INVESTIGATING PROACTIVE BALANCE CONTROL IN INDIVIDUALS WITH INCOMPLETE SPINAL CORD INJURY WHILE WALKING ON A KNOWN SLIPPERY SURFACE

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ABSTRACT

Background: Falls are a growing concern among individuals with incomplete spinal cord injury (iSCI). As many as 83% of individuals with iSCI experience at least one fall per year. Most outdoor falls occur while walking on uneven or slippery surfaces. Individuals with iSCI employ proactive balance strategies to a greater extent than able-bodied (AB) individuals during normal walking, which is effective in reducing the intensity of an unexpected slip. Whether individuals with iSCI can use proactive balance strategies in a feedforward manner to adapt to expected slip perturbations and reduce slip/fall potential while walking has not been assessed.

Methods: 19 individuals with iSCI (AIS D; 14 males; 61.01 ± 17.67 years) and 17 age- and sex-matched AB individuals (13 males; 60.86 ± 17.79 years) were included in the study. Low-friction steel rollers were used to induce a slip in the anterior-posterior (AP) direction. Participants completed three walking conditions: normal walking trials (NW), one unexpected slip trial (US), and four expected slip (ES) trials. Changes in kinematic and electromyography (EMG) data were analyzed to give an indication of feedforward adaptations to the slip. Outcome variables included step width, step length, center of mass (COM) velocity, foot-floor angle, medial-lateral and AP margin of stability (MOS), maximum post-slip velocity (PSV), and integrated EMG for tibialis anterior (TA), soleus (SOL), and gluteus medius (GM) muscles.

Results: Individuals with iSCI used feedforward behavioural strategies to a greater extent than AB individuals while approaching a known slippery surface including walking with shorter steps, a flatter foot-floor angle, a more anteriorly positioned COM, and slower COM velocity. AB and iSCI groups made similar changes in their muscle activity to proactively prepare for the ES trials. The main difference between groups was a reduced ability of individuals with iSCI to proactively modulate the amplitude of the trail SOL muscle compared to AB individuals. Both AB and iSCI groups were able to make significant feedforward adjustments to behaviour and muscle activity within 1-2 trials after experiencing an US. These proactive balance strategies were effective at reducing the maximum PSV and thus the slip/fall potential in both groups.

Conclusions: Results demonstrate that individuals with iSCI maintain the ability to make appropriate feedforward changes in behavior and muscle activity and do so in a similar manner to AB individuals.

PREFACE AND AUTHOR CONTRIBUTIONS

I, Mackenzie Dior Bone, was the primary author of this thesis. Drs. Alison Oates and Gary Linassi were the co-supervisors of this thesis and Drs. Kristin Musselman and Joel Lanovaz were members on the research advisory committee. I thank Dr. Tarun Arora, Janelle Unger, and all of the research assistants working in the BBAM lab who completed data collection for this project. All custom MATLAB scripts were written by Dr. Joel Lanovaz.

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CONCEPTUAL DEFINITIONS

Center of mass: The location of the total body center of mass is the point on the body where the average center of mass for each body segment converges [1].

Center of pressure: The point location where all vertical ground reaction force vectors are acting along the support surface [1].

Base of support: The area of support formed beneath an object or person that includes anything in contact with the support surface and defines the possible range of movement of the center of pressure [1]–[3].

Posture: The physical orientation of body segments in relation to the global reference system, particularly in relation to the vertical gravitational force vector [1].

Balance/Postural Control: A generic term for the complex act of maintaining, achieving, or restoring equilibrium during any posture or activity which relies on the ability to control the position and movement of the center of mass in relation to the base of support [2], [4], [5].

Margin of Stability: The distance between the position of the velocity-dependent center of mass and the boundary of the base of support [3].

Fall: An event that results in a sudden, uncontrollable descent where the individual inadvertently comes to rest on any lower surface including the ground or floor [6].

LIST OF ABBREVIATIONS USED

AB	Able-bodied
ABC Scale	Activity Specific Balance Confidence Scale
AP	Anterior-posterior
APR	Automatic Postural Response
AIS	American Spinal Injury Association Impairment Scale
BOS	Base of Support
COM	Center of Mass
COP	Center of Pressure
CNS	Central Nervous System
EMG	Electromyography
ES	Expected Slip
GM	Gluteus Medius
iEMG	Integrated Electromyography
iSCI	Incomplete Spinal Cord Injury
Mini-BESTest	Mini Balance Evaluation Systems Test
ML	Medial-lateral
MOS	Margin of Stability
NW	Unperturbed Normal Walking
PSV	Post Slip Heel Velocity
PiEMG	Proactive Integrated Electromyography
RM ANOVA	Repeated Measures Analysis of Variance
RiEMG	Reactive Integrated Electromyography
SCATS	Spinal Cord Assessment Tool for Spastic Reflexes
SCI	Spinal Cord Injury
SCIFAP	Spinal Cord Injury Functional Ambulation Profile
SOL	Soleus
TA	Tibialis Anterior
TUG	Timed Up and Go
US	Unexpected Slip
WISCI II	Walking Index for Spinal Cord Injury II
xCOM	Extrapolated Center of Mass

LIST OF APPENDICES

Permission was obtained from Dr. Tarun Arora before using images found in Figure 2.1 showing a schematic representation of the experimental setup and slip device originally used in his study.

1. INTRODUCTION

Falls are a growing concern for ambulatory individuals with spinal cord injury (SCI) who have a greater incidence, prevalence, and consequences of falls compared to the able-bodied (AB) population [7]–[10]. Impairments in sensorimotor conduction due to SCI contributes to impaired balance control and lower extremity strength, which are believed to be main contributing factors to the increased rate of falls in this population [7]–[9]. It has been demonstrated that balance is context-dependent and is a flexible motor skill that can adapt and be modified with prior experience, knowledge, and intention [2], [4], [5], [11]–[13]. Balance has been identified as a modifiable risk factor for falls [4], [12], [13], which has led to a growing interest in efforts to better understand and improve balance control through motor learning and rehabilitation [2], [11], [14]–[16]. However, as we learn more about the control of balance it becomes increasingly clear that each population, and even each individual, with impaired balance control has a unique combination of constraints affecting them (i.e. strategy selection, response latencies, motor coordination, force control, ability to adapt etc.). Therefore, before we can effectively improve balance rehabilitation programs, we must deepen our understanding of the mechanisms behind balance control and identify which sub-components of balance control are most impaired in each population.

1.1 Anatomy of the Spinal Cord

Along with the brain, the spinal cord is part of the central nervous system (CNS) and is a highly organized and complex structure responsible for controlling the activities of the body. The spinal cord is a collection of ascending and descending tracts which integrate and modify inputs from sensory systems to influence motor output signals and regulate autonomic function [17]. Simply put, ascending pathways carry sensory information via afferent inputs in broad categories including pain and temperature, touch, and proprioception. The various sources of sensory information are integrated together via supraspinal centers to provide an indication of body position which is used to generate and continually modify motor commands [12], [18], [19]. These motor commands are carried via descending pathways to modulate motor output signals at the level of interneurons and motoneurons which ultimately influences behaviour [18]. The spinal cord is also responsible for the generation of simple reflexes such as the monosynaptic stretch reflex and more complex polysynaptic reflexes designed to act quickly, without conscious control, and protect us from injury [18].

The spinal cord extends from the brainstem to the first lumbar vertebra and is protected by three membranous layers (dura, arachnoid, and pia mater) and the bony vertebral column [17], [18]. The spinal cord consists of 31 segments and can be divided into cervical (8), thoracic (12), lumbar (5), sacral (5), and coccygeal (1) segments [17], [18]. The most distal part of the spinal cord is called the conus medullaris where lumbosacral nerve roots branch off to create a bundle of paired nerve roots called the cauda equina [17], [18]. The cellular organization of the spinal cord is unique compared to other structures of the CNS. The grey matter (neuronal cell bodies) of the spinal cord forms a distinguishable "H" shape and is surrounded by white matter (myelinated axons of neurons); furthermore; the ratio of grey to white matter varies at different levels of the spinal cord [17], [18]. The grey matter is divided into the dorsal horn, intermediate grey, ventral horn, and central grey matter. At the most distal part, where the white matter has tapered off, the spinal cord ends as a single mass of grey matter (conus medullaris) [18].

1.2 Spinal Cord Injury

The complex circuitry of the spinal cord is critically dependent on its ability to communicate with the brain. Any damage to the spinal cord can disrupt sensory, motor, and autonomic pathways. The disruption of sympathetic control due to SCI often results in features of autonomic dysregulation such as bradycardia, orthostatic hypotension, bowel and bladder dysfunction, and autonomic dysreflexia [20]. In contrast, the disruption of sensorimotor signalling due to SCI is thought to contribute to impaired balance control which has a negative influence on functional mobility and independence in this population [7]–[9]. It is important to note that although individuals with SCI can improve their levels of functioning, particularly in the first three to six months post-injury, the injury to the spinal cord itself is still irreversible [20].

It has been shown that the extent of impairments resulting from a SCI depends on the neurological level and completeness/severity of the injury and that this information can be used to predict motor recovery from the onset of injury [17], [21]. The neurological level of injury is defined by Kirshblum (2011) as the most caudal segment of the spinal cord with normal sensory and antigravity motor function on both sides of the body [17]. The neurological level of injury is used to determine whether an individual is classified as having tetraplegia or paraplegia [17]. Tetraplegia results from damage to cervical segments and involves impairments to both upper and lower extremities and the trunk [17]. In contrast, paraplegia results from damage to thoracic, lumbar, or sacral segments, the conus medullaris, or cauda equina and involves impairments to lower extremities with sparing of upper extremity function [17].

The severity of the injury depends on how complete the damage to the spinal cord is, which can be graded using the American Spinal Injury Association Impairment Scale (AIS). According to the International Standards of Neurological Classification of Spinal Cord Injury, each SCI is given a grade of A, B, C, D, or E that reflects the completeness of both motor and sensory impairments [17]. Table 1.1 summarizes each AIS grade and how they are characterized. A complete SCI is an injury to the spinal cord where there is no remaining motor or sensory function below the neurological level of injury, while an incomplete SCI (iSCI) is an injury to the spinal cord with varying preservation of sensory or motor function below the neurological level of injury which includes sacral sparing of S4-S5 [17]. Presence of sacral sparing can include sensation at the anal musculocutaneous junction or deep anal pressure and/or the presence of voluntary contraction of the external anal sphincter which can be tested using a digital rectal examination [17].

The specific location of the lesion dictates which spinal tracts and types of neurons are damaged after SCI which influences the clinical symptoms that are seen. For instance, damage to the dorsal column, a major ascending tract of the spinal cord, causes impairment in the sensations of light touch, vibration, and proprioception [17], [18], [20]. While damage to the anterolateral or lateral spinothalamic tracts, cause impairments in temperature and pain sensation [17], [18], [20]. In contrast, damage to the corticospinal or rubrospinal tracts, two main descending motor tracts would likely cause impairments in the fine control of movement including muscle tone and postural control. Specifically, damage to the corticospinal tract, affects mainly upper motor neurons while damage to the cauda equina (part of the peripheral nervous system) mainly affects lower motor neurons [17], [18], [20].

Upper motor neurons originate in supraspinal centers such as the motor cortex or the brain stem and carry information to lower motor neurons, which innervate different types of muscle fibers to elicit muscle contraction [18].

Table 1.1 ASIA Impairment Scale (AIS) designation used in grading the degree of impairment

Grade	Type of Injury	Conditions
A	Complete	No sensory or motor function preserved in the sacral segments (S4-5)
B	Sensory Incomplete	Sensory but not motor function is preserved below the neurological level (includes S4-5), AND no motor function is preserved more than 3 levels below the motor level on either side of the body
C	Motor Incomplete	Motor function is preserved below the neurological level More than ½ of key muscles below the level have a muscle grade < 3
D	Motor Incomplete	Motor function is preserved below the neurological level At least ½ of key muscles below the level have a muscle grade > 3
E	Normal	Sensory and motor function are graded as normal in all segments

** For an individual to receive grade C/D they must have sacral sparing which is defined as voluntary anal sphincter contraction or deep anal pressure (DAP) at spinal level S4-5 with sparing of motor function more than 3 levels below the motor level

Note: information provided in table was drawn from Kirshblum and colleagues, 2011

Damage to upper motor neurons results in generalized weakness, spastic paresis, hyperactive stretch reflex, increased tone, clonus, positive Babinski sign (plantar reflex response is upgoing), and absent superficial reflexes [17], [20]. In contrast, damage to lower motor neurons results in weakness that is more notable in distal muscles than in proximal muscles, flaccid paralysis, atrophy of muscles, hypotonic or absent deep tendon reflex and Babinski sign, and the presence of muscle fasciculations [17], [20]. Depending on the location of the SCI, specific syndromes have been identified in which unique sets of symptoms are observed including central cord, Brown-Sequard, anterior cord, posterior cord, conus medullaris, and cauda equina syndrome [17], [20]. A summary of these syndromes including common causes, location of lesion/tracts involved, and characteristic symptoms is provided in Table 1.2.

Table 1.2 Summary of common syndromes after spinal cord injury

Syndrome	Common Causes	Location of Lesion	Characteristic Symptoms
Central Cord Syndrome	Most common, falls which cause hyperextension injuries and pre-existing myelopathies	Center of cervical cord	Greater weakness in the upper extremities than in the lower extremities
Brown-Sequard Syndrome	Traumatic causes such as knife/bullet wounds	Spinal cord hemisection	Ipsilateral paresis and loss of touch, proprioception, vibration, and motor control Contralateral loss of pain and temperature sensation
Anterior Cord Syndrome	Relatively rare, decreased or absent blood supply to the spinal artery	Anterior two thirds of the spinal cord - Dorsal columns are spared but corticospinal and spinothalamic tracts are damaged	Loss of motor function, pain and temperature sensation at and below the injury Light touch and joint position sense are preserved
Posterior Cord Syndrome	Rarely occurs	Dorsal column lesion	Loss of touch, vibration and proprioception Motor function, pain, and temperature sensation are preserved
Cauda Equina Syndrome		Lumbosacral nerve roots (lower motor neurons) - may spare the spinal cord	Flaccid paralysis of the lower extremities, areflexic bowel and bladder, impaired sensation (partial/complete loss), and sacral reflexes (bulbocavernosus and anal wink) will be absent
Conus Medullaris Syndrome		Similar to cauda equina but injury is more rostral (L1/2)	Mixed picture of upper (conus injury) and lower (nerve root injury) motor neuron symptoms

Note: information provided in table was drawn from Kirshblum and colleagues, 2011

1.3 Epidemiology and Consequences of Spinal Cord Injury

Over 86,000 Canadians have a SCI and of that 70-80% have an incomplete spinal cord injury (iSCI) [22], [23]. An estimated 51% of all SCIs are from traumatic causes and 49% are from non-traumatic causes [22], [23]. In 2015, the estimated incidence of traumatic SCI in Canada was 3.6 people per million [24]. Traumatic SCI is most commonly caused by falls, motor vehicle accidents, sporting injuries, acts of violence, and surgical complications [24]–[26], while non-traumatic injuries often result from medical surgical complications, infections, ischemia, tumours, and spondylosis [20], [22]. Binge drinking, prescription medication use, spasticity, and several personality characteristics including impulsivity, risk-taking, neuroticism, and aggression have been identified as risk factors for traumatic SCI and subsequent injuries [9], [25], [27]. Ultimately, the physical, psychological, social, and economic burden of SCI contribute to a reduced quality of life and an increased rate of depression in this population [26], [28].

The medical consequences and complications related to SCI result in a large economic impact on both the individual and their family and on the Canadian health care system. The costs related to SCI result from both direct (treatment of the initial SCI, hospital stays, acute rehabilitation, etc.) and indirect (home care, purchase of assistive devices, treatment of secondary complications, loss of productivity, etc.) expenses [23], [29]. According to a six year follow-up study individuals with SCI are rehospitalized 2.6 times more often and required 30 times more hours of home care services than AB individuals [26]. Additionally, it has been shown that approximately 91% of individuals with iSCI experienced at least one medical complication six months after their initial SCI [7]. Due to these direct and indirect expenses, an individual with SCI in Canada will spend between $1.5 and $3 million on SCI-related costs over their lifetime [29]. Moreover, it is estimated that Canada will spend approximately $3 billion dollars per year on SCI-related costs [23], [29].

The most common complications treated after SCI include urinary tract infections, pressure ulcers, pneumonia, autonomic dysreflexia, neuropathic pain, fractures from falls, and deep venous thrombosis [7], [23], [26], [30]. These secondary health complications, as well as the underlying sensory and motor impairments from a SCI can contribute to diminished balance control and a high incidence of falls seen in the SCI population. Reported fall rates for ambulatory individuals with SCI range from 73-83% having fallen at least once over one year [7]–[10]. Moreover, 28-68% of individuals reported recurrent falls (> 1 fall) in one year and 65%

of individuals experienced at least one fall that caused injury [9], [10], [31]. For ambulatory individuals with SCI, it was shown that indoor and outdoor falls were equally likely to occur [9]. Most outdoor falls occurred while walking on uneven or slippery surfaces [9], [10], [31], which is likely a result of impaired balance control caused by underlying neurological and musculoskeletal problems (internal factors), inattention and distraction (behavioural factors), and destabilizing or unexpected conditions (external factors) [7]–[10], [31].

The incidence and consequences of falls for individuals with iSCI is comparatively greater than fall rates for elderly individuals (65 years and older; 25-35%), frail elderly individuals (80 years and older; 40-50%), individuals with Parkinson's disease (38-62%), and individuals with peripheral neuropathy (50%), which highlights the growing concern for effective falls prevention for individuals with iSCI [8], [10]. Individuals with iSCI perceived that mostly extrinsic factors (i.e. hazards in the environment) contributed to their challenges and falls but that some intrinsic factors (i.e. reduced strength, spasticity, fatigue, inattentiveness, poor balance) also contributed [8], [10], [32]. Impaired balance control, limited mobility, and injuries from falls can result in a fear of falling leading to subsequent limitation of activity, loss of independence, and restriction of community participation [7]–[10], [32], [33]. This fear of falling and loss of independence ultimately contributes to reports of low subjective well-being and overall quality of life compared to AB individuals [28].

Finding strategies to improve functional mobility and balance control, reduce falls, and increase participation in daily life have been identified as top priorities by individuals with iSCI for improving their overall quality of life [8], [32], [34], [35]. Implementing early interventions and perturbation-based balance training into rehabilitation may be an effective way of reducing the prevalence of falls in individuals with iSCI which would improve their quality of life and reduce the economic burden associated with fall-related injuries and rehospitalizations.

1.4 Balance Control

Balance control is a complex motor skill that is essential for the safe execution of daily tasks. The functional goals of the balance system include the maintenance and adjustment of posture, facilitation of voluntary movement, and the ability to recover equilibrium [2], [36], [37]. Additionally, according to the systems framework proposed by Horak (2006), effective postural control requires six essential resources: biomechanical constraints, movement strategies, sensory strategies, orientation in space, control of dynamics, and cognitive processing [12]. Damage to any one of these systems that are involved with the control of these postural resources will result in specific balance impairments. One of the main objectives of this thesis is to compare the balance control of those with iSCI to AB individuals. The underlying neurological and musculoskeletal impairments due to SCI influence many of the postural resources identified by Horak (2006); however, this thesis will mainly provide insight into the way individuals with iSCI adapt their control of dynamic balance, sensory strategies, and movement strategies to compensate for underlying impairments from SCI [12]. Therefore, the remainder of this thesis will focus on these three aspects of balance control.

Control of Dynamics

Standing balance is achieved by voluntarily moving one's center of pressure (COP) underneath the feet to maintain the vertical projection of the center of mass (COM) within the boundaries of the base of support (BOS) [3], [5]. However, this condition of stability does not always apply in dynamic situations such as locomotion where the velocity and position of the COM needs to be considered differently. During forward progression in locomotion, the COM velocity and position is directed outwards which creates a situation in which the COM is constantly 'falling' outside of the BOS [3], [5]. In this situation, to prevent a fall from occurring, the size of the BOS must be increased by taking a step and 'catching' the COM with the placement of the swing foot [3]–[5], [12]. Thus, maintaining balance during dynamic situations such as locomotion relies on the ability to keep the extrapolated, velocity-dependent center of mass (xCOM) within the changing BOS; when this can not be done, a fall is likely to occur [2], [3], [5], [12], [38].

