by walking was a major component of recovery (Anderson, 2004). A similar study by Ditunno et al. used consensus to compare walking functions to other functions to investigate the recovery preferences of individuals with SCI (Ditunno, 2008). Their study developed a survey for the individuals with SCI and as well for the rehabilitation professionals working with people with SCI to evaluate the difference in relative value of various functional activities. The features assessed were 14 modified functional independent measures (MFIM) which included six items of self-care (eating, grooming, bathing, dressing, and toileting), sphincter management (bladder and bowel), wheelchair and walking. The survey asked each panelist to give their highest preference. The study concluded that for most consumer panels walking is a highly desired goal relative to other functions. Eight out of nine consumer panels placed walking at a high level next to bladder function. This supports the hypothesis that consumers with SCI express a preference for restoration of walking over most other functions on the MFIM. It is apparent that as early as stage 3, rehab professionals preferred wheelchairs and not walking, and while consumers ranked walking and not wheelchair use as high priority. Thus the discrepancy related to walk/wheelchair is also reflected between consumers and rehab professionals, where rehabilitation professionals give more importance to wheelchair while consumers prefer ambulation (Ditunno, 2008).

A study by Kilgore, K.L., et al. asked people with SCI to prioritize functions that would improve the quality of life, and the response showed that 'being able to walk' was one of the top priorities of this population. They also believed that being able to stand was alone not important but being able to walk and perform various activities was important. Hence these studies further emphasize the need to restore ambulation (Kilgore, 2001).

1.3 Restoring Ambulation

1.3.1 Functional Electrical Stimulation

Functional electrical stimulation (FES) has been considered as a possible way to restore ambulation. FES involves stimulating the neurons by passing electricity using electrodes. Electrodes can be placed on the surface of the skin or be embedded in the body to achieve highly precise stimulation of the neurons (Kilgore, 2001). A current FES system allows the user to be able to stand upright and be able to walk a few steps with full body support but do not allow the user to control the gait (Kilgore, 2001).

Using FES, walking is deemed approximately 20 to 30 percent achieved. In order to approach 100 percent, improvements include having far fewer wires and electrodes in the system, better noise cancellation, and better balance and coordination, and user control of gait pattern. Also, smoother gait and better energy efficiency need to be attained. Even if the engineering problems are addressed, many consumers have stated that they had become quite proficient using their wheelchair for mobility and any alternative means of mobility would have to be more effective for them to even consider it (Kilgore, 2001). They were not interested in disrupting their lives for little or no practical gains. Yet, even among those with this opinion was heard the willingness to “go for it” if the outcomes could be guaranteed with some high degree of certainty. Users of implantable FES systems identified “reversibility” as an important feature of the systems they selected. Also, downtime that the procedure would require and the risk of surgery were also major deterrents against using FES. (Kilgore, 2001). All the above disadvantages have not made FES not a viable option for mobility.

1.3.2 Wearable Lower Extremity Exoskeletons

Current research on restoring the functionality of gait is focused on using wearable exoskeletons for mobility. Lower extremity exoskeletons are active electromechanical devices that have links and joints corresponding to those of the user, and work in tandem with the user (Dollar, 2008). Lower extremity exoskeletons have actuators that actively produce torque to the joints, thus assisting in movement when not possible by the human body (Anam, 2012).

Wearable lower extremity exoskeleton robots can be classified into three groups: assistive, rehabilitative and power enhancing robots. Power enhancing robots amplify the power (efficiency) of the user, thus enabling the user to perform tasks which otherwise the user finds difficult to perform (Dollar, 2008). Assistive and rehabilitative robots aid the user to perform everyday tasks which the user is not able to perform otherwise (Dollar, 2008). Rehabilitative lower extremity exoskeletons are predominantly used in stroke and incomplete SCI rehabilitation where the exoskeleton helps the user to stand and perform repetitive gait patterns which are initiated by the user, thus aiding in the recovery of functionality over time (Dollar, 2008). This is based on the principles of motor learning and cortical representation that repetitive task oriented movements can improve muscular strength and movement coordination in individuals with impairments such as stroke, I-SCI (Kwakkel, 1999). Assistive lower extremity exoskeletons are predominantly intended for long term use by individuals with complete SCI to perform the gait patterns for them as the user is completely unable to perform the movement. They provide support and perform gait patterns for the user.

Figure 1.2 Power Enhancing Exoskeletons: a) HARDIMAN Exoskeleton (Left) b) MIT Exoskeleton (Right).
Source: Dollar, 2008.

I. Power Enhancing Robots

The earliest device resembling an exoskeleton was built in 1890; the invention used long springs operating in parallel to the leg which helped augment the running and jumping (Dollar, 2008). The first working exoskeleton was designed as a power enhancing robot as part of the first phase of DARPA (Defense Advanced Research Projects Agency) to augment the performance of soldiers; “HARDIMAN” (Human Augmentation research and development investigation) was developed in the 1960’s and included enormous hydraulics (680 kg, 30DOF) to amplify strength of arms and legs (Dollar, 2008). Though HARDIMAN was able to amplify the power of the upper extremities, it could not assist with the lower extremities; hence was never tested with a human subject. In the mid 80’s other exoskeletons were built with similar outcomes (Dollar, 2008).

The second phase of the DARPA project included exoskeletons for human performance augmentation (EHPA). Three working exoskeletons were built under this program: BLEEX, SARCOS and MIT Exoskeleton. The first generation of BLEEX included the first energetically autonomous, load bearing exoskeleton for the lower extremity (Kazerooni, 2006, Zoss, 2000). Each leg included 3 DOF at the hip, 1 DOF at the knee and 3 DOF at the ankle. The hip and the ankle were actuated using linear hydraulic actuators, while the inversion/eversion and the flexion at the ankle were spring loaded (Kazerooni, 2006, Zoss, 2000). The exoskeleton included 8 encoders and 16 linear accelerometers to determine angular velocity and acceleration of all joints. Also, each foot included force sensors to determine the distribution of load in each foot (Kazerooni, 2006, Zoss, 2000). The control mechanism sensed the movement of the user with 8 single axis force sensors and would move the exoskeleton in the direction intended by the user (Kazerooni, 2006, Zoss, 2000). The MIT exoskeleton, on the other hand, did not rely on actuators for adding power. Instead the design used the energy stored in springs during phases of walking to enhance the power. It had 3 DOF at the hip that were spring loaded, 1 DOF at the knee that used a magnetorheological damper, and 2 DOF that employed springs for performing rotation and flexion/extension at the ankle (Dollar, 2008). The control mechanism for the exoskeleton used sensory information from strain gages and potentiometers to sense the force applied by the user in a direction and assist the user with the movement in the direction of the force. The SARCOS exoskeleton is the product of SARCOS Corporation, which uses rotary hydraulic actuators. It is able to successfully support loads up to 84 kgs and walk at a speed of 1.6 m/s. Though there have been numerous improvements to SARCOS from the initial version, their work has been predominantly restricted to load bearing capabilities and not in assistive/rehabilitative technology (Dollar, 2008).

Many other successful exoskeletons have been built as load bearing exoskeletons such as the Hanyang Exoskeleton Assistive Robot (HEXAR) that has 7 DOF per leg, two of which are powered by an electrical motor to assist the user in load bearing capabilities. (Kim, 2014). ExoClimber and ExoHiker by Berkeley and Human Universal Load Carrier (HULC) by Lockheed Martin enhances the user's strength i.e. can help carry load up to 200 pounds as well as decrease the metabolic cost of the user (http://bleex.me.berkeley.edu/research/exoskeleton/hulc).

II. Rehabilitation Robots

The lessons learnt from power enhancing exoskeletons emboldened researchers to design robots for repetitive motions to help with rehabilitation of individuals with disability requiring repetitive exercises to restore mobility. This led to the rehabilitative exoskeletons or gait trainers such as Lokomat, ALEX, LOPES. All the gait trainers have a rigid frame that provide assistance to the users as needed, and help them train by repeating the gait movement over and over again (Dollar, 2008). Some of the above devices have also been integrated with virtual environment to make the task goal oriented which in turn, helps accelerate the rehabilitation of the individual with disability (Dollar, 2008).

The Lokomat is fixed to a rigid frame providing assistance to perform highly repetitive gait using two active joints at the hip and knee and has a passive joint at the ankle (Reiner, 2012). It provides rehabilitative assistance for individuals with stroke or I-SCI.

ALEX is a bilateral exoskeleton for gait rehabilitation for lower extremity. The exoskeleton comprises of a support platform and two robotic legs with 12 DOF providing assistance as required. Its unique characteristic is the possibility to actively control 12 DOF while the user walks on the treadmill when the body weight is supported to provide assistance in sagittal and coronal plane. The stiffness and assistance provided can be varied by the physician to improve the rehabilitation (Zanotto, 2013).

LOPES was designed as a rehabilitative device to be used for training on a treadmill. It has 3 DOF with 2 DOF for flexion/extension at the hip and flexion at the knee and one DOF at the hip for adduction/abduction. The adduction/abduction provides the balance of the hip in the sideways direction. Thus LOPES helps the user in the sagittal plane as well as the coronal plane. LOPES uses cable driven actuators to allow the robot to be back drivable. (Ekkelenkamp, 2007).

ANdROS exoskeleton is a 2DOF wearable and portable gait rehabilitation exoskeleton, and the control mechanism uses impedance control to help individual with lower limb paralysis. The exoskeleton has 2DOF at the hip for flexion/extension and adduction/abduction (Unluhisarcikli, 2011).

Figure 1.3 Rehabilitation Exoskeletons: a) ALEX Exoskeleton b) LOKOMAT c) LOPES.
Sources:http://engineering.columbia.edu/web/newsletter/fall_2014/sunil_agrawal%E2%80%94personalized_medicine, http://www.rcsismj.com/2009-2010-issue/wii-habilitation, http://www.neurocontrol.nl/projects/current.

Table 1.2 Characteristics of Assistive Exoskeletons

Exoskeleton	Active DOF	Pre-Programmed Gait	Control Mechanism	Ankle
ALEX	12	Yes	➢ Force-field controller is used to create a force field around the foot to conform the gait to a selected gait pattern. The gait pattern is either selected by trainer or the gait of the unimpaired leg is used.	Active
LOKOMAT	2	Yes	➢ Gait pattern selected by the Trainer. ➢ Allows the gait of the user but when it deviates away from the reference trajectory corrects the leg to the desired trajectory.	Passive
LOPES	3	Yes	➢ Gait pattern selected by the Trainer. ➢ Complementary Limb Motion Estimation (CLME) to use physiological inter joint couplings to control movement of one leg with the real time motion of the other leg.	Passive
ANdROS	2	Yes	➢ Gait pattern selected by the Trainer.	

Sources:http://engineering.columbia.edu/web/newsletter/fall_2014/sunil_agrawal%E2%80%94personalized_medicine,https://www.utwente.nl/ctw/bw/RESEARCH/PROJECTS/LOPES/INDEX.HTML, Unluhisarcikli, 2011

III. Assistive Robots

Frontiers were further pushed and current research is engaged in developing exoskeletons to assist mobility of those who otherwise cannot perform gait. These exoskeletons can also be used as assistive exoskeletons. Kazeerooni et al. researched the first assistive exoskeleton: BLEEX (Kazeerooni, 2006). The mechanical design was similar to the power enhancing exoskeleton but the control mechanism included using accelerometers and a gyroscope at the arm to measure the swing angle and force sensor embedded on the crutches to measure the landing of the crutch. When the swing angle exceeds a threshold value while both crutches are on ground, a pre-programmed step is initiated on the contralateral leg (Strausser, 2011).

Hybrid Assistive Limb (HAL) was developed as a full body suit but the rehabilitative version of HAL is a lower extremity exoskeleton with 2 DOF with one at the hip and knee, respectively. The control mechanism involves using myoelectric signals from the flexors and extensors to sense the user's intention, and thus assist the user's legs to move forward. The assistance provided by the exoskeleton can be varied to provide assistance as needed to enhance rehabilitation/ assistance. (Lee, 2002, Hayashi, 2005).

Goldfarb et al. built the lightest of all exoskeletons which was later commercialized as the Indego exoskeleton (Farris, 2012). It has two DOF, hip and knee, but no ankle support in the sagittal plane. It uses Hall effect sensors, potentiometers and accelerometers to detect the center of pressure (COP) (Farris, 2012, Quintero, 2012, and Farris, 2012). As the user's intention is detected when the user leans forward, the COP is shifted in the direction of movement, thus instructing the exoskeleton to initiate the gait in the contralateral leg (Quintero, 2012, and Farris, 2012).

Figure 1.4 Current Assistive Exoskeletons: a) BLEEX, b) HAL.
Sources: Kazerooni, 2005, Hayashi, 2005.

Rewalk is the only exoskeleton currently available with FDA approval. It has two active DOF's at the hip and knee, with the ankle consisting of a simple orthotic joint with limited motion and spring assisted dorsiflexion. The control system includes a tilt sensor to determine changes in trunk motion and center of gravity, the tilt sensor determines the angle of torso and initiates the preprogrammed hip and knee displacement in the appropriate leg (Esquenazi, 2012).

Ekso by Ekso Bionics has two options to control the gait cycle. The first is by using a button pad. The user uses buttons to transition between steps or to transition between different states i.e. between sit to stand and vice versa. The other option uses sensors embedded in the suit to detect changes in the hip position. The step is initiated by the user by moving the hip forward and laterally (Strickland, 2012, http://www.eksobionics.com).

In contrast, Rex Bionics has 5 DOF and is the only available assistive exoskeleton to provide support in all planes. It has 3 DOF at the hip, 1 DOF at the knee and 1DOF at the ankle, and uses joysticks to control the gait cycle. Though it provides balance, it is extremely slow and it is the heaviest of all exoskeletons. It does not provide the user with the ability to control the gait cycle or the stride length (http://www.rexbionics.com/).

Most of the above exoskeletons have two active degrees of freedom, one at the hip and the other at the knee.

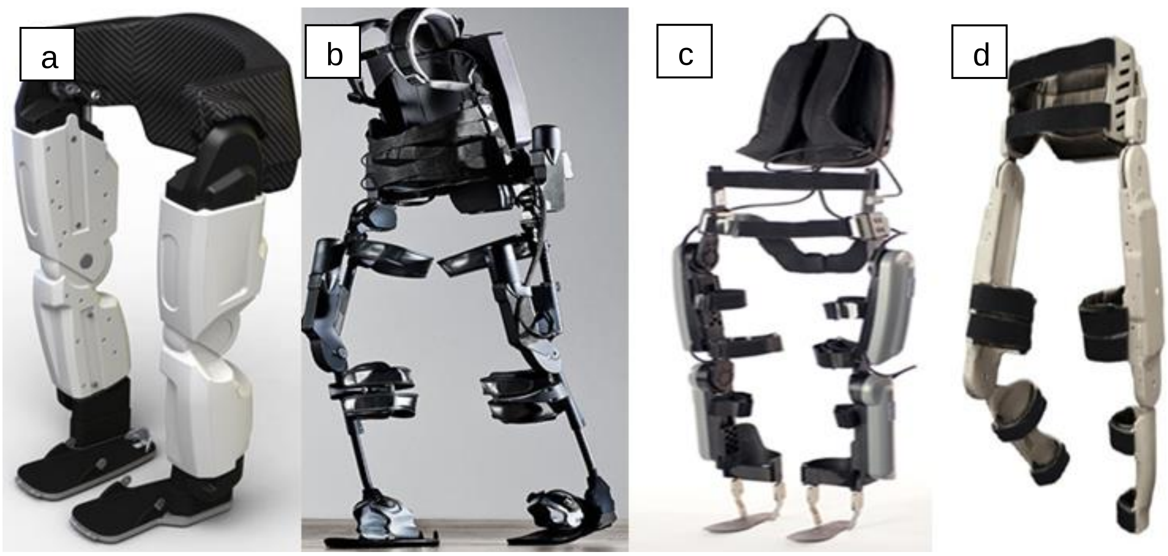


Figure 1.5 Commercial Exoskeletons: a) Rex Exoskeleton b) Ekso, c) Rewalk d) Indego. Sources: Rex bionics Personal Exoskeleton, 2015, Ekso bionics, 2011, Rewalk, 2015, Indego, 2015

Table 1.3 Assistive and Rehabilitative Exoskeletons

Exoskeleton	Active DOF	Pre-Programmed Gait	Control Mechanism	Ankle	Weight /Speed
REX	5	Yes	Joy stick	Active	38 kg/ .05m/s
Rewalk	2	Yes	Trunk movement and center of gravity & Wrist pad controller	Passive	21kg/ 1.4mph
Ekso	2	Yes	Hips forward and shifting them laterally Button commands	Passive	20 kg/ 2 mph
Indego	2	Yes	Trunk movement and location of center of pressure	Passive	11.8kg
BLEEX	5	Yes	Joy stick	Active	38 kg/ .05m/s
HAL	2	Yes	EMG	Passive	21kg/ 1.4mph

Source: Rex bionics Personal Exoskeleton, 2015. Rewalk, 2015. Ekso bionics, 2011. Indego, 2015, Strausser, 2011, Hayashi, 2005

1.3.3 Limitations of Current Assistive Exoskeletons

All the current research and commercial exoskeletons provide the user with the ability to initiate the movement but provide no control over the amplitude of the gait cycle or the length of the gait cycle and do not provide any proprioceptive feedback. Though preprogrammed gait is a big leap towards giving the people the ability to walk, these exoskeletons suffer from the major disadvantage of not being able to provide complete user control of the gait pattern, hence making it difficult to navigate through obstacles, stairs, or uneven surfaces (Ferris,2005, Hasegawa, 2009, Dollar, 2008, Mohammed, 2008). Also, humans constantly use abduction/adduction of hip and ankle to navigate obstacles which is not possible with these exoskeletons (Kobetic, 2009). The user relies on constant visual feedback to control the movement of the exoskeleton as most of the users lack sensory (force or haptic) feedback. As stated by Riener et al. “Rehabilitation devices work with patients in a “master-slave” relationship thus forcing the patients to follow a predetermined motion without consideration for voluntary efforts” (Reiner, 2014). These limitations call for better control mechanism for the user, where the user can not only initiate but also control the foot movement in real time.

1.3.4 Alternate Control Mechanisms

The current control mechanisms use either joy sticks to control the movement or simply provide the user only with the ability to initiate the movement. Any control mechanism for exoskeletons should include complete control of stride length and amplitude at all times to effectively use the exoskeletons. This could either be achieved by using signals from the central nervous system (CNS) or peripheral nervous system (PNS) or from physical interaction by the user using other articulators with sensors (Dellon, 2007, Del-Ama, 2012, and Lee, 2012).

I. Brain Computer Interface

The user intention is detected from EEG (electroencephalogram) signal from central nervous system and is translated to kinematics of the lower extremity. Brain Computer Interfaces (BCI) have been developed over the last decade, where EEG signals are recorded, interpreted and translated to actions. BCI’s have been explored to communicate the intention of the user to the lower extremity exoskeletons, thus providing the user with ability to control the gait cycle. Gancet et al. have tried to interpret EEG signals from the motor cortex to calculate the kinematics of the gait cycle. A dynamic recurrent neural network was used to train the network to detect the gait patterns in the EEG signal as shown in Figure 1.6 (Gancet, 2012).The Walk Again Project by Nicolelis et al. also used BCI to communicate the user's intention to kinematics of gait cycle. The complete gait control was to be demonstrated during the soccer World Cup by Nicolelis et al but when the user with paraplegia was driven into the soccer field, and a soccer ball was placed at his feet, all the user could do was initiate a kick as shown in Figure 1.7 (Nicolelis ,2003). This demonstration is testament to the infancy of BCI's ability to give individuals the ability to walk.

Limitations of BCI Control of Exoskeletons

Both groups have reported numerous challenges with the recording of EEG; identifying the user intention for each joint was not possible, mechanical artifacts due to relative movement of EEG cap producing random noises that are difficult to filter and physiological artifacts due to muscle activity in the vicinity of the cap. Even with extensive filtering, they were not able to completely isolate the relevant signals at all time periods (Gancet, 2012, Contreras-Vidal, 2013). Though BCI would be an ideal solution to control the exoskeleton as the control directly translates the user's intentions, it is in its infancy where the electrodes need to be implanted on the surface of the brain to remove artifacts and requires extensive training to perform the simplest tasks as current algorithms rely on identifying patterns of signals for a task (http://www.nicolelislab.net/?p=584).

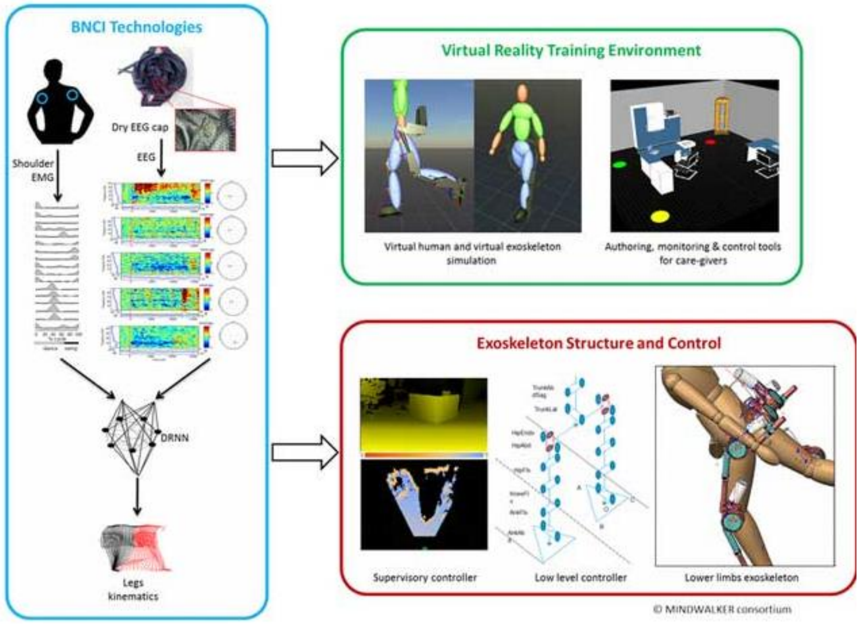


Figure 1.6 BCI control in Mindwalker project.
Source:http://www.robaid.com/bionics/mindwalker-mind-controlled-exoskeleton-could-help-disabled-people.htm

Figure 1.7 BCI control in Walk again project showing a user kicking a soccer ball.
Source: http://neurogadget.com/2014/06/13/paraplegic-man-mind-controlled-robotic-suit-kicks-world-cup-2014-video/10434

II. Electromyography Control

Signals from peripheral nervous system i.e. Electromyography (EMG) has also been used for control of lower extremity exoskeletons. Surface electrodes are attached on the surface of the skin to collect electrical activity due to active motor units in the muscles (Hasegawa, 2009). EMG signals obtained from the lower extremity could provide the kinematic information required to control each joint to produce the movement of the exoskeleton gait. Ferris et al. have used EMG signals as proportional control where the rectified EMG signals when above the set threshold were used to provide torque proportional to the magnitude and direction of EMG to move the joints to the desired position (Ferris, 2009). Hasegawa et al. placed electrodes on multiple muscles in the lower extremity in order to accurately determine the intention of the user’s movement. Further the torque required was determined using neural networks or recursive least square algorithm (Hasegawa, 2009). Yin et al. used neuro fuzzy controller which integrates the EMG signal with joint information to predict movement trajectory (Yin, 2012). Though EMG provides a better mechanism to decode the kinematics, it suffers from the disadvantage such as movement artifacts, cross talk, amount of tissue between the motor units and electrode, inability to accurately decode trajectory from the EMG signal. These disadvantages make it extremely difficult to reliably obtain EMG signals and to further provide complete control of the gait cycle.

III. Control through other articulators

The goal of this dissertation is to develop an intuitive control mechanism to generate real time gait while also providing proprioceptive feedback to the user. The control mechanism includes using other articulators to express the neural encoding of the desired trajectory instead of BCI. The project uses trajectories produced by the hand while performing walking-like movements to control foot movement. The form of neural coding of intended movement signals is not known, thus making it difficult to detect complete user intention (Feldman, 2005). The forces and the torques required to generate the movement of the foot may be computed and developed centrally or in combination with the periphery. The study by Guo et al (2014) has shown evidence that the signals from the spinal cord may contain the reference trajectories from the CNS (Guo, 2014). Churchland et al (2007) have shown that some cortical signals correlate with some movement parameters and that they do not generalize across different tasks (Churchland, 2007). Cisek observed that "the role of the motor system is to produce movement not to describe it". (Cisek, 2006). This suggests that trajectories similar to foot could be generated from other articulators such as hands or fingers allowing a natural, biological decoding of user intention (Karunakaran, 2013).

The control paradigm presented in this work uses force sensors connected to the hands to read the user-intention in real time, with the proprioceptive feedback, provided by a physical link between the foot and the hands, providing information on when the foot makes contact with the ground. An admittance control paradigm is used to translate the forces to Cartesian position of the foot. Admittance control is based on Newton's second law of motion: the relationship between an object's mass m, its acceleration a, and the applied force F is $\sum F = ma$. When a force is applied, an object with a fixed mass will move in the direction of the net force with an acceleration proportional to the force. The acceleration is computed from force by considering the object to be of a constant virtual mass; the double integration of the acceleration provides the position in Cartesian space for each time period. Since the mass of the object can be set to be very low, the user feels the object to be virtually weightless. Admittance control allows the user to input a force and translates the force into motion in the direction of the force. The advantages of using admittance control are force amplification, intuitive control (as the user applies force in the direction of movement desired), back-drivability as the object acts a pseudo-passive object, and proportional control (Van der linde, 2002). Studies have shown that admittance control needs to operate at minimum of 100Hz for optimal human interaction (Van der linde, 2002). Admittance control is exceptional in facilitating human-robot interactions where the person has physical contact with the robot and moves it along a path defined by the user's movements.

1.4 Human Gait

Human gait involves periodic movements of the leg to propel the center of gravity forward in order to move the body forward. The gait of an individual is a highly variable activity that differs from individual to individual based on age, sex, body type, physical condition, fatigue, etc. but there are characteristics in a gait cycle that can be used to define a normal human gait cycle (Hughes, 1979). The description of the gait cycle is confined to a single cycle (Vaughan, 1992) assuming the following cycles are all the same. The gait cycle is characterized by the set of events between heel strike (0%) to heel strike (100%) of the same foot (Dollar, 2008).

1.4.1 Phases of Gait Cycle

The normal human gait cycle has two distinct phases: swing phase and a stance phase. The swing phase and stance phase are defined as periods when the foot is off the ground and on the ground, respectively. The stance phase can be further classified as single stance and double stance. Single stance is when one foot is on the ground while the other foot is off the ground. Double stance is when both feet are on the ground (Vaughan, 1992).

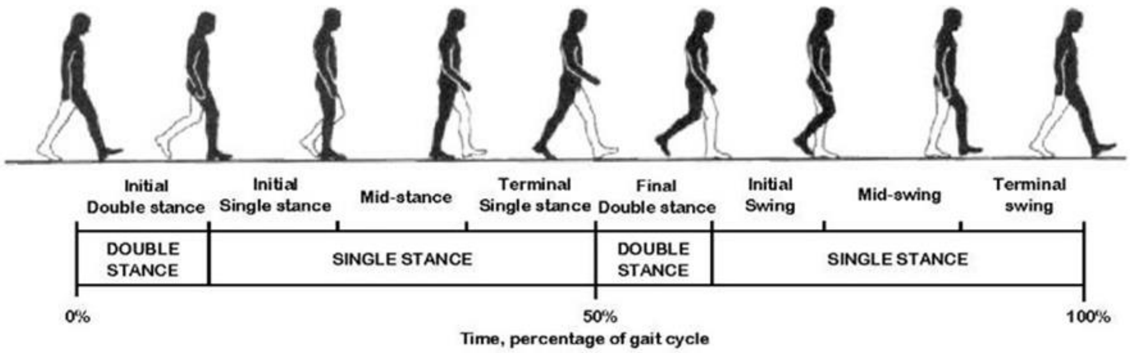


Figure 1.8 Normal human gait cycle.
Source: Vaughan, 1992

1.4.2 Kinematics of Gait Cycle

The gait cycle commences with a double stance phase, and at its half-way mark again the same leg is in double stance phase, and terminates with a swing phase as shown in Figure 1.8. The initial double stance consists of right knee that extends and right heel that makes contact with the ground which is known as heel strike, with the ankle held at a right angle (90 degrees) to the leg. Simultaneously, the left leg is in preparation for toe-off with left heel off the ground with only the toe in contact with ground, in preparation to move to single stance. During the single stance, the weight is shifted to the right leg where the foot is flat on the ground, while the left leg is in its initial swing phase with its knee bent and its toe swinging forward. This period is marked by the clearance of the foot and a period of acceleration of the foot forward and the initial swing phase ends when the foot is apposition to the stance foot (Hughes, 1979). The mid stance occurs at the end of loading response and continues until the body weight is aligned over the forefoot. The terminal of single stance that makes the half-way mark of the gait cycle consists of left heel in contact with the ground, and the right heel lifted off the ground. Next, follows the final double stance, but this time with the left heel that makes contact with the ground, and the right leg is in preparation for toe-off i.e. in pre-swing phase with heel off the ground and right toe on the ground. The faster the speed of walking, the shorter the duration spent on double support phase. During the succeeding swing phase of the right leg, there is a pre-swing phase when the initial contact ends and the rapid unloading of the weight occurs to the left leg. This is followed by the initial swing phase of the right leg where the right knee is bent and the right toe swings forward. The mid swing starts when both ankles are in apposition and terminates when the swinging foot is forward. The swing terminates by decelerating the foot and ends with the right heel on the ground, and the left heel off the ground. This marks a complete gait as shown in Figure 1.9. Though the nomenclature described starts with the right leg, the same could be applied for the left leg (Vaughan, 1992). Dynamic ambulation involves a stance phase of 60 percent at normal walking speed and reduces as speed increases and a swing phase of 40 percent at normal walking speed and increases as speed increases (Vaughan, 1992).

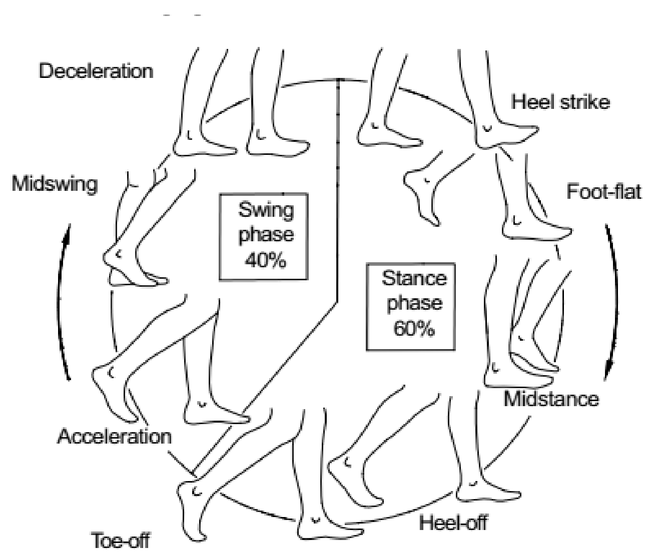


Figure 1.9 Phases of the human gait cycle
Source: Vaughan, 1992

1.4.3 Joint Kinematics

The two phases of stance and swing of the leg are made possible by the DOF available to the hip, knee and ankle joints of human leg (Vaughan, 1992). The angular position of the hip, knee and ankle contribute to each phase of a gait cycle.

a) Double Stance: The right hip during the start of the gait cycle is at 30 degrees of flexion and knee is at 5 degree of flexion.

b) Single Stance: During the single stance phase when the foot is flat on the ground with the loading response from the leg, the hip flexes by 30 degrees and knee continues to flex with it being at max of 20 degrees.

c) Mid Stance: The hip during the midstance extends towards neutral position and the knee during midstance is extending and is at about 8 degree of flexion.

d) Terminal Stance: During the terminal stance phase the hip goes into about 15 degree extension and the knee at the beginning of the terminal stance is at about 5 degrees flexion then the motion is reversed and the knee begins flexing to about 12 degrees. Now, the leg enters the pre-swing phase where the knee begins rapid flexion to about 40 degrees and the hip starting to flex to about 10 degrees.

e) Initial Swing: The pre-swing is followed by initial swing where the hip continues to flex to about 25 Degrees and the knee flexes to another 20 degrees which is followed by extension to 5 degrees.

f) Mid-swing: During the mid-swing the flexion slows down to a stop while the knee flexes to 35 degrees and the hip reaches a flexion of 30 degrees.

g) Terminal Swing: During the terminal swing phase the hip holds the positions and extends to about 5 degree while knee continues to extend and maintain the neutral 0 degree position (Vaughan, 1992).

The human leg has 7 DOF, with 3DOF at the hip, one at the knee and 3 DOF at the ankle. The most important DOF at the hip is in the sagittal plane and provides flexion/extension of the leg during the walking. The kinematics of the hip starts with the hip providing the clearance during the initial swing phase, then helps propel the leg through swing, and finally helps the transition from flexion to extension during swing (Vaughan, 1992). Movements in the coronal and transverse planes of the hip are less compared to movement in the sagittal plane; however, the rotational DOF in coronal plane helps with obstacle navigation (Vaughan, 1992). The user uses the abduction/adduction of the hip to shift the weight and to provide clearance in order to navigate obstacles. The transverse plane helps with changing directions thus helping the user to turn. The knee has one DOF, which is flexion/extension in the sagittal plane. This DOF also helps the leg in clearance of foot, and advance the leg during the swing phase. The ankle joint has 3 DOF with the rotation in sagittal plane (plantar flexion and dorsiflexion) being the most important of them, where it provides torque to toe off the ground just before the swing phase. The rotation about the coronal plane helps with balance (Vaughan, 1992).

From the foregoing, it is clear that to obtain gait cycles similar to human gait cycle, it is essential to have rotations about the hip, knee, and ankle in the sagittal plane and have rotations about the hip and ankle in coronal plane for obstacle navigation and rotation about the hip in transverse plane for turning.

CHAPTER 2

IMPORTANCE OF HAPTICS IN ALTERNATE CONTROL

2.1 Introduction

Rhythmic behaviors are those where parts of the body produce cyclic movements that are repeated over and over again, ex: walking, juggling, tapping, swimming, breathing, etc. Walking is a complicated repetitive motor task requiring multiple joints to work together to produce the required movement and balance (MacKay-Lyons, 2002). For long, scientists have wondered about the working of the central nervous system (CNS) to produce such a complicated motor movement. There were two main hypotheses for producing such movements. The first hypothesis is that of peripheral control, where the sensory feedback provides cues for the rhythmic movement. Each phase of the movement is considered to produce sensory cues necessary for producing the rhythmicity and the phases of the movement. The loss of sensory cues would therefore cause the complete disruption of the movement. The second hypothesis is that there exist neural oscillators or central pattern generators which are specialized neurons that can provide timing for muscle contraction to generate the required rhythmic movement. Studies indicate that both hypotheses seem to be true, where rhythmic movement by neural oscillators or central pattern generators is modulated by sensory cues to produce what is known as human gait. Therefore, the work to replicate human gait pattern also involves capturing the sensory cues for use in the intuitive control mechanism (Pearson, 1993).

2.1.1 Central Pattern Generators

Central pattern generators (CPG) are neural networks that work autonomously to produce rhythmic patterned output without sensory afferent feedback or input from the brain. CPG's are thought to exist in the vertebrates and mammals, and they are believed to produce variant rhythmic motor patterns such as walking, running etc. (Duysens, 1998).

The first evidence of CPG's was shown in the study on decerebrate cats, where the spinal cord was severed at the level of brain stem and the cats could perform movements (Duysens, 1998). This theory was further confirmed by studies by Brown, who made a similar observation in cats that performed simple stepping movements in the hind limbs though they were decerebrate and deafferented (Duysens, 1998). In spite of cutting off the inputs from the brain and afferent feedbacks these cats could still produce movements in the hind limbs. Hence, from this study Brown et al. concluded that CPG's in the spinal cord are sufficient to produce movements in the hind limbs of the cats (Duysens, 1998). This concept of motor program was defined by Marsdeb et al (1984) as "a set of muscle commands which are structured before a movement begins and which can be sent to the muscle with the correct timing so that the entire sequence is carried out in the absence of peripheral feedback".

Currently, there are numerous studies in vertebrates that have validated that there are nerve cells that produce specific autonomous rhythmic movements without afferent signals. Motor control of human walking is also theorized to be using CPG's generated from the spinal cord. The existence of CPG's in human walking has been only indirectly observed as in vivo experiments cannot be performed (Pearson, 1993). Supported treadmill studies provide evidence regarding the existence of CPG's in humans. These studies have shown that subjects with incomplete SCI have been shown to regain some or most of their motor function when trained with body weight support, but the same results could not be replicated with subjects with complete SCI (Duysens, 1998). The fact that researchers have thus been able to obtain movements in subjects with incomplete SCI should not be interpreted as to mean ‘afferent inputs are not important to pattern generation by humans during walking’. Though it has been shown that rhythmic movements are produced by CPG’s in spinal cord as shown in studies involving deafferented nervous system, such movements are not identical to the patterns produced with intact nervous system (MacKay-Lyons, 2002).