Healthy individuals mainly control the position of the COP in the anterior-posterior (AP) direction using the ankle plantar flexors (gastrocnemius and soleus) and dorsiflexors (tibialis anterior) and in the medial-lateral (ML) direction using the hip abductors (gluteus maximus and medius) and adductors (obturator externus, adductor brevis, longus, and magnus) [1], [39]. More specifically, with respect to abductors, the gluteus medius (GM) is the main antigravity muscle during gait [40] and the magnitude of its activity has been shown to be correlated with frontal plane pelvis control and knee stability during transitions from double- to single-leg stance [41]. Therefore, for this thesis, muscle activity was measured in the tibialis anterior (TA) and soleus

(SOL) muscles to give an indication of balance control in the AP direction, and the GM muscles were measured to give an indication of the ML control of balance.

Sensory Strategies

Balance is achieved through the complex integration and coordination of sensory information from the vestibular, visual, and somatosensory systems [12]. Posturography studies have shown that during normal, unperturbed standing, AB individuals predominately rely on somatosensory cues the most for postural orientation (70%), followed by vestibular cues (20%), and visual cues (10%) [12]. In situations where certain sensory information has been removed, environmental characteristics have been changed, or when sensory information is inaccurate or untrustworthy, healthy individuals reweight the contribution of each sensory system [19]. For instance, somatosensory input is rendered unreliable when walking on a compliant surface, so healthy individuals will reweight the contributions of sensory inputs to rely more on visual and vestibular cues to maintain postural orientation [19]. The ability to effectively integrate information from sensory systems and efficiently use this to modify motor outputs, as well as the ability to reweight the contributions of sensory inputs, is critical for making appropriate postural adjustments to maintain equilibrium and achieve safe ambulation.

Movement Strategies

Balance control, including postural responses and movement strategies, can be studied by providing a perturbation such as a slip, which displaces the individual's BOS under their COM [11], [13], [38], [42]. After simple monosynaptic reflexes (30-40 ms), automatic postural responses (APRs) respond to external perturbations and are triggered within 70-180 ms of perceiving the perturbation [4], [11], [12]. An advantage of APRs is that, unlike voluntary motor responses, they are not limited by the inherent time delays caused by the electromechanical conduction velocities needed to send, interpret, and elicit movement based on sensory information [4], [11]. These APRs produce consistent, characteristic patterns of muscle activity that are shaped by the CNS to maintain balance during both self-initiated movements and in situations where external disturbances cause a threat to stability [4], [11].

The hip and ankle strategies are two distinct APRs used in static conditions to control the COM position within a fixed BOS [43]. The ankle and hip strategies are centrally programmed and selected for based on both external factors (i.e. environmental constraints) and internal factors such as previous experience, knowledge, and goals [4], [5], [11], [12], [43]. Healthy individuals are able to scale the size of their APRs to match the intensity of a perturbation [11], [43]. However, many neurological conditions affect the ability to modulate the size or onset of these APRs due to impaired sensory and/or motor signalling [11], [43] so these populations often use alternative strategies to prevent a fall. Moving the trunk or arms with respect to the COM is a common whole-body strategy used to maintain the COM within a fixed BOS [3]. When balance cannot be maintained without changing the size of the BOS the individual could take a compensatory step, grab hold of a nearby stationary object, or use an assistive device such as a cane or walker to increase the size of the BOS [3], [43].

During dynamic situations such as locomotion, where the BOS is constantly moving and changing size, the hip and ankle strategies are not enough to maintain stability on their own. Therefore, dynamic stability is maintained through both feedforward (proactive and anticipatory) and feedback (reactive) control of balance [4], [5], [13], [44]. Reactive balance strategies are employed *in response* to an unexpected perturbation, such as a slip [2], [4], [5], [11]. In contrast, proactive strategies are employed *before* contact with an expected perturbation (slippery surface) and involve anticipatory and predictive changes in muscle activity or gait pattern to prepare for contact with the destabilizing surface [2], [4], [5], [11].

Research has investigated how AB individuals react to an unexpected slip and how they proactively adapt their gait in anticipation of an expected slippery surface to maintain stability and prevent a fall [13], [15], [16], [38], [42]. When exposed to an unexpected slip while walking, common reactive balance strategies for AB individuals include increased arm swing, trunk rotations, a rapid activation of lower extremity muscles that is scaled to the perturbation intensity, and compensatory stepping [13], [15], [45]. However, even for AB individuals, it has been suggested that proactive balance strategies are more efficient than reactive responses since they are not fully reliant on sensory information and thus aren't as prone to electromechanical delay [44].

Proactive behavioural adjustments are dependent on visual information (anticipatory) and prior experience or knowledge with the surface conditions (predictive) [5], [13], [46], [47]. Proactive balance control is thought to be guided by supraspinal signalling from the brain stem, cerebellum, and motor cortex which can be further modulated via descending (i.e. corticospinal) signalling [47], [48]. Knowledge or visual information about the location, size, shape, and surface characteristics of a destabilizing surface can influence which balance strategies an individual will employ [5], [13], [46], [47].

A common proactive strategy is to avoid stepping on an undesirable landing surface altogether by either turning around or using alternate foot placement. It has been shown that even something as seemingly rudimentary as alternate foot placement is guided by specific selection strategies which consistently produce dominant stepping outcomes in a variety of different situations [46]. These selection strategies seek to minimize the effort needed to safely avoid an undesirable surface and the CNS achieves this by preferentially selecting foot placement positions that; (1) maintain the foot on its course in the plane of progression; (2) require longer as opposed to shorter steps; and (3) place the foot more medially as opposed to laterally [46]. However, when it is not possible to avoid stepping on an undesirable surface, feedforward adaptations to gait are made to minimize the threat to dynamic stability and avoid a fall. Feedforward adaptations commonly made by AB individuals before walking on a known slippery surface include reduced walking speed, shortened step length, increased step width, reduced foot-floor angle, an anterior shift in COM position, as well as increased muscle activity and ankle co-contraction [13], [15], [16], [42], [49]–[54].

Both proactive and reactive balance control have been shown to reduce the intensity of a slip perturbation and thus the fall potential in young and older healthy adults [13], [15], [16]. Improvements in pre-slip stability due to feedforward adaptations from a single acquisition session (consisting of five slips) can be retained for up to 12 months [49], [55], [56], while improvements in post-slip stability can be retained up to 4 months later, resulting in a decreased incidence of falls [55]. Additionally, it has been shown that adaptive feedforward responses acquired from one type of task-specific perturbation training such as platform translations, can be applied to reduce the perturbation intensity in a different paradigm such as a slip [50]. Taken together, these findings provide support for the use of task-specific, motor learning-based balance training to facilitate the use of proactive balance strategies and reduce the intensity of unexpected slips and thus the fall potential in vulnerable populations. Preliminary studies have indeed shown that perturbation-based balance training is effective at reducing fall risk among healthy young and older adults, individuals with Parkinson’s disease, and individuals who have had a stroke [56]–[59].

1.5 Balance Control in Individuals with Incomplete Spinal Cord Injury

Impairment in sensory signalling can compromise an individual's ability to perform sensory reweighting which is important for maintaining balance in contexts where certain types of sensory information may be limited or unavailable [12], [19]. For instance, individuals with iSCI often have impaired somatosensation (touch, pressure, proprioception) and thus have a higher risk of falling when visual or vestibular cues become distorted or unavailable [12], [19]. In addition to somatosensory impairments, individuals with iSCI have been shown to have reduced lower extremity strength which makes it more difficult to counteract gravitational forces and implement responsive coordinated multi-joint movements [8], [20]. It is important to determine which aspects of balance control these impairments in somatosensation and strength may impact, and to what extent they limit the functional mobility of individuals with iSCI.

Individuals with iSCI have been shown to have increased postural sway compared to AB individuals during quiet standing tasks [60], [61] and reduced precision during dynamic standing tasks such as reaching or leaning movements [62]. Individuals with iSCI show increased postural sway during quiet standing with the eyes closed, demonstrating a greater reliance on vision compared to AB individuals [61]. This greater reliance on vision is likely a result of impaired somatosensation. Despite increased postural sway during standing tasks, it has been shown that sway can be improved through the addition of haptic input such as the use of light touch (less than 1 N) on a railing [60]. These results suggest that individuals with iSCI have the ability to integrate additional sensory input from intact sensory systems to compensate for reduced inputs from impaired sensory systems. Additionally, individuals with iSCI are capable of quickly (within 1-2 exposures) adapting their muscle activation responses to repeated surface perturbations while standing, similar to AB individuals; however, they showed delayed muscle activation onset and reduced magnitude of muscle activity [63]. Taken together, these results suggest that individuals with iSCI have the ability to make adequate proactive and reactive postural adjustments in response to perturbations while standing, but that the mechanisms (muscle onset times, latencies, and magnitudes) behind these adjustments are different than AB individuals due to underlying sensorimotor impairments.

Recent literature investigating walking balance among individuals with iSCI has shown that this population has a limited ability to regain stability in the lateral plane with a compensatory step, slower onset of TA muscle activity, and reduced magnitude of SOL muscle activity compared to AB individuals when responding to an unexpected slip perturbation [64]. This indicates that individuals with iSCI have impaired reactive balance control compared to AB individuals. Despite an increased incidence of falling, postural instability during quiet and dynamic standing, and impaired reactive balance control while walking, ambulatory individuals with iSCI have been shown to have greater stability during normal walking compared to AB individuals [65]–[67]. It has been demonstrated that individuals with iSCI require greater disturbing forces to become destabilized during normal walking compared to AB individuals, suggesting that they have greater walking stability [67]. This greater walking stability is believed to be mainly modulated by a slower walking speed, shorter steps, and more time spent in the double-support phase of gait [65], [67]. It is possible that this greater stability seen during normal walking indicates that individuals with iSCI use proactive balance strategies to compensate for underlying musculoskeletal and sensorimotor impairments which limit their ability to effectively employ reactive balance strategies.

Although it is known that individuals with iSCI can adapt to perturbations while standing and may use more proactive strategies than AB individuals during normal walking, it is unknown whether this population is able to use proactive balance strategies to adapt to repeated perturbations while walking. Additionally, it is unclear how quickly adaptation would occur compared to AB individuals and whether these adaptive strategies would be effective at preventing a hazardous slip/fall when faced with a destabilizing condition, such as a slippery surface. A slippery surface reduces the ease with which an individual can effectively control the COP position and thus hinders the main mechanism used for maintaining the COM position within the boundaries of the BOS. Therefore, studying the slip response of individuals with iSCI provides insight into whether individuals with iSCI are able to use residual sensorimotor function or compensatory strategies to adapt to a perturbation that directly challenges their walking balance. Addressing this gap in knowledge is key to learning how to improve dynamic balance control and prevent falls in the iSCI population and may help inform therapists as to how we can tailor existing balance training protocols to provide specialized rehabilitation for individuals with iSCI.

1.6 Intervention Programs and Rehabilitation Therapies

Perturbation studies have shown that movement strategies are centrally programmed and can be influenced by past experiences, environmental context, and expectations [4], [5], [11], [12]. Therefore, applying concepts from motor learning including repetition and task-specific experience may be of use for improving balance control and preventing falls in conditions that are commonly hazardous [11], [68], [69]. More specifically, individuals with SCI could be taught to identify and avoid dangerous tasks and situations based on their unique set of balance constraints. Perhaps activities that individuals already limit/avoid can be used as a starting place to guide task-specific rehabilitation. Some difficult tasks identified by ambulatory individuals with iSCI that are often avoided include the negotiation of obstacles, opening/closing doors, carrying objects, walking on uneven, sloped, or slippery surfaces, in crowded areas, narrow spaces, and on steps [33]. However, when faced with a hazardous situation that they are not able to avoid, it is possible that these individuals could be trained how to properly use compensatory movement strategies or assistive devices to control their COM despite underlying sensorimotor impairments. Furthermore, based on the principles of motor learning, individuals should be provided with repeated exposures to a variety of different hazardous situations in rehabilitation so that they can practice using appropriate postural responses and applying these effective movement strategies to many different contexts [50].

The main clinical factors that have been identified as predictors of future walking level and performance for individuals with iSCI include total body strength, balance, spasticity, and age [70]. Most of the functional recovery that is possible occurs within the first 3-6 months after a SCI [20], [71], which makes it of particular importance that individuals begin frequent, personalized, task-specific rehabilitation as early as possible [72]. Currently, the most common rehabilitation techniques being used to improve stepping performance for individuals with iSCI include treadmill training with body weight support [73]–[76], walking with the help of robotic exoskeletons [77], [78], walking with functional electrical stimulation (often of the peroneal nerve or gluteus muscles) [73], [78], and general strength and balance training [79]. Locomotor training in particular has been shown to improve lower extremity strength, functional walking speed, and reduce reliance on assistive devices for individuals with iSCI [73], [75], [77].

Although these intervention programs appear to improve stepping performance, they do not appear to have any significant advantages over conventional physiotherapy [68], [69], [78].

Additionally, improved stepping does not necessarily lead to improvements in functional ambulation since underlying impairments in stability and balance may limit further improvement [4], [68]. Thus, it makes sense that rehabilitation programs which include balance training have been shown to be the most effective at reducing falls in older adults and have been recommended for use in clinical populations with high rates of falling [14], [57], [79]. However, the heterogeneous nature of SCI and our limited knowledge of balance strategies used among individuals with iSCI while walking, constrains our ability to develop effective fall-prevention programs and interventions for this population.

1.7 Balance Assessments

In populations with a high risk of falls such as individuals with iSCI it is crucial to identify balance assessment tools that are able to pinpoint the unique balance constraints of each individual to properly guide rehabilitation programs, track improvements in balance control and mobility throughout rehabilitation, and that are able to reliably predict fall risk. This is especially important for such a heterogeneous population like those with SCI because even two individuals with the same lesion could present with a completely different set of symptoms [21]. The ideal balance assessment tool should be able to comprehensively assess all domains of balance, have good clinical utility (cost effective and easy to administer), and be psychometrically sound (high validity, reliability, and responsiveness) [64], [80]. There are a number of measures validated for the SCI population that are used to quantify different clinical outcomes including balance control, balance confidence, and functional mobility.

Common balance evaluation tests include the Berg Balance Scale (BBS) [81], Mini-Balance Evaluation Systems Test (Mini-BESTest) [82], [83], and the Timed Up and Go (TUG) test. The Activities Specific Balance Confidence Scale (ABC) is the most common test used to evaluate the individual's perception of and confidence in their own balance [84], [85]. Finally, common functional mobility tests include the Functional Reach Test, 10-meter Walk Test (10 MWT), 6-Minute Walk Test (6 MWT), Timed Up and Go (TUG), Spinal Cord Injury Functional Ambulation Profile (SCI-FAP) [86], [87], and the Walking Index for Spinal Cord Injury (WISCI II) [88], [89] [64], [80]. The 10 MWT, 6 MWT, TUG, and the WISCI II have been shown to have the most valid and reliable measures of improvements in ambulation over time [90], [91].

Additionally, a recent systematic review, the MiniBESTest was identified as the single most comprehensive and clinically useful assessment of balance for individuals with iSCI [80].

One drawback to many of the balance assessments mentioned above is their inherent subjectivity due to use of Likert scales and reliance of the assessor on observation. Additionally, many of these assessments are subject to ceiling effects which limits sensitivity to small changes in function. To overcome these limitations, many new objective techniques are being introduced to the field of balance assessment including wearable inertial sensors (gyroscopes and accelerometers) and biomechanical gait analysis which takes advantage of technology like force plates, 3D motion capture, and electromyography (EMG) to identify slight balance impairments or compensatory strategies that may not be visible to the naked eye [36], [68], [92]. That being said, these objective assessment techniques often require expensive equipment and trained personnel which limits their clinical utility. Therefore, a combination of clinical measures and new technology (3D motion capture, force plate, and EMG) were used in this thesis so that results may be translatable to both biomechanical research and clinical applications.

1.8 Objectives and Hypotheses

In summary, the high prevalence and consequence of falling is a growing concern for individuals with iSCI. There are gaps in the literature regarding the understanding of walking balance control in this population which may provide insights into how we can best supplement current rehabilitation programs to minimize the number of falls in this population. The primary objectives of this thesis were: (1) to compare the proactive balance of individuals with chronic iSCI to age- and sex-matched AB individuals when walking over a known slippery surface; (2) to determine how the feedforward adaptations of individuals with iSCI change over repeated exposures to a known slippery surface and how they compare to AB individuals; and, (3) to determine the relationship between slip intensity and various clinical measures of strength, balance, and walking function.

Since individuals with iSCI walk at a slower speed and use more proactive balance strategies than AB individuals during normal walking [65]–[67], it was hypothesized that, compared to their AB peers, individuals with iSCI would continue to employ even more pronounced proactive balance strategies when approaching a known slippery surface. In particular, since a slip directly causes a shift in the BOS, it was hypothesized that their gait would be adapted to increase step width to increase the size of the BOS, reduce step length and foot-floor angle to minimize horizontal shear forces upon contact with the slippery surface, and maintain their COM stability without stopping forward progression of locomotion. The common feedforward behavioural adaptations we expected to see included reduced walking speed, reduced step length, increased step width, reduced foot-floor angle, a greater AP MOS indicating an anterior shift in the xCOM, and increased muscle activity when approaching the known slippery surface [15], [16], [42].

Secondly, similar to balance adaptation seen during standing, it was hypothesized that individuals with iSCI would require at least one to two experiences walking on the known slippery surface before showing significant feedforward adaptations, which would be comparable to the rate of adaptation seen in AB individuals [63]. In line with this, we expected to see a shift from a purely reactive strategy in response to the unexpected slip trial, to more of a proactive strategy during the expected, feedback-based slip trials. Proactive changes to balance would include changes to both the feedforward (before contact with the slippery surface) and feedback (immediately after contact with the slippery surface) control of balance and locomotion [42], [49], [56], [93], [94]. Finally, since previous literature shows a relationship between greater slip/fall potential and impaired sensorimotor functioning, balance capabilities, lower extremity strength, reduced functional mobility, and greater spasticity [7]–[9], [32], [70], [95]; it was hypothesized that clinical scores for cutaneous pressure sensation, proprioceptive ability, muscle strength, spasticity, functional walking, and balance (particularly the sub-component of the mini-BESTest that measures anticipatory balance control) would be significantly able to predict changes in slip intensity, measured functionally as the change in maximum post-slip heel velocity (PSV).

Results from this study will help fill a knowledge gap regarding walking balance control in individuals with iSCI and their ability to adapt to expected slip perturbations. This information will be of critical use for determining which balance strategies can be used effectively by individuals with iSCI and which aspects of balance control are impaired compared to AB individuals. Knowledge of the capabilities and limitations of dynamic balance control for individuals with iSCI could be used as a starting point to guide task-specific training and the development of effective fall-prevention programs and interventions.

2. METHODS

2.1 Participants

Twenty-six individuals with chronic iSCI were recruited through advertisements in regional health centres across Saskatchewan including Spinal Cord Injury Saskatchewan and also through a mail out to patients of local physiatrists. Participants were included if (1) they were 18 years of age or older at the time of the study; (2) their injury was classified as grade C or D on the American Spinal Injury Association Impairment Scale (AIS) [17]; (3) they were a minimum of one year post-SCI; and (4) they were able to walk independently for 10 m (braces used during daily walking were permitted). Able-bodied individuals were matched by sex and age (± 5 years) to the participants with iSCI and were recruited from the local community through advertisements and word of mouth. Participants were excluded if they had any injury, disease, or condition that could affect their walking or balance control (e.g. vestibular impairment, joint pain, fracture, etc.). This study was approved by the University of Saskatchewan's research ethics board and consent was obtained before testing.

2.2 Experimental Setup

Participants wore their own comfortable, closed-toe shoes and were secured in a safety harness for fall-prevention that allowed them to move freely along a 10 m walkway. The height of the harness was modified for each participant so if they were to fall, their knees would not touch the ground. Slips were induced using a set of low-friction steel rollers (0.46 x 0.51 m; coefficient of static friction in unlocked state = 0.09) which were embedded in the middle of the walkway, flush with the floor surface (Figure 2.1). For normal walking, the rollers were locked in place but could be unlocked without any visible changes to promote a slip in the antero-posterior (AP) direction upon foot contact. Two force plates (0.46 × 0.51 m, AMTI OR6-7, Advanced Mechanical Technology, Inc., Watertown, MA) were embedded in the walkway (one under the slip device and the other diagonally adjacent to the slip device) (Figure 2.1) and collected ground reaction forces (fs = 2000 Hz) which were used to detect foot-contact and foot-off events for each gait cycle [96], [97].