2.1.2 Impact of Afferent Feedback on CPG

Although the basic locomotor pattern can be present in fictive locomotion (i.e. in complete absence of afferent input), it is seen that the role of afferents is very important in shaping the rhythmic pattern, to control phase-transitions and to reinforce the ongoing activity in cats (Duysens, 1998). Therefore, a rhythm generating structure without its normal afferent input can be very artificial and, therefore, cannot entirely reproduce the motor output as seen in the intact cat. Pearson described that afferent feedback plays a major role in rhythmic movement. He states that sensory feedback provides information to ensure that the motor output is appropriate for the biomechanical state of the moving body part in terms of position, direction of movement, and force (Pearson, 1993).

Human two legged walking is a complex rhythmic movement that involves phase changes that are influenced by the environment i.e. fast walking/slow walking etc. These rhythmic tasks often consist of discrete events often due to contact with the environment. These events provide the body with information regarding the current state of the body and regarding the environment, thus helping with the transition from one state to another effectively. Pearson also states that the sensory feedback facilitates the transition between the different phases of the rhythmic movements (Pearson, 1993).Thus CPG’s in CNS may be modulated by the sensory feedback from the peripheral receptors to control the frequency and amplitude of the centrally generated motor pattern.

Ivanenko et al. evaluated the importance of ground contact forces as sensory cues in the trajectory of subjects' feet as they walked with reduced body weight. Weight reduction was achieved by mechanical body weight support. In this study, it was observed that leg kinematics were dramatically affected when no ground contact forces were present (i.e. walking in the air), but were largely unchanged during partial body weight support (Ivanenko, 2002). Results show that even though subjects were instructed to make walking movements, including mimicking ground contact patterns, the trajectories of the feet under the condition of complete body weight support (i.e. walking in the air) changed dramatically and were erratic. The frequency of the leg movements became quite variable compared to normal gait, indicating that ground force feedback contributes to the cyclic (i.e. rhythmic) nature of gait. Furthermore, when walking with no ground force feedback, subjects converted their foot movements to a simpler cycling pattern. However, when body weight was not fully supported, the subjects' foot trajectories remained similar to biological gait. This implies that ground force feedback is vitally important for biological gait kinematics, but that the amplitude of the force does not exert significant influence on the kinematics. Hence, to effectively develop an interface to control walking, haptic feedback or tactile cues of ground contact needs to be provided to the nervous system during each phase of gait to modify (i.e. modulate) the central patterns and produce what is known to be biological gait. Ivanenko et al suggest that ground contact feedback could provide a preferred modulation of the cyclic central patterns or could signal a transition between distinct locomotor patterns of stance and swing (Ivanenko, 2002). This also suggests that the gait pattern is not fully specified by higher cortical regions, but is modified (modulated) by important sensory feedback. Further, Ankarali et al. have shown in their study that providing sensory feedback-like force impulse to the hand while performing rhythmic motor tasks, such as virtual paddle juggling, enhances performance by reducing variability in the rhythmic movement (Ankarali, 2013). They state that the haptic feedback produces the necessary feedback to determine the phase transition while performing the rhythmic movement.

Studies of upper extremity training have shown that performance increases when both haptic and visual feedback are present, compared to when haptic-only or Visual-only feedback is present. Feygin et al. concluded that spatial aspects of movement improved with visual training, while temporal aspects improved with haptic training (Feygin, 2002). Gunn et al. observed improved speed and accuracy when haptic feedback was introduced along with visual feedback in a virtual environment (Gunn, 2009). Both studies suggest that combination of haptic and visual feedback improved performance compared to haptic-only or Visual-only feedback conditions.

Proprioceptive, vestibular and visual feedbacks are needed to achieve normal gait trajectories in non-disabled subjects (Karunakaran, 2014). Proprioceptive feedback consists of haptic feedback. While vestibular feedback plays an important role in balance and equilibrium of the system, haptic feedback helps maintain the rhythmic patterned gait trajectory while walking. It is observed that upper extremity control relies heavily on visual feedback for guidance while lower extremity control uses visual feedback predominantly to scan the environment for obstacles without directly influencing the gait (Feygin, 2002)

Koritnik et al. demonstrated that lower limb rehabilitation yielded better spatial and temporal adaptation in the haptic-only when compared to Visual-only mode. Also, visual and haptic feedback can improve performance of lower extremity training more than Visual-only or haptic-only modes (Koritnik, 2010). In a recent study, Turchet et al showed that vibrotactile feedback (haptic feedback) provided to the foot increased the realism of walking when the subjects were asked to walk in a virtual environment (Turchet, 2013).

The control mechanism proposed here for lower extremity exoskeletons uses hand movements as surrogates that express neurally intended foot trajectories (Karunakaran, 2013 and Karunakaran, 2014). This study evaluates the need for haptic feedback in addition to visual feedback to allow hand trajectories to produce exoskeleton gait that resembles biological gait.

In the pilot study, conducted prior to the formal dissertation proposal, the importance of providing both position and ground force feedback to subjects' hands as they employed their hands to operate virtual feet in a graphically-rendered walking task was explored. This study addresses whether hand trajectories could produce trajectories similar to that of foot trajectories and the different feedbacks required to produce such a trajectory.

2.2 Methodology

The experimental setup for the study included a virtual environment (VE) consisting of two feet on an infinite path which was designed and built using Simulink 3d toolbox in MATLAB. The VE provided the users with a 2-dimentional view with depth perception. Two Geometric Phantom Omni, 3- Degree of freedom (DOF) haptic devices were used for rendering haptics. The Ascension Technologies electromagnetic position tracker the Nest of Birds (NOB) was attached to the distal end of the omni. The position of the NOB was mapped to virtual environment, thus moving the hands also moved the feet on the pathway in real time. The Phantom Omni was synchronized with the virtual environment to produce haptic feedback of magnitude 0.88N (maximum allowable force feedback of the Omni) when the feet were in contact with the floor. Eighteen subjects consented to participate in the study approved by the Internal Review Board of NJIT. All subjects selfreported no disabling conditions, and had full function in upper and lower extremities and normal vision. Exclusion criteria included disability to upper or lower extremities or non-correctable visual impairment.

The study consisted of three experimental groups: Visual-only (VO), Haptic-only (HO) and Haptic & Visual (HV). Subjects were randomly assigned to the three groups of six individuals that were age and gender matched. All subjects were under the age of 40. All subjects in each group participated in five sessions, where each session consisted of eleven trials of 60 seconds' ambulation followed by 60seconds' rest. These trial and rest durations minimized muscle fatigue. Subjects were instructed to hold the omni and perform walking like movements using their hands and to walk the feet as far as possible during the trial duration. They were informed that ambulation (forward movement) was possible only when at least one shoe was in contact with the virtual pathway (stance) and the other shoe was in its swing phase. The pathway works as a manual treadmill. Therefore, there was no movement when both shoes were above or below the virtual pathway, or if the stance shoe was below the pathway when the other shoe is in swing phase. Stride length (horizontal distance), vertical height and speed of shoes were controlled by the user's hand movements. A black drape (not shown in Figure 2.1) prevented subjects from seeing their hands. Thus, feedback was limited to proprioceptive sensation from the arms and visual observation of the shoes and moving walkway. The first trial was used to acquaint the subjects with the procedure where all three groups were provided with both forms of feedback. This ensured all three groups trained for the task equally and performed the task with equal ability. Data analysis was performed only on trials two to eleven, with the practice trial omitted.

Figure 2.1 Experimental setup :(Left) A) VE monitor. B) NOB sensor on Phantom Omnis. (Right): Phantom device with NOB sensor.

A) Protocol for Visual-only (VO) Group

Subjects provided with Visual-only feedback saw the shoes on an infinite virtual pathway presented on the computer monitor. Subjects were expected to control the movement of the shoes with visual feedback. Shoes were lost from view when the user's hands placed them below the walkway and appeared to float above the walkway when placed too high; also a visual cue of red dot was provided to let the user know that both their feet were above the ground. The treadmill-like movement did not occur when the stance foot was incorrectly placed.

B) Protocol for the Haptics-only (HO) Group

A haptic surface was rendered to simulate foot contact and provide force feedback from the pathway when the virtual shoes made contact. The force feedback simulated walking on a flat surface. Applied vertical hand forces greater than .88N resulted in the shoe dropping below the virtual pathway (fall throughs), and hence there is no forward movement on the pathway. The shoe needs to be on the floor to move forward. Hence subjects received haptic cues from the Phantoms about their success in providing an appropriate vertical hand force to make contact and prevent falling below the surface. This discouraged subjects from using excessive force while coming in contact with the floor. Cues regarding both feet off the floor were not necessary as the subjects used haptics as cue to place at least one foot on the ground while walking unlike the visual group which could not place their foot on the ground. The monitor was turned off for Haptic-only subjects to prevent them from having visual feedback.

C) Protocol for Haptic & Visual (HV) Group

Subjects in the third group experienced both visual and haptic feedback as described above.

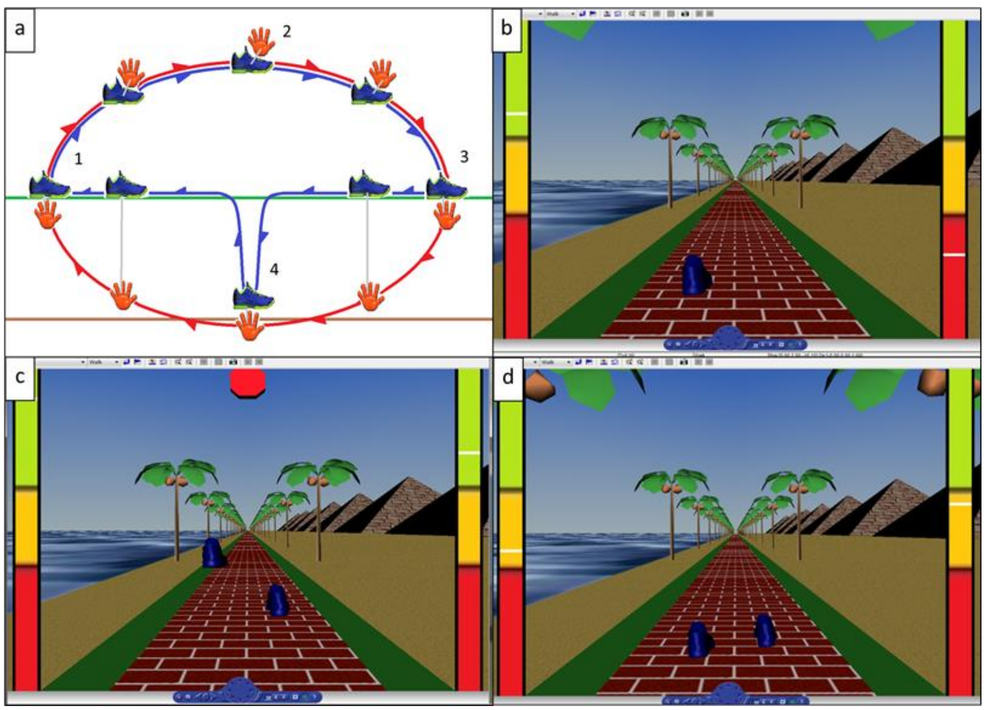


Figure 2.2 a) Typical virtual gait cycle: The orange hand represents the movement of the hand. The blue shoe rises as the hand is elevated and the shoe is returned to the ground as the NOB sensor reaches ground threshold. The ground threshold is defined by the region between the green and red line. The shoe drops below the virtual pathway when the hand goes below the ground threshold, which is referred to as a fall through. b) The shoe drop below the threshold is shown. c) The boot shoes are above the ground, which is cued by a red dot on the top d) VE shoes and virtual pathway: the pathway acts as an infinite treadmill to allow forward progress of the shoe, while remaining within the range of motion of the Phantom Omnis.

2.2.1. Data Analysis

MATLAB was used to analyze the hand trajectory data from the NOB and to evaluate the performance of subjects.

2.2.1.1 Horizontal and Vertical Trajectories. Horizontal and vertical trajectories were collected at 100 Hz and filtered at 25 Hz cut-off frequency using an effective 4^{th} order, zero-lag Butterworth low-pass filter. These filtered data were further analyzed to determine the time synchrony of hands.

The time synchrony was calculated by standard deviation between the inter-peak intervals for the horizontal trajectory of each hand for each trial. An average of the standard deviation across trials for all subjects in each group for each session was computed. The Kruskal-Wallis non-parametric test and Mann-Whitney *U* test with Bonferroni correction for multiple comparisons were performed to determine the statistical difference between the groups.

2.2.1.2 Fall Throughs. The average number of fall throughs (subject misestimate the location of the virtual walkway) per unit distance across subjects for each session in each group was computed. Kruskal-Wallis non-parametric test and Mann-Whitney *U* test with Bonferroni correction was performed to determine the statistical difference between the groups for both hands. Friedman's Test was used to evaluate the performance differences between sessions in each group.

The learning in terms of accuracy was evaluated in each group by comparing between the fall through in each session and in each trial. A repeated measure of one way Analysis of Variance (ANOVA) was used to evaluate performance differences between sessions and trials in each group.

2.2.1.3 Distance Travelled. Average distance traveled was computed for each hand and the performance of each group was assessed. One way ANOVA and Tukey's post hoc test were performed to determine the statistical difference between the groups.

The learning in terms of speed was evaluated in each group by comparing the distance travelled in each session and in each trial. A repeated measure ANOVA was used to evaluate performance differences between sessions and between trials in each group.

2.2.1.4 Duty Cycle. The duty cycle of the gait cycle was calculated. The average relative percentage of stance and swing phase for each session was calculated for all three groups for all sessions using the below formulae:

$$\text{Duty cycle} = \text{Stance Phase} + \text{Swing Phase}$$

$$\text{Stance Phase \%} = 100 * \text{Stance phase} / \text{Duty Cycle}$$

$$\text{Swing Phase \%} = 100 * \text{Swing phase} / \text{Duty Cycle}$$

ANOVA was used to determine the statistical significance between each group. Independent sample t-test with Bonferroni correction was used to determine the significance between groups.

2.2 Results

2.3.1 Horizontal and Vertical Trajectory

Significant difference was observed between all three groups, with Haptics & Visual group doing better than Haptic-only ($p<.017$) and Visual-only ($p<.017$) group and Haptic-only group doing better than Visual-only group ($p<.017$) in both left and right hands as shown in Figure 2.5. Thus groups with haptic feedback showed lower variability in their trajectories, implying that the movements are more consistent when haptics is provided.

2.3.2 Fall Throughs

Figure 2.6 shows average fall throughs for each session for all five sessions. Significant group differences were observed between the Haptic-only and Visual-only group, between the Haptic & Visual and Haptic-only group and between the Haptic & Visual and Visual-only groups for both hands. No significant difference was observed between sessions in all three groups. No significant differences were observed between trials in each session for all three groups except for session 4 in the left hand for the Visual-only group and session 4 in the right hand for the HO group. Figure 2.7 shows that fall throughs for the Visual-only continues to show no improvement between Sessions 1 and 5, indicating no learning within session. Also, there is no significant improvement within trial in session 1 or session 5. These results signify that the user will not learn even with training.

2.3.3 Distance Travelled

Figure 2.8 shows the average distance travelled by each group for each session. Significant difference was observed between Haptic & Visual and Visual-only groups ($p<.05$), and between Haptic-only and Visual-only groups ($p<.05$) for both hands. No significant difference was observed between the Haptic & Visual and Haptic-only groups for both hands. Significant difference was observed among sessions for Haptic & Visual group, but no significant difference was observed among sessions for Visual-only and Haptic-only groups. Further, no significant differences were observed among trials within sessions for all three groups

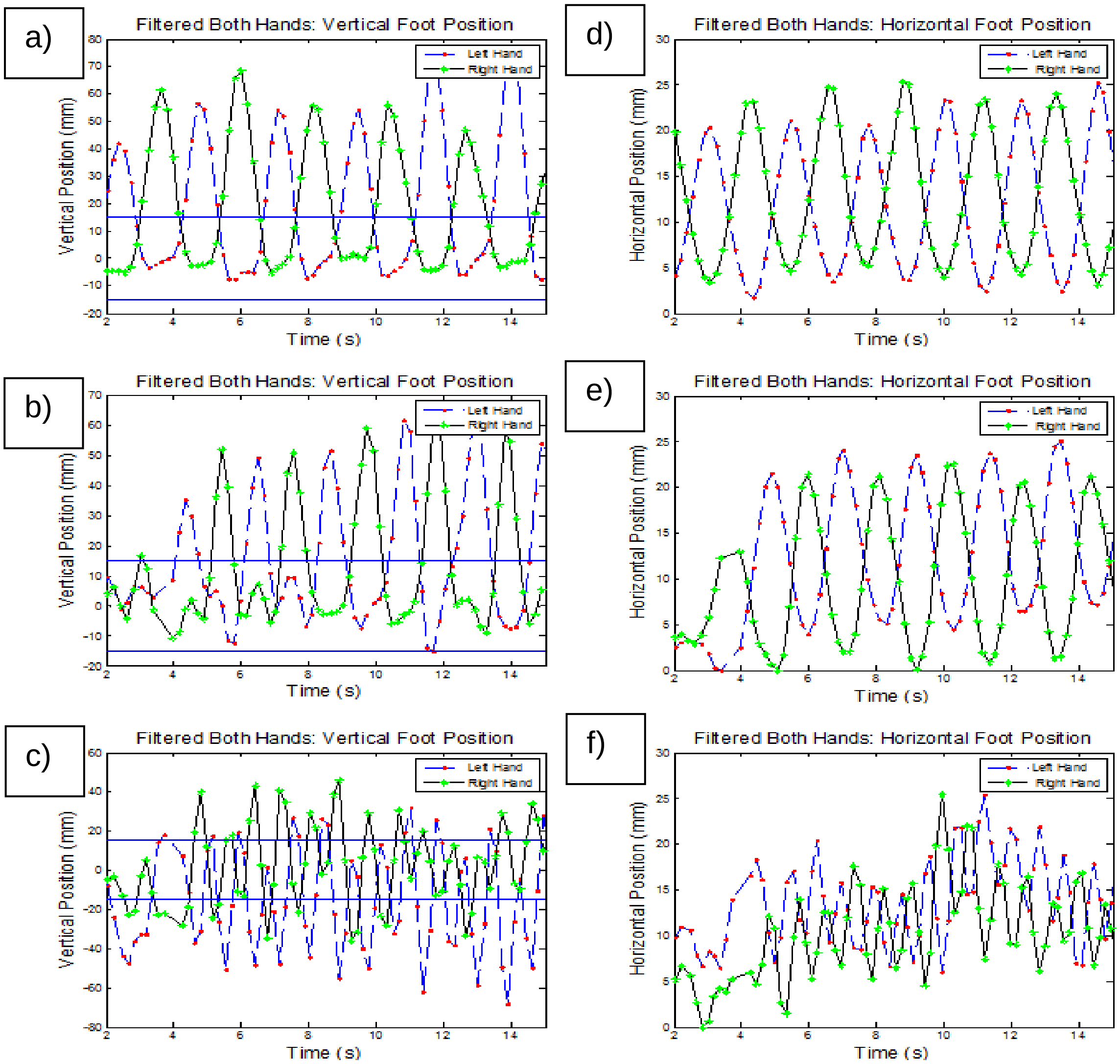


Figure 2.3 Vertical Position of trial 11 in session 5 of subject 5 in each of the following groups: a) Haptic &Visual b) Haptic-only c) Visual-only feedback. The ground threshold is defined by the two blue lines in a, b & c. Shoe dropping below the threshold is referred to as a fall through. Horizontal Position of trial 11 in session 5 of subject 5 in each of the following groups: d) Haptic & Visual e) Haptic-only f) Visual-only feedback.

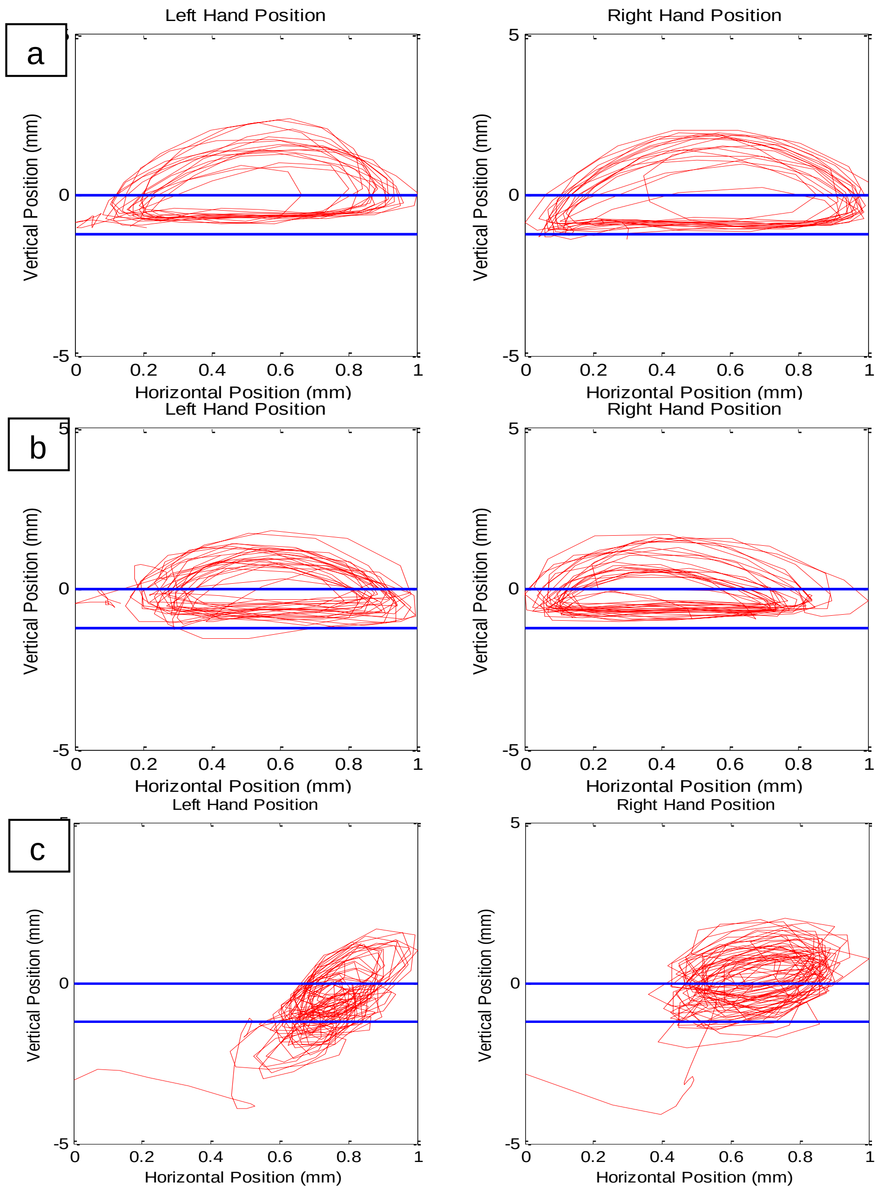
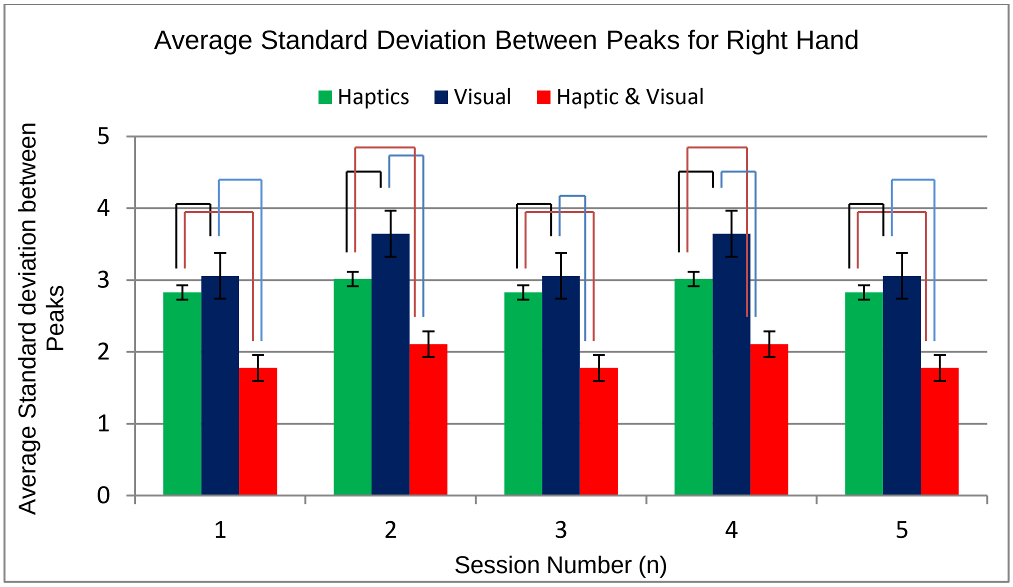


Figure 2.4 x and y position of the left and right hand during a) Haptic & Visual feedback b) Haptic-only feedback c) Visual-only feedback.

a)



b)

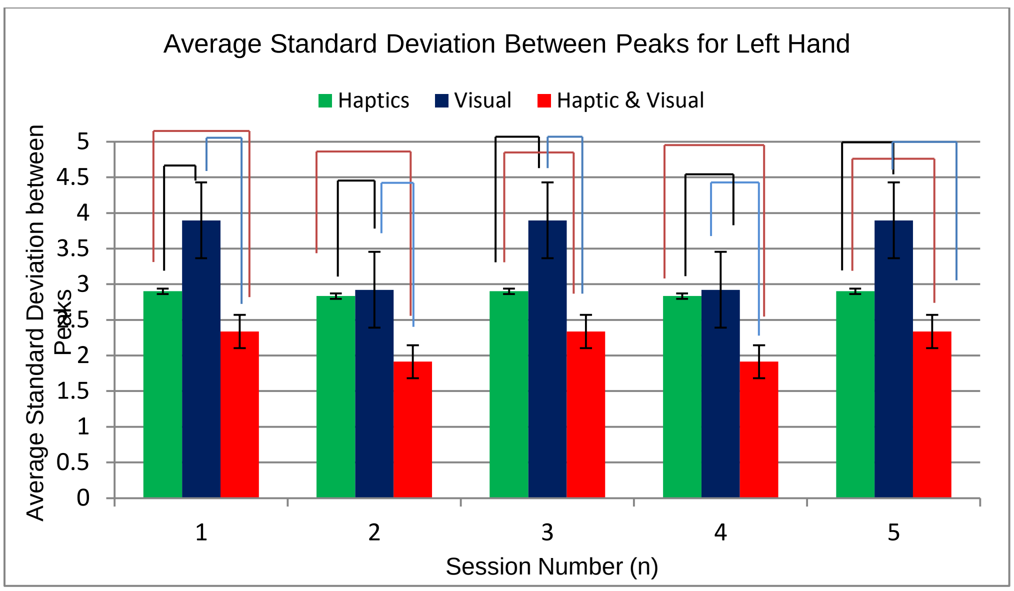
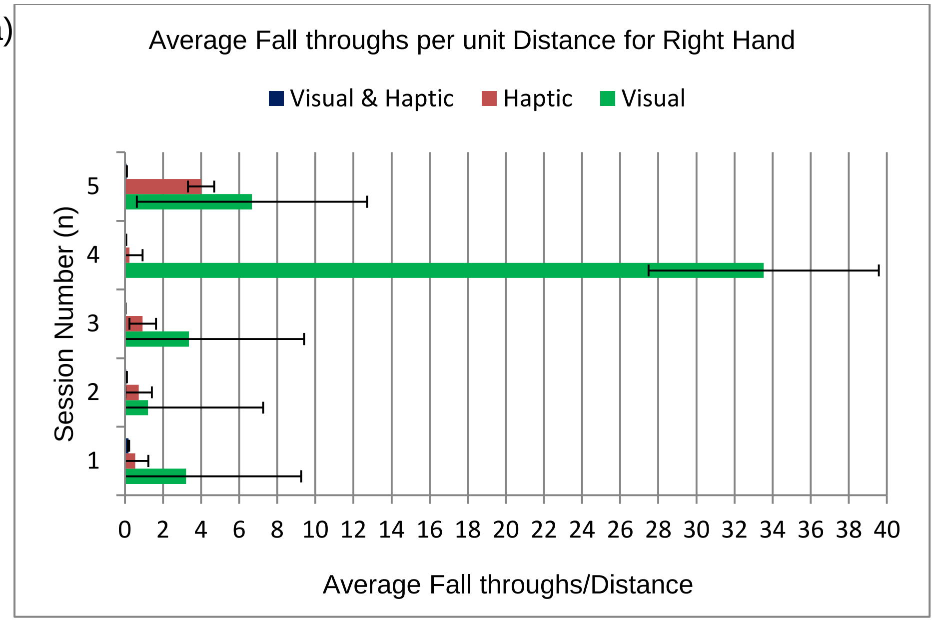


Figure 2.5 Average standard deviation between peaks of horizontal trajectory (SEM) for a) Right hand b) Left hand for with each session. Significant difference was observed between all groups.

a)



b)

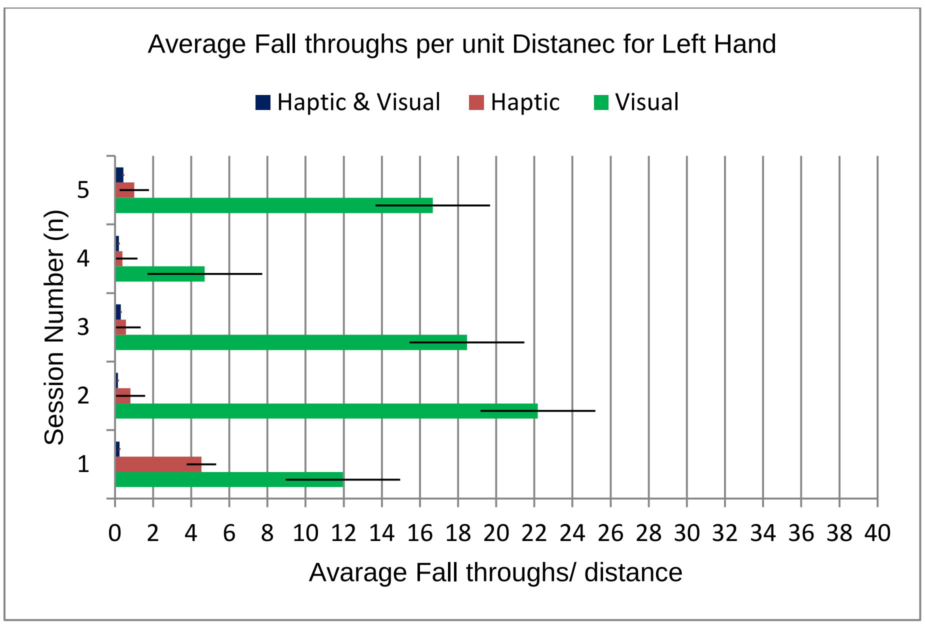
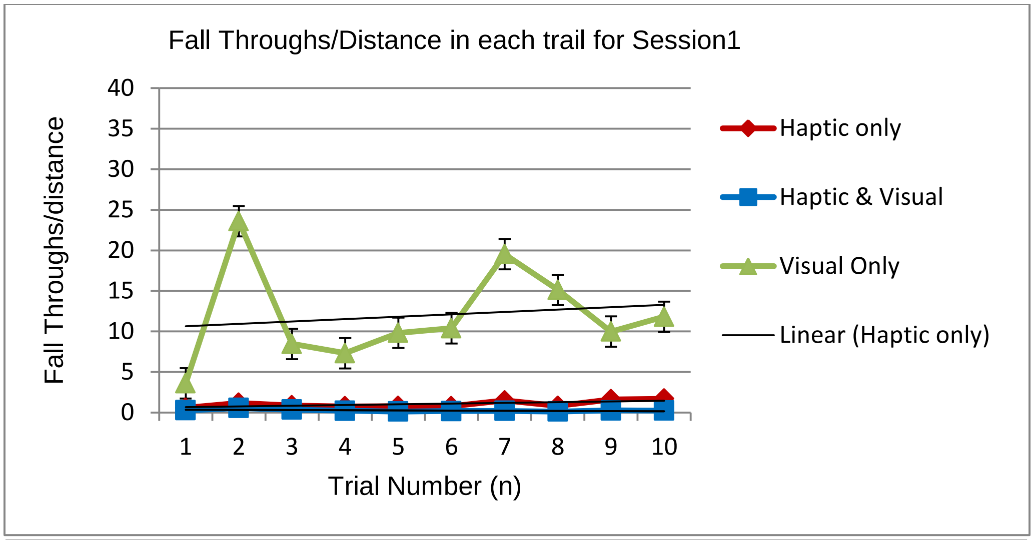


Figure 2.6 Average Fall throughs per unit distance (SEM) for each session for a) Left Hand and b) Right Hand.



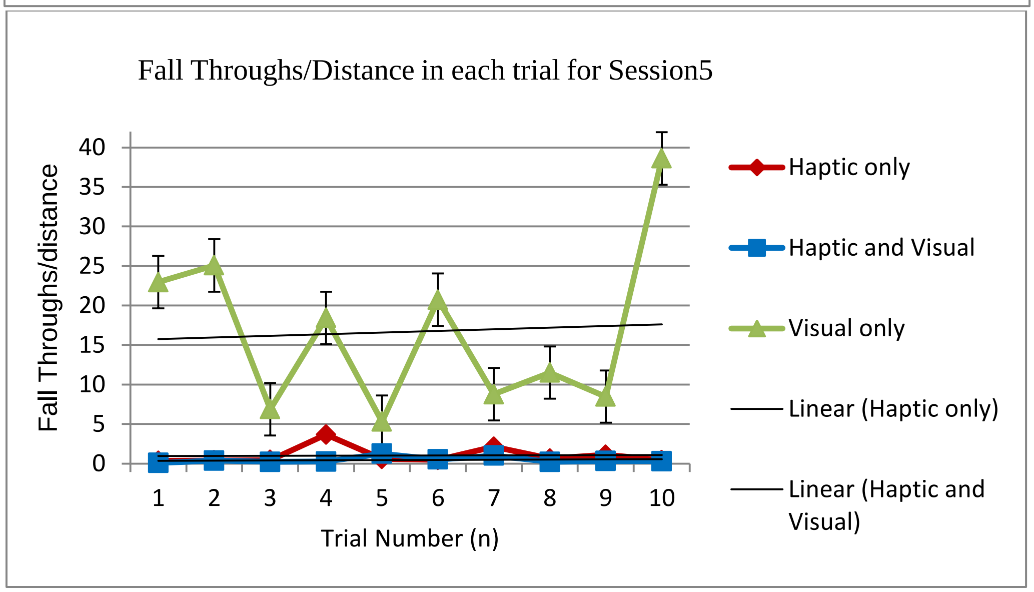
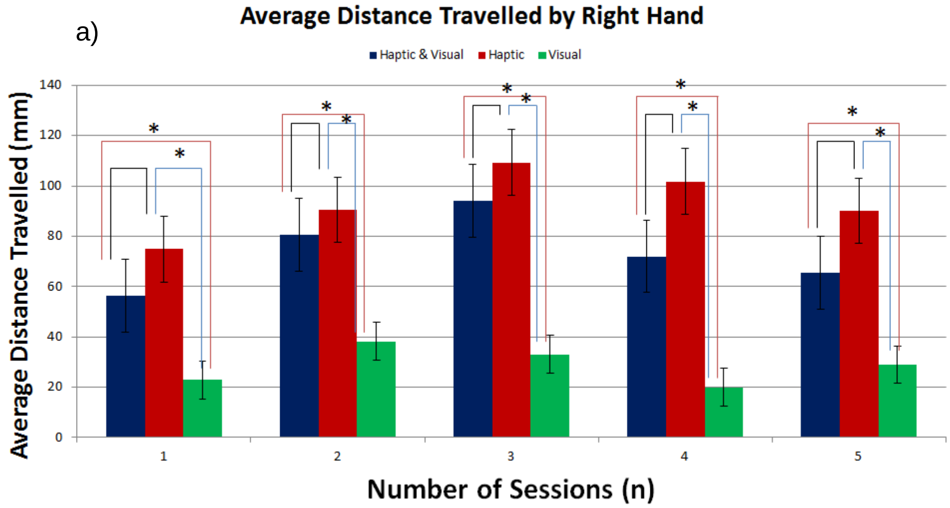


Figure 2.7 Mean Fall throughs/distance travelled per trial for sessions 1 (Top) and 5 (Bottom) by left hand with SEM.

a)



b)

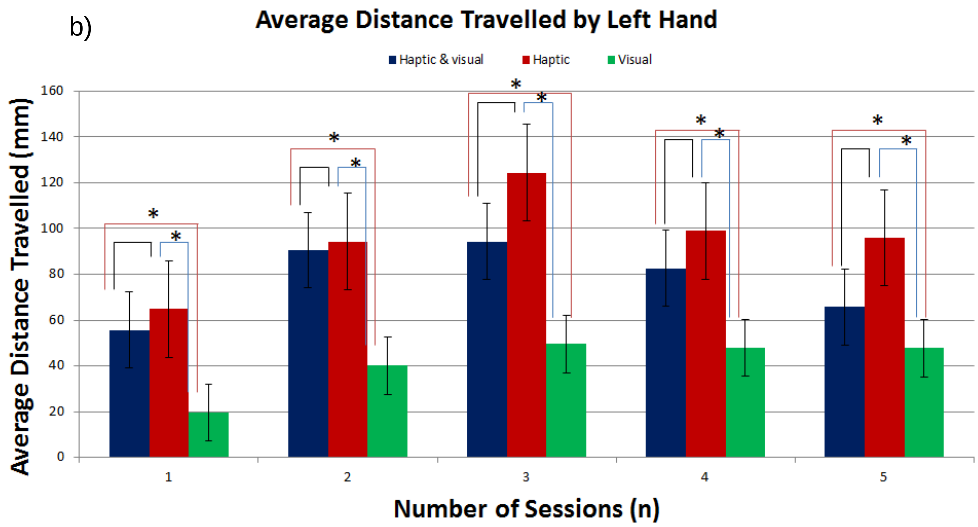


Figure 2.8 Average distance travelled/session by a) Right Hand b) Left Hand with SEM.