A wireless EMG (2400GT2, Noraxon Inc, Scottsdale, AZ) system was used to measure muscle activity (fs = 2000 Hz) to study the neuromuscular characteristics of the tibialis anterior (TA), soleus (SOL), and gluteus medius (GM) muscle responses bilaterally. The skin surface was shaved and cleaned with alcohol swabs to remove as much resistance as possible before placing the electrodes. The bipolar surface electrodes were carefully placed on the center of each muscle belly to minimize cross talk from adjacent muscles. Accurate electrode placement was confirmed with voluntary isometric contractions. Additionally, EMG signals were monitored throughout the experimental protocol to ensure that the EMG data was being properly recorded and that no signals disappeared or looked considerably inaccurate or noisy.

The lab was equipped with an eight-camera 3-D motion capture system (Vicon Nexus, Vicon Motion Systems, Centennial, CO) that collected kinematic data at a sampling rate of 100 Hz (Figure 1). The cameras were calibrated before each data collection to ensure accurate tracking of the kinematic marker set. The marker set consisted of 63 reflective markers (14 mm diameter, 22 calibration and 41 non-calibration) and was used to collect kinematic information from 12 segments: the head, trunk, and the right and left upper arms, forearms, thighs, shanks, and feet. The markers were placed on the participant at various anatomical landmarks, the locations of which are summarized in Table 2.1. This experimental set-up is also described in previous studies by Arora and colleagues (2018, 2019).

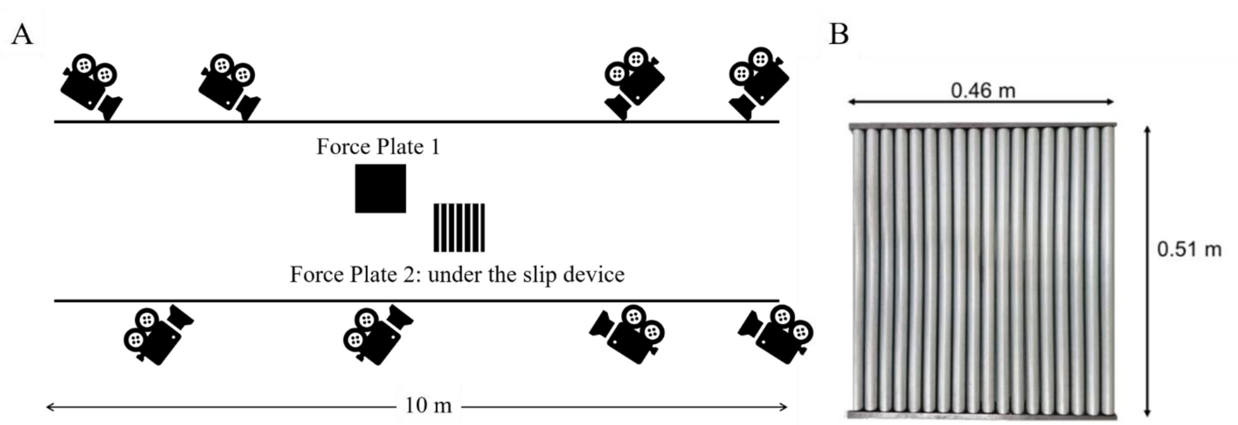

Figure 2.1 Schematic representation of the experimental set-up including the arrangement of eight infrared motion capture cameras, the ten-meter walkway, two force plates, and the slip device (A). Overhead view of the low friction steel rollers used to induce a slip during data collection (B). This figure was used with permission and modified from Arora et al., 2018 (see Appendix B).

Table 2.1 Anatomical landmarks used for placement of reflective markers

Body Segment	Non-Calibration Markers	Calibration Markers
Head	3 markers: center of the forehead and above the left and right ear, level with the forehead marker	
Trunk		
Back	3 markers: C7 vertebral prominens and the inferior angle of the scapula on both sides	
Thorax		5 markers: suprasternal notch, xiphoid process, laterally on the $7^{th}/8^{th}$ rib on both sides in line with the xiphoid process marker, and a marker on the spine level with the lateral rib markers
Waist		3 markers: anterior superior iliac spine on both sides and midway between and level with the iliac spine markers on the anterior surface
Pelvis	4 markers: fixed on a rigid rectangular cluster, fastened in the approximate location of the pelvis using an adjustable belt	
Arms		
Shoulders	2 markers: acromioclavicular joint on each shoulder	
Elbows	2 markers: lateral epicondyle of each humerus	
Wrists	2 markers: styloid process of the ulna of each wrist	
Finger	1 marker: nail of dominant index finger	
Legs		
Thigh	8 markers: cluster of 4 markers fixed on a rigid rectangular cluster, fastened approximately midway down the lateral side of the femur	
Knee		4 markers: medial and lateral femoral epicondyles of each knee
Shank	8 markers: cluster of 4 markers fixed on a rigid rectangular cluster, fastened approximately midway down the lateral side of the tibia	
Ankle		4 markers: medial and lateral malleoli of each ankle
Feet		
Foot Cluster	6 markers: cluster of 3 markers placed on the lateral side of each foot, arranged in a non-collinear manner	
Toe Markers		6 markers: most anterior point on each shoe (tip of 1^{st} distal phalanx), the 2^{nd} metatarsal on each shoe, and the lateral aspect of the 5^{th} metatarsal on each shoe
Heel	2 markers: the calcaneus (heel) of each foot	
Total	**41**	**22**

2.3 Experimental Protocol

Testing was completed over two days to avoid fatigue. The first day the participant came into the lab they completed a series of clinical testing which was administered by a trained researcher with a background in physiotherapy. The participants were asked a series of questions regarding their SCI to determine the neurological level of their injury, if they are classified as paraplegic or tetraplegic, their grade on the AIS, the cause of their iSCI (e.g. traumatic or non-traumatic), and the amount of years since their injury. If they did not remember this information, then the participant's medical records were consulted with their consent. They were also asked "Do you use an assistive device and/or brace while walking indoors? Outdoors?", "Have you fallen in the past year? How many times?", and "Do you have a fear of falling, defined as a lasting concern about falling causing you to avoid or curtail activities you felt you were capable of doing?" [98].

Next, to give an indication of dorsal column functioning and it's effect on walking balance control in individuals with iSCI, proprioceptive ability was assessed in the lower extremities using a passive joint position sense test [99], [100], and cutaneous pressure sensation was assessed using Semmes-Weinstein monofilament testing [101]. Proprioceptive ability was tested bilaterally in the ankle joint and big toe (4 sites with a maximum of 6 points at each site) to obtain a total lower extremity score out of 24 points. Cutaneous pressure was assessed bilaterally on the plantar surface of the big toe using six monofilaments of different thicknesses applied in descending order to obtain a total lower-extremity score of 72.

In addition, manual muscle strength testing was performed by a researcher with a background in physiotherapy on the lower extremities to obtain a total lower-extremity strength score out of 80 (max score of 5 for each muscle on each side). Muscles were tested bilaterally using the qualitative medical research council (MRC) manual muscle testing [102], [103] scale and included the ankle dorsiflexors and plantar flexors, knee extensors and flexors, hip extensors, flexors, abductors, and adductors. The same researcher trained in physiotherapy then performed the Spinal Cord Assessment Tool for Spastic Reflexes (SCATS) on the lower extremity muscles. The SCATS is a valid and reliable tool for assessing the severity of three separate components of spasticity after iSCI including clonus, flexor spasms, and extensor spasms [104], [105]. Each component of spasticity is given a severity score of 0-3 based on the duration of clonus/spasm seen; where 0 is no reaction, 1 the duration is < 3 seconds, 2 the duration is between 3-10 seconds, and 3 the duration is > 10 seconds [105]. This results in a total SCATS score between 0 and 9, where 9 is the most spastic and 0 indicates no spasticity [105].

Finally, clinical tests of balance and mobility were conducted including the mini-Balance Evaluation Systems Test (mini-BESTest), the Activities-specific Balance Confidence (ABC) Scale, the Walking Index for Spinal Cord Injury II (WISCI II), and a modified version of the Spinal Cord Injury Functional Ambulation Profile (SCI-FAP). The mini-BESTest, which is considered the most comprehensive balance assessment for the SCI population [80], was completed for each participant as an indication of functional balance. It consists of 14 items and has a maximum score of 28, with high scores indicating better balance control [82]. The mini-BESTest was designed to assess four of the six components of balance with a focus on dynamic balance and is grouped into four sections: anticipatory postural adjustments, postural responses, sensory orientation, and balance during gait [12], [82]. The anticipatory balance section was of particular interest in this study and was used as a functional indication of proactive balance control. The ABC Scale was used to measure perceived balance confidence while completing specific daily tasks and has been shown to be reliable and valid for use among individuals with iSCI [85]. The ABC scale consists of 16 items where participants rank their perceived confidence in completing the task without losing their balance from 0% (meaning no confidence) to 100% (meaning completely confident) [84]. The WISCI II is a measure of functional walking capacity that ranks walking according to the amount of physical assistance, braces and walking aids required. A perfect score of 20 indicates that no assistance is needed to walk 10 m at a self-selected pace [88], [89].

The SCI-FAP and WISCI II tests were also completed as measures of ambulatory status. The SCI-FAP is scored based on the time it takes for an individual to complete common tasks at a comfortable pace as well as the degree of assistance needed to complete each task [86]. There are seven tasks on the SCI-FAP and the highest total score is 2100 where lower scores reflect greater functional ability [86]. We completed a modified version of the SCI_FAP which included five tasks: carpet, obstacles (one direction only), step (one direction only), stairs, and a 7 m Up and Go (TUG) task. In the carpet task, the participant was timed to see how long it took them to walk from one end of a 7 m long short-pile carpet to the other end of the carpet. The obstacle task measured how long it took for the participant to walk towards and consecutively step over a 5 cm x 5 cm Styrofoam brick, a 10 cm x 10 cm Styrofoam brick, and finally walk around a trash can (diameter 56 cm, height 70 cm) and back. The step task timed how long it took the participant to walk towards, step on and over a step (21 cm x 81 cm x 122 cm) and continue walking for 1 m. The stairs task timed how long it took the participant to walk up and back down a small flight of stairs consisting of four steps with hand railings on both sides (each step was 29 cm depth, 76 cm width, 15 cm height). In the stairs task the participant could use any technique to ascend or descend the stairs but must fully turn around at the top of the stairs so that they approach the descent from the forwards direction. The stairs task was not completed by the AB control group. Finally, in the TUG task, participants were timed to see how long it took to stand up from a seated position, walk 7m, turn around, walk back to the chair and return to a seated position. For more detail on the modified SCI-FAP, please refer to Appendix B.

On the second day participants came into the laboratory, their height was measured using a stadiometer and their mass was measured using a standard weigh scale. Participants were told that they would perform a series of walking trials; however, the amount of trials was not specified. Additionally, in the consent form participants were made aware that at some point in the protocol they may be slipped, it read: “There is a chance that during the balance-challenged scenario you may loose your balance; however, the safety harness and/or spotter will prevent you from falling to the floor. The area of the floor that will be slippery will be minimalized to allow for a non-slippery surface to be within one step”. They were also instructed that if they do lose their balance, they should try and recover their balance and continue walking like normal. This way the slip was consented to but was still unexpected since they were unaware of *when* the slip would occur. To begin the protocol, a standing calibration trial was collected, followed by hip and knee calibration trials to identify joint locations. The standing calibration trials allowed participants time to get used to the feeling of wearing the safety harness, EMG electrodes and reflective markers.

Following the calibration trials, the participants completed three to five normal walking (NW) trials on a non-slippery surface (slip device locked), where they were asked to walk unassisted (ankle braces were allowed) at a self-selected speed for 10 m. NW trials were followed by one unexpected slip (US) trial where the slip device was unlocked without the participant knowing. After the US trial, participants were asked if the slip was truly unexpected or not and data were excluded if the slip was expected. Then, participants were told that the next phase of the protocol would be a series of four expected slip (ES) trials where they would be aware that the slip device was unlocked and would walk as normal as possible over the unlocked steel rollers. If the participant avoided stepping on the slip device, they were directed to step directly on the rollers and that trial was re-done until a trial was obtained with proper foot placement.

2.4 Data Processing and Analysis

Changes in kinematics and muscle activity observed during the ES trials, in comparison to the US trial, were representative of the feedforward adaptations that were made in response to knowledge of and experience with the slippery surface. Additionally, differences in kinematics between NW and ES trials were used to determine whether individuals were using similar amounts of proactive strategies compared to normal walking or if they were making additional adaptations when they knew the surface would be slippery. Changes in balance strategies were examined using a number of measures including step width, step length, foot-floor angle of the slip foot, COM velocity, ML and AP MOS, and post slip heel velocity (PSV) from kinematic data, as well changes in lower limb muscle activity from the TA, SOL, and GM muscles. The kinematic measures were used as a functional indication of the behavioural adaptations that individuals used to improve their proactive balance control, whereas the EMG data were collected to help determine differences in the ability of groups to make feedforward adaptations to muscle activity.

2.4.1 Kinematic Data

Kinematic data were sampled at 100 Hz and low-pass filtered at 8 Hz using a 4th order Butterworth filter [106]. Standing calibration trials were used to set up segmental coordinate systems which allowed for segmental centers of mass to be tracked via the reflective marker set. Kinematic data combined with anthropometric data for older (> 60 years; Jensen & Fletcher, 1994; Pearsall, Reid, & Ross, 1994; Yeadon, 1990) and younger [110] adults were used to calculate the segmental and total body COM during walking trials. The total body extrapolated center of mass (xCOM) was calculated using a formula provided by Hof, Gazendam, and Sinke (2005) intended for dynamical situations which accounts for both the position and velocity of the COM [3], [111]. Additionally, standing calibration trials were used to generate virtual markers on the first and fifth metatarsal, medially on the first metatarsophalangeal joint, and on the heel.

These virtual markers indicating the boundaries of each foot were then projected on the horizontal contact surface and were subsequently used to identify foot-contact and foot-off events as well as to calculate the midfoot location.

An algorithm was used to detect foot-contact and foot-off events for each gait cycle from the ground reaction forces and the progression of foot markers [96], [97]. Foot-contact was defined as the time point when any part of the foot came in contact with the supporting surface and forward foot progression was arrested. Foot-off was defined as the first time point following foot-contact when no part of the foot was in contact with the supporting surface. This algorithm used the resultant velocity signal from the heel and toe to define foot-contact as the beginning of stance phase and foot-off as the end of stance phase. Next, the foot-contact and foot-off values obtained from the algorithms were visually confirmed for each trial by comparing values with the reconstructed video in Vicon software. Finally, if footfall data were inaccurate, foot-off and foot-contact events were manually adjusted using a custom MATLAB script.

Kinematic data were used to calculate the following measures: mean step length, step width, foot-floor angle of the slip foot, COM velocity, MOS in both the AP and ML directions, and maximum post-slip heel velocity (PSV). Kinematic variables were calculated over each step during NW trials, which were used to produce an average step value for each NW trial (range of 1-27 steps/trial). Next, an average NW value was calculated for each participant from three NW trials. All kinematic variables, except for maximum PSV, from US and ES trials were calculated at foot contact with the slip device when the individual began the double support phase. Specific details regarding how each variable was calculated are outlined below.

Step length was calculated as the AP distance between the heel markers on the lead- and trail- feet. Step width was calculated as the ML distance between right and left mid-foot locations. The mid-foot locations were calculated as the average distance between the virtual foot markers obtained from the standing calibration trial. Sagittal plane foot-floor angle at contact with the slip device was calculated as the angle between the long axis of the foot, formed by the ankle-joint center and the second metatarsal marker, and the horizontal contact surface. The forward displacement of the total body COM was used to derive velocity information which, via the central difference method, was used to calculate an instantaneous COM velocity at foot-contact with the slip device. Walking with an increased step width, decreased stride length, slower COM velocity, and reduced foot-floor angle at foot contact are indicative of a cautious gait and have been identified as common proactive, feedforward adaptations [13], [15], [16], [51], [53], [54].

The AP MOS and ML MOS were calculated by comparing the position of the xCOM to the posterior boundary (the heel marker of the trail foot during double support) and lateral boundary (the most lateral aspect of lead foot at the base of the fifth metatarsal) of the BOS, respectively [13]. The AP MOS and ML MOS were normalized to step length and width, respectively, since these parameters have been shown to influence MOS values [111]. MOS was used as an indication of dynamic stability because, from a biomechanical perspective, the distance between the xCOM and the boundaries of the BOS is directly proportional to the impulse needed for an individual to lose their balance [3]. Thus, since the posterior edge of the trail foot was used to calculate AP MOS, an increase in the value represents an anterior shift in the xCOM position which reflects increased stability from backward loss of balance [3]. Similarly, an increase in ML MOS would indicate a shift in the xCOM position towards the trail/stance limb and represents an increase in stability in the medial-lateral plane while preparing to step onto the slip device.

Finally, the maximum post-slip heel velocity was defined as the greatest local horizontal heel marker velocity 50 ms after contact with the slip device and has been used functionally as an indication of slip intensity [15], [16], [112]. It is important to take into consideration that many of the kinematic variables are related to or derived from one another. For instance, as stated before, step length and width have been shown to influence ML and AP MOS variables, respectively [111]. Step width and step length on their own are important indicators of walking performance while ML and AP MOS, although influenced by step length and width, are indicators of dynamic stability and balance control. Although the kinematic measures are highly related, each provides important insight into which specific resources for the dynamic control of balance may be impaired and which resources individuals have the ability to modify when making feedforward- and feedback-based adaptations.

2.4.2 EMG Data

EMG signals were sampled at 2000 Hz, high-pass filtered at 20 Hz, full-wave rectified, and then low-pass filtered at 100 Hz using a 4th order Butterworth filter [106]. Bilateral EMG data from TA, SOL, and GM muscles were used to calculate proactive integrated EMG (PiEMG) and reactive iEMG (RiEMG) variables. iEMG was used as an indicator of the magnitude of the EMG response and was obtained for two different time periods to reflect both feedforward- and feedback-related adaptations to the slippery surface. Proactive, feedforward changes in muscle activity used in anticipation of the slip were calculated by integrating the EMG signal over the step prior to foot contact with the slip device. Reactive, feedback-based adaptations to the slip were calculated by integrating the EMG signal from foot contact on the slip device to 200 ms after contact. All iEMG values from US and ES trials were normalized by dividing them by the EMG values from NW trials, which were integrated over the same time intervals. Therefore, an iEMG value of 2.0 would represent a signal that is twice the amplitude compared to iEMG during normal walking. Additionally, the PiEMG values for each participant were normalized by their step time while stepping onto the slip device. This allowed for between-subjects comparisons to be made.

The time frames for PiEMG and RiEMG were chosen based on previous literature [5] and are meant to represent different types of motor control. Specifically, both predictive and visually guided anticipatory changes to balance control have been shown to occur one to two steps before contact with a perturbation [5]. Therefore, one step (from trail foot contact to lead foot contact on the slip device) was used as the time frame to represent proactive changes in muscle activity. Additionally, it has been shown that muscle activity within 0-100 ms after a perturbation reflects reflexive activity, 100-200 ms reflects functionally relevant behavioural responses, and 200-500 ms reflects voluntary recovery responses [5]. Since feedforward behavioural adaptations would help modulate reflexive iEMG responses to a slip, and would not likely affect the voluntary responses mediated by sensory cues [5], EMG was integrated over a fixed time frame of 200ms after contact with the slip device. This time interval for RiEMG was chosen to represent changes in both reflexive and functionally relevant postural responses that may result from feedforward adaptations after repeated exposures to the slip.

2.4.3 Qualitative Data

The ES trials were categorized by observation from reconstructed Vicon files or video from a digital camera, as either a walkover or skateover adaptive strategy [13], [49]. Walkover strategies included trials where the participant attempted to perform controlled alternating steps with both feet on the unlocked slip device and exhibited minimal heel marker displacement [49].

In contrast, skateover strategies included trials with a large heel marker displacement where the participant attempted to glide over the slip device with the lead-foot and plant the swing foot on the floor just beyond the slip device [13], [49]. The same researcher visually classified all slip trials as either a walkover or skateover strategy to avoid inter-rater differences. Finally, each participant was classified as someone who used a pure-skateover, a pure-walkover, or a mixed (both skateover and walkover) strategy based on performance in all their ES trials.