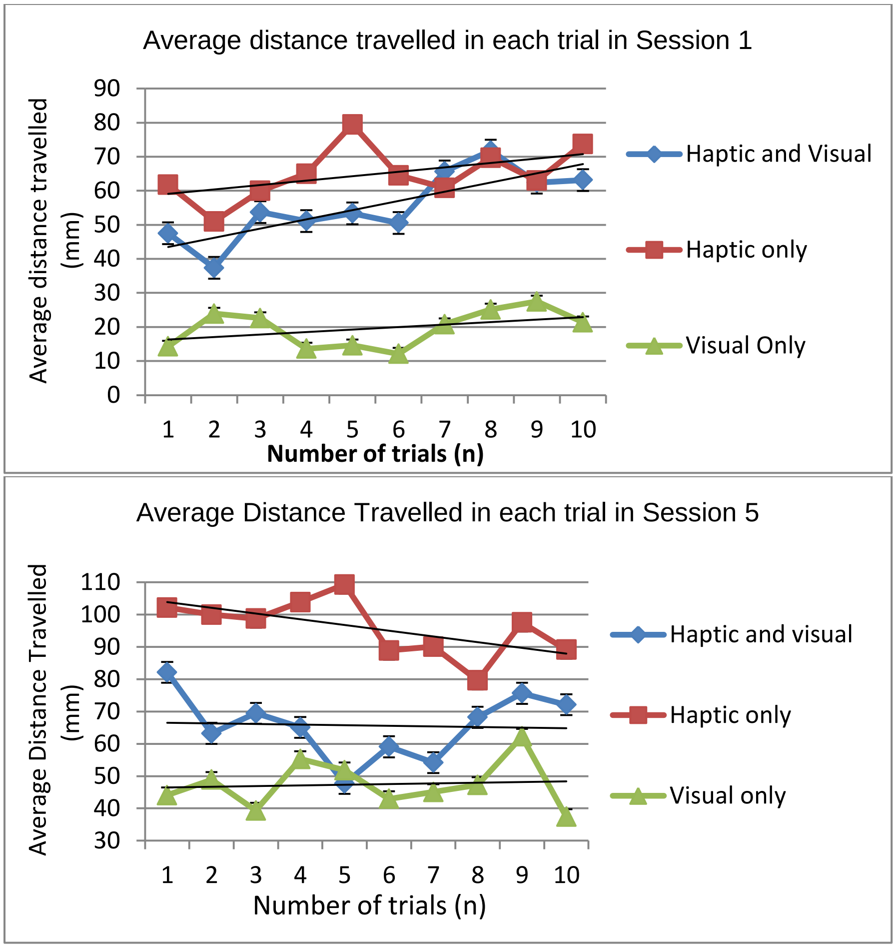


Figure 2.9 Mean distance travelled in Session 1 (top) and 5 (bottom) by left hand with SEM.

2.3.4 Duty Cycle

Average duty cycle demonstrated by the Haptic & Visual group was 66% stance and 34% swing while the Haptic-only group demonstrated an average duty cycle of 70% and 30%, while Visual-only group demonstrated an average duty cycle of 73% and 27% Significant difference was observed between all three groups.

2.4. Discussion

Results show that Haptic & Visual feedback groups achieved better time and amplitude synchrony as demonstrated by gait cycle (Figure 2.3) , fall throughs (accuracy, Figure 2.6) and distance travelled (speed, Figure 2.8) . Normal gait cycle is divided into two phases - swing and stance. Figure 2.3 shows that when one hand is in stance the other is in swing phase, as observed during human gait. A typical gait stance phase represents 60% of the cycle and swing phase 40% of the cycle (Winter, 1990). Analysis of hand trajectories demonstrated groups with haptic feedback fared better than the Visual-only group and also compared well with the natural gait cycle.

The standard deviation between the peaks of the horizontal trajectory was used as a measure to evaluate the time synchrony. A small standard deviation between the peaks signifies synchrony. Figure 2.5 shows Haptic & Visual feedback group have smaller standard deviation than both Visual-only and Haptic-only groups. This signifies greater synchronicity in Haptic & Visual group compared to the other two groups. The Haptic-only group showed better synchrony than the Visual-only group. Our findings are in accordance with Turchet et al. that haptic feedback plays a relevant role in the perception of both real and simulated surfaces during the act of walking (Turchet, 2013). This indicates haptic feedback is essential for motor control of rhythmic movement.

Results also suggest that hand movements, when provided with real-time ground contact information, share considerable similarity with the foot movements of natural human gait as shown in Figure 2.4, where the trajectory conforms to walking patterns while when only visual feedback is provided, it conforms to cyclic pattern. These findings are very consistent with those of Ivanenko et al. in which ground forces applied to the feet result in relatively consistent kinematic trajectories when body weight support is varied from 0% to nearly 100%. However, at 100% body weight support, with no ground contact forces, the movements of the feet (i.e. walking in air) take on a dramatically different kinematic appearance (Ivanenko, 2002). In Figure 2.3, we see that the rhythmicity of the trajectories is well maintained when haptic feedback is present. In panels ‘a’ and‘d’ the timing and the vertical and horizontal excursions of the hands shows very little variability over multiple simulated steps when both Visual & haptic feedback are provided, compared with panels ‘c’ and ‘f’ in which the timing and vertical and horizontal excursions are much more varied in the absence of haptic information (analogous to walking in the air). A striking example is the comparison between panels ‘a’ and ‘c’, where with visual and haptic feedback, there are clearly observed flat portions of the vertical trajectory that correspond to the placement of the hand during ground contact (at the simulated floor height) during stance, while panel ‘c’, without force feedback shows no such flat stance. Also, the phase change from stance phase to swing phase and vice versa between the two legs are more synchronized in groups with haptic feedback than in groups with only visual feedback.

While Ivanenko et al. found significant kinetic changes resulting from the varying magnitude of ground force reactions; this is of less importance to the eventual work. We propose employing hand trajectories only to define the kinematics and not the kinetics of the foot trajectory.

Figure 2.6 shows the average fall throughs per unit distance. The Haptic & Visual feedback group had fewer fall throughs per unit distance, followed by Haptic-only feedback group, followed by Visual-only feedback implying haptic feedback is important in spatial awareness during ambulation. In session 4, subject 4 in the Visual-only group displayed fewer fall throughs per unit distance compared to all other Visual-only subjects. This reduced the mean as observed in Figure 2.6. The fall throughs per unit distance within session and within trial analysis showed no significant change implying that there is no significant learning occurring after successive trials or session. In fact, the curve of fall throughs per unit distance is almost flat as can be observed in Figure 2.7

for the Visual-only group, indicating little learning. This suggests that even with extensive training, the Visual-only feedback group will continue to lag behind other two groups in their performance.

Distance travelled by the Haptic-only and Haptic & Visual feedback groups was significantly better than Visual-only group (Figure 2.8). This implies that Visual-only feedback does not play a role in virtual horizontal movement. Also, it was observed in the Visual-only group, the distance travelled by using left and right hand was different. We believe it is because the Visual-only group were not able to identify the ground and were unaware that they were taking shorter steps with one leg and longer steps with the other. The haptic group performs better with successive sessions with respect to distance travelled but they make the same number of errors per unit distance implying they gain confidence with the task to walk faster without however improvement in accuracy. It is therefore seen that there is learning, not with regard to fall throughs but with regard to improving the distance travelled. Though there was non-significant learning observed for groups, the learning plateaued from session 3 (Figure 2.8). It was also observed that learning was better for the Haptic & Visual feedback group than Haptic-only group between trials in the initial sessions (session 1 & 2) but there was no learning observed between trials in latter sessions (Figure 2.9). However the distance travelled, though not statistically significant, is less in the Haptic & Visual group compared to the Haptic-only group in almost every trial and session. This may be because the visual feedback when added to haptic feedback reduces errors at the expense of speed. Distance travelled by Haptic & Visual group improved in successive trials as shown in Figure 2.9 to almost catch up with Haptic-only group as practice improves performance. Studies by Woodworth et al. have shown that visual feedback reduced the error in movement (Proteau, 1992). Karniel et al. have shown that subjects performed larger and faster movements when not provided with visual feedback as compared to when provided with visual feedback during rhythmic forearm movements (Levy-Tzedek, 2012). This observation is in accordance with the results shown in our study where subjects performed faster movement when provided with only haptic feedback and were able to maintain the frequency of the movement but not the accuracy.

Ivanenko et al. showed the importance of ground force reaction on the feet in modulating the trajectories of the distal portions of the legs. The study has shown that the neural control of hand movements, used to define foot trajectories, requires equivalent force input. Even though human hand movements are not used to control gait, subjects who received visual & haptic, or Haptic-only feedback, quickly adapted to the haptic feedback and produced adequate and reliable simulated gait with their hands. No significant learning period was necessary.

2.5. Conclusion

Haptic and visual feedback is important for production of a normal gait cycle. Haptic feedback is vital to maintain the rhythmicity, pattern, to induce the phase changes of the gait pattern and to trigger changes in the foot trajectory. Visual feedback is required to navigate obstacles and reduce errors, but visual feedback alone cannot provide the necessary feedback required for the normal gait cycle even after extensive training. This study confirms that haptic feedback to the hands is essential to allowing the hand trajectories to be surrogates for neural signals for gait that are otherwise prevented from reaching the legs due to spinal cord injury. Furthermore, the current performance in spite of lack of extensive learning indicates that subjects are likely to adapt well to using their hands to manipulate their legs.

CHAPTER 3

CONTROL ALGORITHM FOR EXOSKELETON

3.1 Introduction

The study from Chapter 2 demonstrated that trajectories from hand could produce trajectories similar to normal gait cycle when provided with appropriate feedback. Proprioceptive feedback, especially relating to position and force is necessary to help produce the desired trajectories. It is therefore necessary to develop a control method that can provide the feedback. The goal of this work was to develop an intuitive control mechanism to translate the intentions of the hand in real time while also providing haptic feedback to the user. This work addresses specific aim 1. Key to the development of the control mechanism is the use of the admittance control paradigm.

Broadly there are two control paradigms in robotics; impedance control and admittance control. Impedance control uses the position detected from the user's intention using sensors to compute the force or torque required to move the robot (Carignan, 2000, Glosser, 1994). The control objective of impedance control paradigm is to impose (i.e. impede), along each direction of the task space, a dynamic relation between the manipulator position and end effector force (Hogan, 1984, Hogan 1985). It can be described as 'displacement in, force out' as the user moves the mechanical device, the device will react with a force (Van der linde 2002). Impedance control devices are very light and achieve back drivability by cable driven mechanism ex. SensAble's PHANToM. An admittance on the contrary represents the inverse of impedance control. It detects the force commanded by the operator and controls (i.e. admits) the velocity and/or displacement of the device (Dautenhahn, 2011). Admittance control can thus be summarized as ‘force in displacement out’. In its simplest form, it is a computer implementation of Newton’s 2nd Law; $\sum$ force= mass X acceleration. Here, the mass can be set to a desired low value, hence the user feels the inertia of the low mass and not of the entire robot. This control mechanism reduces/minimizes the inertia and backlash of the system while allowing higher stiffness and high forces, thus allowing for varying mechanical designs ex. Haptic Master (Dautenhahn, 2011). Thus, admittance control or force control enables human-like compliant motion and manipulation in complex environments. It is also called interaction control, as the algorithms are capable of controlling the robot’s position along the direction of the task space (Haidegger, 2009). Admittance control devices are capable of rendering very high stiffnesses and minimal friction, giving a very free feel to the motion. They are very suitable for larger workspaces, and also for master-slave applications and for carrying complex end effectors with many degrees of freedom. Also, they intrinsically register forces encountered, and are therefore very suitable for haptics and neurological research. Also, with admittance control, the device can be made back drivable, and therefore, like any back drivable device the device can overcome/resist the forces presented by system (Van der linde 2002).

Admittance control is appropriate for use with human/robot interaction as it is a very intuitive control mechanism, where the object or robot will move in the direction of the force applied by the user and also will move proportional to the force applied by the user. The amount of force required to activate the system can be varied to accommodate the user needs. Since the system is back drivable, it is safe and easy for human interaction.

3.2 Robot Design

3.2.1 Mechanical Design

A 1/2 scale biped robot representation for lower extremity exoskeleton was built to test the control mechanism. The robot has 2 links, from hip to knee, and from knee to ankle complying with the anthropometric scale of the human leg. The two legs are joined at the hip. A foot was custom printed in ABS Plastic to be fitted at the ankle (Appendix C). It also has a provision to mount Phidget force sensors to detect ground contact. A 3 DOF force sensor is attached to the ankle of each leg using custom made constructs in ABS (Appendix C). A trekking pole is mounted on the force sensor to be controlled with hand movement as shown in Figure 3.1.

3.2.2 Electrical Design

The robot has 5 DOF in each leg, the hip has 2DOF for flexion/extension and abduction/adduction, knee has 1DOF for flexion and ankle has 2 DOF for plantar flexion/dorsi-flexion and inversion/eversion. The hip and knee flexion/extension are provided by the Dynamixel MX-106 motors and the ankle flexion/extension is provided by Dynamixel MX- 64 motor on each leg of the biped.

The hip abduction/adduction is provided by an additional Dynamixel- MX-106 motor and the ankle inversion/eversion is provided by an additional Dynamixel- MX-64 motor. Since the hip and the knee have higher torque requirements during the swing phase to clear the ground the higher rated torque motor (MX-106) is used at both these joints. Dynamixel motors are fast industrial quality servo motors that have internal microcontrollers that provide proportional/integrative/derivative (PID) control. All motors are daisy chained by a 3-wire bus on which they were individually addressed from MATLAB software using the serial protocol at 1 Mbits/second. The motors are integrated with temperature and torque sensors which turn off the motors when the temperature and torque values exceed the preset values. The maximum speed of the motor is 45 RPM in no load condition and at an operating load of 80%, the speed decreases to 10 RPM, hence the speed of the biped is dependent on the speed of operation of the motor.

3DOF 35 mm Flattop Optoforce sensors are mounted at the ankle of each leg to detect the forces exerted by the user. A carbon fiber rod serves as the trekking pole connecting the sensor to the user's hands. Forces obtained are digitized and preprocessed by the Optoforce's custom DAQ device.

The completed design of the biped is shown in Figure 3.1

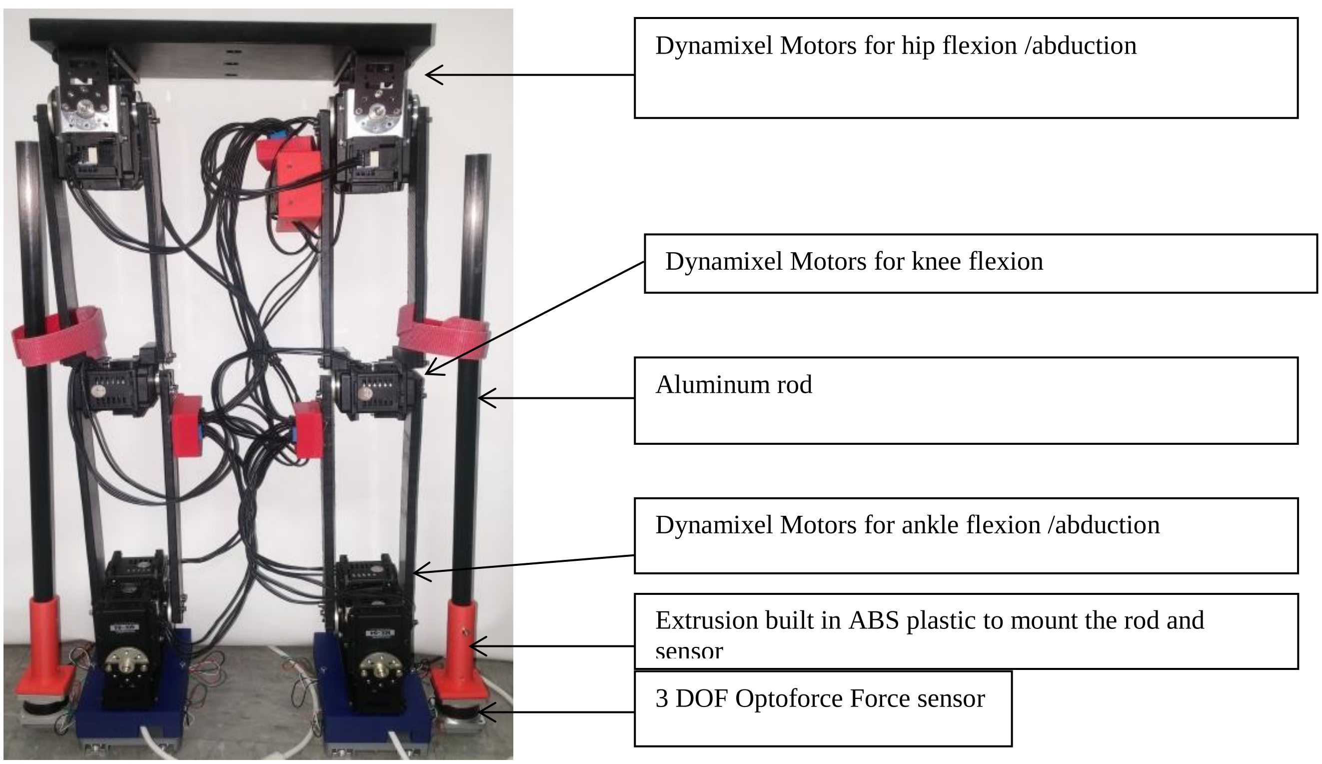


Figure 3.1 Front view of 10 DOF biped robot designed based on anthropometric data.

3.3 Control Algorithm

The control algorithm for each leg consists of three outer admittance loops (one for each force/torque direction) to produce Cartesian position, inverse kinematics to convert Cartesian positions to joint angles and an impedance loop associated with each motor. The admittance loop and inverse kinematics are accomplished in MATLAB on a computer running Windows 7 and the inner impedance loop is executed by the embedded processor in each motor.

3.3.1 Sagittal Plane Control

The forces in the x and y directions are read at 1000 Hz from the 3 DOF Optoforce force sensor. The Cartesian position in each direction is computed by solving the differential equation (Equation 3.1) using CVode (Ordinary Differential Equation Solver) developed at Eindhoven University (Van riel, 2012) for each interval of .01 s.

$$x''(t) = \frac{F(x)}{M} - \frac{B * x'(t)}{M} \tag{3.1}$$

Where F(x) = Force (N), M=Mass (kg), B = damping (Ns/m), x' (t) =velocity (m/s), x'' (t) =acceleration (m/s^2)

The x and y foot position values are converted to the corresponding joint angles of the knee and hip using custom developed inverse kinematics. The joint angle of the ankle is computed from the hip and knee to keep the foot parallel to the floor. The joint angles are in turn converted to hip, knee and ankle motor positions at each .01s time interval in the sagittal plane. These motor values are in turn fed to the corresponding motors to generate the required torque to perform the movement intended by the user. The control paradigm is described by the flow diagram shown in Figure 3.2.

Inverse Kinematics

The inverse Kinematics algorithm uses the positions of the foot and converts to angles of the hip, knee and ankle in sagittal plane. The angles are calculated using the law of cosines (Equations 3.2, 3.3 and 3.4), where X1 and Y1 are the positions obtained from the admittance control paradigm. θ1 is the hip angle, θ2 is the knee angle, and θ3 is the ankle angle.

$$\theta 2 = -2tan^{-1}\sqrt{\frac{(L1+L2)^2-(X1+Y1)^2}{(X1+Y2)^2-(L1+L2)^2}} \tag{3.2}$$

$$\theta 1 = \tan\left(\frac{L2sin\theta 2}{L1+L2cos\theta 2}\right) - \tan\left(\frac{\mathrm{Y1}}{\mathrm{X1}}\right) \tag{3.3}$$

$$\theta 3 = \ \mathrm{\theta 2} - \tan\left(\frac{Y1}{X1}\right) + \tan\left(\frac{L2sin\theta 2}{L1+L2cos\theta 2}\right) \tag{3.4}$$

Where, X1, Y1 are the end effector position, L1 is the link length between hip and knee, L2 is the Link length between knee and ankle. θ1 is hip angle, θ2 is knee angle, θ3 is ankle angle.

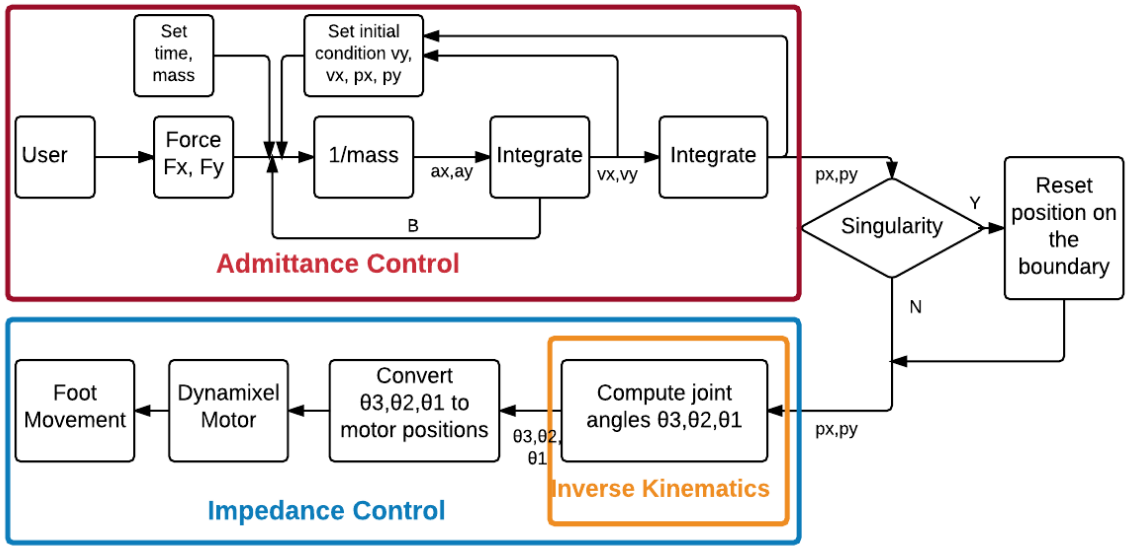


Figure 3.2 Control Algorithm for sagittal plane: The control algorithm consists of the entire control running at 100 Hz. The impedance control running at 1000 Hz. The admittance loop includes our custom admittance control algorithm. Here, θ1-hip angle, θ2-knee angle, θ3-ankle angle. px, py ax, ay vx and vy are position, acceleration and velocity in Cartesian x and y direction, respectively.

3.3.2 Coronal Plane Control

The foot position in the z axis is computed by reading the force in the z-direction from the force sensor at 1000Hz. The force is converted to torque by multiplying the force by the moment arm. Angular acceleration is computed by dividing the torque by inertia (Acceleration =Torque/Inertia); the double integration of the angular acceleration provides the angular position in z direction. The angular position is obtained by solving the differential equation (Equation 3.5) using CVode (Ordinary Differential Equation Solver). The joint angle of the ankle to keep the foot parallel to the floor is computed. The joint angles are in turn converted to hip and ankle motor positions at every point of time. These motor values are in turn fed to the corresponding motors to generate the required torque to perform the movement intended by the user. The ankle angle is computed from hip angle as shown in Equation 3.6. All the algorithms are written and executed using MATLAB.

$$x''(t) = \frac{T(x)}{I} - \frac{B * x'(t)}{I} \tag{3.5}$$

$$\theta 5 = 180 - \theta 4 \tag{3.6}$$

Where, T(x) = Torque (Nm), I=Inertia (kgm^2), B = damping (Ns/m), x' (t) =velocity (m/s), x'' (t) =acceleration (m/s2), θ4 is the hip angle, and θ5 is the ankle angle.

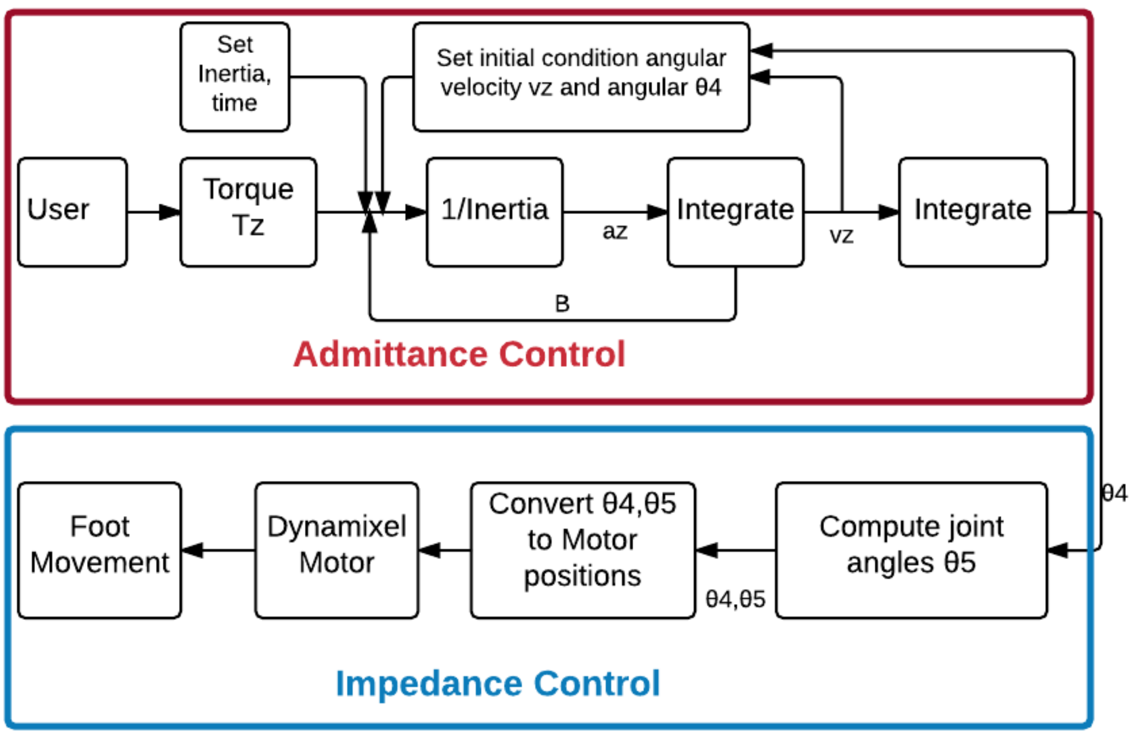


Figure 3.3 Control Algorithm for Coronal Plane: The control algorithm consists of an admittance control running at 120 Hz and an impedance control running at 1000 Hz. Here, θ4 is the hip angle, θ5 is the ankle angle. ay vx and vy are angular acceleration and angular velocity in x and y directions, respectively.

3.4 Gait Control

The gait cycle involves both swing and stance phase, when one leg is in swing the other leg is in stance according to the gait cycle. Swing phase involves the hip, knee, and the ankle joints of the leg. The stance phase involves the hip and ankle joints of the leg. Two control paradigms have been designed to control the gait cycle; Hybrid Control and Complete Control.

3.4.1 Hybrid Control

The hybrid control involves the user controlling only the swing leg of the robot using the admittance control algorithm while the stance foot positions are programmed in real time based on the swing foot positions to aid the robot to move forward.

The hybrid control algorithm computes the joint angle of the swing phase based on the force exerted in x, y, z directions by the user on the rod mounted on each leg. The joint angle of the stance phase is determined based on the stride length and amplitude of the swing phase. These motor values are in turn fed to the corresponding motors to generate the required torque to perform the movement intended by the user. Four strain gage single axis force sensors are mounted on the four corners of the bottom of each of the custom designed feet of the biped robot. The forces from sensors are read every 8 ms into MATLAB. The readings are used to determine if the leg has reached stance from swing phase and vice versa. The swing phase of either foot is possible only if the other foot is on the ground. The haptic feedback is provided by the floor and is felt by the hand when the foot in stance phase is in contact with the floor. The user controls the gait pattern with only the swing leg while the synchronization of the stance leg with the swing leg is performed by the algorithm.

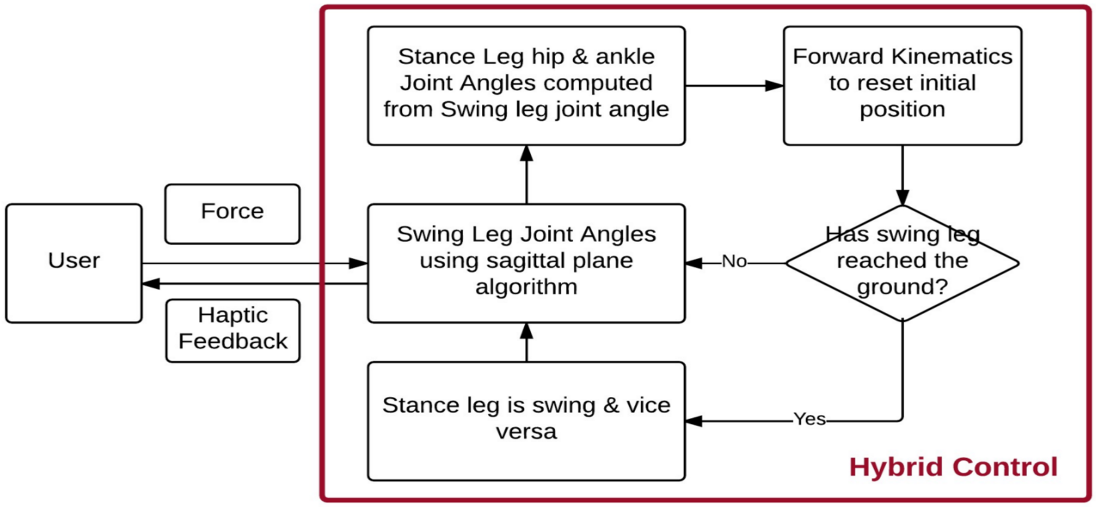


Figure 3.4 Control Flow of Hybrid Control Mechanism.

3.4.2 Complete Control

The complete control involves the user independently controlling both the swing leg and stance leg of the robot using the admittance control algorithm.

The complete control algorithm computes the joint angle of the swing phase and stance phase based on the force exerted in x, y, z directions by the user on the rod mounted on each leg. These angle values are converted into motor values and are in turn fed to the corresponding motors to generate the required torque to perform the movement intended by the user. The haptic feedback is provided by the floor and is felt by the hand when the foot in stance phase is in contact with the floor. The user needs to swing the leg forward while the trekking pole of the stance leg is pushed backwards. Since the stance foot is unable to move, the hip joint, and hence the entire robot is propelled forward. Here, the user needs to synchronize the movement between both the leg as well maintain the gait pattern.

3.5 Evaluation of the Design

A slow gait-like movement was performed for a period of 60 s to demonstrate the user control. The accuracy of the algorithm, time delay, and horizontal trajectories was evaluated to validate the working of the control algorithm.

3.5.1 Accuracy

The accuracy of the algorithm was evaluated by comparing the lag in cm between the desired and the actual position of the feet in x and y direction while performing the gait-like movements using the hand. The accuracy in the sagittal plane was evaluated using the forward kinematics algorithm. The motor angles were read at each time point in the sagittal plane from hip, knee motors. The motor positions were converted to joint angles of the hip and knee. A forward kinematics algorithm was developed and applied to obtain the x and y position reached by the motor in Cartesian space. This position was in turn compared with the desired x and y positions (x and y position computed using the admittance control) to evaluate the accuracy of the inverse kinematics algorithm and accuracy of the motor.

The actual and desired angular position in z axis was also evaluated for the accuracy of algorithm. The hip abduction and adduction motor position were read from hip motor. The motor position was converted to joint angle. The joint angle was in turn compared with the joint angle of the computed angular position in z axis.

3.5.2 Time Period

The time period required to perform each iteration of the algorithm was evaluated. A longer time period leads to poor user experience as the user would experience the movement afresh at each time point; hence it is important to evaluate the lag time for understanding the user experience.

3.5.3 Horizontal Trajectories

The foot trajectories in sagittal plane were compared to the normal foot trajectories to evaluate the effectiveness of the control using hand trajectories by one subject.

3.6 Results and Discussion

3.6.1 Accuracy of the Algorithm

As shown in Figure 3.5, the desired trajectory followed the actual trajectory closely in both x and y direction. The average lag in cm between the actual and desired trajectory was computed to be .1 cm in the x direction and .13 cm in the y direction.

As shown in Figure 3.6, the actual angular position followed the desired angular position closely in z direction. The average lag in radians between the actual and desired trajectory was computed to be .04 radians in z direction.

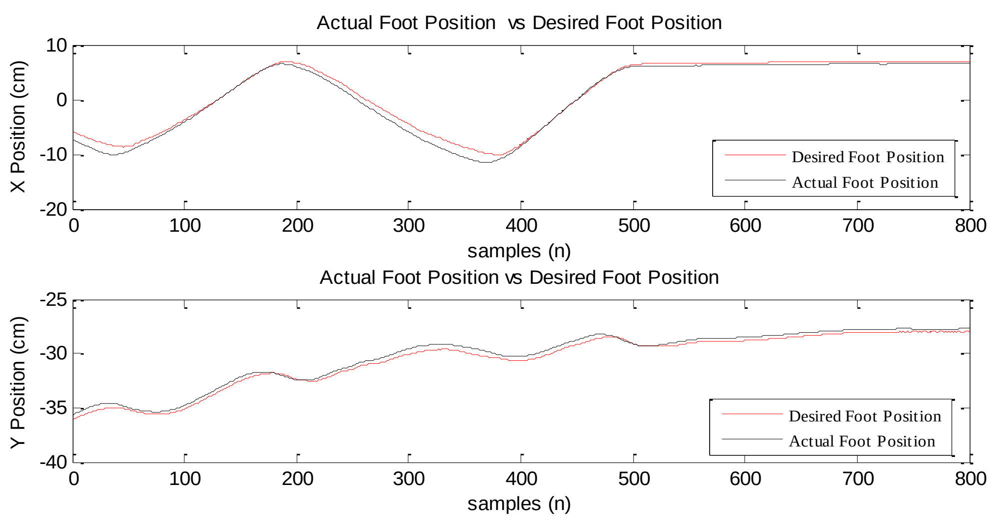


Figure 3.5 a) Desired foot position (red) and actual foot position (black) in x axis. b) Desired foot position (red) and actual foot position (black) in y axis.

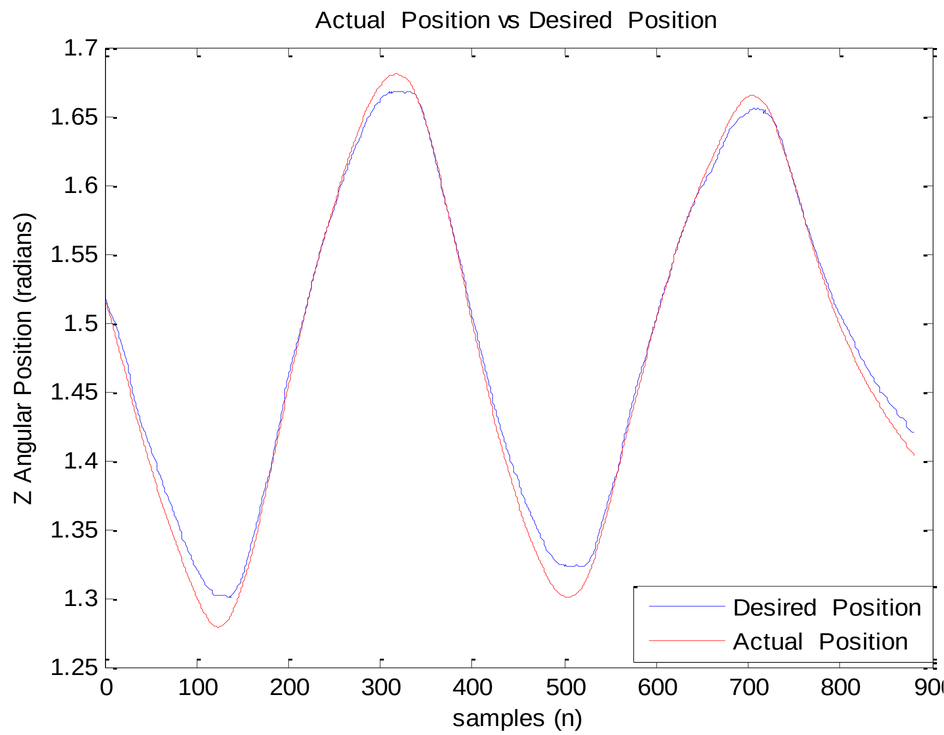


Figure 3.6 Desired angular position (blue) and actual angular position (red) in z axis.

3.6.2 Time Duration of the Algorithm

The time duration was computed for each iteration to be 10 ms for each leg with read function. This time was further reduced to 8 ms for both legs by using sync write function and removing the read function. Since the algorithm to control does not require the position of motors, the read function was not used in the algorithm.