Additionally, for each participant it was determined whether they changed which foot they used to step on the slip device after the unexpected slip. If the participant used the same foot for all four ES trials, it was assumed that they had a preference for that foot when stepping on the slippery surface. The foot placement patterns of individuals who walked with noticeable unilateral impairments that affected one limb more than the other (i.e. foot drag/toe drop) were given special attention. This is because if an individual consistently chose to step on the slip device with their less impaired foot, this could be considered an important feedforward behavioural adaptation in addition to changes in kinematics or muscle activity.

2.5 Statistical Analysis

First, for each variable, all outliers that fell outside of the ± 3SD range were removed from the data. Next, normality was tested using Shapiro-Wilk's test and by visual inspection of histogram plots. If data was non-normally distributed, a two-step data transformation for continuous variables was completed [113]. Height and mass were compared between iSCI and AB groups using separate independent samples *t*-tests. Since no significant differences were found for height and mass between groups, and participants were age- and sex-matched during recruitment, analysis continued without the use of height and mass as covariates. All statistical analyses were completed with a confidence interval of 95% ($\alpha = 0.05$).

2.5.1 Kinematic Data

To investigate whether knowledge of the surface condition had an impact on the extent of feedforward behavioural adaptations made and how quickly each group was able to make a significant adaptation, differences in the kinematic variables between groups (AB and iSCI) and across slip conditions (US, ES1, ES2, ES3, ES4) were examined using 2 x 5 RM ANOVAs. Separate 2 x 5 RM ANOVAs were conducted for each kinematic variable: step width, step length, foot-floor angle, COM velocity, ML MOS, AP MOS, and maximum PSV. The assumption for sphericity was assessed using Mauchly's test and the Greenhouse-Geisser adjustment was used when violations to sphericity were found. Significant interaction effects were further examined using separate one-way ANOVAs and significant main effects were examined via pairwise comparisons with appropriate Bonferonni adjustments to account for multiple comparisons.

Furthermore, since individuals with iSCI have been shown to use more pronounced proactive strategies than AB individuals during normal unperturbed walking [65]–[67]; if any main condition effects were found from the analysis across slip conditions follow-up 2 (AB and iSCI) x 2 (NWavg and ESavg) RM ANOVAs were conducted. These 2 x 2 RM ANOVAs served to distinguish whether individuals were simply using the same amount of proactive strategies that they would use during normal walking, or whether they were able to further increase the extent of proactive strategies being used when approaching the known slippery surface. Once again, separate 2 x 2 RM ANOVAs were conducted for each kinematic variable: step width, step length, foot-floor angle, COM velocity, ML MOS, and AP MOS. Significant main effects were further examined via pairwise comparisons with appropriate Bonferonni adjustments to account for multiple comparisons.

2.5.2 EMG Data

To examine whether feedforward adaptations to muscle activity occurred in response to knowledge of the slippery surface condition, differences in iEMG between groups (AB and iSCI) and slip conditions (US, ES1, ES2, ES3, ES4) were examined using 2 x 5 RM ANOVAs. Separate analyses were run for muscle activity over the step before foot contact with the slip device (PiEMG) and 200 ms after foot contact with the slip device (RiEMG). These separate analyses were used to investigate changes in both proactive and reactive muscle responses over time as a result of prior experience with and knowledge of the surface condition. Additionally, separate RM ANOVAs were run for lead TA, trail TA, lead SOL, trail SOL, lead GM, and trail GM muscles. The assumption for sphericity was assessed using Mauchly's test and the Greenhouse-Geisser adjustment was used when violations to sphericity were found. Significant interaction effects were further examined using separate one-way ANOVAs and significant main effects were examined via pairwise comparisons with appropriate Bonferonni adjustments to account for multiple comparisons.

2.5.3 Qualitative Data

After classifying each participant as having used a pure-walkover, pure-skateover, or mixed strategy for the ES trials, percentages of individuals who used each strategy within each group (iSCI and AB) were calculated. Additionally, a total percentage of individuals in each group who used the same foot to step on the slip device in all four ES trials was calculated. No further statistical analyses were conducted on the qualitative data.

2.5.4 Clinical Data

Since the ability to reduce slip intensity (measured functionally as a change in maximum PSV from the US to the first ES trial) was used in this study as the major outcome variable indicating successful and meaningful adaptation to the known slippery surface, a multiple linear regression was performed to determine whether the scores on clinical assessments were able to significantly predict the change in maximum PSV. The regression analysis was also used to determine whether each clinical assessment was able to explain a significant amount of variance in PSV on their own and whether they were significantly correlated to the change in PSV.

Independent variables included in the regression model used to predict change in PSV were: lower extremity proprioception, cutaneous pressure, strength, functional walking ability (WISCI), balance confidence (ABC Scale), dynamic balance (Mini-BESTest), and the anticipatory balance subtest of the Mini-BESTest. The Durban-Watson statistic was used to test for the assumption of independence of residuals. There was homoscedasticity of residuals as assessed by visual inspection of a scatterplot of standardized residuals and standardized predicted values; the residuals were approximately evenly spread about the line of zero. Visual inspection of a normal probability plot showed that the residuals were normally distributed. Finally, since tolerance values were all greater than 0.1, there was no evidence of multicollinearity showing that none of the predictor variables were significantly correlated to one another.

3. RESULTS

Twenty-six individuals with iSCI participated in the study, but seven were excluded from this analysis: three participants were excluded because they did not complete any slip trials, three other participants were excluded because they could not complete the protocol without significant use of body weight support or aid from another person, and one participant was excluded because their marker set was incomplete or not visible, making it impossible to retrieve kinematic data. Final data were collected from 19 individuals with iSCI (14 males; 61.01 ± 17.67 years). Not all individuals with iSCI had an age- and sex-match, but a total of 17 AB individuals (13 males; 60.86 ± 17.79 years) were included in the study. Finally, of the participants included, one individual with iSCI and one AB individual completed three of the four ES trials and could not complete the fourth slip trial due to fatigue.

A summary of participant demographics is provided in Table 3.1. Thirteen participants with iSCI (68 %) had a traumatic injury and eleven participants (58 %) had an injury that resulted in tetraplegia. All iSCI participants were caucasian and had an AIS grade D which means they had an incomplete motor lesion and were capable of full range-of-motion movement against gravity for at least half of the muscles below the lesion [17], [20]. The average time since spinal cord injury was 8.68 years (SD = 10.49 years). No falls occurred while walking over the slippery surface suggesting that the strategies used to maintain dynamic stability were successful. These strategies will be discussed further below.

Table 3.1 Summary of participant demographics

Participant Demographics	**iSCI (n = 19)**		**AB (n = 17)**	
	Mean ± SD	**Range**	**Mean ± SD**	**Range**
Mass (kg)	84.68 ± 19.75	60.2 - 131.0	80.19 ± 16.15	58.8 - 120.8
Height (m)	1.73 ± 0.11	1.5 - 1.9	1.74 ± 0.09	1.57 - 1.88
Sex (Male : Female)	14 : 5	-	13 : 4	-
Age (years)	61.01 ± 17.67	29.8 – 95.9	60.86 ± 17.79	29.2 - 94.1
Time Since Injury (years)	8.68 ± 10.49	2.01 – 47.94	-	-
Tetraplegia : Paraplegia	11: 8	C1 - L4	-	-
Traumatic : Non-Traumatic	13 : 6			

3.1 Kinematic Data

Separate 2 x 5 RM ANOVA tests were performed for each kinematic variable to identify what behavioural adaptations occurred over repeated slip perturbations as a result of knowledge of the surface condition. The ANOVA tests showed no significant interaction effect between group (AB and iSCI) and condition (US, ES1, ES2, ES3, ES4) for any of the kinematic variables (Table 3.2). All variables, except mean AP- and ML-MOS, showed a significant main effect of group: mean step width ($F(1, 32) = 4.513$, $p = .041$, $\eta^2 = .124$), step length ($F(1, 32) = 5.510$, $p = .025$, $\eta^2 = .147$), foot-floor angle ($F(1, 32) = 5.609$, $p = .024$, $\eta^2 = .149$), COM velocity ($F(1, 32) = 5.223$, $p = .029$, $\eta^2 = .140$), and maximum PSV ($F(1, 32) = 11.757$, $p = .002$, $\eta^2 = .269$) (Table 3.2). Additionally, all kinematic variables, except mean step width and ML MOS, had a significant main effect of condition: mean step length ($F(4, 128) = 43.757$, $p < .001$, $\eta^2 = .578$), foot-floor angle ($F(4, 128) = 12.504$, $p < .001$, $\eta^2 = .281$), COM velocity ($F(2.539, 81.254) = 10.275$, $p < .001$, $\eta^2 = .243$), AP MOS ($F(3.153, 97.737) = 10.113$, $p < .001$, $\eta^2 = .246$), and maximum PSV ($F(2.234, 71.498) = 41.933$, $p < .001$, $\eta^2 = .567$) (Table 3.2).

Table 3.2 Summary of kinematic results from the 2 x 5 RM ANOVA tests

Dependent Variable	**Interaction Effect**			**Group Effect**			**Condition Effect**		
	F	***p***	**η^2**	**F**	***p***	**η^2**	**F**	***p***	**η^2**
Step Width (mm)	.130	.946	.004	4.513	**.041***	.124	.514	.679	.016
Step Length (mm)	2.059	.090	.060	5.510	**.025***	.147	43.757	**<.001***	.578
Foot-Floor Angle (°)	.491	.743	.015	5.609	**.024***	.149	12.504	**<.001***	.281
COM Velocity (m/s)	.018	.993	.001	5.223	**.029***	.140	10.275	**<.001***	.243
ML MOS	.191	.943	.007	.032	.860	.001	1.405	.237	.046
AP MOS	.192	.909	.006	.064	.801	.002	10.113	**<.001***	.246
Max PSV (m/s)	.449	.662	.014	11.757	**.002***	.269	41.933	**<.001***	.567

Main group effects were further investigated using pairwise comparisons to determine where the differences in each kinematic measure existed between groups, independent of condition. Pairwise comparisons showed that the iSCI group had a significantly greater mean step width (156.1 mm ± 9.6 mm) compared to the AB group (126.2 mm ± 10.2 mm). Mean step length was significantly shorter for the iSCI group (496.7 mm ± 32.9 mm) compared to the AB group (609.2 mm ± 34.9 mm). Mean foot-floor angle was significantly smaller for the iSCI group (11.1° ± 1.4°) compared to the AB group (15.9° ± 1.5°). Mean COM velocity at foot-contact was significantly smaller for the iSCI group (.712 m/s ± .072 m/s) compared to the AB group (.951 m/s ± .076 m/s). Finally, maximum PSV was significantly slower for the iSCI group (.389 m/s ± .046 m/s) compared to the AB group (.621 m/s ± .049 m/s).

Table 3.3 Summary of mean kinematic values ± standard deviations for each variable, split by group and condition

Dependent Variable	Group	Unexpected Slip (US)	Expected Slip 1 (ES1)	Expected Slip 2 (ES2)	Expected Slip 3 (ES3)	Expected Slip 4 (ES4)
Step Width (mm)	iSCI	160.8 ± 73.3	150.5 ± 55.6	154.1 ± 47.1	159.8 ± 52.6	155.2 ± 57.8
	AB	127.9 ± 32.6	123.5 ± 33.9	128.3 ± 40.3	130.5 ± 33.0	120.9 ± 37.2
Step Length (mm)	iSCI	646.8 ± 227.6	464.2 ± 206.9	467.2 ± 166.8	434.9 ± 174.9	470.4 ± 197.9
	AB	835.8 ± 122.9	546.5 ± 98.2	567.5 ± 100.1	551.5 ± 133.1	544.5 ± 103.9
Foot-Floor Angle (°)	iSCI	16.7 ± 8.4	9.4 ± 6.4	8.7 ± 8.7	9.7 ± 7.5	11.1 ± 7.8
	AB	21.3 ± 5.1	13.8 ± 6.9	15.1 ± 7.4	12.6 ± 9.0	16.5 ± 7.2
COM Velocity (m/s)	iSCI	.87 ± .38	.66 ± .33	.70 ± .34	.66 ± .36	.66 ± .37
	AB	1.09 ± .24	.90 ± .34	.95 ± .32	.90 ± .31	.91 ± .31
ML MOS	iSCI	130.86 ± 26.32	153.22 ± 31.68	149.96 ± 27.43	155.02 ± 31.15	158.74 ± 39.93
	AB	120.87 ± 22.77	125.93 ± 27.90	121.02 ± 27.53	127.73 ± 27.26	127.24 ± 27.07
AP MOS	iSCI	615.01 ± 214.15	496.08 ± 184.77	520.13 ± 187.99	484.97 ± 202.43	493.37 ± 202.14
	AB	737.37 ± 111.74	593.50 ± 152.31	636.49 ± 141.52	607.56 ± 139.09	607.54 ± 139.73
Maximum PSV (m/s)	iSCI	.83 ± .51	.32 ± .30	.27 ± .25	.25 ± .19	.26 ± .20
	AB	1.1 ± .40	.49 ± .14	.46 ± .19	.55 ± .26	.49 ± .20

Main condition effects were further investigated using pairwise comparisons to determine where the differences in each kinematic measure existed between slip conditions, independent of group. Pairwise comparisons showed that mean step length was significantly longer in the US condition (741.3 mm ± 31.9 mm) compared to all ES conditions: ES1 (505.4 mm ± 28.4 mm, $p < .001$), ES2 (517.3 mm ± 23.9 mm, $p < .001$), ES3 (493.2 mm ± 26.9 mm, $p < .001$), and ES4

(507.4 mm ± 27.6 mm, $p < .001$). Step length was not significantly different between any of the ES conditions, $p > .05$. Mean foot-floor angle during the US condition (19.0° ± 1.2°) was significantly greater compared to all ES conditions: ES1 (11.6° ± 1.1°, $p < .001$), ES2 (11.9° ± 1.4°, $p < .001$), ES3 (11.1° ± 1.4°, $p < .001$), ES4 (13.8° ± 1.3°, $p = .001$). Mean foot-floor angle was not significantly different between any of the ES conditions, $p > .05$.

Mean COM velocity was significantly faster during the US condition (.980 m/s ± .056 m/s) compared to all ES conditions: ES1 (.783 m/s ± .058 m/s, $p = .003$), ES2 (.824 m/s ± .057 m/s, $p = .035$), ES3 (.784 m/s ± .058 m/s, $p = .002$), ES4 (.785 m/s ± .059 m/s, $p = .001$). Mean COM velocity was not significantly different between any of the ES conditions, $p > .05$. Mean AP MOS was significantly smaller during the US condition (.943 ± .023) compared to all ES conditions: ES1 (1.149 ± .045, $p = .001$), ES2 (1.157 ± .034, $p < .001$), ES3 (1.175 ± .041, $p < .001$), and ES4 (1.132 ± .038, $p < .001$). Mean AP MOS was not significantly different between any of the ES conditions, $p > .05$. Finally, maximum PSV was significantly greater during the US condition (.969 m/s ± .079 m/s) compared to all ES conditions: ES1 (.409 m/s ± .041 m/s, $p < .001$), ES2 (.365 m/s ± .039 m/s, $p < .001$), ES3 (.402 m/s ± .039 m/s, $p < .001$), ES4 (.380 m/s ± .035 m/s, $p < .001$). There were no significant differences in maximum PSV between any of the ES conditions, $p > .05$.

Since the 2 x 5 RM ANOVA tests showed a main effect of condition between the US and the ES conditions for all but two of the kinematic variables (step width and ML MOS) and since there were no significant differences found between the four ES conditions, the values for the four ES trials were averaged and compared to the average values from the NW trials. This second set of analyses (2 x 2 RM ANOVAs) served to distinguish whether individuals were simply using the same amount of proactive strategies that they would use during normal walking, or whether they were able to further increase the extent of proactive strategies being used when approaching the known slippery surface. The 2 x 2 ANOVA tests showed no significant interaction effect between group (AB and iSCI) and condition (NWavg and ESavg) for any of the kinematic variables (Table 8).

Step width ($F(1, 34) = 6.848$, $p = .013$, $\eta^2 = .168$), step length ($F(1, 34) = 7.598$, $p = .009$, $\eta^2 = .183$), foot-floor angle ($F(1, 34) = 8.409$, $p = .007$, $\eta^2 = .198$), and COM velocity ($F(1, 34) = 7.455$, $p = .010$, $\eta^2 = .180$) showed a significant main effect of group (Table 8). Main group effects were further investigated using pairwise comparisons to determine where the

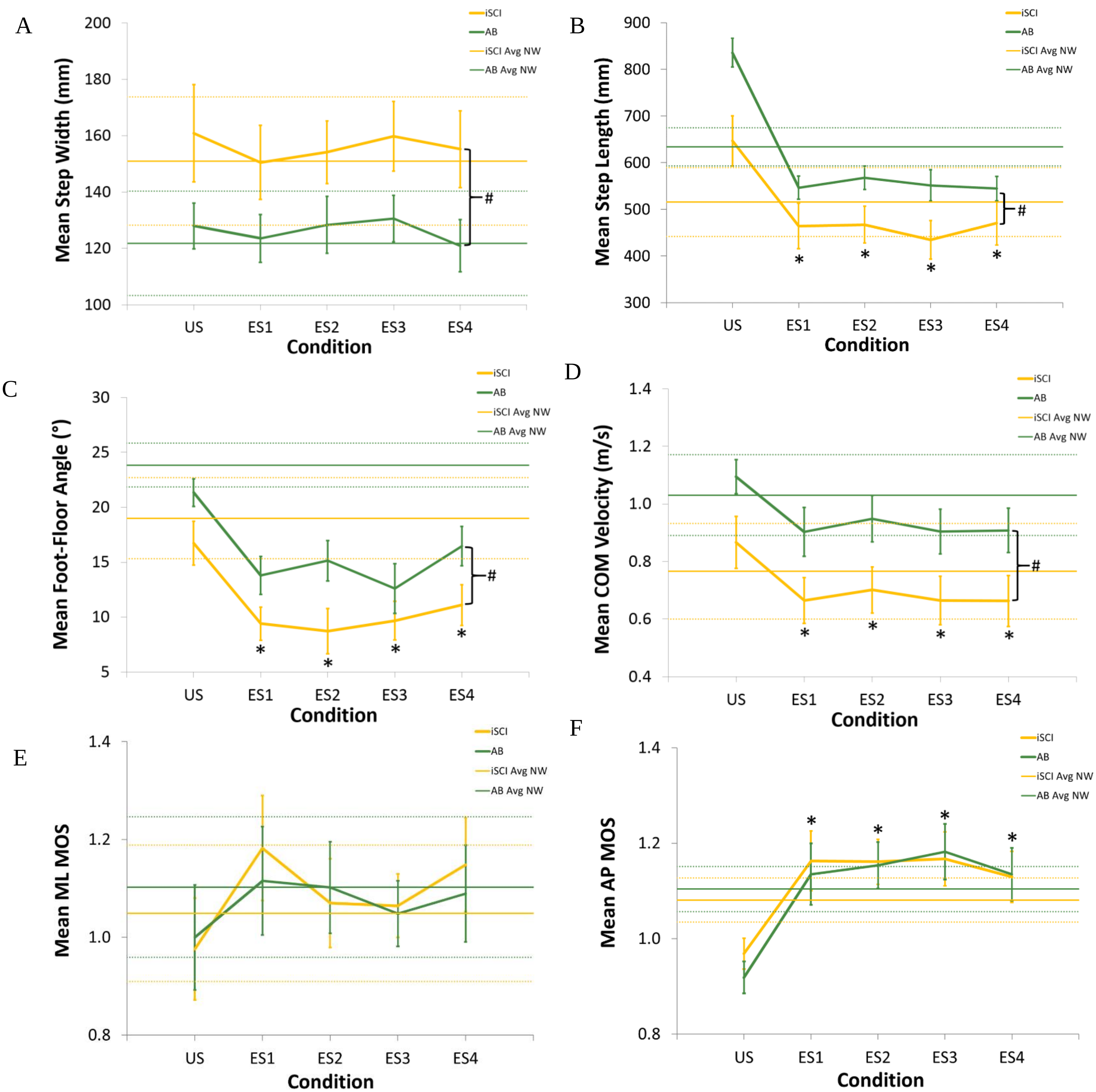

Figure 3.1 Feedforward behavioural adaptations with repeated exposures to a known slippery surface. Error bars represent standard error. Dotted lines represent the 95% confidence interval for the plotted average from the NW trials for each group. A * symbol indicates that the mean of that condition is significantly different compared to the unexpected slip (US) condition, independent of group (α = 0.05). A # symbol indicates that the mean of the iSCI group is significantly different than the mean of the AB group, independent of condition (α = 0.05).