Studies have shown that control loop of 100 Hz is sufficient for human operators to feel smooth, nearly passive movements of a robot (Van der Linde, 2002).

3.6.3 Foot Trajectory

Hybrid Control Trajectories

Figure 3.7 shows the complete gait cycle produced by the control algorithm which resembles the gait trajectory produced during normal walking. Also, it can be observed that there is smooth transition between the stance to swing and vice versa (Figure 3.8).

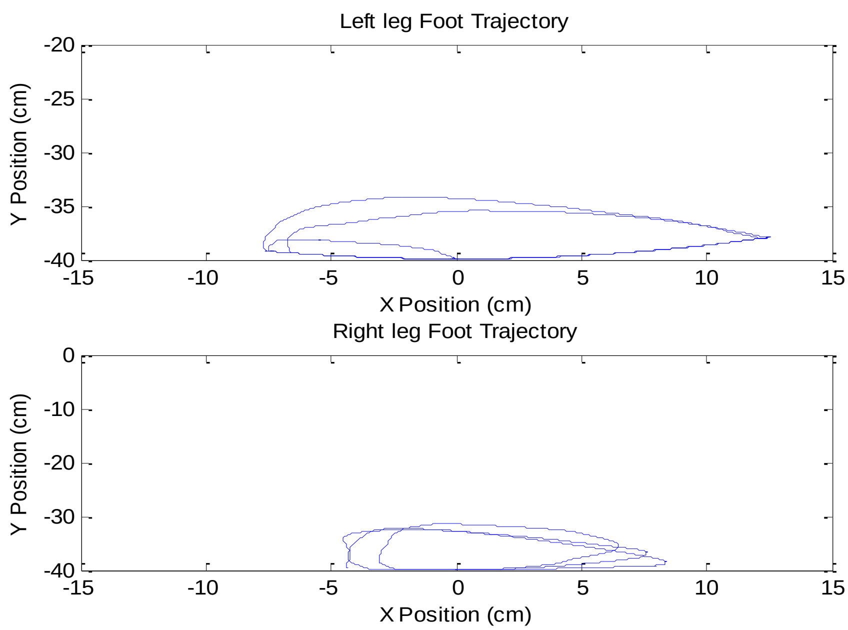


Figure 3.7 Foot trajectory of left leg (top) and the right leg (bottom) during gait cycle.

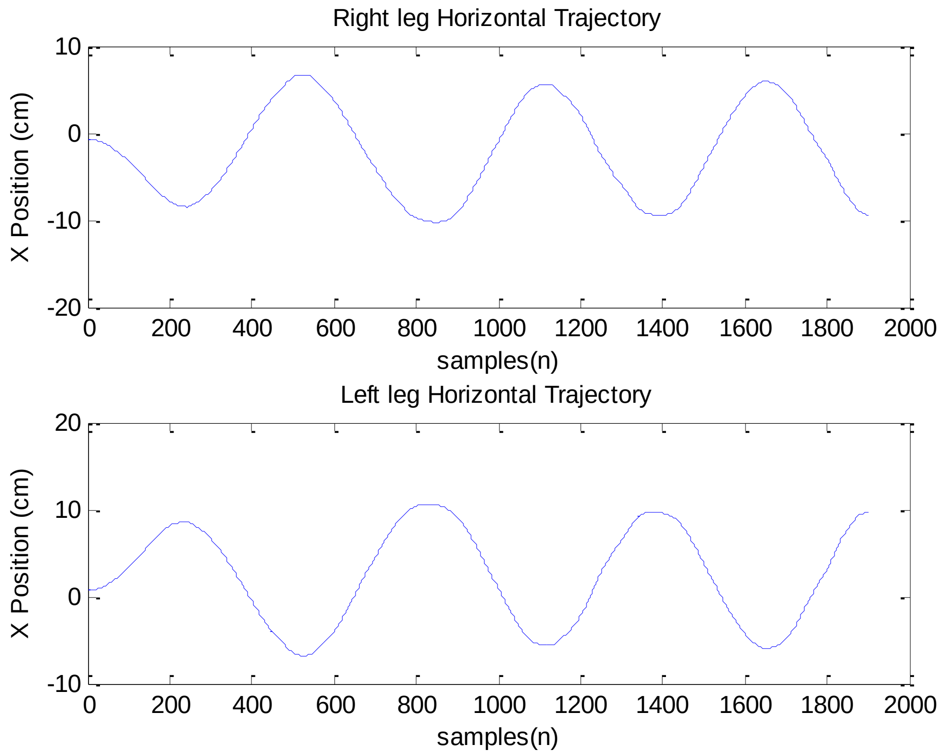


Figure 3.8 Horizontal trajectory of left leg (top) and the right leg (bottom) of the robot during gait cycle.

Complete Control Trajectories

Figure 3.9 shows the complete gait cycle produced by the control algorithm which resembles the gait trajectory produced during normal walking. Also, it can be observed that there is smooth transition between the stance to swing and vice versa (Figure 3.10).

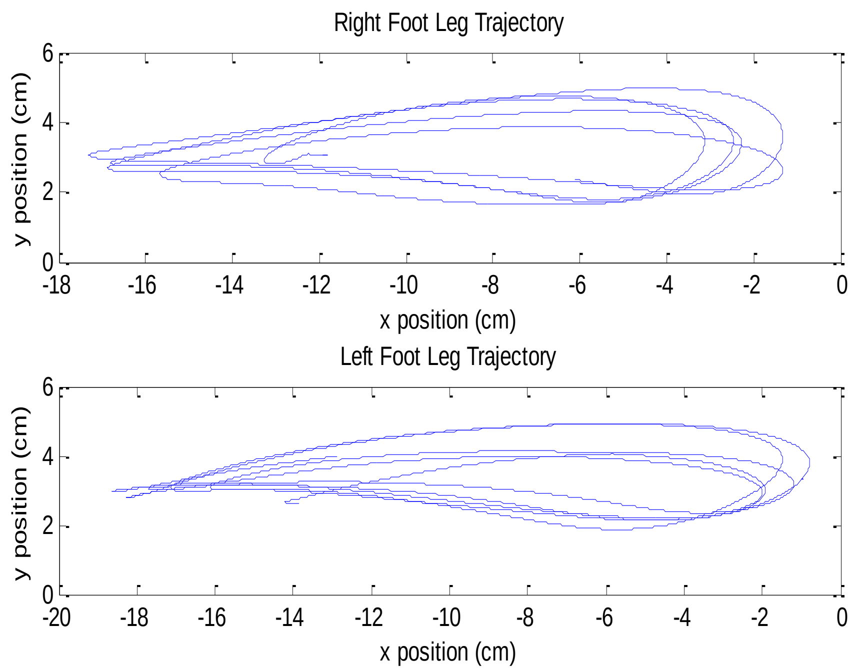


Figure 3.9 Foot trajectory of left leg (top) and the right leg (bottom) during gait cycle.

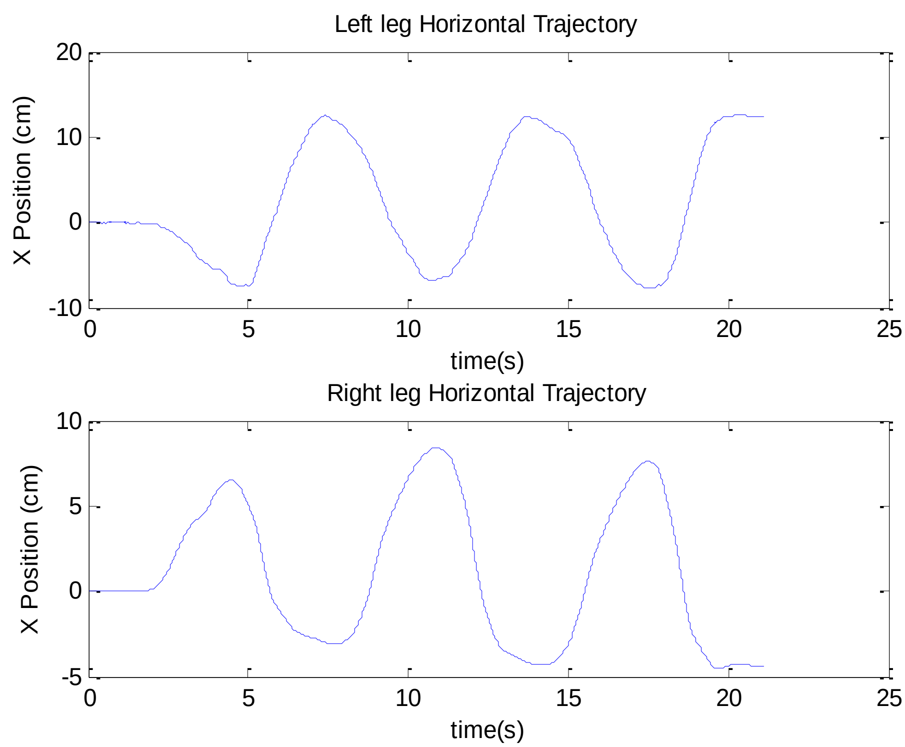


Figure 3.10 Horizontal trajectory of left leg (top) and the right leg (bottom) during gait cycle.

The progression in a normal gait cycle is achieved by two main forces; the swing leg generates the force during the onset of the swing to move the torso forward and the force generated by the contralateral stance leg during single stance helps with the movement of the torso forward, as the ankle dorsiflexes beyond neutral and accelerates with heel rise further propelling the torso forward. The moment generated by these two forces is further used by the next stance phase during the heel contact. Thus, throughout the stance period the heel, ankle, and forefoot serially serve as a rocker that allows the torso to advance over the supporting foot, while the hip helps by providing moment to move the torso forward.

In the case of biped, the flat footed nature of your biped gait greatly reduces the contribution of the swing leg in translating the torso forward. Hence, the motor at the stance ankle has to provide considerably more torque than the biological stance ankle to provide the rocker movement to propel the torso forward. The Hybrid and the Complete control mechanism have actuation about the ankle to provide the torque required to perform the movement to propel forward.

3.7 Conclusion

An admittance control based user-robot control strategy has been developed and tested that allows the user's neurally generated foot trajectories to be redirected through the hands, with hand-generated movements and forces precisely controlling the movement of the biped. Such a system will allow complete user control of real-time ambulation, and will provide haptic (proprioceptive) feedback through the hands, that is essential for modifying gait in the everyday world.

CHAPTER 4

EVALUATION OF THE BIPED CONTROL MECHANISM

4.1 Introduction

Human gait is highly coordinated movement that involves both legs synchronized to propel the body forward. Though gait is a highly rhythmic movement, it is often influenced by the environment; changes in walking conditions can have effects on the patterns. There are many factors in walking activity, such as speed, stride frequency; obstacles etc. that influence the gait pattern (Tanawongsuwan, 2003).

The objective of this study was to evaluate the gait pattern of the biped under all walking conditions when controlled by users using the complete control approach described in Chapter 3. This converts the user intentions of the hand into real time joint angles. The control strategy was tested using naïve healthy subjects on the treadmill to evaluate the kinematics of hand walking control mechanism.

The gait kinematics is defined by the joint angles and duty cycle. Other gait parameters (pattern, speed, time and amplitude synchrony) are used to define user's dynamic stability. The gait can be defined by its pattern, speed, time and amplitude synchrony and inter limb co-ordination. Stride variability in time and amplitude is an important characteristic of stability. Maki et al. found that increased variability of speed and stride length increased the likelihood of a fall (Maki,1997). Hausdorff et. al. found that variability in stride time or decreased time synchrony predicted fall (Hausdorff, 2001). Both studies show that increased variability in time synchrony or stride amplitude lead to instability of gait. Further, it was found the inter limb co-ordination also plays an important role in the stability of gait. Variability in symmetry between the time and amplitude between the limbs also are predictors of instability (Paterson, 2011, Yogev, 2007).

A practiced task inherently produces a smooth trajectory without jerk. Hogan et al. describe smoothness as minimization of mean squared jerk (Flash, 1985).This quantitative measurement of smoothness was utilized to demonstrate that the smoothness of simple, novel, movements increased as the skill level of the task improved, or became better learned (Schneider, 1989). A smooth end point trajectory also defines a planned movement. In gait, this end point is represented by the foot (Hreljac, 2000). Here, the smoothness of the foot trajectory of biped walking was evaluated to understand the jerk associated with performing the movement with different conditions.

The aim here is to show that the robot gait trajectory, produced at varying speeds, with and without obstacles by the real time control of the foot of the biped resembles the normal gait under such varying walking conditions. Described below is the design & methodology of the experimental protocol and the data obtained. Analyses of these results show presence of synchrony, smoothness and similarity to human gait, showing that the users were able to control our biped robot’s complete gait cycle such that the biped gait compared well with normal human gait. This Chapter addresses specific aim 2.

4.2 Design and Methodology

4.2.1 Experimental Setup

Biped Walking

The experimental setup included a Pro-form J6 treadmill around which custom frame was built at NJIT using 80-20 aluminum to support the biped on the treadmill. The custom frame allows the users to have complete view of the treadmill and the robot. Optitrak motion capture system Trio was mounted on the treadmill to record the movement of the biped. The treadmill speed was reduced by adding a high voltage resistor in series with the motor. Optitrak markers were placed on the hip, knee and ankle of both the legs to track the movement of legs.

Human Walking

The experimental setup included the same Pro-form J6 treadmill. An Optitrak motion capture system Trio was placed behind the subject to record the movement of the subject on the treadmill. A custom construct was designed and printed in ABS to strap the Trakstar (Appendix C) and Optitrak markers at hip, knee and ankle of both the legs. A Data Acquisition system from National instruments was used to sync the Optitrak cameras with MATLAB.

Subject population

The study included fourteen subjects divided equally into two groups, with one group controlling the biped on the treadmill and the other group performing normal human walking on the treadmill. The subjects were randomly assigned to the groups. All subjects were able bodied subjects under the age of 35, with fully functional upper and lower extremities. The groups were age and gender matched. Exclusion criteria included any disability to the upper or lower extremities or inability to perform normal gait. The study was approved by the NJIT IRB and the experiment was performed with the subjects' consent.

4.2.2 Experiment

The experiment was designed to measure kinematics, synchrony, smoothness under varying walking conditions such as speed and obstacles so as to evaluate the biped robot's gait in comparison to normal human gait. The subjects who participated in the experiment were divided into two major groups: Biped control group and the Human walking group. The biped control group controlled the gait of the biped on a treadmill using their hand movements while the human walking group performed normal gait on a treadmill. The study consisted of a total of eight sessions. Each session comprised of eight trials. Each trial was for duration of one minute followed by a thirty second rest. The speed of the treadmill during each trial was as shown in Table 4.1. The speed was varied after thirty seconds in Sessions 3, 4, 7 & 8. Sessions 5-8 included users stepping over obstacles. The number of obstacles presented to each leg was the same, though not in the same sequence. Twice the numbers of obstacles were included for the biped group in order to offset for their reduced speed. Both groups avoided obstacles with only one foot during a trial. The obstacle avoidance was alternated between both feet in the successive trials.

Table 4.1 Speed of Treadmill for Each Trial During Each Session

	Sessions1&2	Sessions 3 &4	Sessions 5&6	Sessions 7&8
Trial 1	Medium	Medium to High	Medium	Medium to High
Trial 2	Medium	Medium to Low	Medium	Medium to Low
Trial 3	High	High to Low	High	High to Low
Trial 4	Low	Low to High	Low	Low to High
Trial 5	Medium	High to Low	Medium	High to Low
Trial 6	High	Low to Medium	High	Low to Medium
Trial 7	Low	High to Low	Low	High to Low
Trial 8	Medium	Low to High	Medium	Low to High

I. Biped Control Walking

All subjects of the biped control group control the gait of the biped using the trekking pole extending from the foot of the robot as shown in Figure 4.1 for eight trials in each of the eight sessions. The biped control group performed two additional familiarization sessions before the start of the actual sessions, these sessions were not included as part of the analysis. The first familiarization session was performed without the treadmill. The subjects controlled the leg of robot in the air to get accustomed to kinematics of the leg. Following this, the second familiarization session included eight trials which were performed on the treadmill with the lowest speed (.5 mph).

The biped was placed on the treadmill and the user was seated in a comfortable chair behind the treadmill. The users were instructed to control gait pattern of the biped during each trial by applying small forces to the pole extending from the sensor on each leg in the direction of the intended movement. The control involved ipsilateral control, i.e. the right hand of the subject controlled the movement of the right leg and vice versa. They were instructed to walk the robot on a straight path. The first four sessions involved no obstacles and last four sessions involved navigating obstacles. Obstacles in the form of visual markers were placed in the path of the biped at random intervals on the treadmill, requiring the user to change the stride length of a leg. The subjects were instructed to step over the obstacles. During the obstacle navigation session, a black tarp covered part of the treadmill to keep the subjects from seeing the obstacles in advance. The speed of walking was varied between the trials to test the smoothness of the gait with varying speed. The speed variations for the biped were 0.1, 0.2 and 0.3 mph (0.04 to .16 m/s).

Figure 4.1 User control of biped a & b) User controlling the biped on treadmill. C) User controlling the biped in air during familiarization session.

II. Human Walking

Subjects were instructed to perform normal ambulation for eight sessions with eight trials each on a treadmill, with each trial lasting one minute followed by thirty seconds rest. Not more than one session was performed on a given day. Subjects were instructed to walk in a straight path. The speed of treadmill was varied between the trials (1-2 miles/hr) depending on the subject's normal walking speed. All the subjects' speed was calibrated before the beginning of the first session. This was performed by asking the user to walk at their normal walking speed, then the high and low speeds were derived by adding and subtracting .4mph respective to their normal walking speed. Similar to the biped walking protocol, the first four sessions involved no obstacles and last four sessions involved navigating obstacles. Obstacles in the form of visual markers were placed at random intervals on the treadmill, requiring the user to change the stride length. The subjects were instructed to step over the obstacles. One subject was asked to walk at low speed with a foot orthosis. Due to the orthosis, the subject performed a flat footed gait. The speed was a constant low speed and no obstacles were present.

The data was processed using MATLAB and all statistical analysis was performed using SPSS.

4.2.3 Data Analysis

Horizontal and vertical trajectories collected at 120 Hz of the ankle, hip and knee were filtered using an effective 4th order, zero-lag Butterworth low-pass filter. The cutoff frequency was determined using the power spectrum. The filtered data were used for further analysis.

4.2.3.1 Foot Trajectory.

i. Root Mean Square Error

The Root Mean Square Error (RMSE) was determined as a measure to evaluate the differences in gait cycle between trials for Sessions 1 and 2, in other words to evaluate the synchrony between steps. The trajectories were divided into individual complete gait trajectories of all steps during a trial. The gait cycles were further normalized based on the toe off. Following normalization, the first gait cycle (in other words, first step) was excluded and the RMSE distance between the following complete gait cycles (was considered the first step) and all other remaining steps were computed. By comparing with the first gait cycle, we can determine how the subsequent steps differ during the trial with respect to start of the trial. To do so, we computed the standard deviation of RMSE for each trial of each session for all subjects. Greater the standard deviation of RMSE between first step and remaining steps, lower is the ability to generate a synchronous trajectory during the duration of a trial.

In order to identify any possible learning, we compared Sessions 1 and 2 using Paired Sample T-test. Further, the influence of speed was analyzed to determine, if the speed affected RMSE, i.e. the ability to produce synchronous steps. Independent sample t-test was used to compare between biped and human walking in left hand in session2.

ii. Amplitude and Frequency Synchrony

The rhythmicity of the movement was evaluated using amplitude and frequency synchrony. The time taken for a gait cycle was computed for all the gait cycles in a trial for Sessions 1 and 2 for the human walking and biped walking. Human walking and biped walking differ by their physical size, and would therefore result in different time and distance. Hence the measures were made comparable through normalization by dividing the collected values by the maximum value during a particular trial for human walking and biped walking independently.

The amplitude synchrony was computed by calculating the inter peak distance between the horizontal and vertical trajectory with similar normalization as frequency synchrony. The standard deviations of amplitude and frequency synchrony were computed to determine the variation in the amplitude and frequency of the gait cycles in a trial.

A) Human vs. Biped walking

The amplitude and frequency synchrony were compared between the normal human walking and the biped control group to determine synchronicity differences between the two groups. Based on the assumptions of statistics, Independent sample t- test was employed to compare the values of time synchrony and amplitude synchrony between the two groups at Sessions 1 and 2.

B) Biped Walking Session 1 vs. Session2

The amplitude and frequency synchrony in the biped groups were also compared between Sessions 1 and 2. As synchrony is a measure of differences in reproducible actions, synchrony in gait amplitude and frequency was therefore used as a measure to determine the improvement in periodicity of the movement with practice. Paired sample t-test was used to compare the values of synchrony between Sessions 1 and 2 for biped walking. Pearson's correlation was used to correlate the left and right leg synchrony.

4.2.3.2 Smoothness of the Foot Trajectory. The smoothness of the trajectory can be calculated as a function of jerk. Hogan et al. quantified smoothness of a trajectory as a function of jerk, which is the time derivative of acceleration. The smoothness can be quantified as a square of the third derivative of the position as shown by Equation 4.1. Hence the jerk was computed for a position in x and y plane and further the normalized smoothness of the trajectory were calculated per trial. The measure of the jerk of movement should be without any dependency on the duration and amplitude of the measure i.e. must be dimensionless. The integrated squared jerk has dimensions of length squared divided by the 5th power of time (Hogan, 2009). Hence a dimensionless measure is used as shown in Equation 4.1.The smoothness of the trajectory was analyzed between sessions to determine the influence of speed and obstacles.

The smoothness of trajectory was also compared with the computed smoothness of trajectory during human walking.

$$Smoothness = \sqrt{\frac{time^5}{2*Length^2}\int_0^T jerk^2 dt} \qquad (4.1)$$

Where, Length is the total length of trajectory.

A) Smoothness differences between Human vs. Biped walking

The jerk of Sessions 1 and 2 of human walking was compared with biped walking.

B) Constant Speed

The jerk of Sessions 1 and 2 of biped walking was compared to determine learning between sessions.

C) Effect of speed change in a trial on Smoothness

The Smoothness of Sessions 2 and 3, and Sessions 2 and 4 of biped walking were compared to determine influence of speed on jerk. Also, Sessions 3 and 4 were compared to determine learning between sessions.

D) Effect of obstacle navigation with constant speed in a trial on Smoothness

The Smoothness of Sessions 2 and 5, and Sessions 2 and 6 of biped walking were compared to determine the difference in Smoothness due to obstacles. Also, Sessions 5 and 6 were compared to determine learning between sessions.

E) Effect of obstacle navigation with varying speed in a trial on Smoothness

The Smoothness of Sessions 2 and 7, and Sessions 2 and 8 of biped walking were compared to determine the difference in Smoothness due to obstacles and speed. Also, Sessions 7 and 8 were compared to determine learning between sessions.

As the data violated assumptions of normality for parametric tests, Man Whitney U test was used to compare the values of jerk between Human and biped and Wilcoxon Paired test was used to compare values between different sessions.

4.2.3.3 Joint Angles. The joint angles of hip and knee of both the legs of the biped robot walking for all subjects were computed from the filtered Cartesian position from the Optitrak data using inverse kinematics for Sessions 1 and 2. The joint angles were also computed for single human subject who performed the flat footed gait on the treadmill and also for all other normal human walking subjects**.**

4.2.3.4 Duty Cycle. The duty cycle of the gait cycle was calculated. The average relative percentage of stance and swing phase was calculated for Sessions 1 and 2 using the formula given below:

Duty cycle= Stance Phase + Swing Phase

Stance Phase % = 100 * Stance phase/ Duty Cycle

Swing Phase % = 100 * Swing phase/ Duty Cycle

4.2.3.5 Frequency Variation. Sessions 3 and 4 included trials with change in speed after thirty seconds. The hypothesis is that as the speed changes the length of steps and the number of steps should change. The trial was divided into two by dividing the trial at thirty seconds. The frequency with maximum power was determined for each trial from absolute square of fast Fourier transform (fft) signal. This frequency is the frequency of walking during the first thirty seconds and last thirty seconds. Correlation between the speed of the treadmill and the frequency of walking during each time period was determined. The Pearson's r was determined for the correlation values**.**

4.2.3.6 Accuracy of the Foot trajectory in Obstacle Navigation. The number of obstacles navigated and missed was determined for every trial for all subjects. The percentage of obstacles successfully navigated was evaluated as a measure to determine the accuracy of obstacle navigation, which is an indicator of accuracy of the foot trajectory. Mann Whitney U test was used for statistical analysis.

4.3 Results

4.3.1 Foot Trajectory

The foot trajectory between the human ground walking and biped walking was found to be very similar as seen in Figure 4.2.

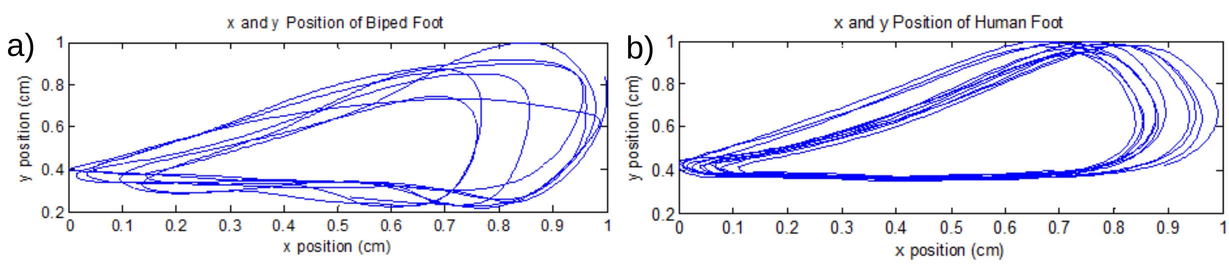


Figure 4.2 Foot trajectory of a) Biped b) Human on treadmill.

4.3.2 Synchrony

The time between steps of a trial between the right and left leg for Sessions 1 and 2 was found to be highly correlated. This implies that the left and right leg had the same time duration for steps for a subject during a trial (Figure 4.3). Also, time synchrony between human and biped walking showed that human walking had statistically significant better synchronicity than biped walking (Figure 4.3c). Human walking showed that synchronicity increased with speed but no such influence was observed with biped walking. The amplitude synchrony results showed no difference between human walking and biped walking and high correlation between left and right leg as shown in Figure 4.3 d and e.

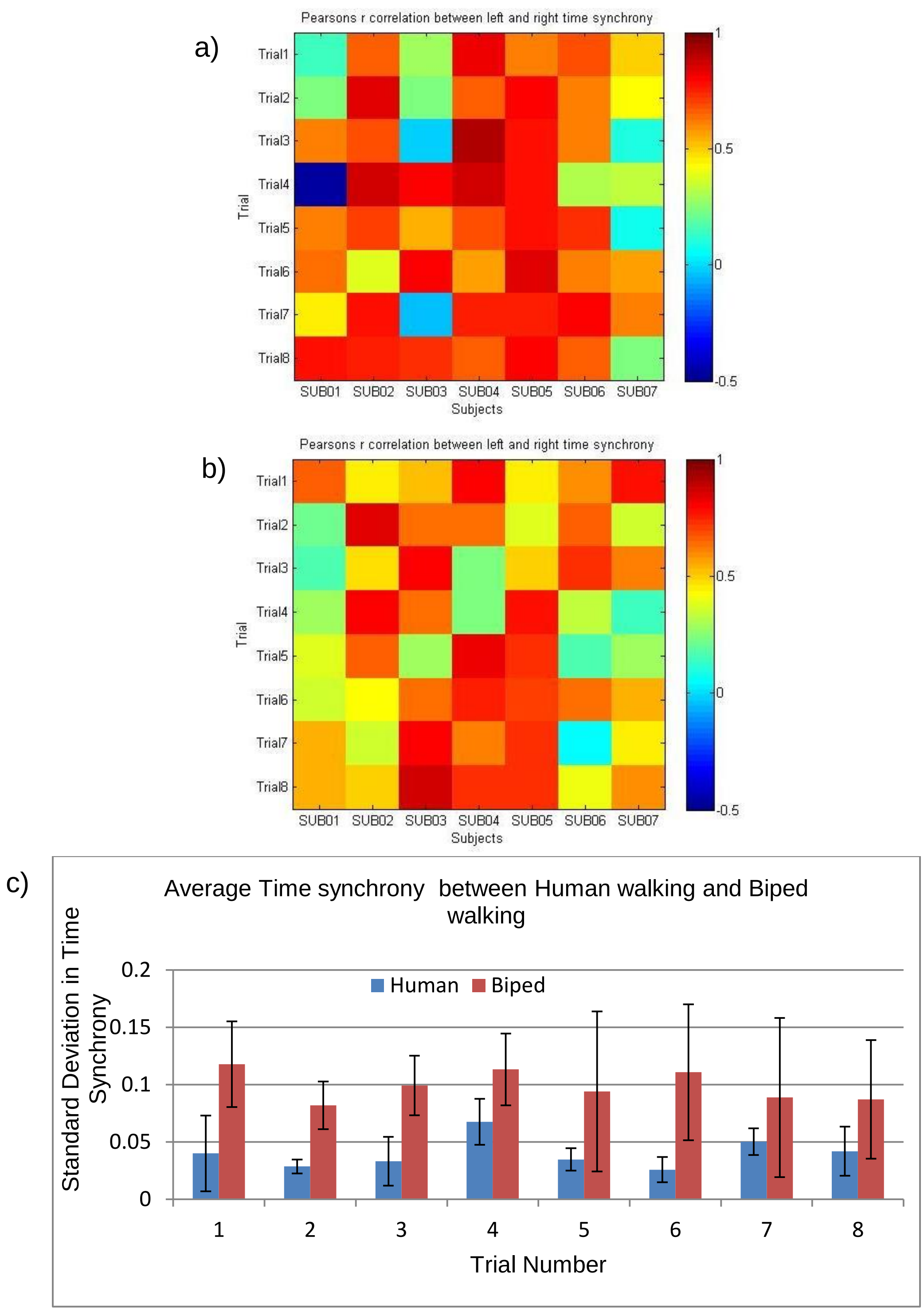
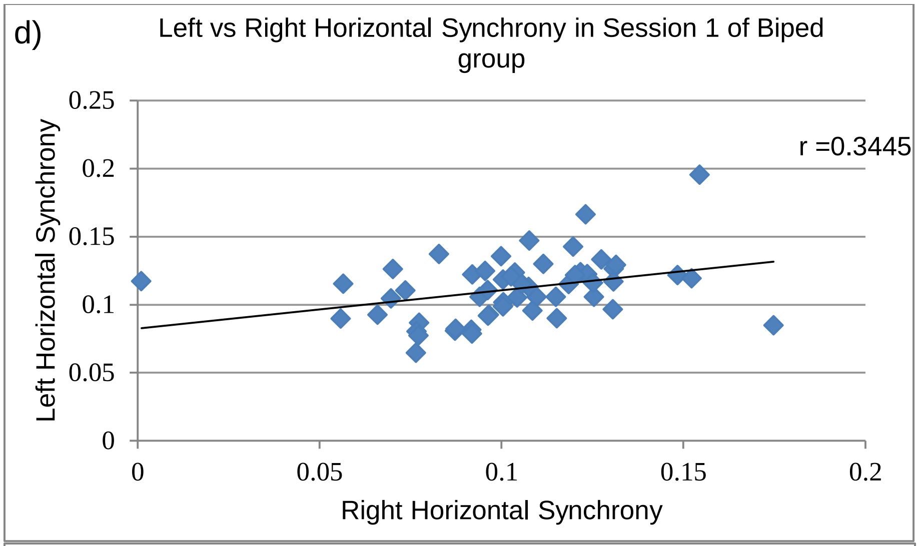


Figure 4.3 Correlation between right and left time of each step for each subject for a) Session 1 and b) Session 2 for biped walking. c) average time synchrony (SEM) of human walking and biped walking by right hand in Session 1.



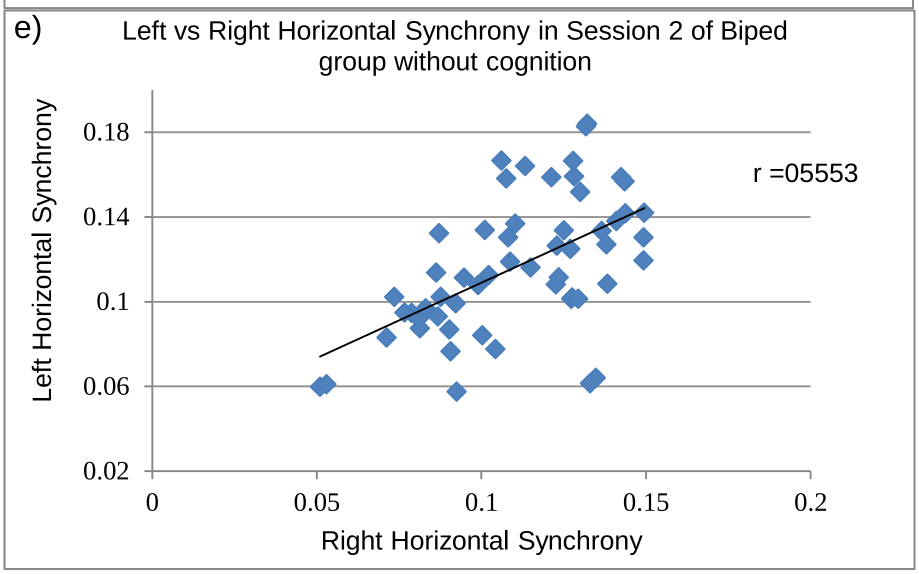


Figure 4.3 (continued) Correlation between right and left amplitude of each step for each subject for d) Session 1 and e) Session 2 for biped walking.

4.3.3 Root Mean Square Error

The RMSE showed statistically significant differences between the Sessions 1 and 2 for both left and right hand as shown in Figure 4.4. No significant difference was observed between human walking and biped walking in Session 2 in left hand. Statistically significant differences were also observed between slow and fast speeds in Session 1 as shown in Figure 4.6. Similar effect of speed on the RMSE was observed in the earlier trials of Session 2 but no significant difference was observed in the later trials of Session

2 between the high and low speed. Also, when the sessions were compared with respect to the speeds, fast speed showed significant difference between the two sessions in left hand as shown in Figure 4.5.

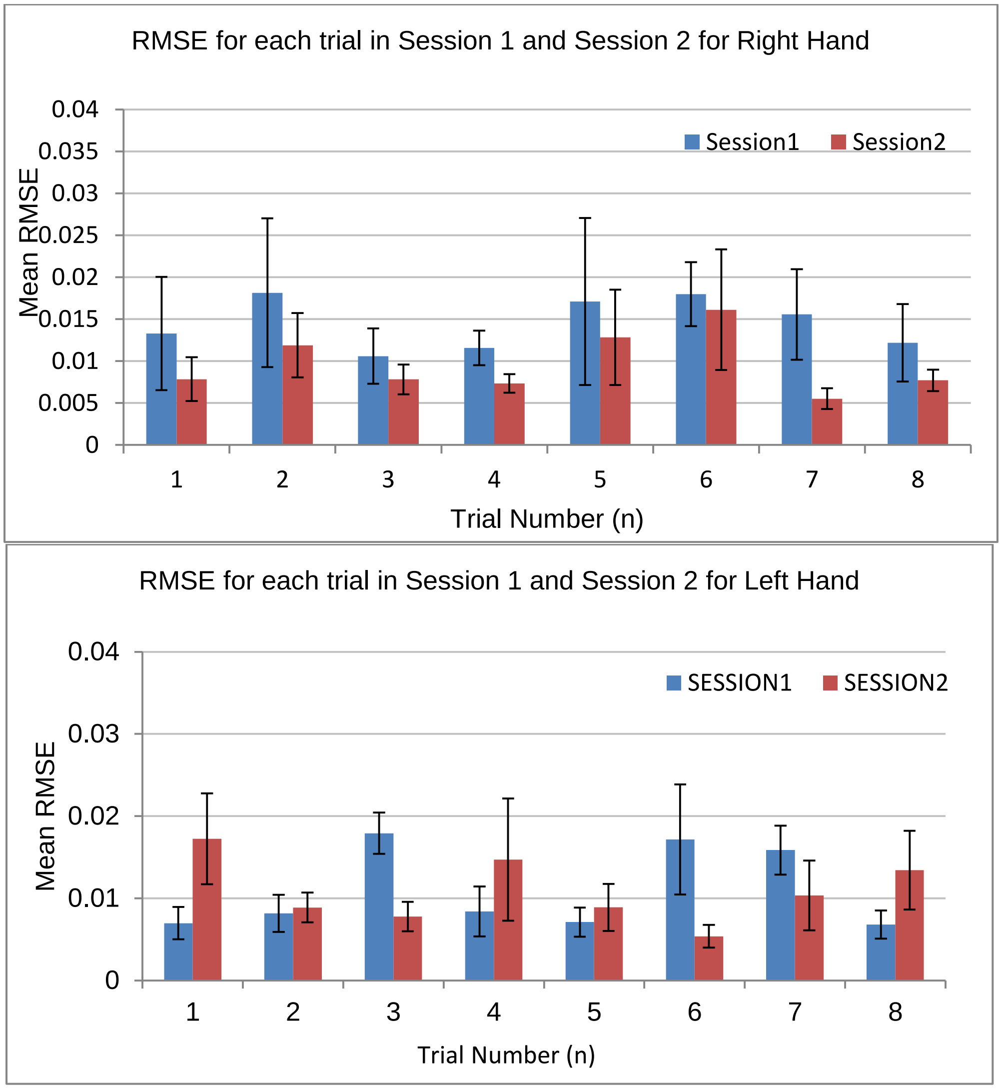


Figure 4.4 Mean Root Mean Square difference between Sessions 1 and 2 for right (Top) and left hand (Bottom).