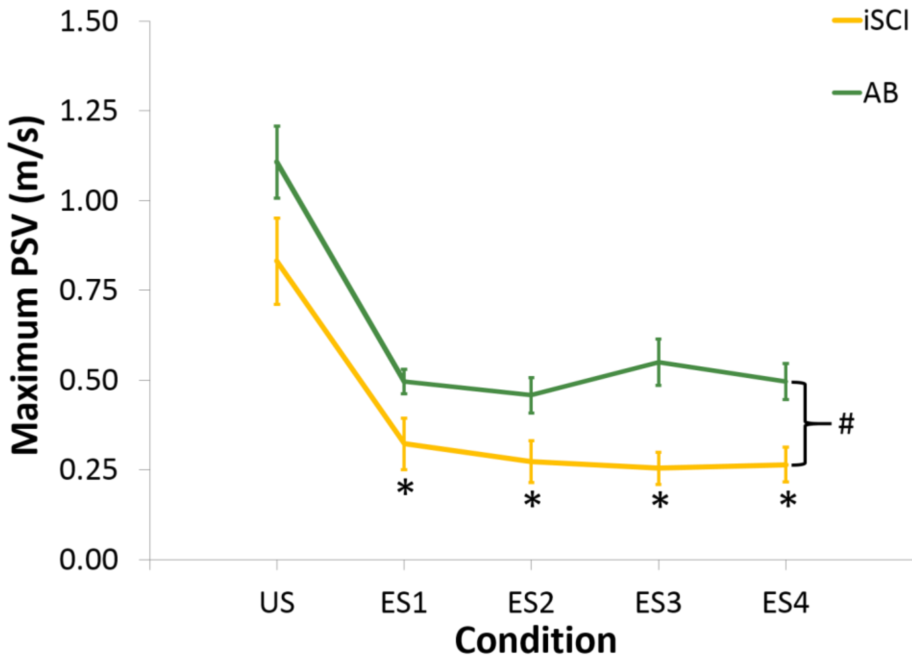

Figure 3.2 Changes in maximum post-slip heel velocity (PSV) with repeated exposures to a known slippery surface. Error bars represent standard error. A * symbol indicates a significant difference in PSV in that condition compared to the unexpected slip (US), independent of group (α = 0.05). A # symbol indicates that the PSV of the iSCI group is significantly different than the AB group, independent of condition (α = 0.05).

Table 3.4 Summary of mean kinematic values ± standard deviations for each variable, averaged over normal walking trials and expected slip trials for each group

Dependent Variable	Group	Normal Walking Average	Expected Slip Average
Step Width (mm)	iSCI	157.3 ± 52.3	161.2 ± 53.2
	AB	120.4 ± 34.2	124.3 ± 31.3
Step Length (mm)	iSCI	500.6 ± 158.4	445.9 ± 175.7
	AB	627.9 ± 78.7	553.3 ± 81.3
Foot-Floor Angle (°)	iSCI	18.2 ± 7.9	9.1 ± 6.7
	AB	23.6 ± 3.8	14.8 ± 5.5
COM Velocity (m/s)	iSCI	.74 ± .34	.64 ± .35
	AB	1.02 ± .26	.92 ± .29
ML MOS	iSCI	1.02 ± .18	1.07 ± .27
	AB	1.09 ± .29	1.06 ± .26
AP MOS	iSCI	1.08 ± .10	1.13 ± .16
	AB	1.08 ± .09	1.12 ± .15

differences in each kinematic measure existed between groups, independent of condition. Pairwise comparisons showed that the iSCI group had a significantly greater mean step width (159.3 mm ± 9.7 mm) compared to the AB group (122.3 mm ± 10.2 mm). Mean step length for individuals with iSCI was significantly shorter (473.2 mm ± 29.2 mm) than the AB group (590.6 mm ± 30.9 mm). The mean foot-floor angle was significantly smaller for the iSCI group (13.7° ± 1.3°) compared to the AB group (19.2° ± 1.4°). Lastly, the mean COM velocity was significantly slower for the iSCI group (.694 m/s ± .069 m/s) compared to the AB group (.968 m/s ± .073 m/s).

Finally, step length ($F(1, 34) = 24.086$, $p < .001$, $\eta^2 = .415$), foot-floor angle ($F(1, 34) = 103.547$, $p < .001$, $\eta^2 = .753$), and COM velocity ($F(1, 34) = 9.822$, $p = .004$, $\eta^2 = .224$) showed a significant main effect of condition (Table 8). Main condition effects were further investigated using pairwise comparisons to determine where the differences in each kinematic measure existed between conditions, independent of group. Pairwise comparisons showed that mean step length was significantly longer in the NW conditions (564.5 mm ± 21.2 mm) compared to the ES conditions (499.6 mm ± 23.3 mm). Mean foot-floor angle was significantly larger in the NW conditions (20.9° ± 1.1°) compared to the ES conditions (11.9° ± 1.0°). Finally, mean COM velocity was significantly faster in the NW conditions (.881 m/s ± .051m/s) compared to the ES conditions (.781 m/s ± .055 m/s).

Table 3.5 Summary of kinematic results from the 2 x 2 RM ANOVA tests

Dependent Variable	**Interaction Effect**			**Group Effect**			**Condition Effect**		
	F	***p***	**η^2**	**F**	***p***	**η^2**	**F**	***p***	**η^2**
Step Width (mm)	.000	.998	.000	6.848	**.013***	.168	.725	.400	.021
Step Length (mm)	.573	.454	.017	7.598	**.009***	.183	24.086	**<.001***	.415
Foot-Floor Angle (°)	.024	.879	.001	8.409	**.007***	.198	103.547	**<.001***	.753
COM Velocity (m/s)	.008	.929	.000	7.455	**.010***	.180	9.822	**.004***	.224
ML MOS	1.542	.223	.045	.132	.719	.004	.051	.823	.002
AP MOS	.143	.708	.002	.068	.796	.002	3.003	.092	.081

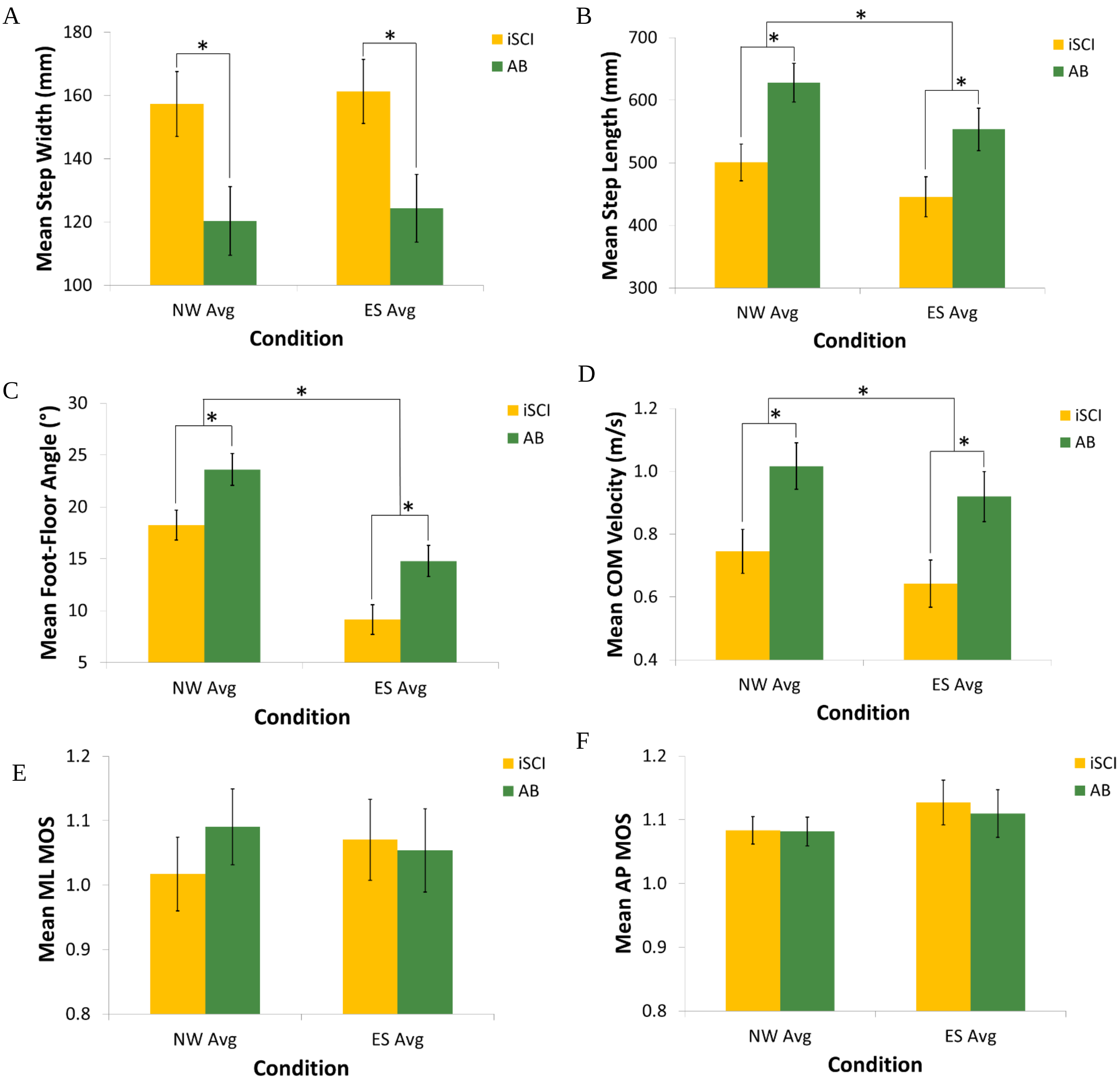

Figure 3.3 Differences between the average behavioural adaptations observed in the expected slip conditions compared to the average behaviour during normal walking trials. Error bars represent standard error. A * symbol indicates that the means of the indicated values are significantly different from one another ($\alpha = 0.05$).

3.2 EMG Data

Pre-Slip, Proactive iEMG (PiEMG)

Separate 2 x 5 RM ANOVA tests were performed for lead and trail muscles to determine changes in PiEMG over repeated slip perturbations. The RM ANOVA tests indicated that only the trail SOL muscle had a significant interaction effect between group (AB and iSCI) and condition (US, ES1, ES2, ES3, ES4), ($F(2.670, 77.421) = 3.928$, $p = .015$, $\eta^2 = .119$) (Table 3.7). Two of the muscles showed a significant main group effect of PiEMG: lead SOL ($F(1, 30) = 4.572$, $p = .041$, $\eta^2 = .132$) and lead GM ($F(1, 30) = 5.289$, $p = .029$, $\eta^2 = .150$) (Table 3.7). All muscles except for the lead GM had a significant main effect of condition: lead TA ($F(2.160, 64.79) = 5.689$, $p = .004$, $\eta^2 = .159$), lead SOL ($F(2.226, 66.773) = 7.391$, $p = .001$, $\eta^2 = .198$), trail TA ($F(4, 124) = 9.010$, $p < .001$, $\eta^2 = .225$), and trail GM ($F(2.369, 68.703) = 12.293$, $p < .001$, $\eta^2 = .298$) (Table 3.7).

Table 3.6 Summary of mean PiEMG values ± standard deviations for each muscle, split by group and condition

Dependent Variable	**Group**	**Unexpected Slip (US)**	**Expected Slip 1 (ES1)**	**Expected Slip 2 (ES2)**	**Expected Slip 3 (ES3)**	**Expected Slip 4 (ES4)**
Lead TA	iSCI	1.09 ± .18	1.06 ± .22	1.52 ± 1.03	1.02 ± .31	1.14 ± .68
	AB	.98 ± .24	.93 ± .31	1.21 ± .45	.88 ± .25	1.40 ± .53
Lead SOL	iSCI	.87 ± .17	1.17 ± .42	1.16 ± .39	1.05 ± .34	1.34 ± .65
	AB	.98 ± .17	1.12 ± .29	1.37 ± .26	1.07 ± .36	1.38 ± .51
Lead GM	iSCI	1.12 ± .17	1.25 ± .49	1.17 ± .51	1.15 ± .41	1.02 ± .41
	AB	1.13 ± .49	1.19 ± .45	1.41 ± .53	1.22 ± .42	1.32 ± .58
Trail TA	iSCI	1.06 ± .35	1.56 ± .71	1.62 ± .62	1.28 ± .43	1.44 ± .45
	AB	.94 ± .32	1.48 ± .65	1.50 ± .50	1.24 ± .52	1.57 ± .54
Trail SOL	iSCI	1.17 ± .31	.98 ± .26	1.23 ± .36	1.07 ± .36	1.35 ± .57
	AB	1.05 ± .14	.94 ± .15	1.66 ± .71	1.06 ± .14	1.82 ± .76
Trail GM	iSCI	.94 ± .14	.99 ± .21	1.48 ± .76	1.03 ± .26	1.47 ± .47
	AB	.95 ± .23	.99 ± .19	2.39 ± 1.99	1.04 ± .22	2.24 ± 1.63

Table 3.7 Summary of PiEMG results from 2 x 5 RM ANOVA tests

Dependent Variable	**Interaction Effect**			**Group Effect**			**Condition Effect**		
	F	***p***	**η^2**	**F**	***p***	**η^2**	**F**	***p***	**η^2**
Lead TA	.209	.828	.007	.676	.418	.022	5.689	**.004***	.159
Lead SOL	1.233	.300	.039	4.572	**.041***	.132	7.391	**<.001***	.198
Lead GM	2.551	.067	.078	5.289	**.029***	.150	1.516	.219	.048
Trail TA	.779	.541	.025	.914	.347	.029	9.010	**<.001***	.225
Trail SOL	3.928	**.015***	.119	5.223	**.030***	.153	13.005	**<.001***	.310
Trail GM	2.553	.076	.081	2.847	.102	.089	12.293	**<.001***	.298

The interaction effect of trail SOL muscle PiEMG was further investigated using follow-up post-hoc analyses. A one-way ANOVA test was conducted and PiEMG from the trail SOL muscle was only significantly different between the iSCI and AB groups during the second ES trial ($F(1, 34) = 8.508$, $p = .006$), with the AB group having a larger muscle amplitude (2.687 ± .342) than the iSCI group (1.729 ± .163). Differences in trail SOL PiEMG between slip conditions were also investigated for each group using separate one-way ANOVAs. Within the AB group, trail SOL PiEMG was significantly greater in the ES2 condition (2.687 ± .342) compared to the US (1.607 ± .082, $p = .043$), ES1 (1.543 ± .089, $p = .023$), and ES3 (1.582 ± .107, $p = .047$) conditions. Additionally, within the AB group, trail SOL PiEMG was significantly greater in the ES4 condition (2.586 ± .261) compared to the US ($p = .019$), ES1 ($p = .007$), and ES3 ($p = .010$) conditions. Finally, within the AB group there were no significant differences in trail SOL PiEMG between the ES2 and ES4 conditions or between the US, ES1, and ES3 conditions, $p > .05$. Within the iSCI group no significant differences were found in trail SOL PiEMG between any of the slip conditions, $p > .05$.

Main group effects were further investigated using pairwise comparisons to determine where the differences in PiEMG existed between groups, independent of condition. Pairwise comparisons showed that the AB group had a significantly greater PiEMG amplitude for the lead SOL muscle (1.960 ± .132) compared to the iSCI group (1.574 ± .124). Secondly, the AB group had a significantly greater PiEMG amplitude for the lead GM muscle (2.017 ± .143) compared to the iSCI group (1.565 ± .135).

Main condition effects were further investigated using pairwise comparisons to determine where the differences in PiEMG existed between slip conditions, independent of group. Pairwise comparisons revealed that there were no significant differences in PiEMG for the lead TA muscle between the US and any of the ES conditions, $p > .05$. The only significant increase in PiEMG for the lead TA muscle was found between ES3 (1.369 ± .105) and ES4 (2.209 ± .282, $p = .048$). Significant differences were found for PiEMG of the lead SOL muscle which showed a smaller amplitude during the US condition (1.370 ± .059) compared to the ES2 (1.968 ± .139, $p = .005$), and ES4 (2.232 ± .224, $p = .006$) conditions. There were no significant differences in PiEMG between any other slip conditions for the lead SOL muscle, $p > .05$.

Significant differences were found for PiEMG of the trail TA muscle which showed a smaller amplitude during the US condition (1.463 ± .091) compared to the ES1 (2.357 ± .197, $p = .001$), ES2 (2.317 ± .168, $p < .001$), and ES4 (2.294 ± .175, $p = .002$) conditions. There were no significant differences in PiEMG between any other slip conditions for the trail TA muscle, $p > .05$. Significant differences were found for PiEMG of the trail GM muscle which showed a smaller amplitude during the US condition (1.410 ± .070) compared to the ES1 (1.639 ± .085, $p = .017$), ES2 (2.907 ± .339, $p = .001$), and ES4 (2.887 ± .368, $p = .002$) conditions. Significant differences were found for PiEMG of the trail GM muscle which showed a greater amplitude during the ES2 condition compared to the ES1 ($p = .002$) and ES3 (1.743 ± .187, $p = .005$) conditions. Significant differences were also found for PiEMG of the trail GM muscle which showed a greater amplitude during the ES4 condition compared to the ES1 ($p = .005$) and ES3 ($p = .042$) conditions. There were no significant differences found for PiEMG of the trail GM muscle between the ES2 and ES4 conditions, $p > .05$.

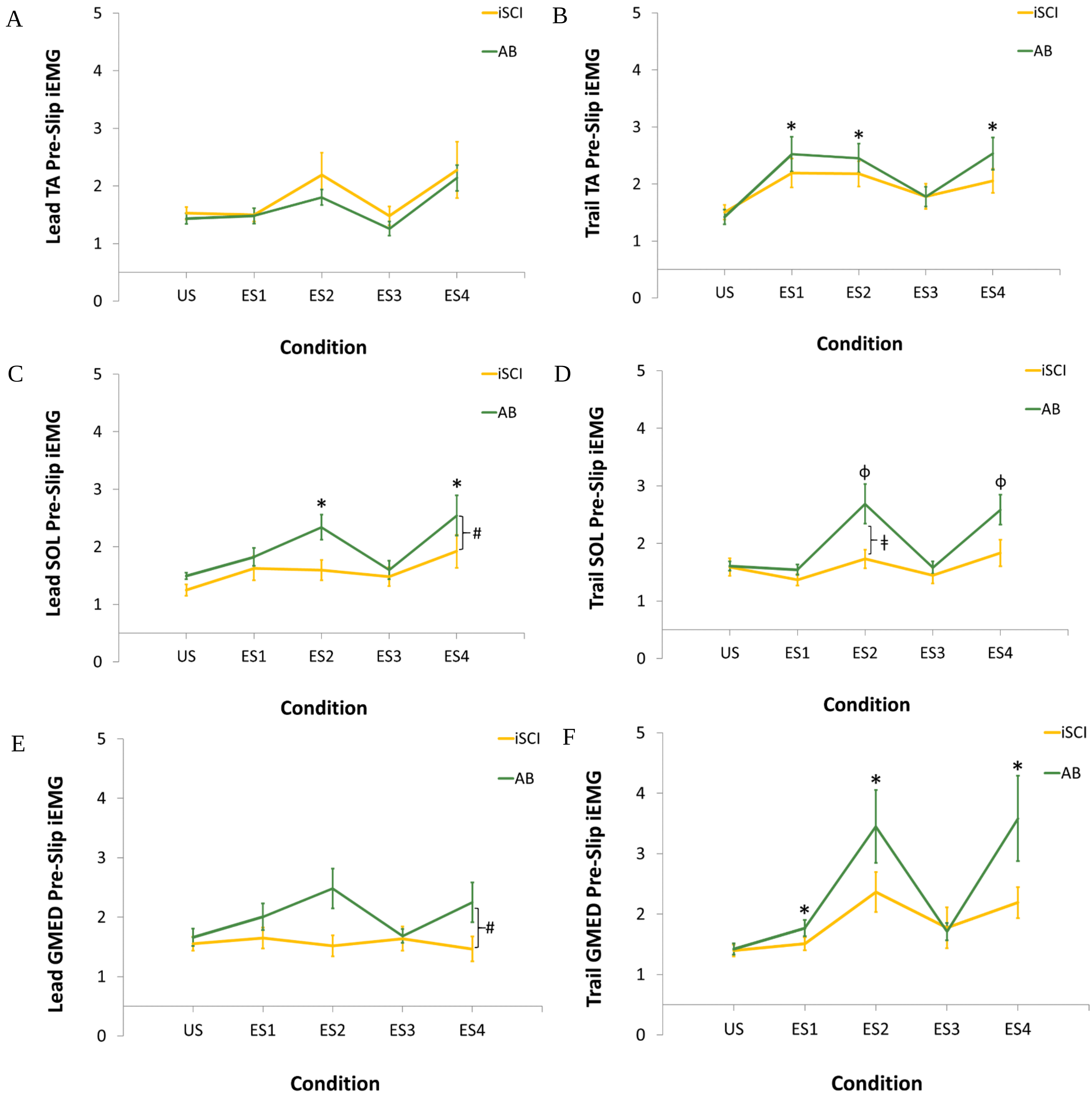

Figure 3.4 Changes in pre-slip, proactive iEMG (PiEMG) with repeated exposures to a known slippery surface. Error bars represent standard error. A * symbol indicates that the mean of that condition is significantly different compared to the unexpected slip (US) condition, independent of group ($\alpha = 0.05$). A # symbol indicates that the PSV of the iSCI group is significantly different than the AB group, independent of condition ($\alpha = 0.05$). A ϕ symbol indicates that the mean of that condition is significantly different compared to the unexpected slip (US) condition, within that group ($\alpha = 0.05$). A ǂ symbol indicates that the mean of the iSCI group is significantly different than the mean of the AB group, within that condition ($\alpha = 0.05$).

Post-Slip, Reactive iEMG (RiEMG)

Separate 2 x 5 RM ANOVA tests were performed for lead and trail muscles to determine changes in RiEMG over the slip perturbations. The RM ANOVA tests indicated that the lead SOL muscle ($F(4, 116) = 4.037$, $p = .004$, $\eta^2 = .122$) and the trail TA muscle ($F(4, 120) = 5.407$, $p < .001$, $\eta^2 = .153$) had a significant interaction effect between group (AB and iSCI) and condition (US, ES1, ES2, ES3, ES4) (Table 3.8). Two of the muscles showed a significant main group effect of RiEMG including lead TA ($F(1, 30) = 6.211$, $p = .018$, $\eta^2 = .172$) and trail GM ($F(1, 28) = 9.596$, $p = .004$, $\eta^2 = .255$) (Table 3.8). Additionally, two of the muscles showed a significant main effect of condition including the lead TA ($F(2.293, 68.790) = 6.431$, $p = .002$, $\eta^2 = .177$) and trail SOL ($F(3.194, 95.819) = 7.651$, $p < .001$, $\eta^2 = .203$) (Table 3.8).

Table 3.8 Summary of mean RiEMG values ± standard deviations for each muscle, split by group and condition

Dependent Variable	Group	Unexpected Slip (US)	Expected Slip 1 (ES1)	Expected Slip 2 (ES2)	Expected Slip 3 (ES3)	Expected Slip 4 (ES4)
Lead TA	iSCI	1.54 ± .72	1.03 ± .23	1.10 ± .48	1.08 ± .21	.98 ± .43
	AB	1.19 ± .51	.89 ± .34	1.05 ± .34	.79 ± .18	.96 ± .29
Lead SOL	iSCI	1.15 ± .41	1.19 ± .44	1.13 ± .62	1.44 ± .84	1.27 ± .64
	AB	1.11 ± .36	1.76 ± .63	1.89 ± .86	1.31 ± .28	1.61 ± .62
Lead GM	iSCI	1.11 ± .32	1.13 ± .35	.86 ± .38	1.10 ± .33	1.16 ± .58
	AB	1.07 ± .27	1.37 ± .73	1.14 ± .32	1.49 ± 1.10	1.06 ± .41
Trail TA	iSCI	2.38 ± 1.49	1.29 ± .48	1.50 ± .78	1.45 ± .64	1.60 ± .74
	AB	1.47 ± .53	2.19 ± 1.17	1.71 ± .81	1.67 ± .90	1.69 ± .69
Trail SOL	iSCI	.92 ± .48	1.13 ± .49	1.20 ± .50	.96 ± .19	1.30 ± .49
	AB	1.00 ± .56	1.00 ± .17	1.51 ± .71	1.04 ± .10	1.51 ± .59
Trail GM	iSCI	1.43 ± .72	1.01 ± .26	1.10 ± .46	1.10 ± .22	1.16 ± .41
	AB	1.52 ± .92	1.43 ± .53	1.88 ± .88	1.42 ± .46	1.72 ± .83

The interaction effect of lead SOL muscle RiEMG was further investigated. A one-way ANOVA test showed that RiEMG values from the lead SOL was significantly different between the iSCI and AB groups during the ES1 ($F(1, 33) = 4.274$, $p = .047$, $\eta^2 = .118$) and ES2 conditions ($F(1, 35) = 4.996$, $p = .032$, $\eta^2 = .128$), with the AB group having a larger muscle amplitude than the iSCI group. RiEMG from the trail TA was only significantly different between the iSCI and AB groups during the ES1 condition ($F(1, 35) = 11.512$, $p = .002$, $\eta^2 = .253$) with the AB group having a larger muscle amplitude than the iSCI group. Differences in lead SOL RiEMG between slip conditions were investigated for each group using separate one-way ANOVAs. Within the AB group, no significant differences were found in lead SOL RiEMG between any of the slip conditions, $p > .05$. Within the iSCI group, no significant differences were found in lead SOL RiEMG between any of the slip conditions, $p > .05$. Next, differences in trail TA RiEMG between slip conditions were investigated for each group using separate one-way ANOVAs. Within the AB group, no significant differences were found in trail TA RiEMG between any of the slip conditions, $p > .05$. Within the iSCI, group no significant differences were found in trail TA RiEMG between any of the slip conditions, $p > .05$.

Main group effects were further investigated using pairwise comparisons to determine where the differences in RiEMG existed between groups, independent of condition. Pairwise comparisons showed that the iSCI group had a significantly greater RiEMG amplitude for the lead TA muscle (1.147 ± .048) compared to the AB group (.976 ± .048). In contrast, the iSCI group had a significantly smaller RiEMG amplitude for the trail GM muscle (1.158 ± .093) compared to the AB group (1.595 ± .106).

Main condition effects were further investigated using pairwise comparisons to determine where the differences in RiEMG existed between slip conditions, independent of group. Pairwise comparisons showed that RiEMG for the lead TA muscle had a significantly greater amplitude during the US condition (1.367 ± .111) compared to the ES1 (.959 ± .051, $p = .035$), ES3 (.938 ± .035, $p = .008$), and ES4 (.968 ± .065, $p = .049$) conditions. There were no significant differences in RiEMG between the US and ES2 (1.076 ± .074) conditions for the lead TA muscle, $p > .05$. The RiEMG of the trail SOL muscle was shown to be significantly greater during the ES2 condition (1.358 ± .108) compared to the US (.961 ± .092, $p = .010$), and ES3 (1.004 ± .027, $p = .032$) conditions. Additionally, RiEMG of the trail SOL muscle was shown to be significantly greater during the ES4 condition (1.406 ± .096) compared to the US ($p = .004$), and ES3 ($p = .002$) conditions. There were no significant differences in RiEMG of the trail SOL muscle between the ES1 (1.065 ± .066) and any other slip condition, between the US and ES3 conditions, or between the ES2 and ES4 conditions, $p > .05$.

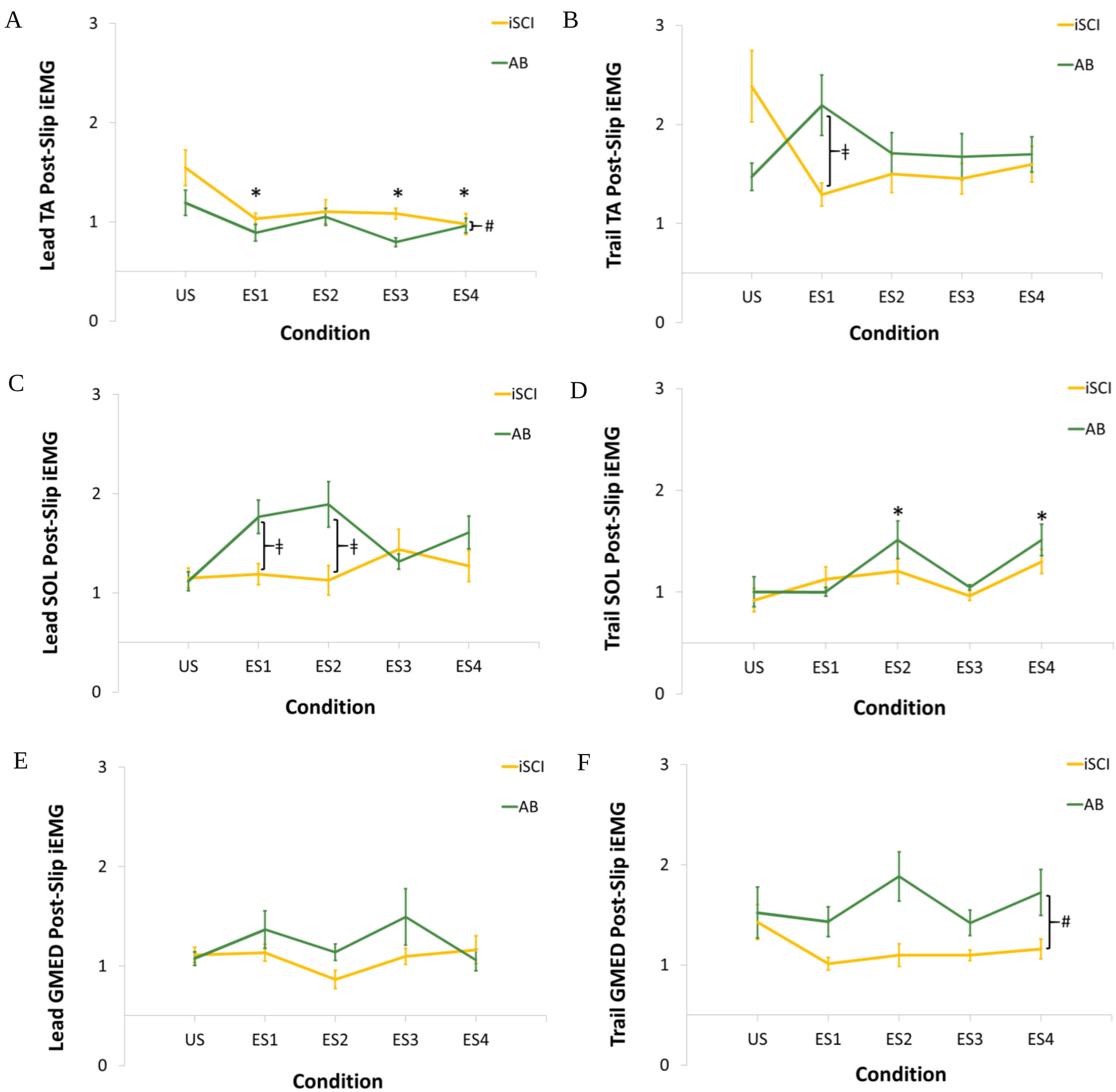

Figure 3.5 Changes in post-slip, reactive iEMG (RiEMG) with repeated exposures to a known slippery surface. Error bars represent standard error. A * symbol indicates that the mean of that condition is significantly different compared to the unexpected slip (US) condition, independent of group (α = 0.05). A # symbol indicates that the PSV of the iSCI group is significantly different than the AB group, independent of condition (α = 0.05). A ϕ symbol indicates that the mean of that condition is significantly different compared to the unexpected slip (US) condition, within that group (α = 0.05). A ǂ symbol indicates that the mean of the iSCI group is significantly different than the mean of the AB group, within that condition (α = 0.05).

Table 3.9 Summary of RiEMG results from the 2 x 5 RM ANOVA tests

Dependent Variable	Interaction Effect			Group Effect			Condition Effect		
	F	*p*	η^2	F	*p*	η^2	F	*p*	η^2
Lead TA	1.088	.349	.035	6.211	**.018***	.172	6.431	**.002***	.177
Lead SOL	4.037	**.004***	.122	5.179	**.030***	.152	2.503	**.046***	.079
Lead GM	1.515	.219	.048	2.031	.164	.063	1.937	.134	.061
Trail TA	5.407	**<.001***	.153	.327	.572	.011	1.075	.372	.035
Trail SOL	1.181	.322	.038	1.328	.258	.042	7.651	**<.001***	.203
Trail GM	1.902	.149	.064	9.596	**.004***	.255	1.824	.162	.061

3.4 Qualitative Data

Of the 17 AB participants, 82.4% (14) used a pure-walkover strategy during all the ES trials while 17.6% (3) used a mixed strategy, employing both skateover and walkover strategies in different ES trials. None of the AB individuals used a skateover strategy during all the ES trials. Of the 19 participants with iSCI, 68.4% (13) used a pure-walkover strategy during all the ES trials, 26.3% (5) used a mixed strategy, employing both skateover and walkover strategies in different ES trials, and 5.3% (1) used a pure-skateover strategy during all the ES trials.

A total of 29.4% (5) AB participants and 26.3% (5) iSCI participants used the same foot to step on the slip device for all four ES trials. Of the iSCI participants who consistently used the same foot to step on the slip device, only two individuals had an apparent foot/leg that had more impaired function than the other. One such participant walked with a noticeable right foot drag (even while using an ankle brace on the right side) but interestingly, they used the more impaired right limb to step on the slip device every time. The other participant walked while dragging their left foot and used the less impaired right foot to step on the slip device every time. There were only three other iSCI participants that had one noticeably impaired limb (usually affected by toe-drop), but this did not appear to influence which foot they stepped on the slip device with.

3.3 Clinical Data

Participants with iSCI had good ambulatory status as indicated by high scores on the WISCI II scale (average = 18.81 ± 2.93, median = 20; Table 3.10). Despite their good ambulatory status, participants with iSCI had balance confidence scores on the ABC scale (average = 73.80 ± 16.41, median = 76.25; Table 3.10) that were approximately 20% lower than their age- and sex- matched AB peers (average = 94.85 ± 5.54, median = 97.65; Table 3.10). Individuals with iSCI took 21.55 seconds longer (52.22 seconds total) to complete four tasks from the SCIFAP (the carpet, TUG, obstacle, and step tasks) compared to AB individuals (30.67 seconds total). Additionally, individuals with iSCI had lower scores on the MiniBESTest (average = 19.26 ± 6.36 out of 28 total; Table 3.10) compared to previously reported normative values for healthy individuals between the ages of 60-69 years old (22.4 ± 6.3 out of 28 total) [114].

Table 3.10 Summary of scores on clinical assessments

Clinical Assessments	iSCI (n = 19)			AB (n = 17)		
	Mean ± SD	**Median**	**Range**	**Mean ± SD**	**Median**	**Range**
WISCI II (/20)	18.81 ± 2.93	20	9 - 20	-	-	-
SCIFAP (seconds)						
Carpet	5.43 ± 1.77	5.69	3.33 – 8.71	3.52 ± 0.62	3.56	2.64 – 5.03
Up and Go (7m)	21.97 ± 8.21	22.14	13.51 – 37.32	13.94 ± 2.20	14.10	11.1 – 19.71
Obstacles (1 way)	16.26 ± 6.93	14.16	10.02 – 32.66	10.46 ± 1.84	10.08	7.84 – 14.61
Step (1 way)	8.56 ± 11.44	5.62	2.35 – 49.62	2.75 ± 0.74	2.44	2.02 – 4.93
Stairs	12.86 ± 8.78	9.70	4.79 – 29.84	-	-	-
ABC (/100)	73.80 ± 16.14	76.25	36.87 – 95.31	94.85 ± 5.54	97.65	79.87 - 100
MiniBESTest (/28)	19.26 ± 6.36	20	8 - 27	-	-	-
L/E Proprioception						
Right (/12)	10.52 ± 2.46	12	3 - 12	-	-	-
Left (/12)	10.42 ± 2.32	12	3 - 12	-	-	-
L/E Cutaneous Sense						
Right L/E (/36)	14.53 ± 8.20	16	0 - 28	-	-	-
Left L/E (/36)	15.84 ± 8.34	18	0 – 28	-	-	-
L/E Muscle Strength						
Right L/E (/40)	32.66 ± 4.47	34	21.5 - 39	-	-	-
Left L/E (/40)	32.55 ± 4.76	33	17.5 - 39	-	-	-

Individuals with iSCI showed similar lower extremity proprioception, cutaneous pressure, spasticity, and muscle strength scores between their left and right limbs (Table 3.10). To receive a mean muscle strength score of approximately 32/40 on each side (R: 32.66 ± 4.47, L: 32.55 ± 4.76), the individuals with iSCI demonstrated a mean score of approximately 4/5 on each of the 8 lower extremity muscles tested which is indicative of the ability to produce moderate to strong pressure and to perform full range of motion tasks against gravity [102], [103]. Additionally, the iSCI individuals received nearly full points on the joint position sense test (R: 10.52/12, L: 10.42/12) indicating mostly intact proprioceptive abilities in the lower limbs. However, cutaneous pressure appeared to be impaired in individuals with iSCI as they received less than half of the total possible score on each side (R: 14.53/36, L: 15.84/36). Unfortunately, scores for muscle strength, proprioception, and cutaneous pressure were not collected for the AB matches so no between-group comparisons could be made.

Finally, the majority (78.9% or 15/19) of iSCI participants had a SCATS score of zero on both left and right sides indicating that most participants were not affected by any type of spasticity: clonus, extensor, or flexor spasms. Only four participants received a score greater than zero; three participants had mild (< 3 seconds) lasting spasticity and one participant had severe (> 10 seconds) lasting spasticity. Therefore, the SCATS scores were not included in the multiple regression model because the nature of the data set produced non-normally distributed data in which any participant who received a score greater than zero was flagged as an influential or leverage point.

The multiple regression model did not significantly predict the change in maximum PSV ($F(7,11) = .618$, $p > .05$) and none of the clinical assessment scores showed a significant correlation with the change in PSV (see Pearson's R values from Table 3.11, $p > .007$). Furthermore, none of the clinical assessment scores added significantly to the prediction model on their own (see *t*-values from Table 3.11, $p > .05$).

Table 3.11 Summary of results from multiple regression between change in maximum post slip heel velocity (PSV) and scores on clinical assessments

Independent Variables	**Pearson's R**	**p-value**	***t*-value**	**p-value**
L/E Proprioception	-.418	.038	-.747	.471
L/E Cutaneous Pressure	-.217	.186	-.569	.581
L/E Strength	.135	.291	.307	.765
WISCI II	.112	.324	.749	.470
ABC	.180	.230	1.018	.331
Mini BESTest	.094	.350	-.081	.937
Anticipatory Sub-Test	-.046	.425	-.688	.506

Note: A Bonferroni adjustment for multiple comparisons was used for the correlations (p = .05/7 = .007).

4. DISCUSSION

The primary objectives of this thesis were: (1) to compare the proactive balance of individuals with chronic iSCI to age- and sex-matched AB individuals when walking over a known slippery surface including both feedforward and feedback-related changes to kinematic and EMG variables; (2) to determine whether individuals with iSCI are able to make feedforward adaptations to repeated exposures with the known slippery surface, and if so, how many trials it takes for adaptation to occur compared to AB individuals; and, (3) to determine the relationship between slip intensity and various clinical measures of strength, balance, and functional walking. The results of this study and whether they support the hypotheses will be discussed in detail below. Additionally, limitations of the study will be summarized along with future directions.

Group Differences in Kinematic Data

The hypothesis that individuals with iSCI would employ more pronounced proactive balance strategies compared to their AB peers when approaching a known slippery surface was partially supported. The main group effect found for all kinematic variables (except the MOS variables) indicates that, independent of condition, iSCI and AB individuals had significantly different group means. In other words, individuals with iSCI demonstrated a significantly more cautious walking strategy in all surface conditions (NW, US, and ES trials) compared to AB individuals and achieved this by walking with a greater step width, shorter step length, flatter foot-floor angle, and a slower COM velocity. Since a slip perturbation directly causes a shift in the BOS, it makes sense that individuals would walk with a larger step width to increase the size of the BOS; and that they would walk with shorter steps, a slower COM velocity, and a reduced foot-floor angle to limit horizontal shear forces and maintain stability while preparing to step on to the known slippery surface.