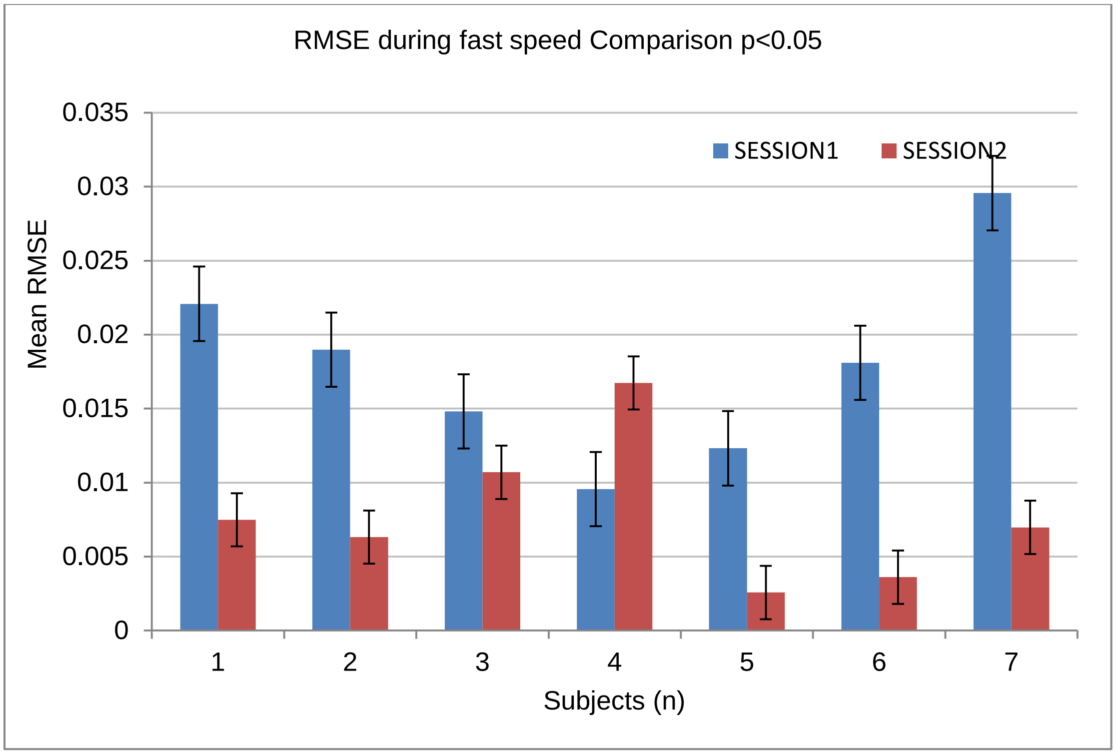


Figure 4.5 Mean Root Mean Square difference of all subjects between Sessions 1 and 2 for left hand with respect to fast speed (trial 3). Significant Difference (p<.05) was observed.

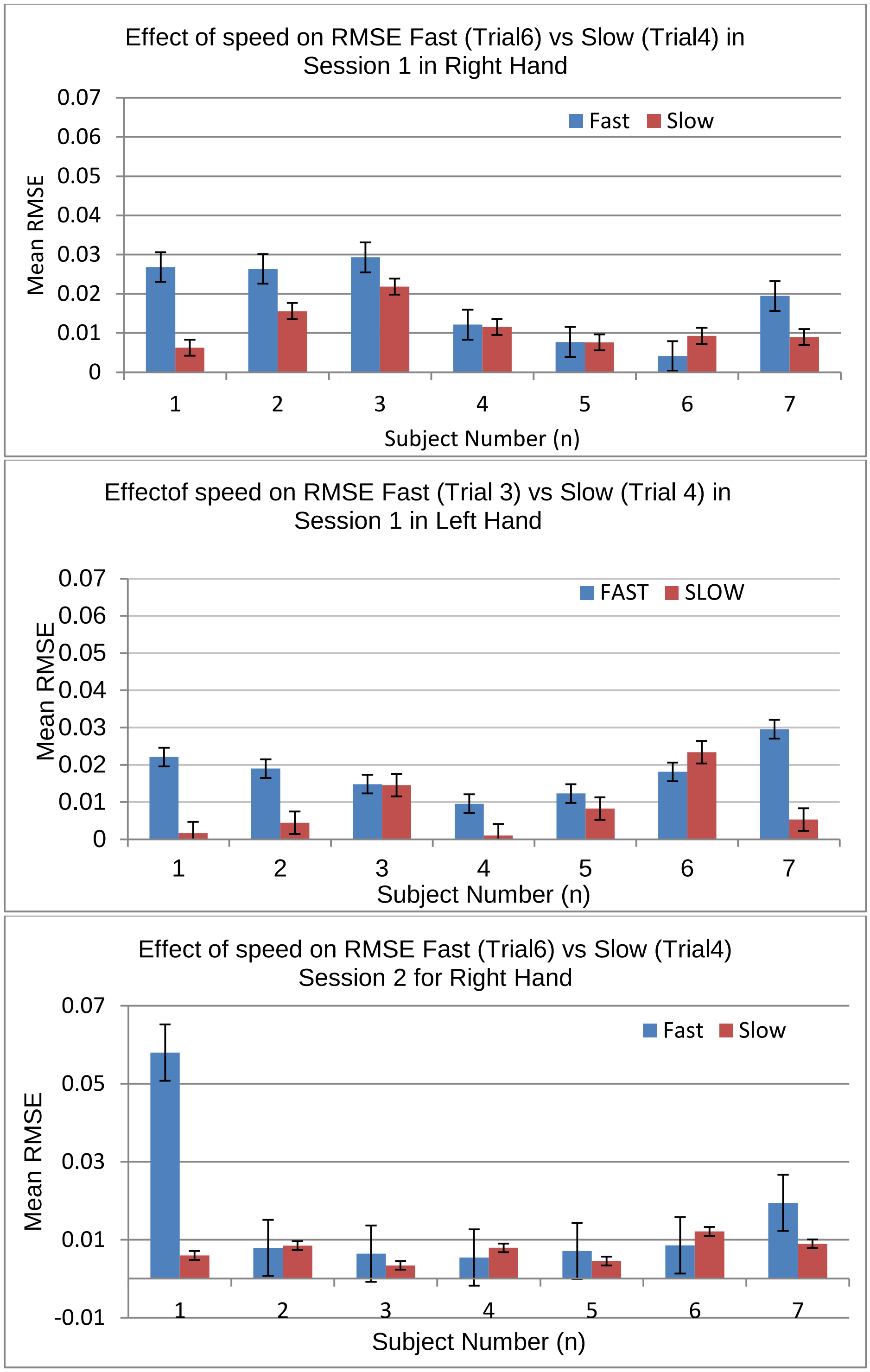


Figure 4.6 Mean Root mean square difference between fast and slow speed in Session 1 for right hand (Top) and left (Middle) hand. Significant Difference (p<.05) was observed. Mean Root mean square difference between fast and slow speed in Session 2 for right hand (Bottom). Significant Difference (p<.05) was observed.

4.3.4 Smoothness of the Foot Trajectory

i) Human Vs Biped Comparison

The Smoothness of trajectory showed that there was no difference between the smoothness of the trajectory in the stance phase but the swing phase showed statistically significant difference in smoothness with the biped trajectories having smoother movement.

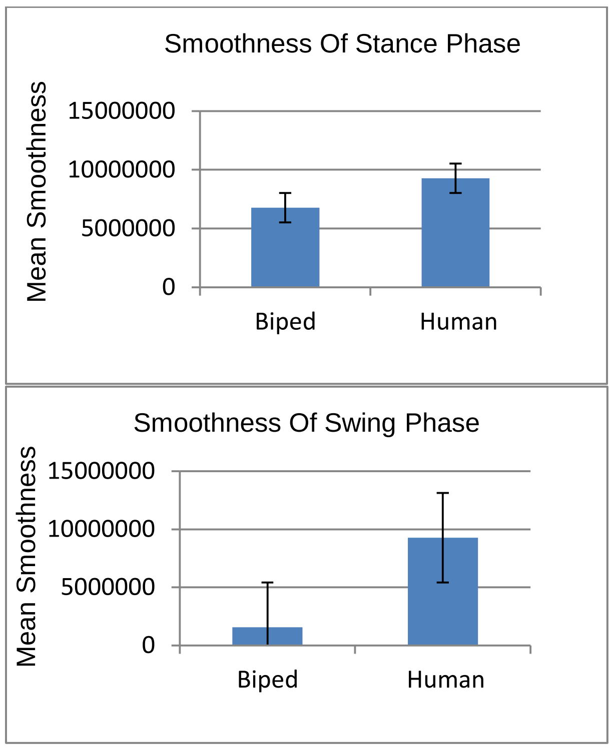


Figure 4.7 Mean Smoothness of stance and swing phase between Human righ leg and biped walking by right leg (SEM).

ii) Biped Between Session Comparison

a) Smoothness during constant speed trials

The smoothness of the trajectory between Sessions 1 and 2 showed lower jerk with increased session, though not statistically significant in both right and left hand as shown in Figure 4.8.

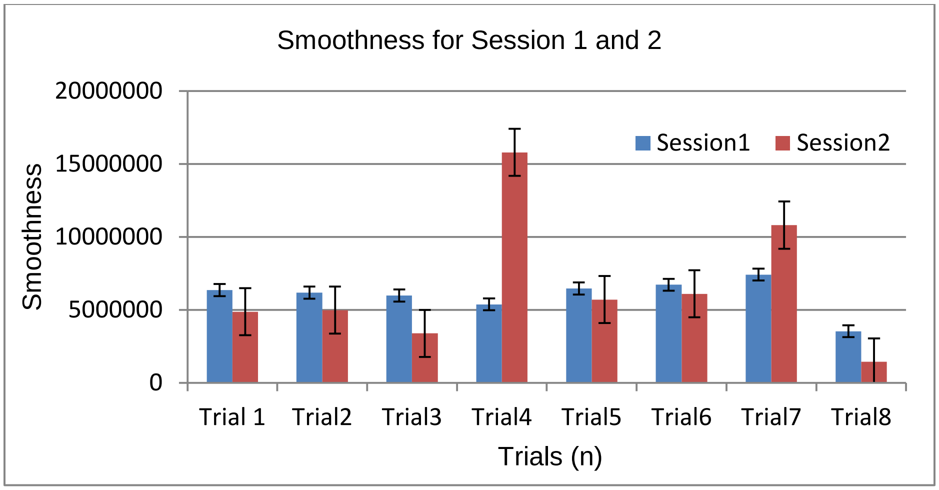


Figure 4.8 Mean Smoothness of biped walking in Sessions 1 and 2 for all trials by right hand (SEM).

b) Effect of change in Speed on jerk during a trial

The smoothness of the trajectory increased from Session 3 to Session 4, though not statistically significant in both right and left hand. The smoothness of the trajectory was further compared with the smoothness of the trajectory from Session 2 to determine the influence of speed on the smoothness. There was no statistically significant difference in the value of smoothness between Sessions 3 and 2 or Sessions 2 and 4 in all trials except trial 8 in right hand, but as shown in Figure 4.9 the average value of smoothness was lower with session even with varying speed of right leg.

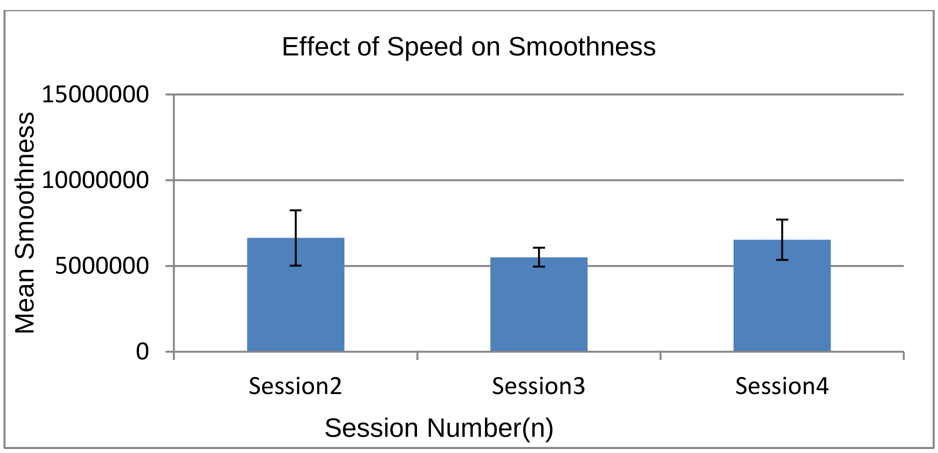


Figure 4.9: Mean Smoothness of biped walking in Session 2, Session 3 and Session 4 for all trials by right hand(SEM).

c) Effect of obstacle with constant speed on jerk

The smoothness of the trajectory showed improvement from Session 5 to Session 6 though it was not statistically significant. The smoothness was further compared with the smoothness from Session 2 to determine the effect of obstacles on the smoothness. There was statistically significant difference in the value of smoothness in trials 3 and 4 in left hand and in trial 8 in right hand between Sessions 2 and 5 but no statistical significance was observed between Sessions 2 and 6 in left hand but there was significant difference observed in in trial 8 in right hand. Though the smoothness of the trajectory decreased with obstacle in Session 5, with practice they were able to learn to navigate obstacles in Session 6 as shown by the decreased jerk as shown in Figure 4.10.

d) Effect of obstacle with varying speed on jerk during trial

The smoothness of the trajectory showed improvement from Session 7 to Session 8 though not statistically significant in both hands. The smoothness was further compared with the smoothness from Session 2 to determine the effect of obstacle and change in speed on the jerk. Statistically significant difference was observed in trial 7 and 8 in right hand and in trial 8 in left hand between Sessions 2 and 7 and no statistically significant difference in the value of smoothness between Sessions 2 and 8 in left and right hand. As shown in Figure 4.11, the average value of smoothness was lower with obstacles but increased with learning.

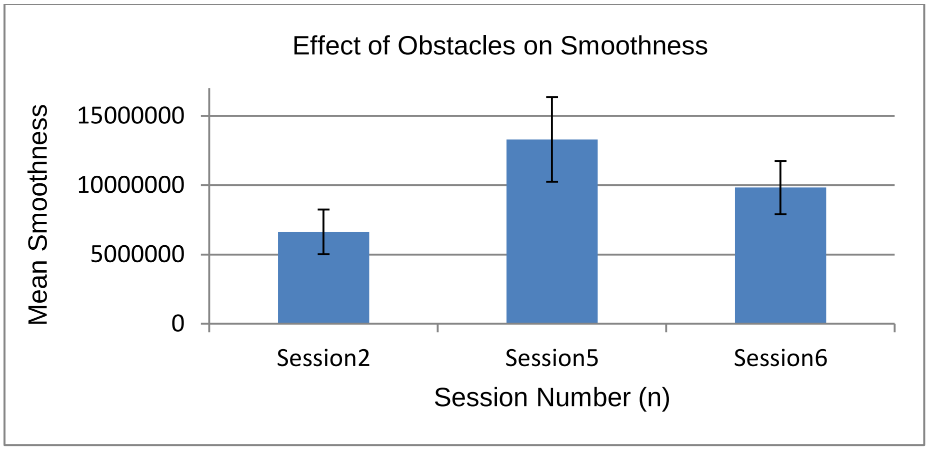


Figure 4.10 Mean Smoothness of biped walking in Sessions 2, 5 and 6 for all trials by right hand (SEM).

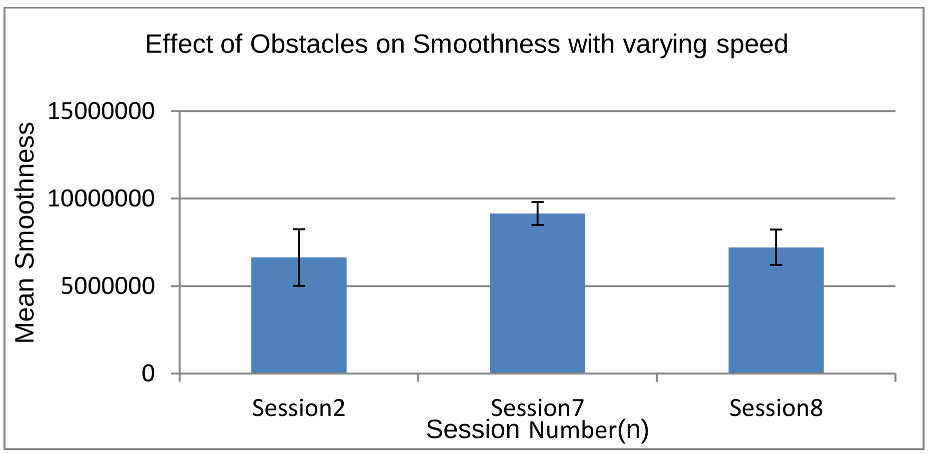


Figure 4.11 Mean Smoothness of biped walking in Sessions 2, 7 and 8 for all trials by right hand (SEM).

4.3.5 Joint Angles

The joint angles computed for the biped walking, human walking, and human walking with flat foot showed that knee and hip angles of biped walking were similar to the angles of human walking with flat foot as shown in Figures 4.12 and 4.13.

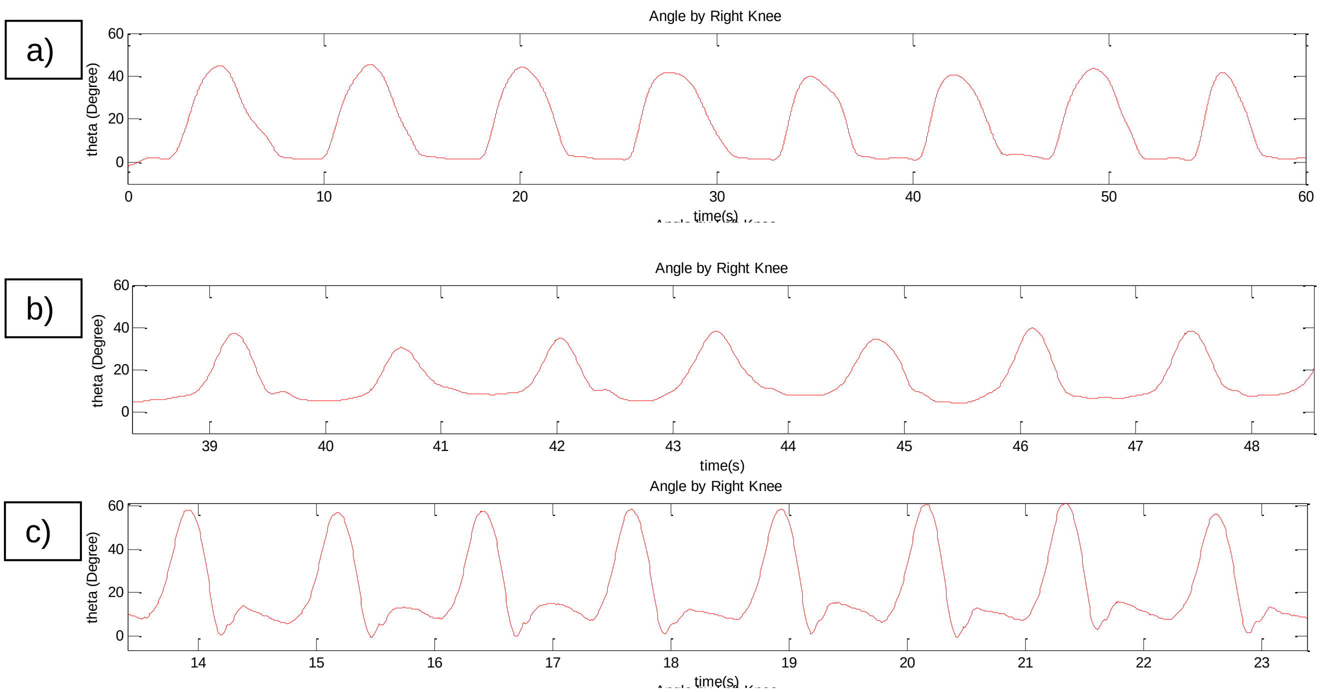


Figure 4.12 Knee Joint angles a) Biped b) Human with flat foot walking c) Typical Human walking

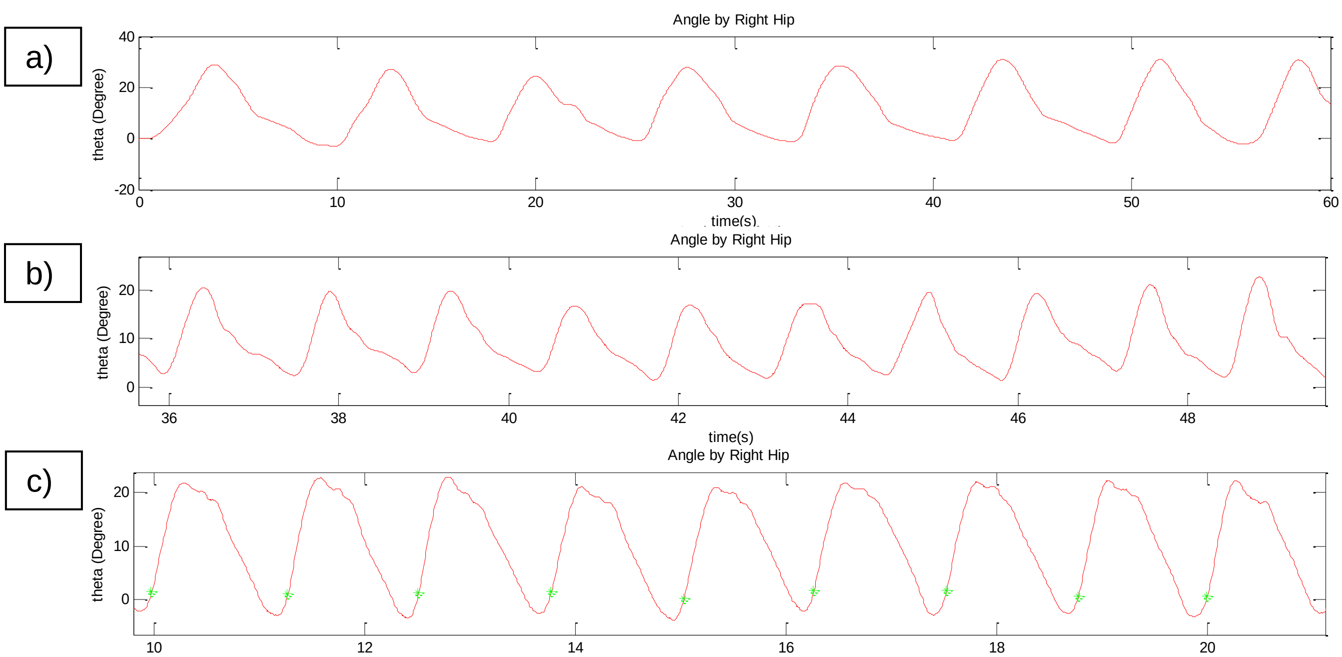


Figure 4.13 Hip Joint angles a) Biped b) Human with flat foot walking c) Typical Human walking

4.3.6 Duty Cycle

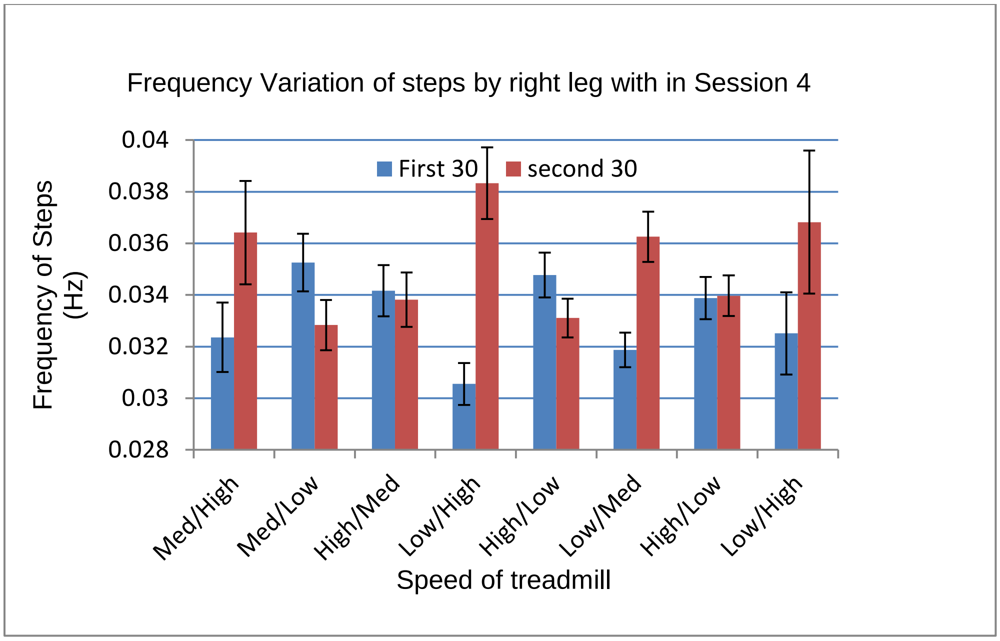
The duty cycle for Session 1 and of biped walking was computed to be an average of 58.02 for the stance phase and 42.08 for the swing phase for Session 1 and 54.18 for the stance phase and 45.80 for the swing phase for all subjects in Session 2.

4.3.7 Frequency Variation

High correlation was observed between speed and frequency of steps for all trials in Session 4 for 5 of the seven subjects in right hand and left hand as shown in Figure 4.14 and high correlation was observed between speed and frequency of steps for all trials among all subjects in Session 3 as shown in Figure 4.15.

4.3.8 Accuracy of the Foot Trajectory in Obstacle Navigation

The percentage of total number of successfully navigated obstacles was evaluated and the results in Figure 4.16 show that there was no significant difference in the success percentage between human and biped walking. Though the average percentage of number of obstacles navigated successfully showed that the biped walking trailed the human walking (Figure 4.16), Session 2 trial 6 and Session 3 trial 4 showed significant differences where the biped walking performed better than human walking as shown in Figure 4.17.



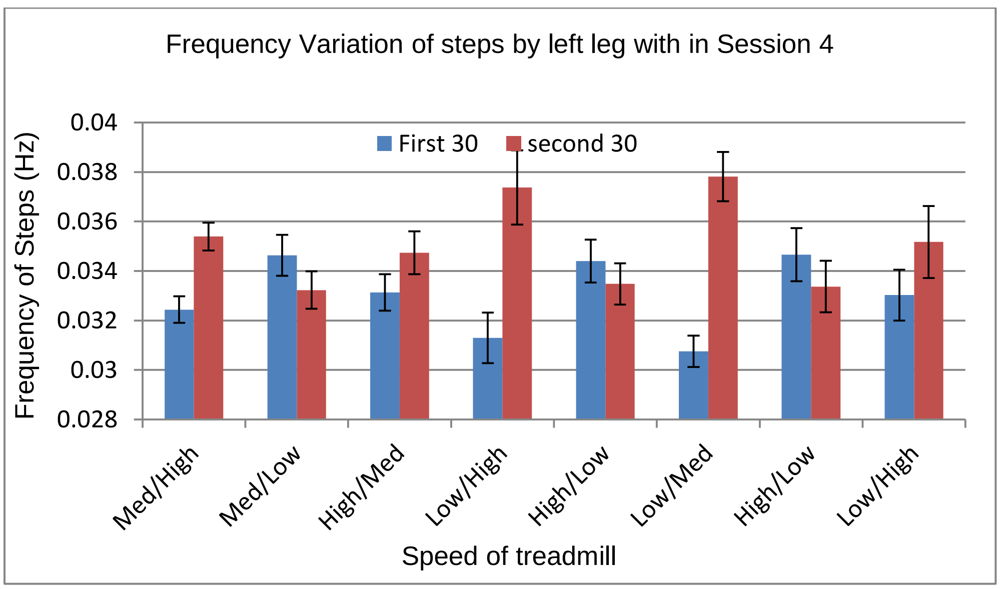
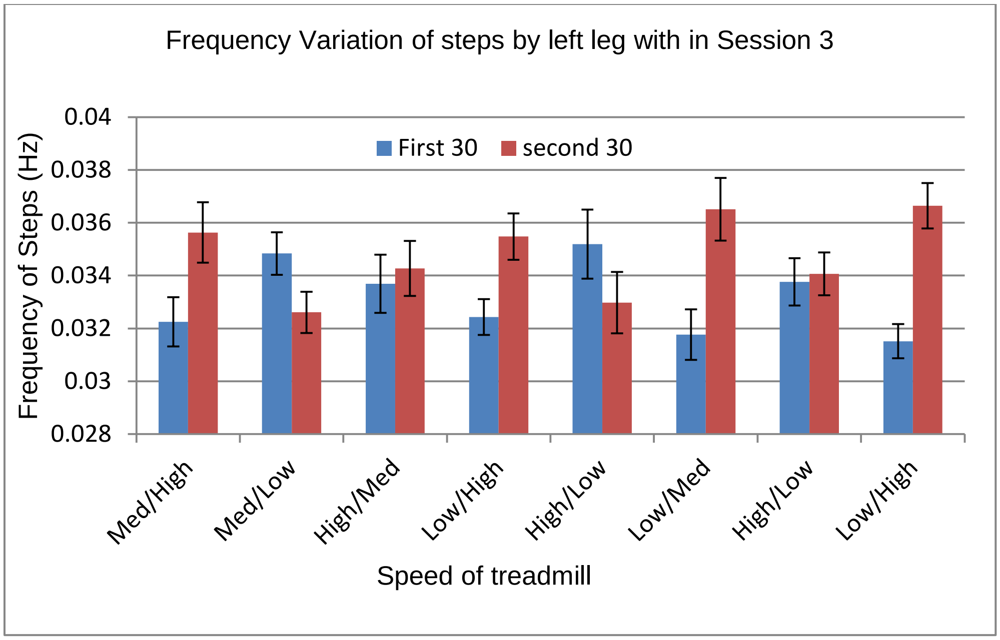


Figure 4.14 Frequency of steps in comparison with the speed of treadmill in Session 4 for right (top) and left (bottom) hand.



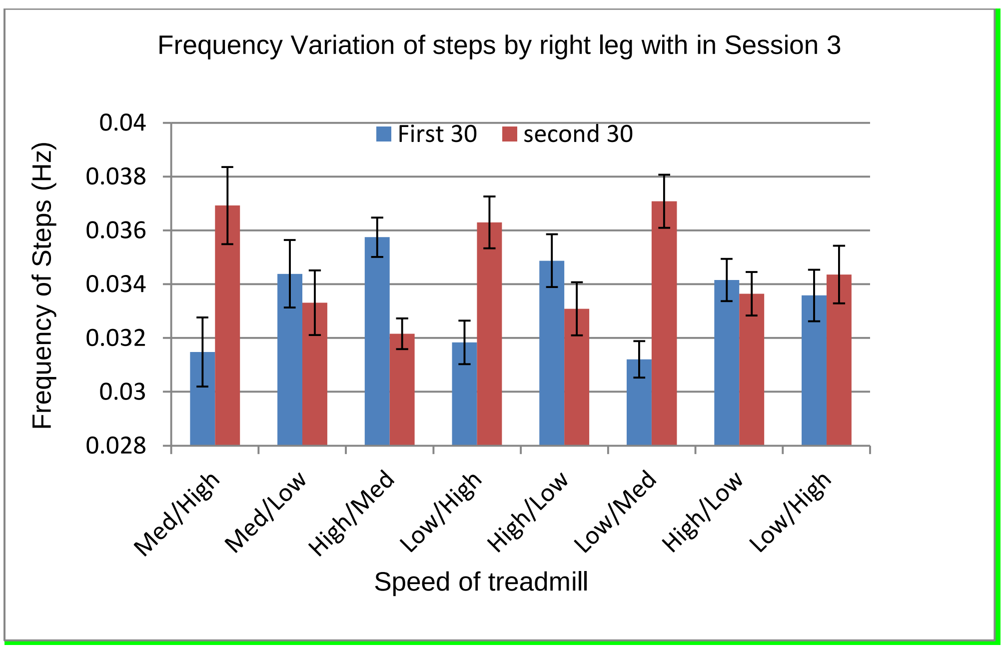


Figure 4.15 Frequency of steps in comparison with the speed of treadmill in Session 3 for right (bottom) and left (top) hand.

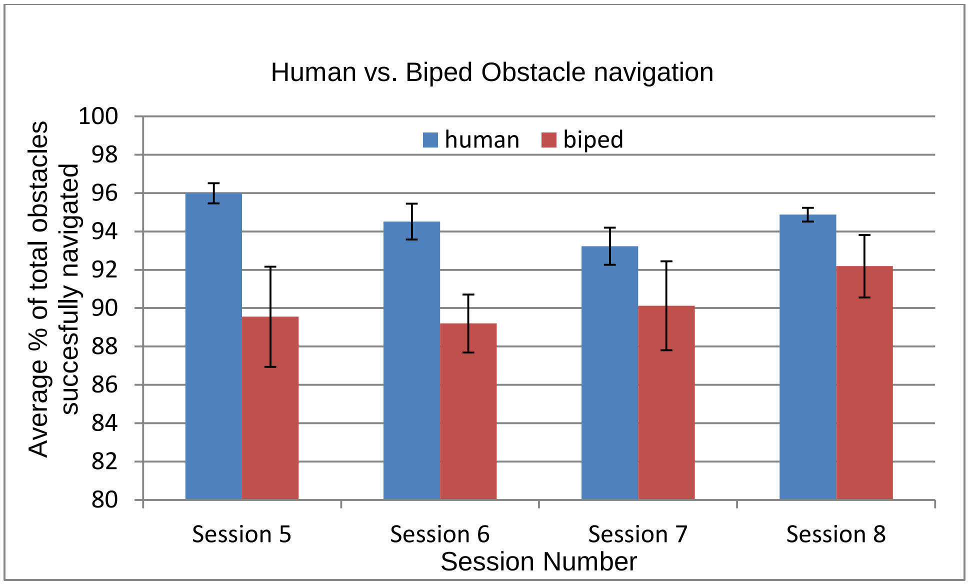


Figure 4.16 Average percentage of obstacles successfully navigated during human walking and biped walking.

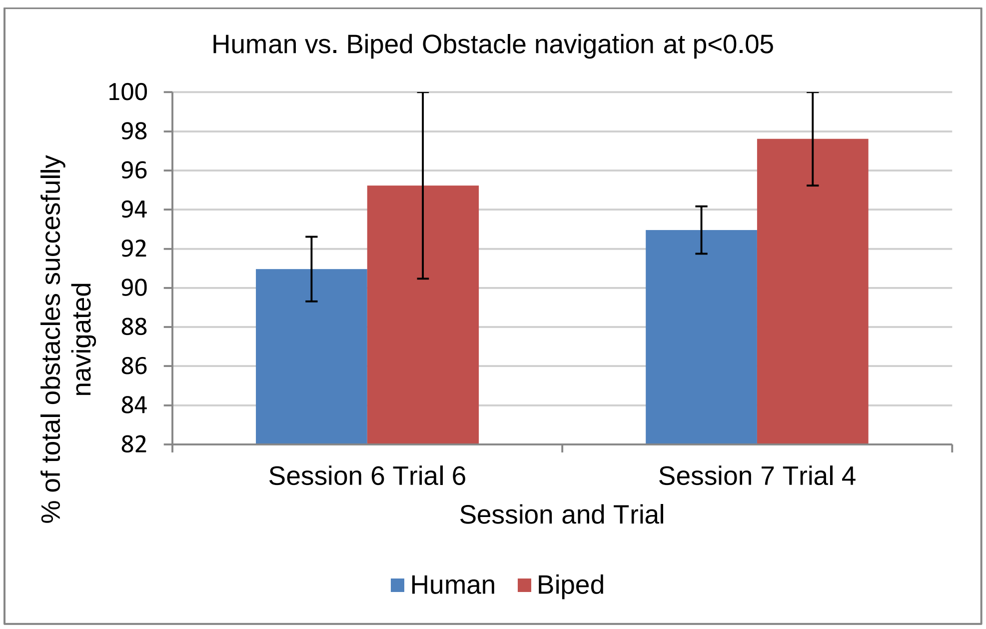


Figure 4.17 Average percentage of obstacle successfully navigated during human walking and biped walking.

4.4 Discussion

A gait pattern is a rhythmic movement of each leg that is influenced by the phase of the other leg to obtain a gait cycle. Hence, an effective exoskeleton control method should be able to reproduce the same pattern of the movement over a period of time, minimize the time variation between the movement as well as coordinate inter limb movement (Pearson,1993 Vaughan, 1992). Inter limb coordination is defined as the ability to assemble and maintain a series of proper relations between the movement of both legs to produce a sequential movement (Forte, 2002).

Differences observed during gait cycles in each of the following four parameters, gait time variability or gait synchrony, inter limb variability (between left and right leg), smoothness of trajectory (with and without obstacles), and gait speed (the correlation between gait speed and gait frequency and between gait speed and stride length) was used as a measure to evaluate the above mentioned criteria for gait cycle. Joint angles and duty cycle were also compared to determine the accuracy of the gait cycle.