One interesting finding from the kinematic data was that, contrary to our hypothesis, individuals with iSCI did not have a significantly larger ML or AP MOS compared to the AB individuals during the US or ES trials. Although we were expecting to see a main group effect for ML and AP MOS like the other kinematic variables indicating a more cautious strategy used by individuals with iSCI, this finding was also found in previous literature [64], [65]. In a study by Arora and colleagues (2019) it was shown that during normal unperturbed walking individuals with iSCI took significantly shorter steps and walked with a slower velocity compared to AB individuals; however, despite the group differences seen in these kinematic variables individuals with iSCI showed similar AP MOS values compared to AB controls [65]. Perhaps group differences in ML and AP MOS are not seen because the greater use of proactive balance strategies among individuals with iSCI (i.e. decreased step length, slower walking velocity) are used to achieve a similar MOS value as AB individuals who are not limited by sensorimotor impairments. Alternatively, perhaps group differences were not seen in the MOS variables because the sample of individuals with iSCI in this study was quite high functioning (AIS D) making group differences less evident.

Overall, the results of this study are consistent with previous literature showing an increased use of proactive strategies of individuals with iSCI during normal walking compared to AB individuals [65]–[67]. The proactive strategies employed in this study by both groups are commonly reported strategies that are used in response to slippery surfaces and have previously been shown to be effective at reducing slip intensity in young and older healthy adults [15], [16], [42].

Group Differences in Muscle Magnitude

Typical proactive changes in muscle activity observed in young and older healthy adults involve an increase in muscle magnitude in preparation for walking on a known destabilizing surface [15], [38], [44], [51], [54], [112]. The data showed that individuals with iSCI had a significantly reduced proactive muscle magnitude (PiEMG) in the lead SOL and lead GM muscles as well as reduced reactive muscle magnitude (RiEMG) in the trail GM muscle compared to AB individuals. These results agree with previous literature that shows individuals with iSCI (AIS D) have a significantly reduced magnitude of EMG compared to AB individuals during both voluntary movements [115] and in response to unexpected perturbations while standing and walking [63], [64] which is likely due to underlying sensorimotor impairments from the SCI.

Since individuals with iSCI showed impaired muscle magnitude of the lead SOL, lead GM muscles (PiEMG), and the trail GM muscle (RiEMG) compared to AB individuals throughout the ES trials, this could infer that individuals with iSCI may use feedforward behavioural strategies to a greater extent than AB individuals to compensate for underlying motor impairments. Although we saw a reduced muscle magnitude in a few of the muscles tested, we were expecting to see a reduced magnitude for individuals with iSCI in most/all of the muscles tested but did not see this pattern. An explanation for the lack of group differences could be that the sample of individuals with iSCI in this study was quite high functioning (AIS D) and therefore group differences may not be as evident as they might be among individuals with a greater impairment.

Ability to Make Feedforward Adaptations to Behaviour

The hypothesis that individuals with iSCI would require at least one to two experiences walking on the slippery surface before showing significant feedforward adaptations, and that they would show a comparable rate of adaptation to AB individuals was partially supported. The lack of interaction effect found for all kinematic variables (mean step width, step length, foot-floor angle, COM velocity, ML MOS, and AP MOS) suggests that both iSCI and AB groups were able to make feedforward behavioural adaptations in a similar manner across the repeated slip trials. The main condition effect for mean step length, foot-floor angle, COM velocity, and AP MOS indicates that independent of group, both iSCI and AB individuals made significant adaptations to their behaviour due to knowledge that the surface would be slippery. Moreover, the results of post-hoc analyses showed that both groups were able to make appropriate feedforward behavioural adaptations after just one experience with the slippery surface (by ES1), which were maintained throughout all four expected slip trials. Specifically, both groups showed a significant increase in AP MOS and significant decreases in mean step length, foot-floor angle, and COM velocity when approaching the known slippery surface compared to the US trial.

There were a few findings from the kinematic data that did not follow the same trend as the other variables. Firstly, contrary to our hypothesis, neither group (AB or iSCI) increased their step width or ML MOS while approaching the known slippery surface compared to the US trial. The lack of feedforward changes found for step width and ML MOS results compared to the other kinematic variables could be attributed to the characteristics of the slip elicited from our device. The steel rollers were oriented such that they elicited a slip solely in the AP direction. Therefore, it is possible that we did not see as strong of an effect in the feedforward adaptations made to step width and ML MOS variables because these specific proactive strategies are mainly used to improve stability in the ML plane and thus, changes in these variables were not as appropriate for adapting to a slip perturbation in the AP direction.

Secondly, although no group differences were found for AP MOS, both groups showed significant feedforward changes to this variable during all four of the ES trials. The significant increase in AP MOS supports the hypothesis that individuals would proactively position their COM more anteriorly while approaching the known slippery surface. An anterior shift in the COM position is a commonly reported proactive balance strategy that has been shown to improve dynamic stability and protect against a slip-related backward loss of balance [3], [42], [49], [94]. This anterior shift in the COM-position was likely achieved through a combination of reducing step length and COM velocity when approaching the known slippery surface. A reduction in step length would make the length of the BOS smaller which would change the relationship between the boundary of the BOS and the xCOM. A reduction in step length would also help reduce the foot-floor angle which would ultimately reduce horizontal sheer forces at foot contact with the slip device and improve stability. A reduction in COM velocity would make it more feasible for the swing limb to catch up to the COM, thus making it more likely that the individual could recover if their balance was perturbed. Unfortunately, due to the complex nature of a dynamic MOS measure it is not possible to elucidate the precise mechanisms by which the AP MOS was regulated or how exactly the anterior shift in the COM was achieved with the data available from this study.

Finally, based on the main effect of group and condition found for step length, foot-floor angle, and COM velocity between the average NW and average ES trial values (2 x 2 ANOVA), it can be inferred that individuals with iSCI not only use more proactive strategies than AB individuals during normal walking [65]–[67], but that they continue to employ a greater magnitude of proactive strategies than AB individuals when approaching a known slippery surface. Specifically, the results show that when approaching a known slippery surface, individuals with iSCI achieve an increase in pre-slip stability by walking slower, with shorter steps, and a smaller foot-floor angle compared to their normal walking behaviour. These results are of particular importance because they indicate that, like AB controls, individuals with iSCI were able to further increase the extent of the proactive strategies they were using when approaching the known slippery surface, despite the fact that they were already using these strategies to a greater extent than AB individuals during normal walking [65], [67]. However, it is important to note that despite the greater use of proactive strategies seen in this study, individuals with iSCI still experience a higher fall rate than AB individuals. Thus, it is likely that individuals with iSCI use a greater amount of proactive strategies than AB individuals to try and compensate for impairments in their reactive balance control; but once their balance is perturbed they are more likely to experience a fall than AB individuals.

Ability to Make Feedforward Adaptations to Muscle Activity

Similar to the kinematic results, the majority of muscles measured (except for trail SOL) showed no interaction effect. This indicates that both iSCI and AB groups made feedforward adjustments in their proactive muscle activity in a similar manner across the repeated slip trials. One exception to this finding is that a significant interaction effect was found for the muscle activity of the trail SOL muscle. This suggests that the feedforward changes in muscle activity in the trail SOL muscle specifically were modulated differently over conditions and between the AB and iSCI groups. More specifically, the results of post-hoc analyses indicate that AB individuals significantly increased the amplitude of their trail SOL muscle activity during the second and fourth ES trials, while the individuals with iSCI were not able to proactively change the amplitude of their trail SOL muscle activity during any of the ES trials. These results agree with past research showing that typical proactive changes in muscle activity observed in young and older healthy adults involve an increase in muscle magnitude in preparation for walking on a known destabilizing surface [15], [38], [44], [51], [54], [112]. The differences in ability to proactively modulate the amplitude of the trail SOL muscle activity between the groups may be a result of impaired sensorimotor signaling due to the iSCI which likely contributes to the functional balance differences seen between the iSCI and AB populations and their slip/fall potential.

The plantar flexor muscles (i.e. soleus, lateral and medial gastrocnemius) of the trail limb have been shown to be involved in weight bearing and the control of foot placement which is important for stability when preparing to step onto a known slippery surface and in recovering balance after a slip has occurred [15], [54], [59], [112], [116]. In line with the results of this study, Marigold and Patla (2002) found that the medial gastrocnemius (MG) was the only muscle to show an effect of prior knowledge among healthy controls while walking on a known slippery surface [13]. Similarly, the main distinguishing feature of the feedforward changes to muscle activity seen between AB individuals and individuals with iSCI in this study was the impaired modulation of proactive trail SOL muscle activity. Impairments in modulation of this muscle specifically may limit the ability of individuals with iSCI to efficiently control their foot placement which may cause individuals with iSCI to rely more on the activation of analogous muscles or the increased use of behavioral strategies to compensate and properly prepare for a slip perturbation.

Finally, all muscles (except for the lead GM) showed a significant main effect of condition which indicates that independent of group, both iSCI and AB individuals were able to make appropriate feedforward adaptations to their muscle activity due to knowledge that the surface would be slippery. As stated previously, due to previous research on young and older healthy adults, it was hypothesized that individuals would proactively increase their muscle magnitude in preparation for walking on a known destabilizing surface [15], [38], [44], [51], [54], [112]. The results of post-hoc analyses showed that both individuals with iSCI and AB individuals were able to make similar adaptations to their PiEMG across conditions. Moreover, if a significant increase in the amplitude of proactive muscle activity did occur, both AB and iSCI groups were able to make these feedforward changes by the first (trail TA and trail GM) or second (lead and trail SOL) ES trial. However, unlike the kinematic data and contrary to our hypothesis, the adaptations to proactive muscle activity were not maintained throughout all four ES trials. The pre-slip muscle adaptation response (increase in PiEMG) consistently disappeared during the third ES trial for both the AB and iSCI groups only to return during the fourth and final ES trial. Since this drop in PiEMG was consistently seen in all the muscles and was not statistically significant on a group level there may be something unique about the third ES trial that influences the motor response produced by the CNS in a different way compared to the other ES trials.

Unfortunately, the data available from this study cannot elucidate the mechanisms responsible for the reduction in PiEMG during the third ES trial, but some speculation can be made. It is interesting to note that this pattern is predominantly seen in the PiEMG data but is not seen in any of the kinematic results. Another interesting thing to note is that the loss of effect in the third ES trial occurs for both the AB and iSCI groups, but the change appears to be more extreme in the AB group. Therefore, this reduction of EMG activity in the third ES trial is more likely a result of changes in natural walking behaviour due to the laboratory-based conditions or experimental protocol design, rather than differences in impairments between the SCI and AB groups. For instance, one possible explanation for the reduction in muscle activity seen in the third ES trial is that perhaps the slip was not challenging enough which caused the participants to become over confident after a couple of exposures to the slippery surface, hence reducing muscle activity in the third trial only to realize they still prefer a more cautious strategy and increasing their muscle activity again in the fourth ES trial.

The results from both kinematic and PiEMG data suggest that the ability of individuals with iSCI to make feedforward adaptations in response to an expected slippery surface is comparable to AB individuals. These findings agree with previous research that has shown that individuals with iSCI are able to adapt to standing balance perturbations much like their AB peers [63]. The results from this study suggest that the ability to use proactive balance control strategies to make adaptive, feedforward changes to behaviour and muscle activity remain partially functional after an iSCI through mechanisms of motor learning. This could potentially be attributed to sensorimotor tract sparing, neural plasticity, or regeneration. However, the main difference in function found between groups in this study was the ability to make proactive changes in the amplitude of the trail SOL muscle. Based on these results, special attention could be given to muscles of the trail/stance limb during motor rehabilitation, with a focus on improving function of the plantar flexor muscles (i.e. soleus) when preparing an individual to walk on a destabilizing surface.

Impact of Feedforward Adaptations on Post-Slip Parameters

It was hypothesized that, as a result of using proactive balance strategies, changes in both the feedforward (before contact with the slippery surface) and feedback (immediately after contact with the slippery surface) control of balance and locomotion would occur during the ES trials compared to the US trial [42], [49], [56], [93], [94]. Maximum PSV was used as the main post-slip outcome measure for this study and served as an indicator of the impact proactive balance strategies had on slip intensity. The PSV data showed a significant main group effect where during every slip condition, including the US, individuals with iSCI had less severe slipping speeds. This finding is consistent with previous research showing that individuals with iSCI who walked with more proactive balance strategies, particularly a slower walking velocity, during normal walking experienced less severe unexpected slips [65].

Additionally, PSV showed a significant main condition effect. Post-hoc analysis showed that both AB and iSCI groups were able to significantly reduce their maximum PSV within just 1 slip (by ES1) and maintain this adaptation over all ES trials. This quick adaptation of PSV and the strong maintenance of the response aligns with the proactive changes seen in the kinematic and PiEMG data. This suggests that both AB individuals and individuals with iSCI were able to effectively regulate slip intensity via proactive, feedforward adaptations. However, once again we feel that is important to recognize that the greater use of proactive kinematic strategies and the smaller post-slip velocity of individuals with iSCI does not necessarily mean that they are less likely to experience a slip/fall than their AB peers. These results may simply indicate that proactive strategies are being used to a greater extent by individuals with iSCI to compensate for impaired sensorimotor functioning and to reduce reliance on their impaired reactive balance control [64], [65].

In addition to the reduction of PSV, the feedforward behavioural adaptations that were made in preparation for walking on the slippery surface seem to have also had a positive influence on the reactive muscle responses. Although previous studies have reported a decreased magnitude of both reactive muscle responses while standing [63] and while walking [64] in the iSCI population, this study only found a significantly reduced magnitude of reactive muscle activity in the trail GM muscle of individuals with iSCI compared to AB individuals. Unexpectedly, post-hoc analysis of the main group effect from the lead TA revealed that the reactive muscle amplitude in the lead TA was actually higher among individuals with iSCI compared to AB individuals. Lead TA muscle activation has been shown to be important for slip recovery in healthy young and older adults [13], [15] and among individuals with iSCI [64]. Thus, the increased magnitude of lead TA muscle response in individuals with iSCI compared to AB controls may suggest that activity in this specific muscle has been primed by the CNS due to prior experience with slippery surfaces to compensate for reduced reactive muscle power due to iSCI [63], [64].

The lack of group and interaction effects found for RiEMG for most of the muscles investigated may indicate that the proactive adaptations that were employed were effective at mitigating the reactive balance impairments of individuals with iSCI [64]. This may lend an explanation as to why individuals with iSCI showed similar changes in RiEMG responses compared to AB controls over the ES trials. That being said, the results did show an interaction effect for the reactive muscle response of the lead SOL and trail TA muscles which indicates that individuals with iSCI and AB controls modulate the magnitude of these muscles differently in response to a slip. More specifically, individuals with iSCI had a smaller reactive muscle magnitude for lead SOL (ES1 & ES2) and trail TA (ES1) muscles compared to AB individuals. Finally, there was a main condition effect found for lead TA and trail SOL muscles but there was no clear pattern seen for the changes in magnitude over the ES trials indicating that reactive muscle activity was not consistently modulated across the repeated slip perturbations. Taken together, these results suggest that although individuals with iSCI are able to effectively use proactive balance strategies to reduce the intensity of a slip, they still have difficulties modulating the reactive muscle response of their ankle dorsi- and plantar-flexor muscles to a slip perturbation which may contribute to their greater fall risk compared to AB individuals.

In light of the results showing a reduced ability to modulate the reactive control of lead SOL and trail TA muscle activity and the critical role distal muscles of the ankle have been shown to have in reactive balance control [45], it is important to train individuals with iSCI how to use proactive balance strategies in everyday life to minimize the intensity of unexpected perturbations and reduce reliance on their impaired reactive balance control [64]. Additionally, this information may be able to help guide rehabilitation therapists to train individuals how to use more effective movement strategies that do not rely on specific sets of muscles that are often impaired as a result of iSCI (i.e. lead SOL and trail TA). For instance, if human movement can be roughly modelled as a series of linked body segments, a logical way to compensate for the reduced functional capacity of a particular segment (i.e. muscles about the ankle joint), may be to train the muscles in an adjoining segment to compensate for the restricted ROM or power at the compromised joint. For instance, therapists could encourage the use of reactive strategies that involve increased knee or hip excursion, or even the use of upper body movements, to help compensate for the impaired reactive muscle responses at the ankle.

Relationship Between Clinical Scores and Slip Intensity

The final hypothesis that the clinical measures for cutaneous pressure sensation, proprioceptive ability, lower extremity muscle strength, functional walking, and balance (particularly the sub-component of the mini-BESTest that measures anticipatory balance control) would be related to the ability to reduce slip intensity (measured as the change in maximum PSV from the US to the first ES trial) was not supported. The multiple regression model was not able to significantly predict the change in maximum PSV and none of the clinical assessment scores showed a significant correlation with the change in PSV. Furthermore, none of the clinical assessment scores added significantly to the prediction model on their own. This contrasts with previous literature that supports a relationship between slip/fall potential and a number of clinical measures including level of sensorimotor functioning, balance capabilities, lower extremity muscle strength, and functional mobility [7]–[9], [32], [70].

However, the fact that none of the scores on the clinical assessments were significantly able to predict the change in maximum PSV may suggest that the significant reduction in maximum PSV seen in the results from the US to the first ES trial can be attributed to the feedforward changes in gait characteristics and muscle activity and not to baseline differences in mobility, balance control, or sensorimotor functioning. The lack of relationship between clinical scores and changes in maximum PSV in combination with the significant reduction in PSV observed during the ES trials may provide preliminary support for the effectiveness of proactive balance strategies in reducing slip intensity in individuals with iSCI, similar to research conducted on young and older healthy adults [15].

Since both iSCI and AB groups were able to significantly reduce slip intensity (measured functionally as a change in maximum PSV from the US to the first ES trial) this indicates that successful and meaningful adaptation to a known slippery surface can be achieved through the use of appropriate proactive changes to behaviour and muscle activity, despite underlying sensorimotor impairments in the iSCI group. However, it is important to remember that even though individuals with iSCI demonstrated a greater use of proactive strategies and subsequently a lower PSV; that this does not necessarily mean they are less likely to experience a slip-related fall. In fact, due to their impairments in reactive balance control, reductions in strength and coordination [32], [63], [64], if individuals with iSCI did experience a loss of balance they may be more likely to experience a fall compared to AB controls.

4.1 Limitations

The results of this study should be interpreted carefully after considering the limitations involved. Firstly, this study had a relatively small sample size ($n_{iSCI} = 19$, $n_{AB} = 17$) which may have contributed to a lack of significant differences in the iSCI vs AB group comparisons or in follow-up post-hoc testing. The participants were all caucasian which is not diverse or representative of the entire iSCI population. Additionally, since all the participants with iSCI were a grade D on the AIS impairment scale, the results of this study can only be generalized to individuals with iSCI who have high functioning levels.

Secondly, there were some limitations in the techniques used for data processing and analysis. Some of the measures for participants were excluded as outliers or could not be calculated due to errors in the collected data (noisy/missing EMG signals or incomplete kinematic data). These errors resulted in some missing values in the data sets that may have affected the power of the statistical analyses. Another factor that would have affected the power of statistical analyses was that multiple individual RM ANOVA's were conducted. Although necessary to track the feedforward changes in multiple kinematic and EMG variables over repeated slip perturbations, this would have resulted in an increased risk of type I error. Moreover, the second set of analyses (2 x 2 RM ANOVA) used to compare average behaviour during the expected slip trials to the average normal walking behaviour could not be conducted for the EMG data because the values from normal walking had already been used to normalize the EMG data so that between-group comparisons could be made. Additionally, the qualitative analysis was subjective making the classifications of walkover versus skateover slip strategies subject to researcher bias.

Thirdly, the nature of the experimental protocol may have caused inherent constraints to the reliability and generalizability of the results. For instance, to avoid fatigue among individuals with iSCI, no walking familiarization trials were provided. Therefore, the only time the participant had to become comfortable with the laboratory set-up and wearing the equipment was during the standing, hip, and knee calibration trials. The limited familiarization period may have affected the first few NW trials by allowing for the possibility of a learning effect. However, to minimize the effects of any learning , only the last three of all available NW trials before the US perturbation were used in analysis to calculate the NW average. Another limitation of the techniques used during data collection is that during the joint position sense test it is impossible to avoid stimulation of cutaneous receptors. To minimize the effects of tactile information influencing participants responses, the examiner held onto a bony prominence, moved the joints slowly, and was careful not to pull on the skin too much to avoid giving directional cues.