The **RMSE** was used to identify the differences between the gait cycles during each trial. The RMSE showed that the trajectories produced by the hand are of similar pattern within a trial signifying that hands can reproduce the movement consistently over a trial period. The RMSE showed that variation between the gait cycles in a given trial was low and that with increased sessions (i.e. more practice), the subject continued to perform better i.e. the RMSE continued to decrease.

Gait time variability, that is, the variability of gait timing in a trial was evaluated. Though it was significantly different from the human walking, the results showed that the variability decreased in successive sessions indicating the subjects improved with successive sessions. An explanation of the observation is that the subjects are still in the learning phase and that with continued practice they would improve with time. Another possibility is that the observed time variability could be due to the decreased speed of operation of the biped. Beauchet et al. have shown that gait time variability was more at lower gait speeds and it improved with higher speeds suggesting walking-speed related changes influence stride time variability. Hence, the variability observed could be because of the low operating treadmill speed for biped walking (Beauchet, 2009). There was, however, no difference in amplitude synchrony implying that the subjects were able to maintain the amplitude of the stride consistently at different speeds.

Inter limb gait Synchrony was measured as a correlation of the time duration and amplitude of steps between right leg and left leg for each subject in all trials. The results showed that both the right and left leg were correlated implying that there was synchrony between the right leg and left leg. The average duty cycle for Sessions 1 and 2 was also observed to be respectively 54% and 58% for the stance phase, and 46% and 42% for the swing phase of the gait cycle. This is almost similar to the normal gait cycle where a typical stance phase was 60 % and swing phase was 40 % (Vaughan, 1992). The slight variation in the biped cycle is attributable to the biped having a flat footed gait lacking therefore initial double stance in the gait cycle and consequently having a comparatively lower stance phase. As such, the biped's whole phase of gait cycle matched well to the normal human gait from the toe off to heel strike. The gait synchrony in time and amplitude and inter limb coordination show that the gait produced by the biped walking is a stable gait.

The stance and swing involve changes in the angles of the hip, knee and foot. The joint angles of the hip and knee showed that the joint angles closely resembled the joint angles of a flat footed human gait. The knee angle did not show flexion during the end of stance phase as there is no initial double stance in a flat footed gait.

Smoothness of trajectory was better than the human trajectory. The smoothness of the trajectory was not influenced by the change in speed; this indicates that the subject was able to transition their speed of walking without influencing the smoothness of the trajectory. The results however showed that there was an effect of obstacles on the smoothness.

Gait Speed: Gait maintains its speed by either varying the frequency of the steps (number of steps taken) or by varying the step length of the gait cycle (Vilas-boas, 2004, Salo, 2012 and Bezodis, 2012). Here, high correlation was observed between the gait speed and frequency of the gait, implying that the subjects were able to vary their gait speed or adjust to the varying speed of the treadmill by varying the number of the steps.

Obstacle Avoidance: The percentage of successful obstacle navigation showed that there was no significant difference between the human walking and biped walking group.

4.5 Conclusion

The results show that the users were able to control through their hands the biped robot's complete gait cycle such that the biped gait compared well with normal human gait. This also implies that the reaction of users of the biped robot to sensory inputs compared well to that of the normal human gait walking group. The users were able to produce synchronous gait rhythm under all walking conditions by the biped robot that was similar to that of normal human gait. As the hand was required to exert only little force to move the biped robot, the task was not strenuous and hence fatigue did not set in as was evident from performance not deteriorating in successive trials. Subjects successfully walked the robot on a treadmill with very little learning period and they were also able to adapt quickly the gait of the robot to varying speed and obstacle conditions ascertaining, therefore, that the biped control was simple and easy. Control task being simple, repetitive, and not strenuous will make the user have in due course absolute control over walking with our biped robot under all walking conditions with more ease. This is the telling and welcome difference from pre-programed exoskeletons.

CHAPTER 5

EFFECT OF COGNITIVE LOAD ON CONTROL OF BIPED WALKING

5.1 Introduction

Every day walking requires only little conscious attention, despite the complex movement and synchronization involved in performing the movement. It is considered a relatively autonomous task (Sparrow, 2002). Smyth et al. described an autonomous process as a task that does not inhibit the capacity to do other processes. In other words, the capacity to perform an autonomous process cannot be inhibited by other processes when performed simultaneously (Smyth, 1996). Studies on older individuals have shown that performing cognitive task while walking impacts the speed of walking (Mulder, 1993). Hence, walking cannot be concluded as an completely autonomic process as per the derivation from the definition by Smyth et. al. This capacity inhibition can be explained as a consequence of competing claim simultaneously over attention resources of the brain by different processes when they are not autonomous and hence require conscious attention (Pashler, 1998, Patel, 2013). Performing two such different 'attention demanding tasks' simultaneously is referred to as dual tasking. Dual task performance causes competition for attention resources and the brain decides on the prioritization of the task (Yogev-Seligmann, Galit, 2010). Deterioration of either motor or cognitive performance when performing a dual task due to task prioritization is known as cognitive motor interference (Plummer-D'Amato, 2012). Based on capacity sharing theory in dual-task control paradigm (Pashler, 1994), the performance of an additional task during walking may alter (impede) gait properties (e.g., speed and variability) or the execution of the cognitive task across domains such as visuomotor processing, verbal fluency, working memory etc. (Dubost, 2008). In particular, two closely related cognitive domain’s, executive function and attention influence gait (Beauchet, 2005, Beauchet, 2002). It is especially relevant to our study because both limb co-ordination and dual-tasking controls are processed centrally in the nervous system. Moreover, attention is an important mediator for motor coordination. Thus, it seems inducing cognitive load on motor tasks such as walking would affect motor coordination.

In the context of walking, prioritization has been said to be given to gait stability over other tasks. It has been shown in many studies with healthy adults, where they give priority to stability and balance of the gait while performing cognitive task when no explicit instruction on prioritization is given. This “posture-first” strategy, a concept originally introduced by Shumway-Cook et al in 1997, suggests that balance, stability and less gait variability gets priority (Shumway-Cook, 1997). On the contrary, studies have shown that dual task increases the gait variability in older adults as compared to single task while such gait variability was not observed in young adults suggesting that young adults are able to compensate for the cognitive load during task prioritization (Al-Yahya, 2011, Holtzer, 2012, Schaefer, 2014), though reduced speed in young and old adults was observed with cognitive load. Yogev-Seligmann et al. recently expanded this model of ‘posture first’ strategy to include “cognitive first” strategy as an equal substitute (Yogev-Seligmann, 2012). This new strategy postulates that postural reserve and hazard estimation is a significant intrinsic factors contributing to the selection of the task prioritization strategy. This was validated by Liston et al. where they demonstrated that older adults did not prioritize postural tasks while dual tasking, in contrast to younger adults who did adhere to the “posture first” paradigm (Liston, 2014). Though, posture first strategy might be more appropriate from the ecologic perspective as it ensures safety.

The aim here is to study the effect of cognitive load on the user’s ability to control the walking of the biped. Cognitive tasks such as serial subtraction and word list generation task have shown to have an effect on the gait speed in young adults and on gait time and amplitude variability in older adults suggesting that locomotion shares central processing resources with semantic memory task and working memory task. The study described here used serial subtraction, a working memory task to evaluate the cognitive motor interference in biped walking control. The effect of performing a cognitive task on biped walking was evaluated by comparing the time and amplitude synchrony and RMSE with biped walking with no cognitive load and the cognitive performance was evaluated by comparing the cognitive task performed by human walking with biped walking.

This Chapter addresses Specific aim 3.

5.2 Design and Methodology

5.2.1 Experimental Setup

The experimental setup for biped walking and human walking are same as the experimental setup described in Figure 4.2.1.

Subject population

The study included 14 subjects divided equally into two groups, with one group controlling the biped on the treadmill and the other group performing normal human walking on the treadmill. The subjects were randomly assigned to the groups. All subjects were able bodied subjects between ages of 21 and 35 with fully functional upper and lower extremities. The groups were matched based on age, gender, educational qualification and socio-economic status. Exclusion criteria included any disability to the upper or lower extremities, inability to perform normal gait or any cognitive disability. The study was approved by the NJIT IRB and the experiment was performed with the subjects' consent.

5.2.2 Experiment

The experiment was designed to measure kinematics, synchrony, smoothness under varying walking conditions such as speed and obstacles when provided with cognitive load. The subjects who participated in the experiment were divided into two major groups: Dual task (Biped and Human walking groups) and Single Task (Biped and Human walking groups). The biped control group controlled the gait of the biped on a treadmill using their hand movements while the human walking group performed normal gait on a treadmill. Dual task groups performed a cognitive task of counting backward by seven from a random three digit number below 500 during each trial. All subjects started from the same number in a trial but the starting number was varied between trials. The single task group is the same group from Chapter 4. The study consisted of a total of eight sessions. Each session comprised of eight trials. Each trial was for duration of one minute followed by a thirty second rest. The speed of the treadmill during each trial was as shown in Table 4.1. The speed was varied after thirty seconds in Sessions 3, 4, 7 & 8. Sessions 5-8 included users stepping over an obstacle. The number of obstacles presented to each leg was the same, though not in the same sequence. Twice the number of obstacles was included for the biped group in order to offset for their reduced speed. Both groups avoided obstacles with only one foot during a trial and the obstacle avoidance was alternated between both feet in the successive trials.

The dual task biped walking and human walking groups performed the task of walking as described in Section 4.2.2. In addition, both these groups performed the cognitive task of counting backwards by 7 from a three digit number below 500. The data was processed using MATLAB and all statistical analysis was performed using SPSS.

5.2.3 Data Analysis

Horizontal and vertical trajectories collected at 120 Hz of the ankle, hip and knee were filtered using an effective 4th order, zero-lag Butterworth low-pass filter. The cutoff frequency was determined using the power spectrum. The filtered data were used for further analysis.

5.2.3.1 Foot Trajectory.

i. Root Mean Square error:

The RMSE was determined as a measure to evaluate the differences in gait cycle between trials for Sessions 1 and 2, in other words, to evaluate the synchrony between steps. The RMSE was calculated as described in Section 4.2.3.1. The RMSE was also compared with gait cycle of the single task group (biped control with no cognition task).

Mann Whitney U test was used to compare between single task group and dual group.

In order to identify possible learning, we compared Sessions 1 and 2 using Wilcoxson non parametric measure. Further, the influence of speed was analyzed to determine, if the speed affected RMSE.

ii. Amplitude and Frequency Synchrony

The rhythmicity of the movement was evaluated using amplitude and frequency synchrony. The time taken for a gait cycle was computed for all the gait cycles in a trial for Sessions 1 and 2.

a) Biped walking with and without cognition task in Sessions 1 and 2

The amplitude and frequency synchrony of dual task group was compared with that of single task group.

b) Biped walking between Sessions 1 and 2

The amplitude and frequency synchrony in the biped groups were also compared between Sessions 1 and 2. As synchrony is a measure of differences in reproducible actions, synchrony in gait amplitude and frequency was therefore used as a measure to determine the improvement in periodicity of the movement with practice. Paired sample t-test was used to compare the values of time synchrony between Sessions 1 and 2 for biped walking. Pearson's correlation was used to correlate the left and right leg synchrony. The standard deviations of amplitude and frequency synchrony were computed to determine the variation in the amplitude and frequency of the gait cycles in a trial.

5.2.3.2 Smoothness of the Foot Trajectory. The smoothness of trajectory was calculated as described in Section 4.2.3.2

I. Constant Speed

The Smoothness of Sessions 1 and 2 of biped walking with cognitive task was compared with that of biped walking without cognitive task to determine the influence of cognitive task on trajectory smoothness.

II. Effect of speed change in a trial on Smoothness between Dual task and Single Task

The Smoothness of Sessions 3 and 4 of biped walking with cognitive task was compared with that of biped walking without cognitive task to determine the influence of cognitive task on trajectory smoothness when speed is varied.

III. Effect of obstacle navigation with constant speed in a trial on Smoothness between Dual task and Single Task

The Smoothness of trajectory in Sessions 5 and 6 of biped walking with cognitive task was compared with that of biped walking without cognitive task to determine the influence of cognitive task on trajectory smoothness due to obstacle navigation.

IV. Effect of obstacle navigation with varying speed in a trial on Smoothness between Dual task and Single Task

The Smoothness of trajectory of Sessions 7 and 8 of biped walking with cognitive task was compared with that of biped walking without cognitive task to determine the influence of cognitive task on trajectory smoothness due to obstacles and speed. Man Whitney U test was used to compare the values of jerk between Human and biped as the data was not normal and Wilcoxon Paired test was used to compare values between different sessions.

5.2.3.3 Accuracy of the Foot Trajectory in Obstacle Navigation. The number of obstacles navigated and missed was determined for every trial for all subjects. The Percentage of obstacles successfully navigated was evaluated as a measure to determine the accuracy of obstacle navigation, which is an indicator of efficiency of the foot trajectory. The percentage of obstacles successfully navigated by Biped group was compared to Human group in conjunction with presence or absence of a cognitive task. We applied a Kruskal-Wallis ANOVA with one factor at four levels to compare obstacle navigation performance in a) Human group during cognitive load, b) Human group with no cognitive load, c) Biped group with cognitive load and d) Biped group with no cognitive load.

5.2.3.4 Cognitive Load. The cognitive load's impact on controlling the biped was evaluated by comparing the number of responses in serial subtraction task in 60s with that of the number of responses given during normal human walking. Independent samples t-test was used to calculate the statistical difference between the biped walking group and human walking group.

A correlation of the number of responses for all trials for all sessions for human and biped group was performed to evaluate the effect of speed and obstacles while human walking and biped walking on cognitive task.

5.3 Results

5.3.1 Foot Trajectory

The foot trajectory between the human ground walking and biped walking while performing cognitive task was found to be very similar as shown in Figure 5.1. The x and y trajectory plot while walking on the treadmill shows that the pattern of the trajectory is similar to normal human walking.

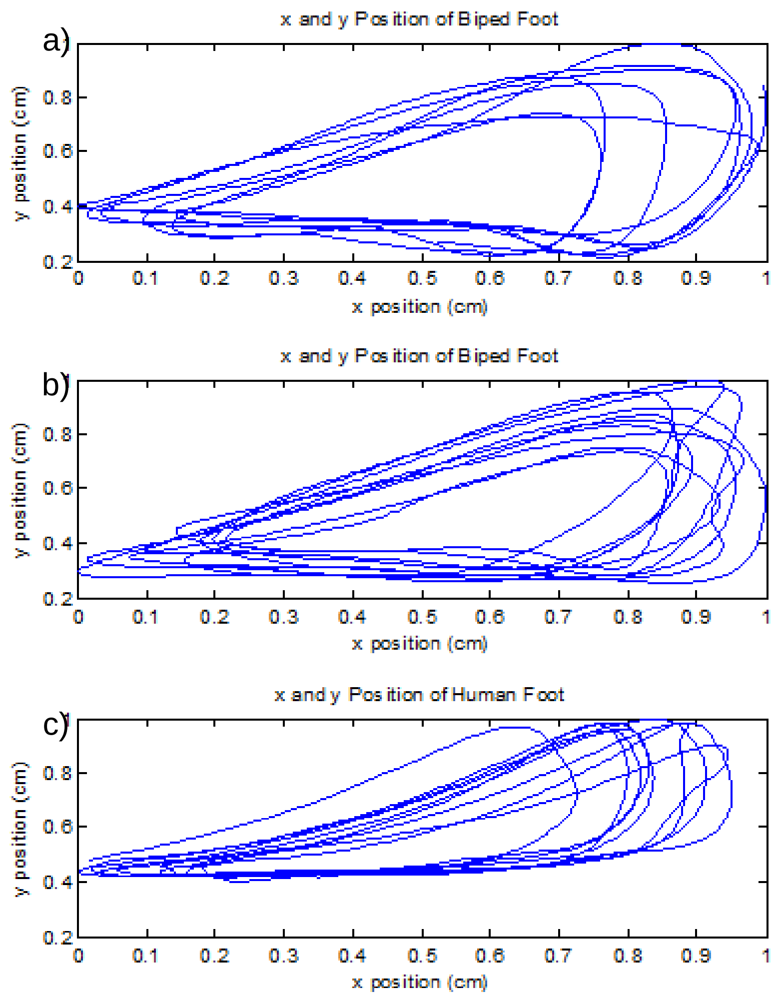


Figure 5.1 Foot trajectory of a) Biped (single task) b) Biped (Dual task) c) Human (Dual task) on treadmill.

5.3.2 Synchrony

The time between steps of a trial between the right and left leg for Sessions 1 and 2 was found to be highly correlated for each subject. This implies that the left and right leg had the same time duration for steps for a subject during a trial (Figure 5.4). Also, time synchrony between biped walking with and without cognition (i.e., cognitive task/load) showed that there was no statistical difference between the synchronicity of biped walking with and without cognitive load (Figures 5.2 and 5.3). In fact, Session 1 showed that with cognitive load group had better synchrony that without cognitive load group. Synchrony between Sessions 1 and 2 showed no statistical difference.

The horizontal and vertical amplitude synchrony results showed no difference between biped walking between single task and dual task groups. The correlation between horizontal synchrony of the right leg with the left leg showed high correlation. This shows the step length of right leg and left leg was same. The time and amplitude correlation between the legs shows inter limb coordination.

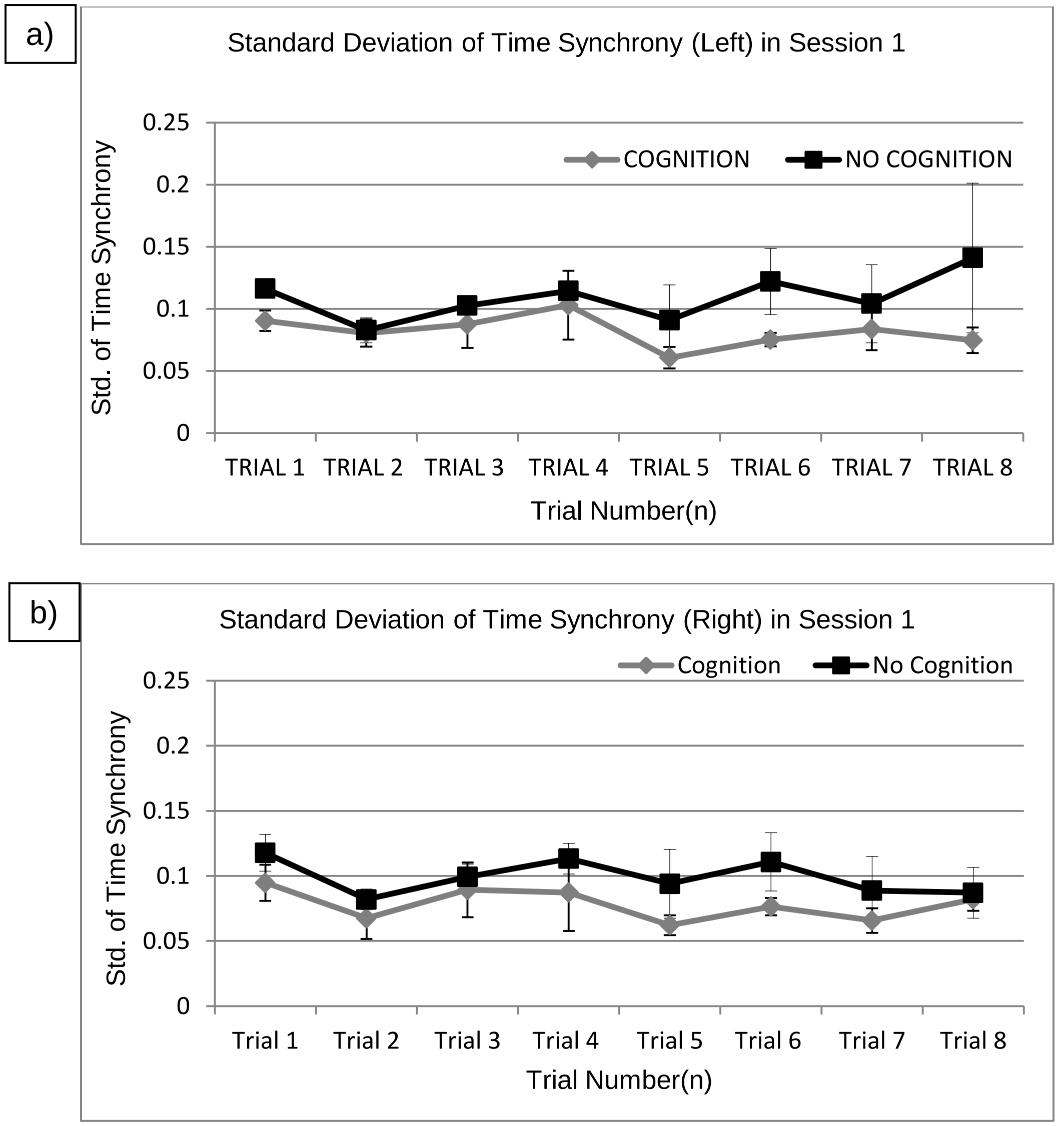
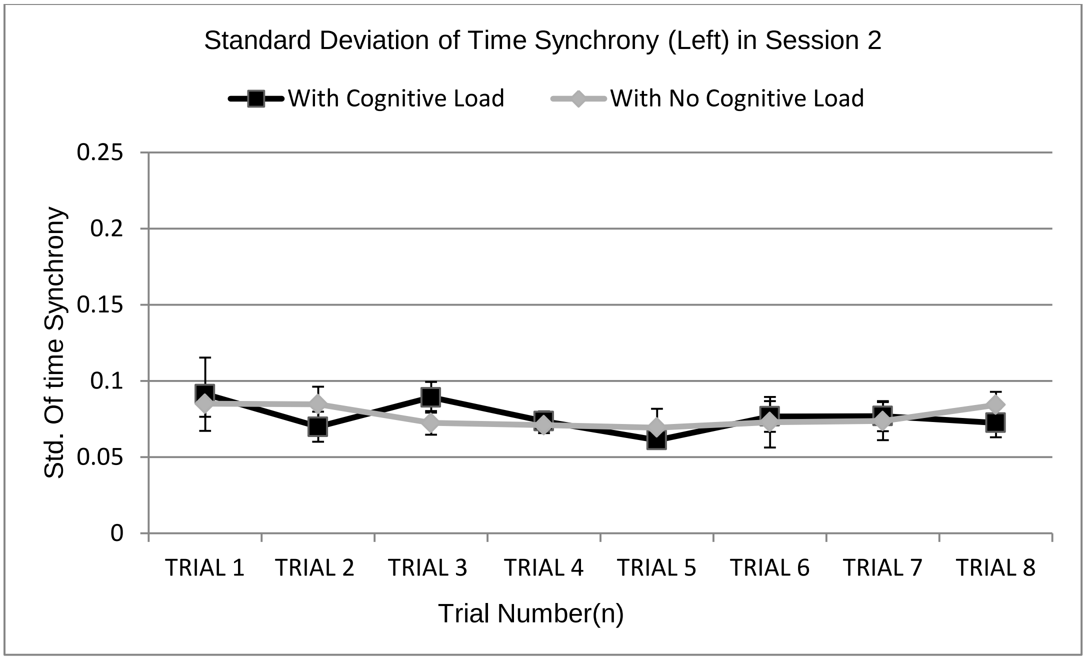


Figure 5.2 Standard deviation (SEM) in time synchrony in Session 1 for a) left and b) right leg of biped with and without cognition.

a)



b)

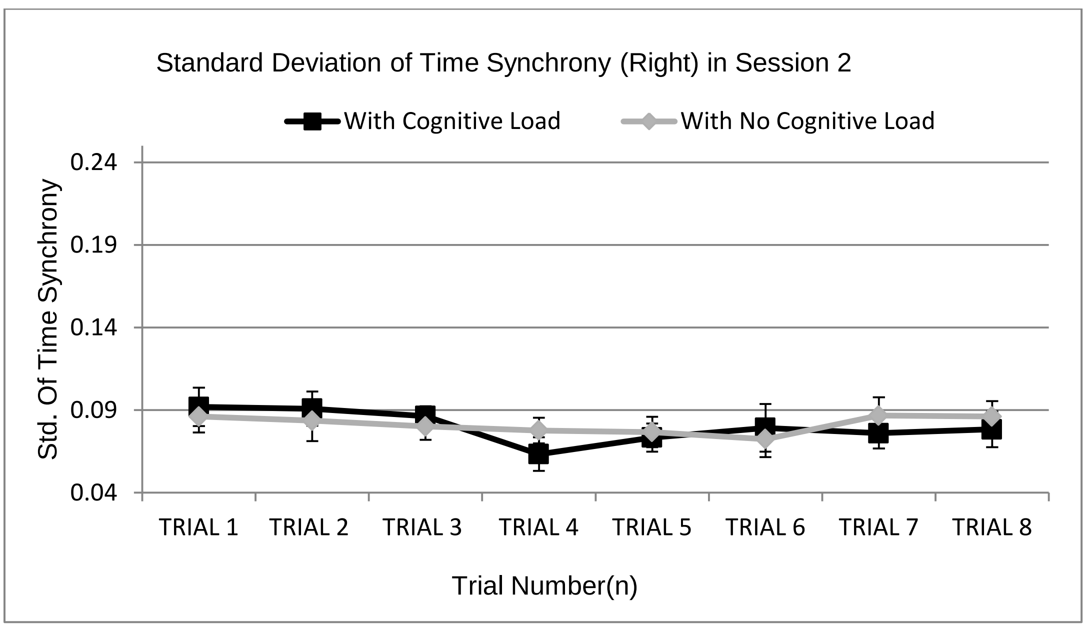


Figure 5.3 Standard deviation (SEM) in time synchrony in Session 2 for a) left leg and b) right leg of biped with and without cognition.

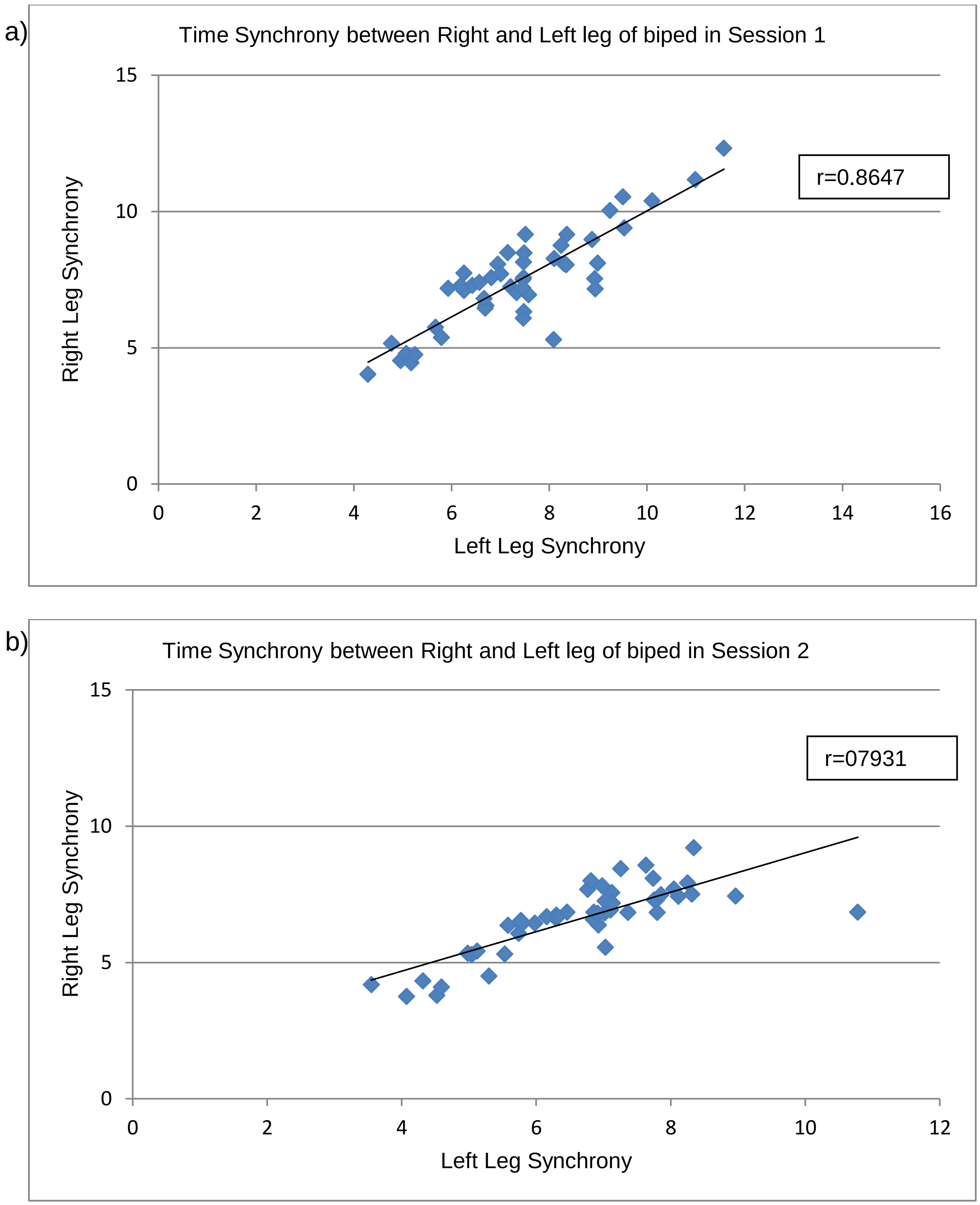
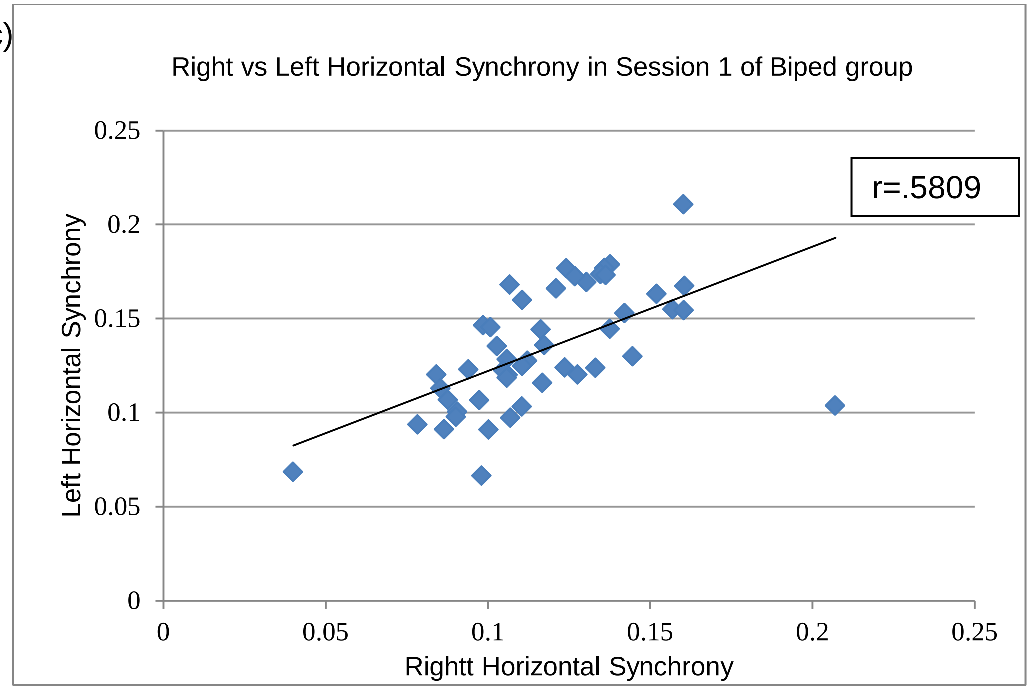


Figure 5.4 Correlation between right leg and left leg time synchrony of biped while performing cognitive task in a) Session 1 b) Session 2.

c)



d)

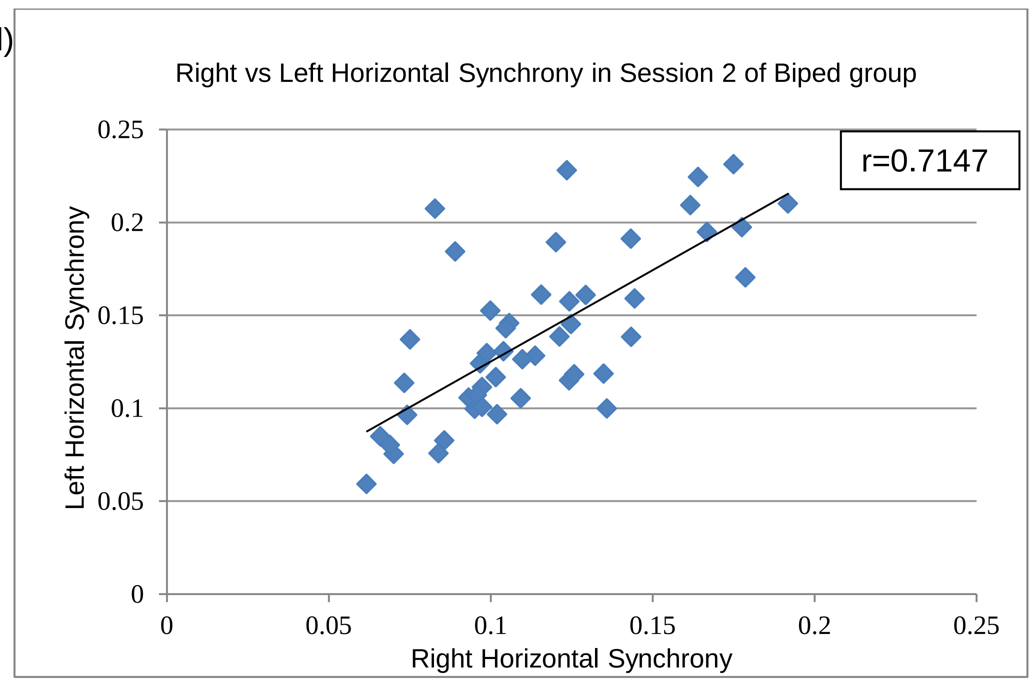


Figure 5.4 (continued) Correlation between right leg and left leg amplitude synchrony of biped while performing cognitive task in c) Session 1 d) Session 2.

5.3.3 Root Mean Square Error

The RMSE showed no significant differences between the dual task group (group with cognitive load) and single task group (group with no cognitive load) in Sessions 1 and 2 across all trials for both legs as shown in Figure 5.5. There was no significant difference between Sessions 1 and 2. Also, there was no effect of speed on the RMSE.

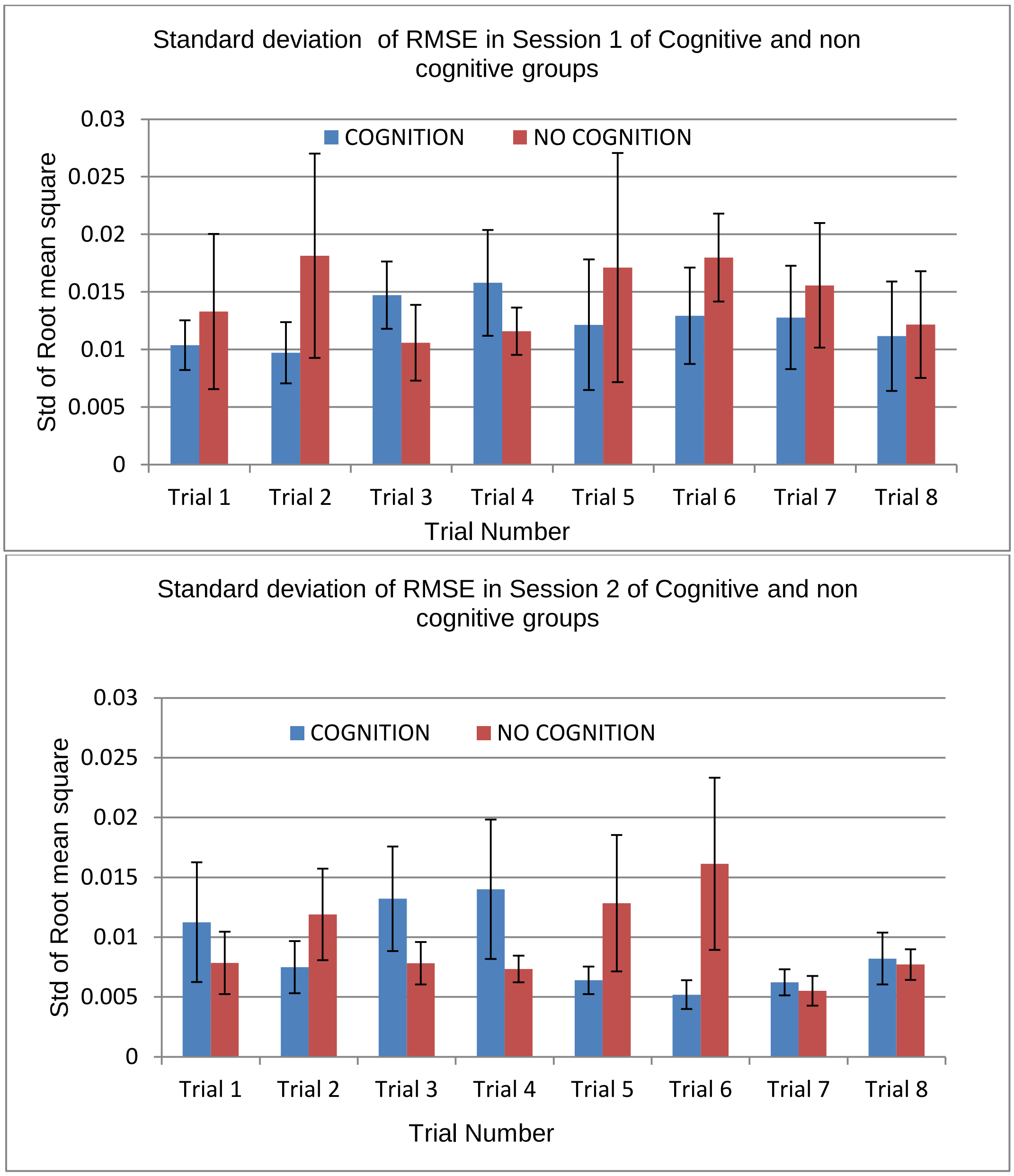


Figure 5.5 Standard deviation Root mean square error between with cognitive load and with no cognitive load group for a) Session 1 (top) and b) Session 2 (bottom) for right leg.