Although we used the SCATS as a measure of spasticity, this assessment does not provide detailed information on which muscles specifically were most affected by spasticity. For instance, if flexor spasms were observed when the knee was moved, we were unable to determine whether the spasticity was specifically affecting the hamstrings, the gastrocnemius, the sartorius, or a combination thereof. Variability could have been introduced into the EMG data depending on which specific muscles were most affected and which type of spasticity (clonus, flexor, or extensor spasms) was most prevalent in each participant as a result of iSCI. This inter-subject variability could have affected results regarding the ability of the iSCI group as a whole to proactively and reactively modify the amplitude of certain muscles during certain phases of the gait cycle. Considering spasticity is a velocity-dependent phenomenon [117], high velocity joint movements such as knee flexion/extension during gait or ankle/knee movements in response to a perturbation, are most likely to have been negatively impacted. Bravo-Esteban and colleagues (2013) showed that different symptoms of spasticity have specific effects on function [95]. For instance, individuals with iSCI and hypertonia showed reduced voluntary flexor muscle activity, while extensor spasms contributed strongly to impaired gait and daily activities [95]. Unfortunately, since the level of detail regarding spasticity was limited to the SCATS measurement in this study and most participants (except for four) showed no spasticity, this data was not able to be included in the regression model. Therefore, any predictions as to how the degree of spasticity is related to the ability to adapt to a known slippery surface can not be made. This may be an interesting area of study for future research.

Some other factors that may have influenced the results of this study were the conditions in which the participants were asked to walk. The experiment was conducted in an indoor laboratory-based setting where participants walked while wearing a safety harness on a relatively small (10 m) walkway with researchers watching them. These factors may have influenced their natural walking behaviour from the onset of the experiment. For instance, the average COM velocity during the NW trials (1.02 ± .26 m/s) for AB participants in this study was notably slower than previously reported normal ranges of walking velocity for AB (1.2 - 1.4 m/s; [53], [93], [118], [119]). Although the average walking velocity in this study was slower than previously reported normative values, it remained relatively consistent with NW averages from previous studies that have been conducted in the same laboratory environment; .94 m/s for AB individuals age 18-65 years [106] and .95 m/s for AB individuals age 29-94 years [65]. The average walking velocity found in this study may also be slower than normative values due to the wide range of ages we included in this study (29.2 - 94.1 years old); however, we are still unable to account for the influence that walking in an indoor laboratory setting may have on the natural walking behaviour of the participants.

Another environmental factor that may have negatively influenced both the NW and slip trials was the size and nature of the slip device. The fact that the rollers were visible may have influenced the participants gait from the onset of the experiment and may have caused them to slowly build suspicion after repeated trials even though they reported that they experienced a true unexpected slip. Additionally, since the slip device was only .46 m in length it is possible that the participant's foot could have slid and hit the floor tiles at the edge of the rollers which could have stopped the forward acceleration of their foot prematurely. Although having a small slip distance is good for the safety of the participants, this may have limited the severity of the slip and thus the extent of proactive strategies used, and the size of adaptation observed. Moreover, since participants were wearing a safety harness that was not instrumented with force sensors and prevented them from making contact with any lower surface, if a loss of balance did occur it was difficult to quantify whether that loss of balance would have been severe enough to elicit a fall or not if they were in a real-world scenario.

Furthermore, the steel rollers used did not create an entirely frictionless system. Although the small amount of available friction may have helped in slowing the forward acceleration of the COM, we believe that the results are generalizable to real-life slip scenarios because previously reported static coefficient of friction for rubber-soled shoes on ice ($\mu_s = .17$), and leather-soled shoes on ice ($\mu_s = .09$) are very similar to that of the steel rollers used in this study ($\mu_s = .09$) [120]. Another limitation to the slip device used in this study is that it only elicited a slip in the AP direction. This controlled, unidirectional slip served as a good starting point for studying walking balance adaptation in individuals with iSCI; however, this may not be representative of a real-world slip. Therefore, the investigation of adaptation to an omni-directional slip in the iSCI population may be an interesting area of study for future research.

Although the constraints of the lab setting made it so that the slip did not exactly resemble a slip in real-world scenarios, we are confident that we were able to elicit a genuine unexpected slip response and appropriate feedforward- and feedback-based adaptations to the expected slip trials seen previously in literature [13], [15], [16], [38], [42], [49], [52]. However, one result that remains unexplained was the consistent drop in PiEMG amplitude observed in the third ES trial. An interesting area for future research could be to try and determine why the adaptive muscle response consistently disappeared in the third ES trial. It would be useful to determine what makes the third ES trial different from the other ES trials (i.e. distinct CNS mechanisms, environmental factors, psychological factors etc.) and whether this pattern would continue to occur over many repeated trials.

5. CONCLUSION

Both AB individuals and individuals with iSCI were able to make significant feedforward changes to their balance strategies when approaching a known slippery surface within one to two trials after experiencing an unexpected slip. Individuals with iSCI used proactive balance strategies to a greater extent than AB individuals while approaching the known slippery surface. These proactive balance strategies were indicative of more cautious walking and included walking with shorter steps, a flatter foot-floor angle, a slower COM velocity, an anterior shift in the COM, and increased muscle activity of the lead SOL, trail TA, and trail GM muscles. Additionally, both AB individuals and individuals with iSCI were able to maintain the use of the feedforward behavioural adaptations during all four ES trials which were effective at reducing maximum PSV and thus the slip/fall potential. The results of this study alone can not directly provide evidence for the effectiveness of perturbation-based balance training in the iSCI population. However, the findings that individuals with iSCI were able to adapt their behaviour and muscle activity in a feedforward manner to reduce slip intensity, similar to AB individuals, provides preliminary support that these types of balance training protocols might be effective at reducing the prevalence of slip-related falls in this population.

The results of this study in combination with the body of literature investigating the effectiveness of perturbation-based balance training in other populations [50], [55]–[59] suggest that perturbation-based balance training could be used in populations where slips and falls occur more frequently (e.g. iSCI, elderly, individuals with stroke) to reduce fall potential in a variety of destabilizing conditions including slippery surfaces. Specifically, balance training could be combined with motor learning techniques to train individuals how to use appropriate proactive strategies to minimize the likelihood of slipping and falling based on prior experience with specific surface conditions [13], [44], [49], [50], [56], [121]. Next steps would be to conduct perturbation-based balance training interventions in the iSCI population to see if benefits in balance and stability are translated to real-life and whether this type of training can have long-term benefits for individuals with iSCI.

Currently, we recommend that clinicians use the results from this study to inform and educate individuals with iSCI about their balance constraints and abilities. Making individuals with iSCI self-aware that although they have impaired reactive balance [63], [64], they are able to use proactive strategies much like AB individuals to prepare themselves and adapt to destabilizing surfaces. We hope that by improving the self-awareness of their balance abilities and constraints that this will give individuals with iSCI (AIS D) the confidence they need to participate in balance training programs and implement strategies into real-life; which will hopefully lead to improvements in dynamic balance control and reductions in slip/fall potential.

Additionally, one of the main findings from this study was that compared to AB individuals, individuals with iSCI exhibited an impaired ability to proactively modulate the amplitude of the trail SOL muscle in preparation for a slippery surface and exhibited impaired reactive motor responses in the lead SOL, trail TA, and trail GM muscles. Therefore, we recommend that existing motor rehabilitation programs focus on improving the strength and function of the plantar flexor muscles (i.e. soleus) when preparing an individual to walk on a destabilizing surface. We also recommend that rehabilitation programs provide individuals with repeated practice walking on many different types of surfaces. With sufficient practice on a wide variety of surfaces, it is possible that individuals with iSCI could prime their CNS programs to be more efficient at employing an appropriate motor response to better prevent a hazardous slip despite their reactive balance impairments [49], [50], [56], [59].

In conclusion, it has been demonstrated that individuals with iSCI are capable of using proactive balance strategies to adapt to a known slippery surface in a similar manner to AB individuals. Additionally, we know that the proactive strategies these individuals use are effective at reducing maximum post slip velocity and thus slip intensity. This provides hope that, although individuals with iSCI have impaired reactive balance control, we may be able to use motor learning principles and perturbation-based training in rehabilitation to improve their feedforward control of balance and help compensate for their impaired reactive balance control. Future studies are needed to establish the ecological validity and effectiveness of perturbation-based training protocols in the iSCI population to see if they would provide benefits that can be translated to real world situations and reduce the risk of falls.

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APPENDIX A.
The Modified Spinal Cord Injury Functional Ambulation Profile (SCI-FAP)

The modified SCI-FAP is composed of 4 tasks: (1) Carpet, (2) Up & Go, (3) Obstacles, and (4) Step. A fifth task, (5) Stairs, will also be tested. Each participant is given a rest period between tasks long enough for the tester to explain and demonstrate the next task. Each participant is instructed to use an assistive device and/or brace(s) as needed. The tester provides instructions and answers the participant's questions. The tester provides physical assistance if needed. The tester times the participant during each task. The tester walks behind the subject, not beside, to prevent affecting the participant's speed. The tester provides feedback/encouragement only after the task is completed. The tester records the performance time for all 5 tasks on a data collection form. If the participant cannot attempt a task, or does not complete a task, he/she is assigned the maximum time for that task, and an assistance rating of 6 ('unable to complete') (see scoring table). If the participant takes longer than the maximum time to complete a task, he/she is assigned the maximum time, and the assistance rating that corresponds to the devices/assistance used for that task.

1) Carpet **Max time**: 220 seconds

Setup: Carpeted area or a piece of short pile carpet, no less than 7-m long and 2-m wide. A 2-cm strip of masking tape is placed 1-m from one edge of the carpet. Another 2-cm strip of masking tape is placed exactly 5-m from the first 2-cm piece of masking tape. Both pieces of tape are at least 1-m from the edge of the carpet. The starting point is a 1-m strip of masking tape placed 1-m before the first 2-cm piece of masking tape (this may be at the edge of the carpet).

1. Tester explains while demonstrating the Carpet task: "When you are ready, walk at your normal, comfortable pace until I say 'stop.' "

2. Tester assists participant as needed in placing toes on starting line tape.

3. Participant starts walking. Once participant's first foot crosses the first 2-cm piece of tape, tester presses stopwatch to begin timing.

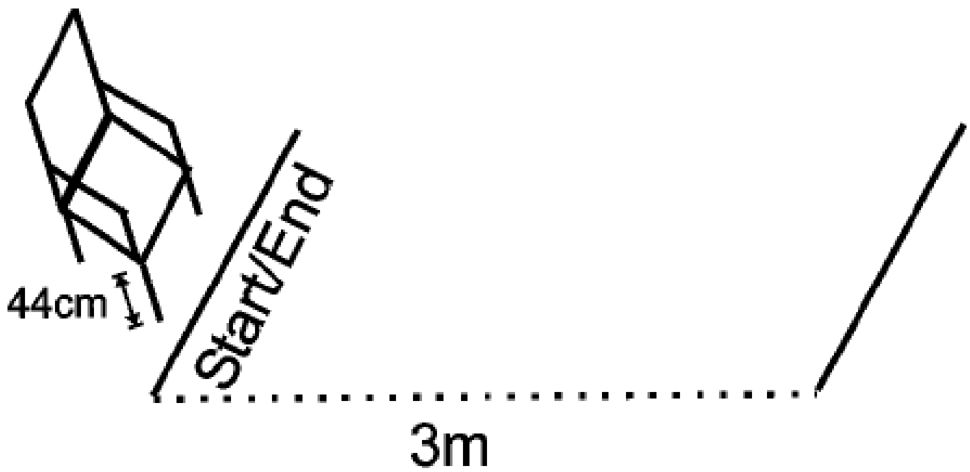
4. Tester walks behind the participant as the participant traverses the 5-m distance.

5. Tester presses stopwatch to stop timing once both of the participant's feet have crossed the second 2-cm piece of tape. Tester tells the participant to stop when he or she is at least 1-m beyond the second 2-cm piece of tape.

6. Tester records time and assistance required on data collection form.

2) Up & Go **Max time**: 455 seconds

Setup: A chair with a backrest and armrests is placed on the hard, non-carpeted floor. The seat height should be about 44cm. A 1-meter long piece of tape is placed 3 meters away from the start line. The participant's toes should touch the start line when seated.



1. Tester explains while demonstrating the Up & Go task: "You will sit in this chair with your back against the back of the chair. When you are ready, you will stand up from the chair, walk at your normal comfortable pace to the wall, touch the wall, turn around, walk back to the chair, and sit down, making sure your back is against the back of the chair."

2. Participant assumes sitting position in the chair. Tester helps participant place toes on starting line. Tester stands beside the chair and prepares to walk with the participant.

3. Tester presses stopwatch to begin timing once the participant initiates task by moving back away from backrest.

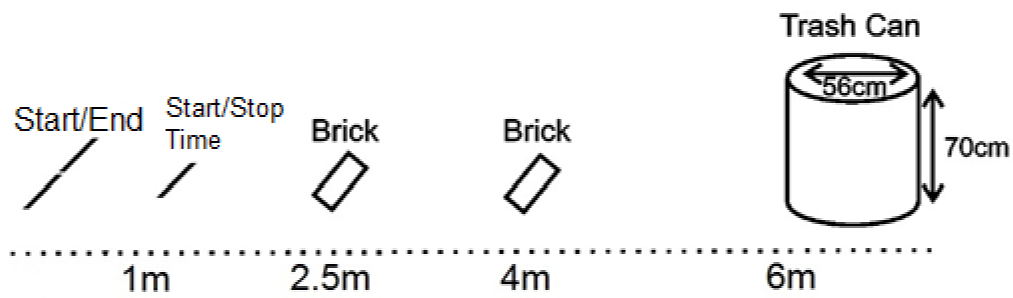
4. Tester monitors 3-meter line to ensure participant's feet cross the line before turning around.

5. Tester stops timing when participant is fully seated with back against the chair.

6. Tester records time and assistance required on data collection form.

3) Obstacles* **Max time**: 570 seconds

Setup: A 1-m piece of masking tape is placed on a hard, non-carpeted floor to mark the starting point. A 2-cm piece of masking tape is placed 1m from the start line. A 5cmx5cm Styrofoam is placed on the floor at the 2.5-m mark and a 10cmx10cm Styrofoam at the 4-m mark. A trash can (diameter 56cm, height 70cm) is placed at the 6-m mark. The end point is the same as the starting point.



1. Tester explains while demonstrating the Obstacles task: "When you are ready, walk forward at your normal, comfortable pace and step over each Styrofoam. Then, walk around the trash can from either the left or right. Then walk back stepping over the Styrofoam again. Continue to walk until I say 'stop'. Do not hit the bricks or bin with your body or walking aid, if possible."

2. Tester assists participant as needed in placing toes on starting line.

3. Participant starts walking. Tester presses stopwatch to begin timing once the participant's first foot crosses the first 2-cm piece of tape.

4. Tester walks with participant.

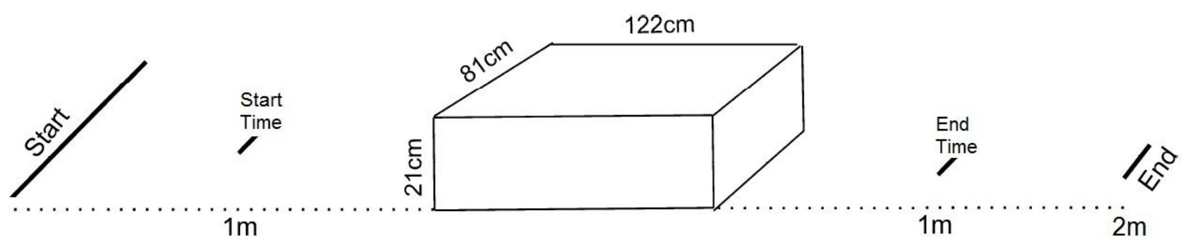
5. When both of the participant's feet have crossed the end line, tester presses stopwatch to stop timing. Tester tells the participant to "stop" when he or she is beyond the end line.

6. Tester records time and assistance required on data collection form.

* If the participant hits one or more of the obstacles with his/her body or walking aid, 1 is added to the factor chosen for this task (e.g., if participant completed task with '1 cane/crutch' – a factor of 2, but he/she hits 1 or more obstacles, he/she is assigned a factor of 3).

4) Step **Max time**: 185 seconds

Setup: A step with the measurements shown in the diagram below is used on hard, non-carpeted floor. Two pieces of masking tape are placed on the floor to indicate the start and finish points. The first, 1-m in length, is placed 2-m in front of the step. The second piece, shorter in length, is placed 2-m behind the step. Two 2-cm pieces of masking tape are placed on the floor to indicate when to start and stop timing. The first is placed 1-m from the start line. The second is placed 1-m behind the step.



1. Tester explains while demonstrating the Step task: “When you are ready, walk towards the step, up and over, and continue walking until I say stop. Walk at your normal, comfortable pace.”

2. Tester assists participant as needed in placing toes on the starting point.

3. Participant starts walking. Tester presses stopwatch to begin timing once the participant’s first foot crosses the first 2-cm piece of tape.

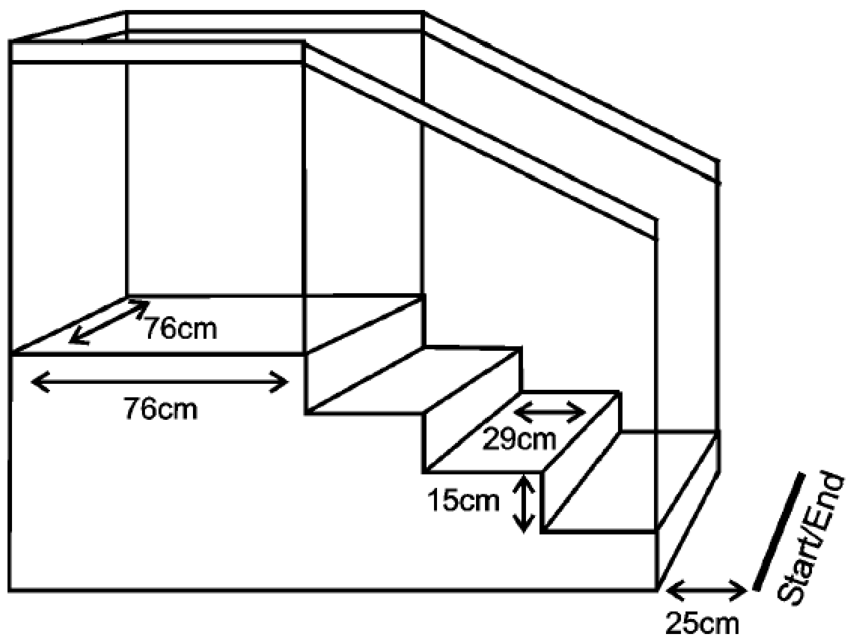
4. Participant walks toward the end point. Tester follows participant through the task for safety.

5. Tester presses stopwatch to stop timing when both of the participant’s feet have crossed the second piece of tape. Tester tells the participant to “stop” when he or she is beyond the end line.

6. Tester records time and assistance required on data collection form.

5) Stairs*   **Max time:** 310 seconds

Setup: Stairs with 4 steps, hand railings on both sides, and the following measurements are utilized: 29-cm stair depth, 76-cm stair width, 15-cm stair height, 76-cm platform depth, and 76-cm platform width. A 1-m piece of masking tape is placed 25 cm from the base of the first step.



1. Tester explains while demonstrating the Stairs task: “When I say ‘go,’ walk up the stairs at your normal, comfortable pace to the top of the stairs, turn around, and come back down. You may use the handrails if needed, but try to use them as little as possible.”

2. Tester assists participant as needed in placing toes on starting line.

3. Tester says “go,” and presses stopwatch to begin timing.

4. Tester follows participant up stairs to guard.

5. Tester presses stopwatch to stop timing when both of the participant’s feet are in firm contact with the floor.

6. Tester records time on data collection form.

*Participant may use any technique to ascend and descend stairs (i.e., forwards, backwards, sideways), but must turn around at the top of the stairs so that he/she approaches the descent from the forwards direction. The technique used is recorded under the “Comments” sectio

APPENDIX B.
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Permission was obtained from Dr. Tarun Arora before using images found in Figure 2.1 showing a schematic representation of the experimental setup and slip device originally used in his study.

TA

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2019-06-17 8:35 PM

To: Bone, Mackenzie; Mackenzie Bone

Hi Mackenzie
Please consider this email as official permission to use the diagram of the schematic layout of the experimental set-up from my thesis.
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PhD Health Sciences
University of Saskatchewan