5.3.4 Smoothness of the Foot Trajectory

Smoothness across sessions did not show any difference between the single task group and dual task group as shown in Figure 5.6

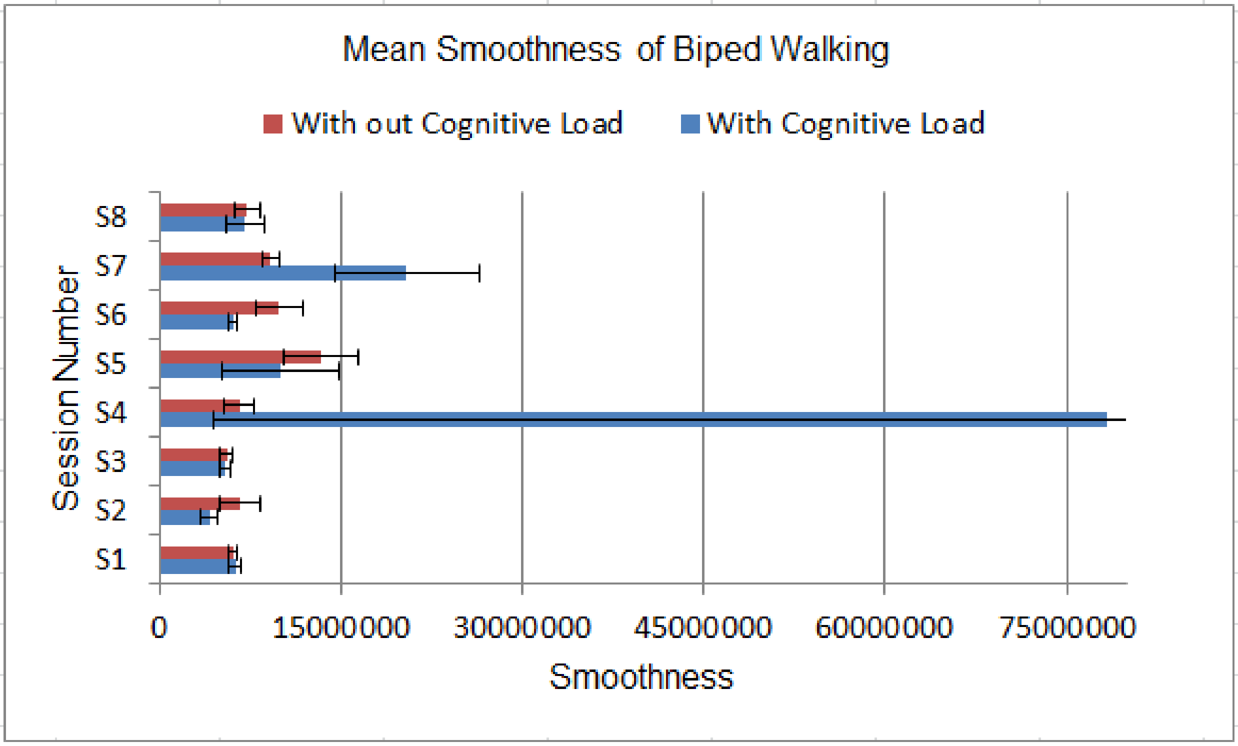


Figure 5.6 Mean Smoothness of biped walking with and without cognitive load for all sessions by right leg (SEM).

5.3.5 Number of Responses

A correlation of the number of responses for all trials for all sessions for human walking and biped walking group was performed. A high correlation was observed in the serial subtraction task performed by the humans and biped walking across all trials and subjects as shown in Figure 5.7.

The results for serial subtraction showed that there is no statistical difference between the human walking and biped walking in the number of responses given in 60s across all sessions and subjects, though the average number of responses was less for the biped group with cognitive load in all sessions as shown in Figure 5.8.

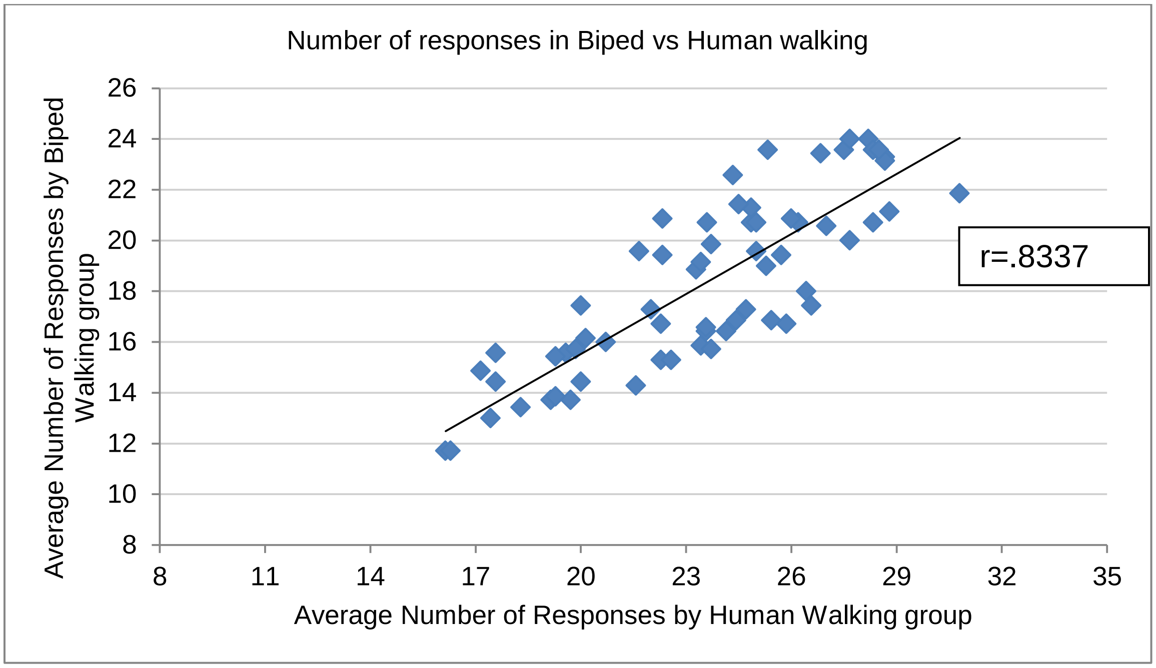


Figure 5.7 Correlation between average number of responses by the biped walking group and human walking group for all trials in all sessions.

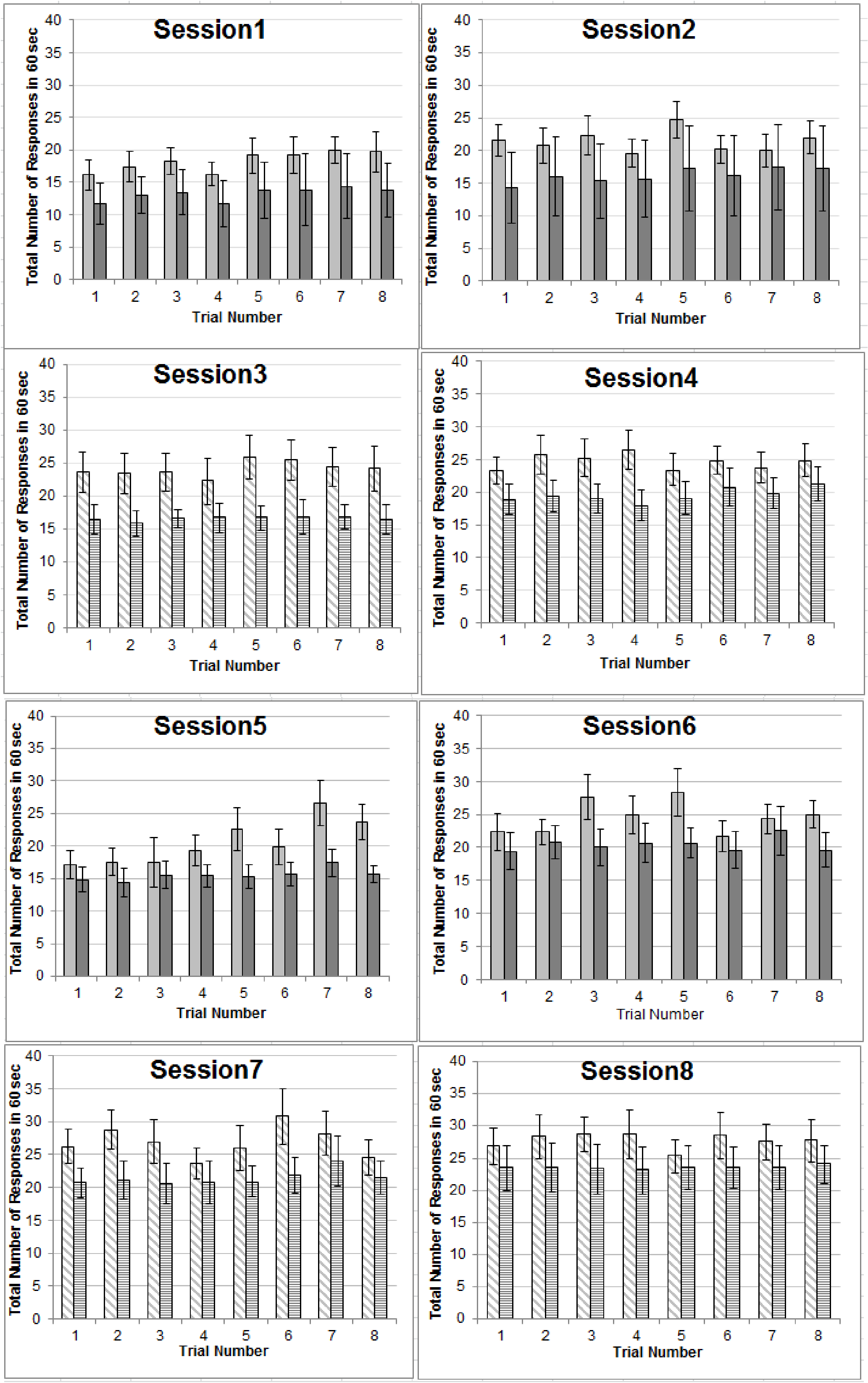
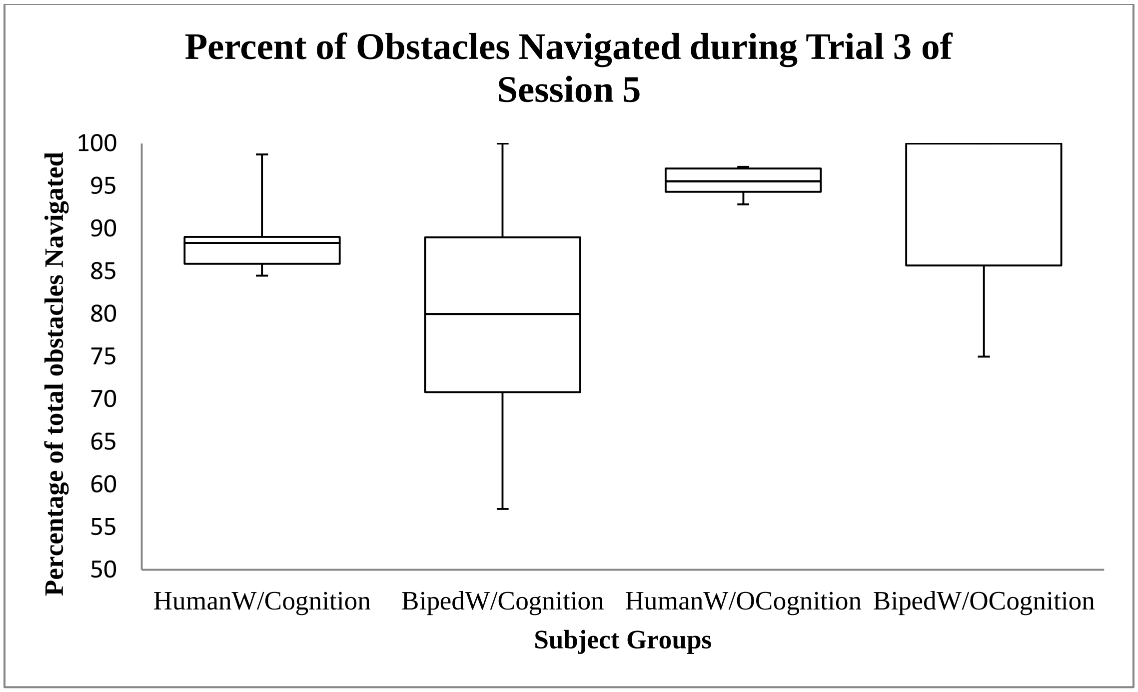


Figure 5.8 Total number of responses during serial subtraction in 60s by Human and biped walking in all sessions. Solid plots show sessions without Speed Change.

5.3.6 Accuracy of the Foot Trajectory in Obstacle Navigation

The percentage of total number of successfully navigated obstacles was evaluated. Significant difference between four groups (Single task (Human walking and biped walking) and Dual task (Human walking and biped walking)) in Trial 3 (trial with highest speed) of Session 5 and Trial 4 (slowest speed) of Session 7. By further performing post-hoc analysis using Mann-Whitney U test on trial 3 of 5th session we observed significant difference between Human group with cognition and Human group without cognition with better navigation observed during the absence of cognitive load. Likewise, Biped group during cognition and Biped group without cognition showed a similar effect with absence of cognition resulting in better navigation among subjects, however, failed to pass the statistical threshold possibly due to small sample size. Next, post-hoc analysis using Mann-Whitney U test on trial 4 of 7 th session showed significant difference between Human group without cognition and Biped group without cognition.



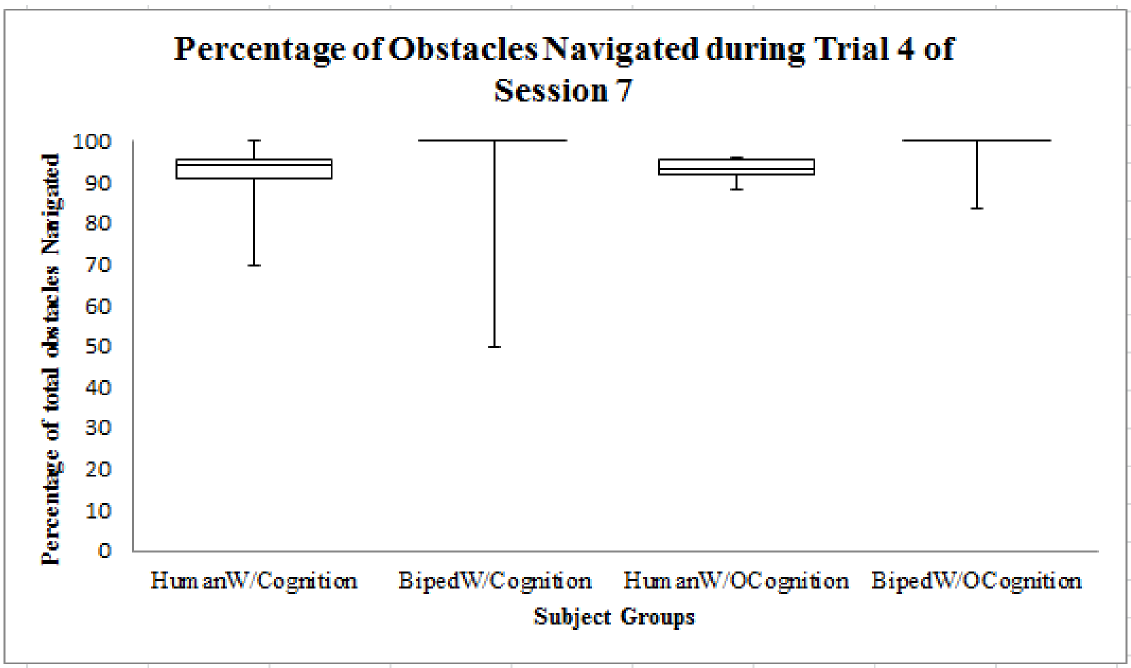


Figure 5.9 Percentage of Obstacle navigated in Sessions 5 (top) and 7(bottom).

5.4 Discussion

The results show that there is no difference in the time and amplitude synchrony or the root mean square error or the smoothness of the trajectory of the gait cycle produced by the biped walking with the cognitive load when compared with the gait produced by the biped walking with no cognitive load in the conditions without obstacles. Increased gait variability have shown to result in decline in balance and leading to fall (Young, 2011, Hausdorff,2001, Maki, 1997), hence the results showing that time and amplitude synchrony and RMSE showing comparable results with Biped walking with no cognitive load shows that the cognitive load did not affect the gait synchrony. Further, study by Paterson et al. have shown that reduced inter limb co-ordination is an early indicator for falling and that lower inter limb co-ordination resulted in falls even when change in other measures of physical function, balance and gait were not present (Paterson, 2011). Our results show that there was high correlation in time and amplitude synchrony of the gait between the right and left leg of the biped walking even with cognitive load showing that there was lower inter limb variability, hence, providing the user with more optimal gait performance.

The smoothness is a measure to evaluate the jerk of a trajectory. A planned trajectory will have lower jerk (Hreljac, 2000). Smoothness comparison between biped walking with and without cognitive load showed no difference in the smoothness. Thus, planning of the trajectory was not affected by the cognitive load.

The number of responses to the serial subtraction for the dual task group was less than human walking group for all trials suggesting that there was cognitive load due to biped control. The results show that subjects prioritized the biped walking task over the cognitive task of serial subtraction. This implies that the cognitive task does not affect the gait trajectory and that like a typical dual task involving gait, the subjects prioritized the walking over the cognitive task even though the task of controlling the gait was performed by the hands. This result is in accordance with the finding of Bloem et al. who postulated that both young and old healthy adults prioritized gait stability over success of secondary task when no specific instruction is given on the prioritization of the task.

Also, the average number of responses to subtraction by seven showed a high correlation between the human walking and biped walking suggesting that though the cognitive load was higher for biped walking than the human walking but the effect of speed variations between trails and in trials and obstacles on the cognitive task was similar to that of human walking. This shows that the prioritization of the task while performing a dual task was very similar to young adults performing treadmill.

The mean obstacle navigation performance was in fact better in Biped group as compared to Human group however the subject variability is much greater in Biped as seen by interquartile range shown in Figure 5.9. Further, the absence of difference due to cognitive load between these groups beyond Session 5 raises the possibility that this variability in biped walkers may be due to the fact that they encountered a novel task; and may have become proficient in the subsequent sessions.

The results show that hands can produce gait trajectories similar to that of the typical human gait when provided with the haptic feedback. The algorithm was able to translate the trajectories produced by the hand to joint angles in the sagittal plane for further control of the joint angles of biped. The evaluation of the biped walking using hand control showed that the control was intuitive, easy to learn and required very little cognitive attention, unlike BCI which requires extensive training and attention to perform the simplest task and currently cannot translate the EEG signals to kinematics.

The study further showed that the prioritization of the task by the user performing obstacle navigation and serial subtraction while controlling the biped walking was similar to human walking leading to the theory that there might be some similarities in higher cortical planning strategies between the hand walking and human walking. Although the current study did not prove that there are similar motor control strategies between the human walking and biped walking, the similarity between the human walking and hand walking raises the question of whether they share the same motor control strategies.

5.5 Conclusion

The biped walking kinematics, synchrony and smoothness was not affected by the cognitive load, though the performance of the cognitive task was comparatively lower than normal human walking. This proves that with increased cognitive load the priority is given to the walking of the biped over the cognitive task. This is very similar to our young adults walking, where adults perform varying cognitive task while walking and still be able to perform the task of walking with smoothness, maintaining less variability in gait.

CHAPTER 6

CONCLUSION

Currently, wheelchairs provide mobility to the user, and any alternate device would have to be very effective in giving complete independence and mobility in order to replace wheelchairs. Current exoskeletons, though a big leap towards giving users mobility, it falls short in granting the user the independence of control over their stride length, height and frequency.

An alternate control mechanisms using brain computer interface to translate intensions to provide control over stride length, height and frequency is a cognitively intensive and, is not efficient or intuitive.

The control mechanism described in this dissertation will, on the other hand, provide users with a realistic, intuitive, real time control of the leg with minimum cognitive load.

The control mechanism was developed using an admittance control paradigm where trajectories from hands were employed as surrogates to control the foot trajectory of the biped. The algorithm was able to convert the forces applied by the user into Cartesian positions. The inverse kinematics converted those Cartesian positions into hip and knee angle that resemble those of human gait. The control also provided haptic feedback to the hands to produce the required gait trajectory.

The control mechanism was evaluated for the synchronicity in time and amplitude, inter limb coordination, smoothness, and gait kinematics in varying speeds and with obstacles, and with cognitive task.

The results show that the complete gait cycle produced by the control algorithm is rhythmic and follows the kinematics of normal human gait cycle. Naïve subjects were able to successfully control the gait of the biped by applying small forces in the intended direction and were able to control the walking of the biped on a treadmill with little to no learning period. They produced consistent ankle trajectories with changes in treadmill speed, in the presence of obstacles, and while also performing a working memory cognitive task (counting backwards by 7). The evaluation proves that hands could be used as an alternate control mechanism to control the gait of an exoskeleton.

Future Directions

The limitations of the study is that it was evaluated only in the sagittal plane; further exploration should involve adding other degrees of freedom for turning and abduction/adduction to evaluate if the user can control the gait of the biped with the added degrees of freedom.

Future enhancement to the control mechanism should include toe off and heel strike at the ankle to provide the torque for the swing cycle. It would also need to address balance of the exoskeleton while walking. Future studies should include enhancing the algorithm to accommodate for higher speeds and further test with full scale exoskeletons.

APPENDIX A

COGNITIVE RESPONSES AND OBSTACLES NAVIGATED

The data and plots for cognitive responses obstacle navigated are shown below.

Table A.1 Average Number of Cognitive Responses from Human Walking Group During Each Trial of Each Session.

Session	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 6	Trial 7	Trial 8
1	16.14286	17.42857	18.28571	16.28571	19.14286	19.28571	20	19.71429
2	21.57143	20.71429	22.28571	19.57143	24.71429	20.14286	20	22
3	23.57143	23.42857	23.57143	22.28571	25.85714	25.42857	24.42857	24.14286
4	23.28571	25.71429	25.28571	26.42857	23.42857	24.85714	23.71429	24.85714
5	17.14286	17.57143	17.57143	19.28571	22.57143	19.85714	26.57143	23.71429
6	22.33333	22.33333	27.66667	25	28.33333	21.66667	24.33333	25
7	26.2	28.8	27	23.6	26	30.8	28.2	24.5
8	26.83333	28.33333	28.66667	28.66667	25.33333	28.5	27.5	27.66667

Note Sessions 5, 6, 7 and 8 included obstacle navigation.

Table A.2 Average Number of Cognitive Responses from Biped Walking Group During Each Trial of Each Session.

Session	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 6	Trial 7	Trial 8
1	11.71429	13	13.42857	11.71429	13.71429	13.85714	14.42857	13.71429
2	14.28571	16	15.28571	15.57143	17.28571	16.14286	17.42857	17.28571
3	16.42857	15.85714	16.57143	16.71429	16.71429	16.85714	16.85714	16.42857
4	18.85714	19.42857	19	18	19.14286	20.71429	19.85714	21.28571
5	14.85714	14.42857	15.57143	15.42857	15.28571	15.71429	17.42857	15.71429
6	19.42857	20.85714	20	20.71429	20.71429	19.57143	22.57143	19.57143
7	20.71429	21.14286	20.57143	20.71429	20.85714	21.85714	24	21.42857
8	23.42857	23.57143	23.28571	23.14286	23.57143	23.57143	23.57143	24

Note Sessions 5, 6, 7 and 8 included obstacle navigation.

Figures A.1 and A.2 are the percentage of total obstacles successfully navigated by single task groups.

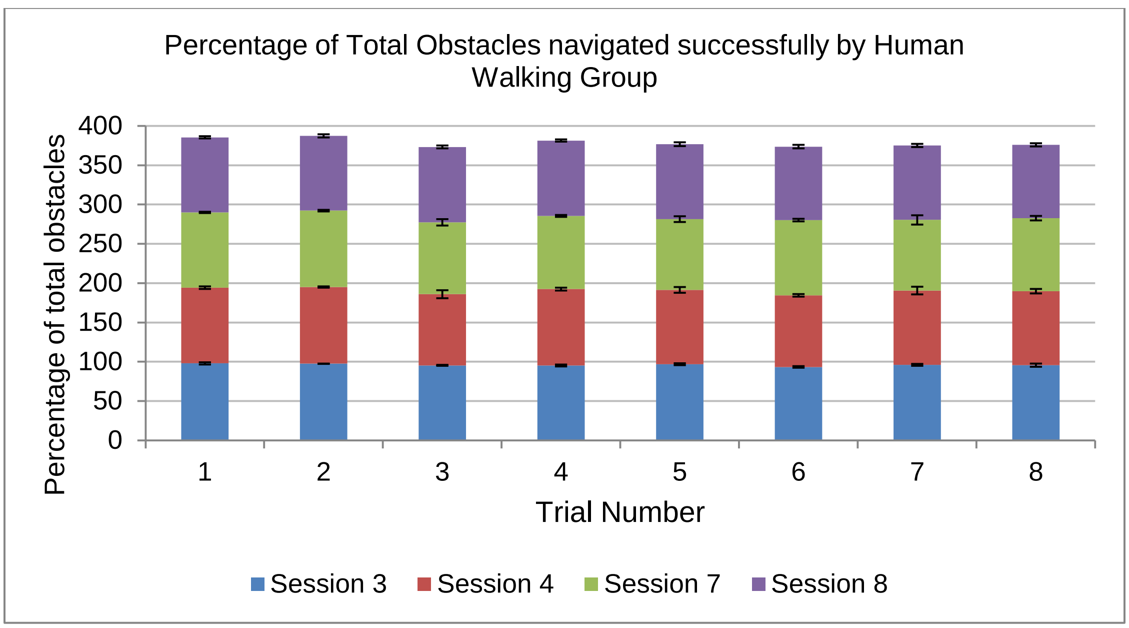


Figure A.1 Percentage of total obstacles navigated successfully by Human Walking Group during each trial of Sessions 5, 6, 7 and 8. Error bars indicate standard error of mean.

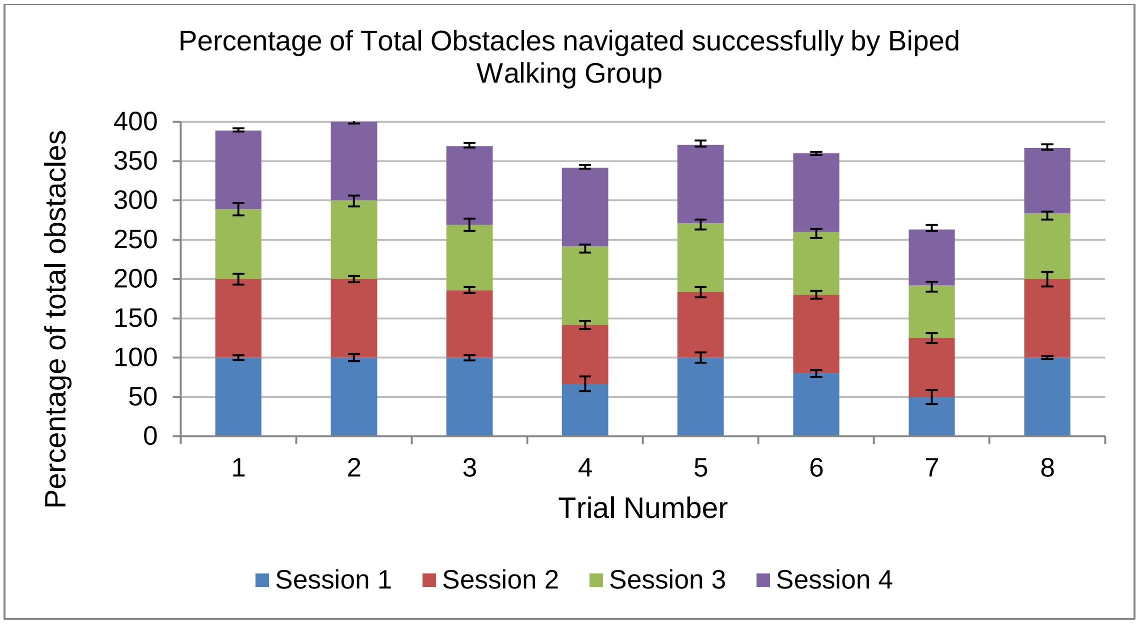


Figure A.2 Percentage of total obstacles navigated successfully by Biped Walking Group during each trial of Sessions 5, 6, 7 and 8. Error bars indicate standard error of mean.

Figures A.3 and A.4 are percentage of total obstacles successfully navigated by dual task groups.

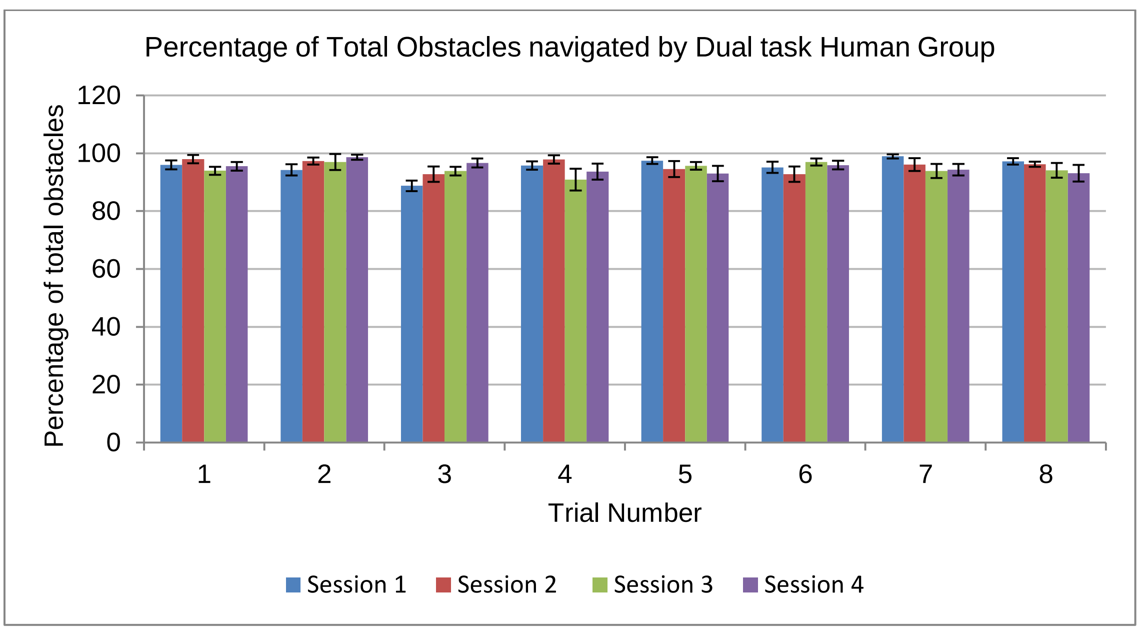


Figure A.3 Percentage of total obstacles navigated successfully by dual task Human Walking Group during each trial of Sessions 5, 6, 7 and 8. Error bars indicate standard error of mean.

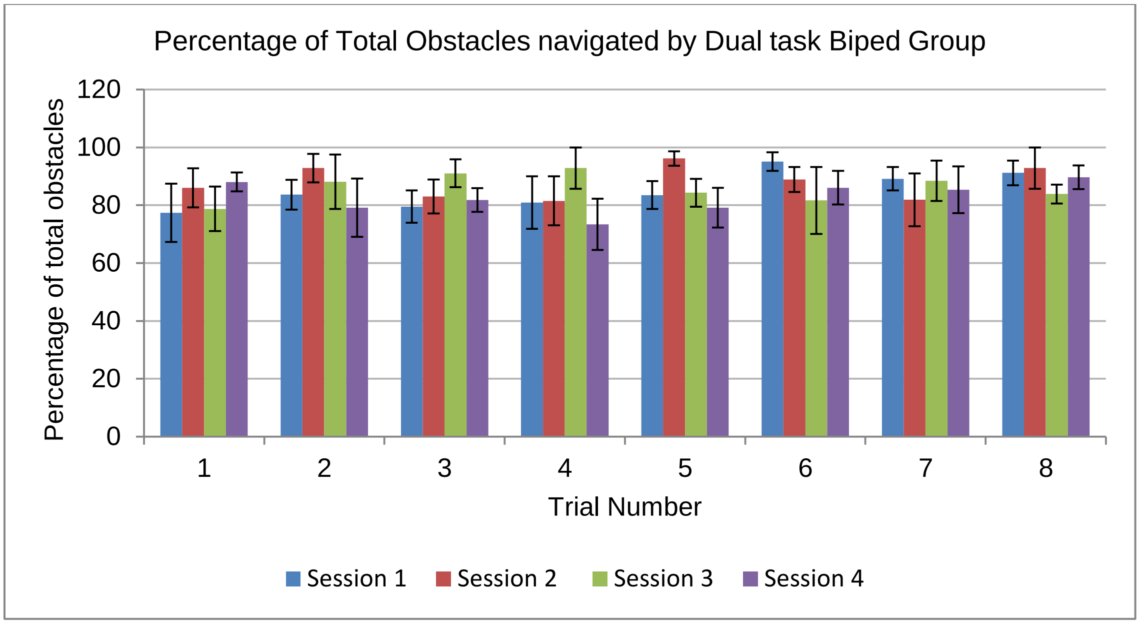


Figure A.4 Percentage of total obstacles navigated successfully by dual task Human Walking Group during each trial of Sessions 5, 6, 7 and 8. Error bars indicate standard error of mean.

APPENDIX B

COMPLETE CONTROL ALGORITHM

Appendix B is the code in MATLAB for the complete control algorithm to control the leg in sagittal plane.

```
%ODE for x & y direction Sagittal control
    [t,positionyLeft] = odeF(tspan, y0,[Fy(1,i),.06,.05],[0],[1e-6, 1e-
8, 10]);% ODE for x
    sizes=size(positionyLeft);
    positionsArray(2,i)=(positionyLeft(1,sizes(1,2)));%initial
condition for y
    y0=[positionyLeft(1,end) positionyLeft(2,end)];
    velocity(2,i)=positionyLeft(2,end);

    [t,positionzLeft] = odeF(tspan, z0,[Fz(1,i),.05,.05],[0],[1e-6, 1e-
8, 10]);% ODE for x
    sizes=size(positionzLeft);
    positionsArray(3,i)=(positionzLeft(1,sizes(1,2)));% initial
condition for y
    z0=[positionzLeft(1,end) positionzLeft(2,end)];
    velocity(3,i)=positionzLeft(2,end);



    % calculate thetas for hip knee and ankle

positionSqr1(1,i)=sqrt((positionsArray(2,i))^2+(positionsArray(3,i))^2)
;
    positionSqr=((positionsArray(2,i))^2+(positionsArray(3,i))^2);
    positionsArrayOld(2,i)=positionsArray(2,i);
    positionsArrayOld(3,i)=positionsArray(3,i);
    if(i>50 && (positionSqr1(1,i) == 39.95 || positionSqr1(1,i) >
39.95))

        ratio(1,i)= 39.94/positionSqr1(1,i);
        positionsArray(2,i)=positionsArray(2,i)* ratio(1,i);
        positionsArray(3,i)=positionsArray(3,i)* ratio(1,i);
        y0=[positionsArray(2,i) positionyLeft(2,end)];
        z0=[positionsArray(3,i) positionzLeft(2,end)];
        positionSqr=((positionsArray(2,i))^2+(positionsArray(3,i))^2);
    end

positionSqrss(1,i)=sqrt((positionsArray(2,i))^2+(positionsArray(3,i))^2
);
%ODE for z direction

    [t,positionx] = odeF(tspan2,
x0,[Fx(1,i)*positionSqrss(1,i)/100,3,.8],[0],[1e-6, 1e-8, 10]);% ODE
for x
    sizes=size(positionx);
    positionsArray(1,i)=(positionx(1,sizes(1,2)));% abduction angle
for hip
    x0=[positionx(1,end) positionx(2,end)];% Setting initial condition
for x
    velocity(1,i)=positionx(2,end);
    positionsArray(5,i)=pi-positionsArray(1,i);%Setting abduction angle
for

    %%Inverse Kinematics
    Ttheta2(1,i) =(((l1+l2)^2)-positionSqr)/(positionSqr-((l1-l2)^2));
    % there are two solution while using the inverse of tan
    theta2Up(1,i)= 2*atan(sqrt(Ttheta2(1,i)));% knee angle
```

```
    theta2Down(1,i)=-2* atan(sqrt(Ttheta2(1,i)));
    % calculating the hip angle using the knee angle
    phi(1,i)=atan2(positionsArray(2,i),positionsArray(3,i));


uptao(1,i)=atan2((l2*sin(theta2Up(1,i))),(l1+(l2*cos(theta2Up(1,i)))));

downtao(1,i)=atan2((l2*sin(theta2Down(1,i))),(l1+(l2*cos(theta2Down(1,i
)))));
        theta1Up(1,i)=phi(1,i)-uptao(1,i);
        theta1Down(1,i)=phi(1,i)-downtao(1,i);
        % code to keep the foot horizontal to the ground
        test1(1,i)=phi(1,i)-pi/2;
        test2(1,i)=pi-(downtao(1,i)+ pi/2-theta2Down(1,i));
        footAngle(1,i)=test1(1,i)+test2(1,i);
```

APPENDIX C

3D DESIGN AND SCHEMATICS

Pro-E designs for the foot and extrusions.

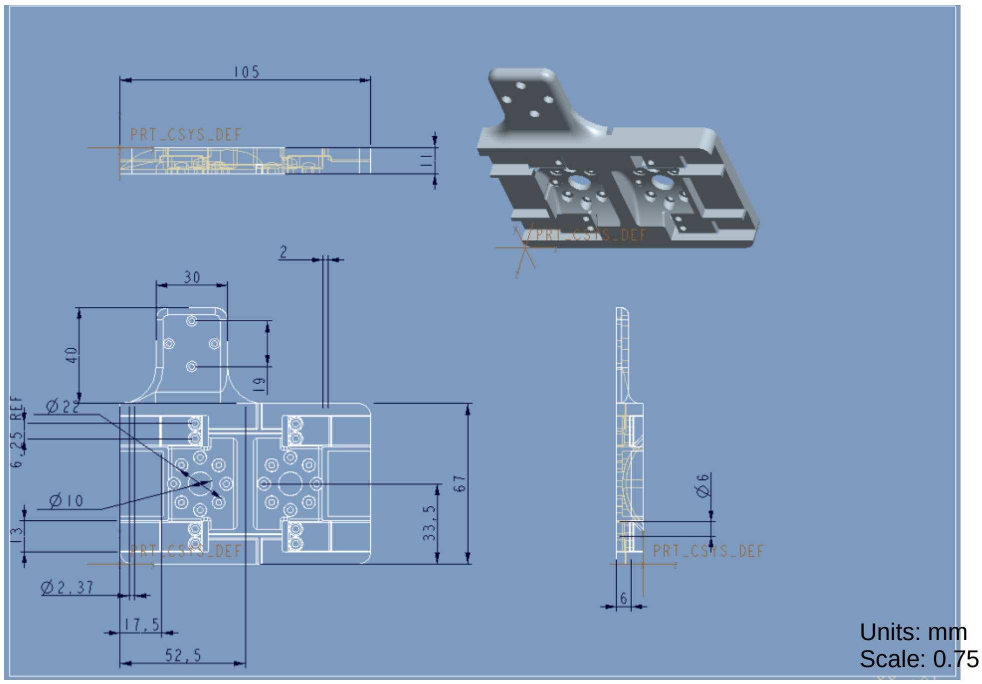


Figure C.1 Schematic of the foot.

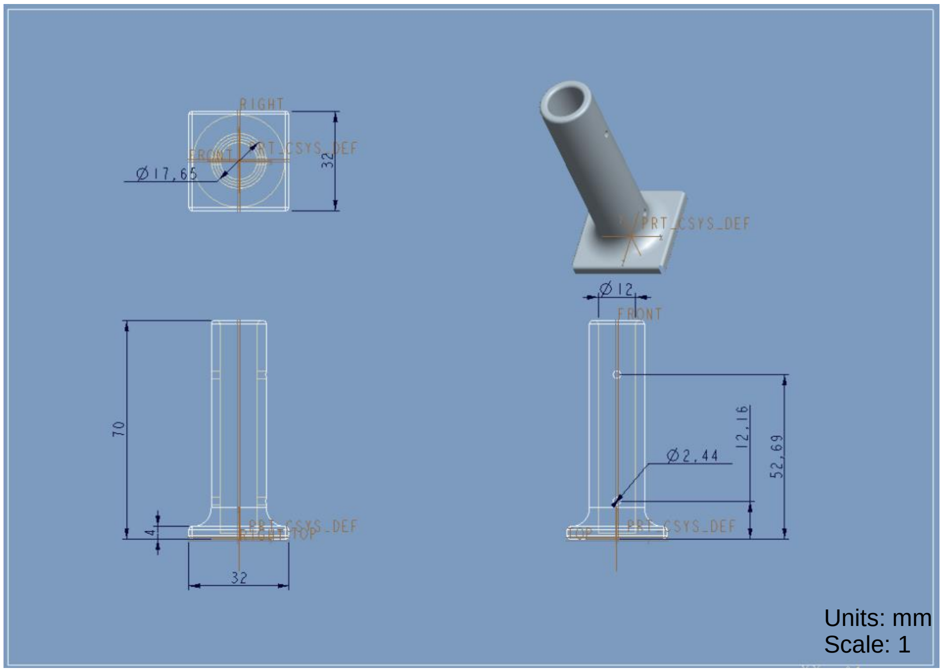


Figure C.2 Schematic of the trekker pole mount.

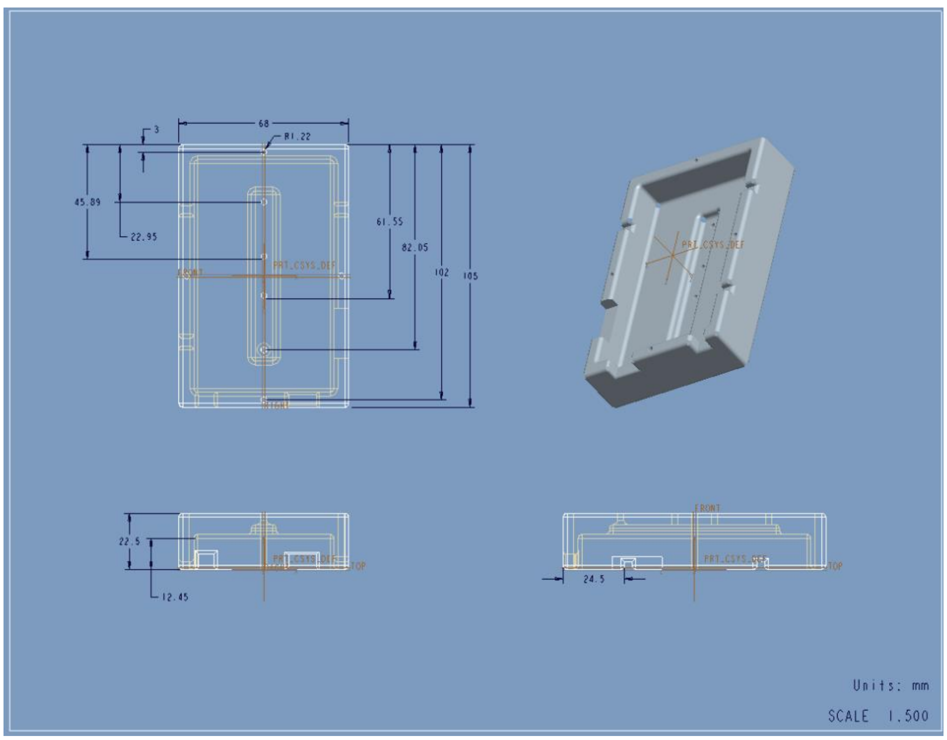


Figure C.3 Schematic of the mount for Phidget force sensor.

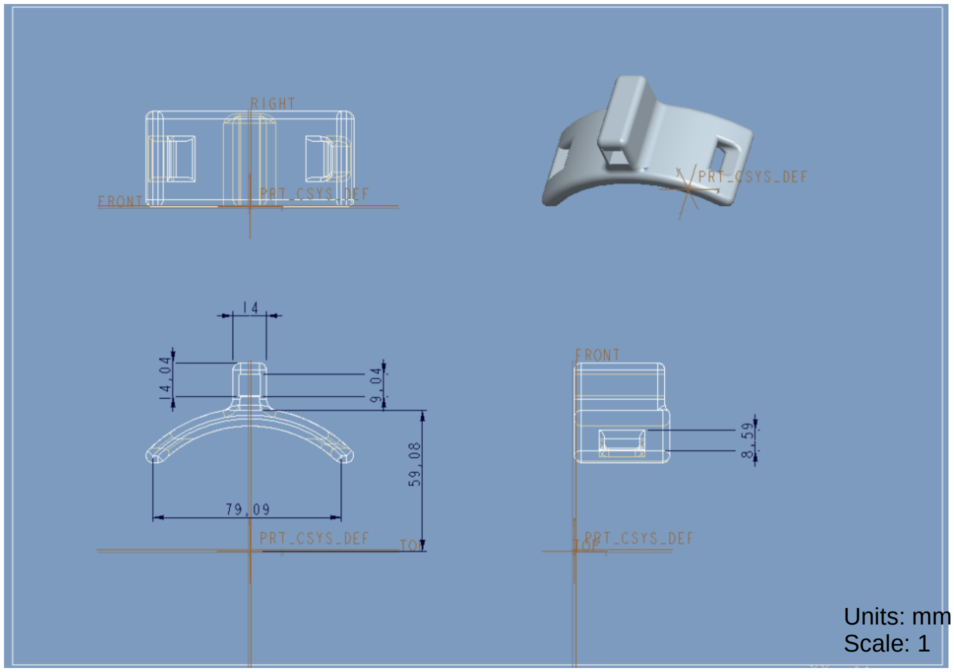


Figure C.4 Schematic of the Optitrak marker mounts.

REFERENCES

1. "NINDS Spinal Cord Injury Information Page." Spinal Cord Injury Information Page: National Institute of Neurological Disorders and Stroke (NINDS). Accessed April 08, 2016. http://www.ninds.nih.gov/disorders/sci/sci.htm

2. "Basic Facts - Understanding Spinal Cord Injury." Basic Facts - Understanding Spinal Cord Injury. Accessed April 08, 2016. http://www.spinalinjury101.org/details.

3. "Atlanta Injury Lawyer - Shigley Law, LLC." Atlanta Spinal Cord Injury Attorney. Accessed April 08, 2016. http://www.atlantainjurylawyer.com/spinal-cord-injury.html.

4. "Spinal Cord Injury." - Bel13ve in Miracles Foundation. Accessed April 08, 2016. http://www.bel13vefoundation.org/spinal-cord-injury.

5. Mckinley, William O., Amie B. Jackson, Diana D. Cardenas, and Michael J. De Vivo. "Long-term Medical Complications after Traumatic Spinal Cord Injury: A Regional Model Systems Analysis." Archives of Physical Medicine and Rehabilitation 80, no. 11 (1999): 1402-410. doi:10.1016/s0003-9993(99)90251-4.

6. Giangregorio, Lora, and Neil McCartney. "Bone loss and muscle atrophy in spinal cord injury: epidemiology, fracture prediction, and rehabilitation strategies." The journal of spinal cord medicine 29, no. 5 (2006): 489.

7. "Spinal Cord Injury Facts & Statistics." Spinal Cord Injury Facts & Statistics. Accessed April 08, 2016. http://www.sci-info-pages.com/facts.html.

8. Anderson, Kim D. "Targeting Recovery: Priorities of the Spinal Cord-Injured Population." Journal of Neurotrauma 21, no. 10 (2004): 1371-383. doi:10.1089/neu.2004.21.1371.

9. Ditunno, P. L., M. Patrick, M. Stineman, and J. F. Ditunno. "Who Wants to Walk? Preferences for Recovery after SCI: A Longitudinal and Cross-sectional Study." Spinal Cord 46, no. 7 (2008): 500-06. doi:10.1038/sj.sc.3102172.

10. Kilgore, Kevin L., Marcia Scherer, Rod Bobblitt, and Jocelyn Dettloff. "Neuroprosthesis consumers' forum: consumer priorities for research directions." Journal of rehabilitation research and development 38, no. 6 (2001): 655.

11. Dollar, Aaron M., and Hugh Herr. "Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art." IEEE Trans. Robot. IEEE Transactions on Robotics 24, no. 1 (2008): 144-58. doi:10.1109/tro.2008.915453.

12. Anam, Khairul, and Adel Ali Al-Jumaily. "Active Exoskeleton Control Systems: State of the Art." Procedia Engineering 41 (2012): 988-94. doi:10.1016/j.proeng.2012.07.273.

13. Kwakkel, Gert, Robert C. Wagenaar, Jos WR Twisk, Gustaaf J. Lankhorst, and Johan C. Koetsier. "Intensity of leg and arm training after primary middle-cerebral-artery stroke: a randomised trial." The Lancet 354, no. 9174 (1999): 191-196.

14. Kim, Wansoo, Heedon Lee, Donghwan Kim, Jungsoo Han, and Changsoo Han. "Mechanical design of the Hanyang Exoskeleton Assistive Robot (HEXAR)." In Control, Automation and Systems (ICCAS), 2014 14th International Conference on, pp. 479-484. IEEE, 2014.

15. "Berkeley Robotics & Human Engineering Laboratory." Berkeley Robotics Human Engineering Laboratory RSS. Accessed April 08, 2016. http://bleex.me.berkeley.edu/research/exoskeleton/hulc/.

16. Riener, Robert. "Technology of the robotic gait orthosis Lokomat." InNeurorehabilitation Technology, pp. 221-232. Springer London, 2012.

17. Zanotto, Damiano, Paul Stegall, and Sunil K. Agrawal. "ALEX III: A novel robotic platform with 12 DOFs for human gait training." In Robotics and Automation (ICRA), 2013 IEEE International Conference on, pp. 3914-3919. IEEE, 2013.

18. Ekkelenkamp, Ralf, J. Veneman, and Herman van der Kooij. "LOPES: a lower extremity powered exoskeleton." In Robotics and Automation, 2007 IEEE International Conference on, pp. 3132-3133. IEEE, 2007.

19. Unluhisarcikli, Ozer, Maciej Pietrusinski, Brian Weinberg, Paolo Bonato, and Constantinos Mavroidis. "Design and control of a robotic lower extremity exoskeleton for gait rehabilitation." In Intelligent Robots and Systems (IROS), 2011 IEEE/RSJ International Conference on, pp. 4893-4898. IEEE, 2011.

20. "Engineering Newsletter." Engineering Newsletter. Accessed April 09, 2016. http://engineering.columbia.edu/web/newsletter/fall_2014/sunil_agrawal—personalized_medicine.

21. "'Wii-habilitation' and Robotic Exoskeletons: Technology in Physiotherapy | RCSI Student Medical Journal." 'Wii-habilitation' and Robotic Exoskeletons: Technology in Physiotherapy | RCSI Student Medical Journal. Accessed April 09, 2016. http://www.rcsismj.com/2009-2010-issue/wii-habilitation/.

22. "NeuroControl." NeuroControl. Accessed April 09, 2016. http://www.neurocontrol.nl/projects/current/.

23. Strausser, Katherine A., and H. Kazerooni. "The Development and Testing of a Human Machine Interface for a Mobile Medical Exoskeleton." 2011 IEEE/RSJ International Conference on Intelligent Robots and Systems, 2011. doi:10.1109/iros.2011.6095025.

24. Lee, Suwoong, and Yoshiyuki Sankai. "Power assist control for walking aid with HAL-3 based on EMG and impedance adjustment around knee joint." InIntelligent Robots and Systems, 2002. IEEE/RSJ International Conference on, vol. 2, pp. 1499-1504. IEEE, 2002.

25. Hayashi, Tomohiro, Hiroaki Kawamoto, and Yoshiyuki Sankai. "Control method of robot suit HAL working as operator's muscle using biological and dynamical information." In Intelligent Robots and Systems, 2005.(IROS 2005). 2005 IEEE/RSJ International Conference on, pp. 3063-3068. IEEE, 2005.

26. Quintero, Hugo A., Ryan J. Farris, and Michael Goldfarb. "A method for the autonomous control of lower limb exoskeletons for persons with paraplegia."Journal of medical devices 6, no. 4 (2012): 041003.

27. Farris, Ryan James. Design of a Powered Lower-limb Exoskeleton and Control for Gait Assistance in Paraplegics. PhD diss., Vanderbilt University, May, 2012.

28. Farris, Ryan J., Hugo A. Quintero, and Michael Goldfarb. "Performance evaluation of a lower limb exoskeleton for stair ascent and descent with paraplegia." In Engineering in medicine and biology society (EMBC), 2012 annual international conference of the IEEE, pp. 1908-1911. IEEE, 2012.

29. Esquenazi, Alberto, Mukul Talaty, Andrew Packel, and Michael Saulino. "The ReWalk powered exoskeleton to restore ambulatory function to individuals with thoracic-level motor-complete spinal cord injury." American journal of physical medicine & rehabilitation 91, no. 11 (2012): 911-921.

30. Strickland, Eliza. "Good-bye, wheelchair." Spectrum, IEEE 49, no. 1 (2012): 30-32.

31. "Ekso Bionics - An Exoskeleton Bionic Suit or a Wearable Robot That Helps People Walk Again." Ekso. Accessed April 09, 2016. http://www.eksobionics.com/.

32. "REX Bionics - Step into the Future." Rex Bionics. Accessed April 09, 2016. http://www.rexbionics.com/.

33. Ferris, Daniel P., Gregory S. Sawicki, and Antoinette Domingo. "Powered lower limb orthoses for gait rehabilitation." Topics in spinal cord injury rehabilitation 11, no. 2 (2005): 34.

34. Kobetic, Rudi, Curtis S. To, John R. Schnellenberger, Thomas C. Bulea, Richard Gaudio CO, and Gilles Pinault. "Development of hybrid orthosis for standing, walking, and stair climbing after spinal cord injury." Journal of rehabilitation research and development 46, no. 3 (2009): 447.

35. Hasegawa, Yasuhisa, Junho Jang, and Yoshiyuki Sankai. "Cooperative walk control of paraplegia patient and assistive system." In Intelligent robots and systems, 2009. IROS 2009. IEEE/RSJ international conference on, pp. 4481-4486. IEEE, 2009.

36. Mohammed, Samer, and Yacine Amirat. "Towards intelligent lower limb wearable robots: Challenges and perspectives-State of the art." In Robotics and Biomimetics, 2008. ROBIO 2008. IEEE International Conference on, pp. 312-317. IEEE, 2009.

37. Reiner, Robert. "Proceedings of Design Criteria for the Development of Wearable Robot." International Workshop on Wearable Robotics,WeRob, 2014

38. Dellon, Brian, and Yoky Matsuoka. "Prosthetics, exoskeletons, and rehabilitation." IEEE Robotics and Automation magazine 14, no. 1 (2007): 30.

39. Del-Ama, Antonio J., Aikaterini D. Koutsou, Juan C. Moreno, Ana De-Los-Reyes, Ángel Gil-Agudo, and José L. Pons. "Review of hybrid exoskeletons to restore gait following spinal cord injury." J Rehabil Res Dev 49, no. 4 (2012): 497-514.

40. Lee, Heedon, Wansoo Kim, Jungsoo Han, and Changsoo Han. "The technical trend of the exoskeleton robot system for human power assistance." International Journal of Precision Engineering and Manufacturing 13, no. 8 (2012): 1491-1497.

41. Gancet, Jeremi, Michel Ilzkovitz, Elvina Motard, Yashodhan Nevatia, Pierre Letier, David De Weerdt, Guy Cheron et al. "MINDWALKER: going one step further with assistive lower limbs exoskeleton for SCI condition subjects." InBiomedical Robotics and Biomechatronics (BioRob), 2012 4th IEEE RAS & EMBS International Conference on, pp. 1794-1800. IEEE, 2012.

42. Nicolelis, Miguel AL. "Brain–machine interfaces to restore motor function and probe neural circuits." Nature Reviews Neuroscience 4, no. 5 (2003): 417-422.

43. Contreras-Vidal, Jose L., and Robert G. Grossman. "NeuroRex: A clinical neural interface roadmap for EEG-based brain machine interfaces to a lower body robotic exoskeleton." In Engineering in medicine and biology society (EMBC), 2013 35th annual international conference of the IEEE, pp. 1579-1582. IEEE, 2013.

44. "News from The Walk Again Project." Laboratory of Dr Miguel Nicolelis. Accessed April 09, 2016. http://www.nicolelislab.net/?p=584.

45. "Mindwalker – Mind-controlled Exoskeleton Could Help Disabled People | RobAid." RobAid. 2013. Accessed April 09, 2016. http://www.robaid.com/bionics/mindwalker-mind-controlled-exoskeleton-could-help-disabled-people.htm.

46. "Paraplegic Man in Mind-controlled Robotic Suit Kicks off World Cup 2014 (VIDEO) - Neurogadget." Neurogadget. 2014. Accessed April 09, 2016. http://neurogadget.com/2014/06/13/paraplegic-man-mind-controlled-robotic-suit-kicks-world-cup-2014-video/10434.

47. Hasegawa, Yasuhisa, Junho Jang, and Yoshiyuki Sankai. "Cooperative walk control of paraplegia patient and assistive system." In Intelligent robots and systems, 2009. IROS 2009. IEEE/RSJ international conference on, pp. 4481-4486. IEEE, 2009.

48. Feldman, Anatol G., and Mark L. Latash. "Testing hypotheses and the advancement of science: recent attempts to falsify the equilibrium point hypothesis." Experimental Brain Research 161, no. 1 (2005): 91-103.

49. Guo, Yi. "Characterizing Motor Control Signals in the Spinal Cord." PhD diss., New Jersey Institute of Technology, Department of Biomedical Engineering, 2014.

50. Churchland, Mark M., and Krishna V. Shenoy. "Temporal complexity and heterogeneity of single-neuron activity in premotor and motor cortex." Journal of neurophysiology 97, no. 6 (2007): 4235-4257.

51. Cisek, Paul. "Preparing for speed. Focus on “Preparatory activity in premotor and motor cortex reflects the speed of the upcoming reach”." Journal of neurophysiology 96, no. 6 (2006): 2842-2843.

52. Karunakaran, Kiran K., and Richard Foulds. "Control of Foot Trajectory in Biped Robots." In Bioengineering Conference (NEBEC), 2013 39th Annual Northeast, pp. 203-204. IEEE, 2013.

53. Hughes, J., and N. Jacobs. "Normal human locomotion." Prosthetics and orthotics international 3, no. 1 (1979): 4-12.

54. Vaughan, Christopher L., Brian L. Davis, and Jeremy C. O'Connor. Dynamics of Human Gait. Champaign, IL: Human Kinetics Publishers, 1992.

55. MacKay-Lyons, Marilyn. "Central pattern generation of locomotion: a review of the evidence." Physical therapy 82, no. 1 (2002): 69-83.

56. Duysens, Jacques, and Henry W.a.a Van De Crommert. "Neural Control of Locomotion; Part 1: The Central Pattern Generator from Cats to Humans." Gait & Posture 7, no. 2 (1998): 131-41.

57. Delcomyn, F. "Neural Basis of Rhythmic Behavior in Animals." Science 210, no. 4469 (1980): 492-98.

58. Pearson, K. G. "Common Principles of Motor Control in Vertebrates and Invertebrates." Annu. Rev. Neurosci. Annual Review of Neuroscience 16, no. 1 (1993): 265-97

59. Ivanenko, Y. P., 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 87, no. 6 (2002): 3070-3089.

60. Ankarali, Mert M., H. Tutkun Sen, Avik De, A. , Okamura, Allison and N. J. Cowan. "Haptic Feedback Enhances Rhythmic Motor Control by Reducing Variability, Not

Improving Convergence Rate." Journal of Neurophysiology 111, no. 6 (2013): 1286-299.

61. Koritnik, T., A. Koenig, T. Bajd, R. Riener, and M. Munih. "Comparison of Visual and Haptic Feedback during Training of Lower Extremities." Gait & Posture 32, no. 4 (2010): 540-46.

62. Turchet, Luca, Paolo Burelli, and Stefania Serafin. "Haptic Feedback for Enhancing Realism of Walking Simulations." IEEE Trans. Haptics IEEE Transactions on Haptics 6, no. 1 (2013): 35-45.

63. Winter, David A., and David A. Winter. Biomechanics and Motor Control of Human Movement. New York: Wiley, 1990.

64. Proteau, Luc, and Digby Elliott. Vision and Motor Control. Amsterdam: North-Holland, 1992.

65. Levy-Tzedek, Shelly, M. Ben Tov, and Amir Karniel. "Rhythmic movements are larger and faster but with the same frequency on removal of visual feedback." Journal of neurophysiology 106, no. 5 (2011): 2120-2126.

66. Feygin, David, Madeleine Keehner, and Frank Tendick. "Haptic guidance: Experimental evaluation of a haptic training method for a perceptual motor skill." In Haptic Interfaces for Virtual Environment and Teleoperator Systems, 2002. HAPTICS 2002. Proceedings. 10th Symposium on, pp. 40-47. IEEE, 2002.

67. Gunn, Chris, Warren Muller, and Amitava Datta. "Performance improvement with haptic assistance: A quantitative assessment." In EuroHaptics conference, 2009 and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems. World Haptics 2009. Third Joint, pp. 511-516. IEEE, 2009.

68. Karunakaran, Kiran, Kevin Abbruzzese, Hao Xu, Naphtaly Ehrenberg, and Richard Foulds. "Haptic proprioception in a virtual locomotor task." InEngineering in Medicine and Biology Society (EMBC), 2014 36th Annual International Conference of the IEEE, pp. 3594-3597. IEEE, 2014.

69. Carignan, Craig R., and David L. Akin. "Using robots for astronaut training."Control Systems, IEEE 23, no. 2 (2003): 46-59.

70. Van der Linde, Richard Q., Piet Lammertse, Erwin Frederiksen, and B. Ruiter. "The HapticMaster, a new high-performance haptic interface." InProc. Eurohaptics, pp. 1-5. 2002.

71. Hogan, Neville. "Impedance control: An approach to manipulation." InAmerican Control Conference, 1984, pp. 304-313. IEEE, 1984.

72. Hogan, Neville. "Impedance control: An approach to manipulation: Part II—Implementation." Journal of dynamic systems, measurement, and control107, no. 1 (1985): 8-16.

73. Dautenhahn, Kerstin, and Joe Saunders. New Frontiers in Human-robot Interaction. Amsterdam: John Benjamins Pub., 2011.

74. Haidegger, Tamas, Balázs Benyó, Levente Kovács, and Zoltán Benyó. "Force sensing and force control for surgical robots." In 7th IFAC Symposium on Modeling and Control in Biomedical Systems, vol. 7, no. 1. 2009.

75. Glosser, Gregory D., and Wyatt S. Newman. "The implementation of a natural admittance controller on an industrial manipulator." In Robotics and Automation, 1994. Proceedings., 1994 IEEE International Conference on, pp. 1209-1215. IEEE, 1994.

76. van Riel, Natal. "Speeding up simulations of ODE models in Matlab using CVode and MEX files.", 2012.

77. Tanawongsuwan, Rawesak, and Aaron Bobick. A study of human gaits across different speeds. Georgia Tech, Tech. Rep, 2003.

78. Beauchet, Olivier, Cedric Annweiler, Yhann Lecordroch, Gilles Allali, Veronique Dubost, François R. Herrmann, and Reto W. Kressig. "Walking speed-related changes in stride time variability: effects of decreased speed."Journal of neuroengineering and rehabilitation 6, no. 1 (2009): 1.

79. Vilas-boas, João Paulo. "The influence of ambulatory speed on gait biomechanical parameters." RPcd 11, no. 3: 64-87.

80. Salo, Aki IT, Ian N. Bezodis, Alan M. Batterham, and David G. Kerwin. "Elite sprinting: are athletes individually step-frequency or step-length reliant?."Medicine and science in sports and exercise 43, no. 6 (2011): 1055-1062.

81. Bezodis, I. N. "Investigations of the step length-step frequency relationship in sprinting: applied implications for performance." In Proceedings of the 30th Annual Conference of Biomechanics in Sports, pp. 43-49. 2012.

82. Sparrow, W. A., Elizabeth J. Bradshaw, Ecosse Lamoureux, and Oren Tirosh. "Ageing effects on the attention demands of walking." Human movement science 21, no. 5 (2002): 961-972.

83. Smyth, Mary M., Alan F. Collins, Peter E. Morris, and Philip Levy. Cognition in Action. Hove: Psychology Press, 1996.

84. Mulder, Theo, Herman Berndt, John Pauwels, and Bart Nienhuis. "Sensorimotor adaptability in the elderly and disabled." In Sensorimotor impairment in the elderly, pp. 413-426. Springer Netherlands, 1993.

85. Pashler, Harold E. Attention. Hove, East Sussex, UK: Psychology Press, 1998.

86. Patel, Prakruti, Melissa Lamar, and Tanvi Bhatt. "Effect of type of cognitive task and walking speed on cognitive-motor interference during dual-task walking." Neuroscience 260 (2014): 140-148.

87. Plummer-D'Amato, Prudence, Briana Brancato, Mallory Dantowitz, Stephanie Birken, Christina Bonke, and Erin Furey. "Effects of gait and cognitive task difficulty on cognitive-motor interference in aging." Journal of aging research2012 (2012).

88. Yogev-Seligmann, Galit, Yael Rotem-Galili, Anat Mirelman, Ruth Dickstein, Nir Giladi, and Jeffrey M. Hausdorff. "How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability." Physical therapy 90, no. 2 (2010): 177-186.

89. Dubost, Véronique, Cédric Annweiler, Kamiar Aminian, Bijan Najafi, François R. Herrmann, and Olivier Beauchet. "Stride-to-stride variability while enumerating animal names among healthy young adults: result of stride velocity or effect of attention-demanding task?." Gait & posture 27, no. 1 (2008): 138-143.

90. Beauchet, Olivier, Véronique Dubost, François R. Herrmann, and Reto W. Kressig. "Stride-to-stride variability while backward counting among healthy young adults." Journal of neuroengineering and rehabilitation 2, no. 1 (2005): 1.

91. Beauchet, O., V. Dubost, F. Stierlam, M. A. Blanchon, F. Mourey, P. Pfitzenmeyer, R. Gonthier, and R. W. Kressig. "[Influence of a specific cognitive task on spatial-temporal walking parameters in elderly frail individuals]." Presse medicale (Paris, France: 1983) 31, no. 24 (2002): 1117-1122.

92. Shumway-Cook, Anne, Marjorie Woollacott, Kimberly A. Kerns, and Margaret Baldwin. "The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences 52, no. 4 (1997): M232-M240.

93. Al-Yahya, Emad, Helen Dawes, Lesley Smith, Andrea Dennis, Ken Howells, and Janet Cockburn. "Cognitive motor interference while walking: a systematic review and meta-analysis." Neuroscience & Biobehavioral Reviews 35, no. 3 (2011): 715-728.

94. Holtzer, Roee, Cuiling Wang, and Joe Verghese. "The relationship between attention and gait in aging: facts and fallacies." Motor control 16, no. 1 (2012): 64.

95. Schaefer, Sabine. "The ecological approach to cognitive–motor dual-tasking: findings on the effects of expertise and age." Frontiers in psychology 5 (2014).

96. Hogan, Neville, and Dagmar Sternad. "On rhythmic and discrete movements: reflections, definitions and implications for motor control." Experimental Brain Research 181, no. 1 (2007): 13-30.

97. Mertz, Leslie. "The next generation of exoskeletons: Lighter, cheaper devices are in the works." Pulse, IEEE 3, no. 4 (2012): 56-61.

98. Ferrati, Francesco, Roberto Bortoletto, Emanuele Menegatti, and Enrico Pagello. "Socio-economic impact of medical lower-limb exoskeletons." InAdvanced Robotics and its Social Impacts (ARSO), 2013 IEEE Workshop on, pp. 19-26. IEEE, 2013.

99. Esquenazi, Alberto, Mukul Talaty, Andrew Packel, and Michael Saulino. "The ReWalk powered exoskeleton to restore ambulatory function to individuals with thoracic-level motor-complete spinal cord injury." American journal of physical medicine & rehabilitation 91, no. 11 (2012): 911-921.

100. Díaz, Iñaki, Jorge Juan Gil, and Emilio Sánchez. "Lower-limb robotic rehabilitation: literature review and challenges." Journal of Robotics 2011 (2011).

101. Zoss, Adam B., Hami Kazerooni, and Andrew Chu. "Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX)." Mechatronics, IEEE/ASME Transactions on 11, no. 2 (2006): 128-138.

102. Kazerooni, H., and R. Steger. "The Berkeley lower extremity exoskeleton."Journal of dynamic systems, measurement, and control 128, no. 1 (2006): 14-25.

103. Kazerooni, Hami, Ryan Steger, and Lihua Huang. "Hybrid control of the Berkeley lower extremity exoskeleton (BLEEX)." The International Journal of Robotics Research 25, no. 5-6 (2006): 561-573.

104. Low, K. H. "Robot-assisted gait rehabilitation: From exoskeletons to gait systems." In Defense Science Research Conference and Expo (DSR), 2011, pp. 1-10. IEEE, 2011.

105. Wang, Letian, Shiqian Wang, Edwin HF van Asseldonk, and Herman van der Kooij. "Actively controlled lateral gait assistance in a lower limb exoskeleton." In Intelligent Robots and Systems (IROS), 2013 IEEE/RSJ International Conference on, pp. 965-970. IEEE, 2013.

106. Patla, Aftab E., Allan Adkin, and Tonya Ballard. "Online steering: coordination and control of body center of mass, head and body reorientation." Experimental brain research 129, no. 4 (1999): 629-634.

107. Gérin-Lajoie, Martin, Carol L. Richards, and Bradford J. McFadyen. "The negotiation of stationary and moving obstructions during walking: anticipatory locomotor adaptations and preservation of personal space." MOTOR CONTROL-CHAMPAIGN- 9, no. 3 (2005): 242.

108. Takeda, Ryo, Giulia Lisco, Tadashi Fujisawa, Laura Gastaldi, Harukazu Tohyama, and Shigeru Tadano. "Drift removal for improving the accuracy of gait parameters using wearable sensor systems." Sensors 14, no. 12 (2014): 23230-23247.

109. Colombo, Gery, Matthias Joerg, Reinhard Schreier, and Volker Dietz. "Treadmill training of paraplegic patients using a robotic orthosis." Journal of rehabilitation research and development 37, no. 6 (2000): 693.

110. Yogev- Seligmann, Galit, Jeffrey M. Hausdorff, and Nir Giladi. "Do we always prioritize balance when walking? Towards an integrated model of task prioritization." Movement Disorders 27, no. 6 (2012): 765-770.

111. Liston, Matthew B., Jeroen H. Bergmann, Niamh Keating, David A. Green, and Marousa Pavlou. "Postural prioritization is differentially altered in healthy older compared to younger adults during visual and auditory coded spatial multitasking." Gait & posture 39, no. 1 (2014): 198-204.

112. Young, Patricia M. McAndrew, and Jonathan B. Dingwell. "Voluntarily changing step length or step width affects dynamic stability of human walking." Gait & posture 35, no. 3 (2012): 472-477.

113. Maki, Brian E. "Gait changes in older adults: predictors of falls or indicators of fear?." Journal of the American geriatrics society 45, no. 3 (1997): 313-320.

114. Paterson, Kade, Keith Hill, and Noel Lythgo. "Stride dynamics, gait variability and prospective falls risk in active community dwelling older women." Gait & posture 33, no. 2 (2011): 251-255.

115. Hausdorff, Jeffrey M., Dean A. Rios, and Helen K. Edelberg. "Gait variability and fall risk in community-living older adults: a 1-year prospective study."Archives of physical medicine and rehabilitation 82, no. 8 (2001): 1050-1056.

116. Hreljac, Alan. "Stride smoothness evaluation of runners and other athletes."Gait & posture 11, no. 3 (2000): 199-206.

117. Pashler, Harold. "Dual-task interference in simple tasks: data and theory."Psychological bulletin 116, no. 2 (1994): 220.

118. Flash, Tamar, and Neville Hogan. "The coordination of arm movements: an experimentally confirmed mathematical model." The journal of Neuroscience5, no. 7 (1985): 1688-1703.

119. Schneider, K., and R. F. Zernicke. "Jerk-cost modulations during the practice of rapid arm movements." Biological Cybernetics 60, no. 3 (1989): 221-230.

120. Yogev, Galit, Meir Plotnik, Chava Peretz, Nir Giladi, and Jeffrey M. Hausdorff. "Gait asymmetry in patients with Parkinson’s disease and elderly fallers: when does the bilateral coordination of gait require attention?."Experimental brain research 177, no. 3 (2007): 336-346.

121. Hogan, Neville, and Dagmar Sternad. "Sensitivity of smoothness measures to movement duration, amplitude, and arrests." Journal of motor behavior 41, no. 6 (2009): 529-534.9

122. Yin, Yue H., Yuan J. Fan, and Li D. Xu. "EMG and EPP-integrated human–machine interface between the paralyzed and rehabilitation exoskeleton."Information Technology in Biomedicine, IEEE Transactions on 16, no. 4 (2012): 542-549.

123. Ferris, Daniel P., and Cara L. Lewis. "Robotic lower limb exoskeletons using proportional myoelectric control." In Engineering in Medicine and Biology Society, 2009. EMBC 2009. Annual International Conference of the IEEE, pp. 2119-2124. IEEE, 2009.