CONTROL STRATEGIES DURING GAIT TERMINATION: ELUCIDATING THE MECHANISMS OF ANKLE INSTABILITY

By

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To my parents who taught me to dream and my husband who gave me the strength and support to fulfill those dreams.

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Chronic ankle instability (AI) is a functionally debilitating condition that occurs after the first episode of injury to the ligaments of the ankle joint. Both feedback and feedforward mechanisms of neuromuscular control have been reported to be affected in people suffering from AI. However no previous researchers have examined these deficits in the same population. In this study we used the gait termination model to challenge both the feedback and feedforward mechanisms of neuromuscular control in the same population of subjects suffering from AI and compared the results to controls. Planned gait termination was used to reveal deficits in feedforward neuromuscular control whereas unplanned gait termination was used to reveal the deficits in the feedback neuromuscular control. Thus the purpose of this study was to reveal the type of alterations in neuromuscular control that exist in AI. Force, stability and EMG were used as outcome measured to reveal these deficits. The study was a single session, single subject mixed model design. Twenty participants were recruited in the AI group (age 20.2±1.2years, height 169.8±9.7cm and mass 74.2 ±20.2kg) and 20 participants were recruited in the control group (age 20.4± 1.6years, height164.3 ±7.9cm and mass 64.2 ±10.6kg). The Ankle Injury Questionnaire was used to screen the participants. Each eligible participant performed both unplanned and planned gait termination trials. Normal walking trials were interspersed to avoid

anticipation of stop. Ten trials of both planned and unplanned gait termination were captured with each limb serving as a lead limb in 5 trials. Ground reaction forces were collected using two force plates (0.4m x 0.6m) (Type 4060-10 Bertec Corporation, Columbus, Ohio). The GRF data along X, Y and Z axes were use to calculate dynamic postural stability index (DPSI) and its subcomponents (APSI, MLSI and VSI). A KONIGSBERG T-42AL-8T (Konigsberg Instruments Inc, California) telemetric electromyography unit was used to record muscle activity of Tibialis Anterior (TA), Soleus (Sol) and Gluteus Medius (GM) using bipolar 1-mm x 10-mm Ag/AgCl surface electrodes. A 3-way MANOVA (2:group x 2:limb x 3:condition) with repeated measures on the last factor revealed significance differences in propulsive force and braking force. Both propulsive force and braking force in the AI group were higher than the control group during unplanned gait termination and the AI group relied more on lead limb strategy during gait termination. A 4-way MANOVA (2:group x 2:limb x 6:phases x 2:condition) with repeated measures on the last factor revealed significance differences in average amplitude of TA, Sol and GM. The average amplitude of TA for the involved limb was less than the uninvolved limb during all four subphases of stance. The average amplitude of TA, Sol and GM was higher during unplanned gait termination than during planned gait termination. Average amplitude of Sol in the AI group was less than that in the control group. This finding failed to explain the result that the AI group generates a higher braking force and suggested that another muscle might be responsible for producing the braking force in the AI group during gait termination. A 3-way MANOVA (2:group x 2:limb x 2:condition) with repeated measures on the last factor revealed higher DPSI and APSI scores in the AI group and involved limb during unplanned gait termination than during planned gait termination. All these finding suggest that feedback and feedforward deficits of neuromuscular control coexist in AI.

CHAPTER 1
INTRODUCTION

Chronic Ankle Instability

A lateral ankle sprain is one of the most commonly occurring musculoskeletal injuries which affects both athletes and non athletes (Osborne & Rizzo, Jr., 2003). The incidence of ankle sprains is extremely high, with approximately 23,000 ankle sprains occurring each day in the United States (Hertel, 2002). In addition, this incidence could be an underestimate as 56.8% of individuals do not seek conventional medical treatment (Gerber, Williams, Scoville, Arciero, & Taylor, 1998).

The process of healing from this injury may be complete in three to six weeks, however, the functional recovery of the injured ankle may be difficult and sometimes impossible to achieve (McKay, Goldie, Payne, & Oakes, 2001). Injury of the ankle ligament commonly leads to a condition called ‘Chronic Ankle Instability’ (CAI) in which the person complains of weakness and ‘giving way’ or ‘rolling over’. Fifty to 75% of the people who sprain the lateral ligaments of their ankle suffer chronically from pain and instability (Yeung, Chan, So, & Yuan, 1994).

While the exact cause of incomplete functional recovery of the ankle is unknown, preliminary evidence suggests that both feedback and feed forward neuromuscular control may play a role (Delahunt, Monaghan, & Caulfield, 2006; Freeman, Dean, & Hanham, 1965; Monaghan, Delahunt, & Caulfield, 2006). “Stimulation of a corrective response within the corresponding system after sensory detection is often considered as a feedback control mechanism”. In contrast, feedforward control mechanisms have been described as “anticipatory actions occurring before the sensory detection of a homeostatic disruption” (Johansson & Magnusson, 1991; Riemann & Lephart, 2002). The injury to the ligament complex that occurs during trauma to the ankle may affect the mechanoreceptors and the proprioceptors in the ankle joint (Hertel, 2002). This may result in decreased afferent input through these receptors. Thus, during gait the proprioceptive input is decreased causing an alteration in the feedback neuromuscular control of the ankle. In addition, researchers have examined different movement and muscle recruitment patterns in patients with CAI. Changes in the movement patterns and muscle recruitment after the first episode of ankle injury appear to exist (Delahunt et al., 2006). As the pre programmed movement patterns demonstrate alterations, the feed forward mechanism of neuromuscular control may also be a contributing factor towards Ankle Instability (AI).

Significance of gait termination as a model for studying AI: To date, researchers have tried to uncover the mechanism of AI by using a jump landing protocol (Ross & Guskiewicz, 2004b; Wikstrom, Tillman, Chmielewski, Cauraugh, & Borsa, 2007). However, the activity of jump landing has multiple degrees of freedom (due to multiple lower extremity joints) and a person may be successful in maintaining balance in spite of alterations in normal ankle strategies. Alterations in the neuromuscular control of the ankle if any, could be revealed by an experimental model involving gait termination. Gait termination involves a rapid deceleration of the forward momentum of the body during steady gait. A safe termination of gait requires a complex interaction of the neuromuscular system (Hase & Stein, 1998). Furthermore, the gait termination model replicates a functional activity and does not allow for multiple degrees of freedom. It also possesses a known and repeatable set of neuromuscular responses (Bishop, Brunt, & Marjama-Lyons, 2006; Bishop, Brunt, Kukulka, Tillman, & Pathare, 2003; O'Kane, McGibbon, & Krebs, 2003) and gait termination experiments can be constructed to challenge both feedforward and feedback neuromuscular control (Bishop et al., 2006). Changes in neuromuscular controls in an unstable ankle may be studied by designing experiments to modulate the style of gait termination. Moreover a better understanding of these changes could be possible by specifically studying the force and muscle activity of the lead limb (the first limb to land on the second force plate) and the trail limb (the first limb to land on the force plate) during gait termination.

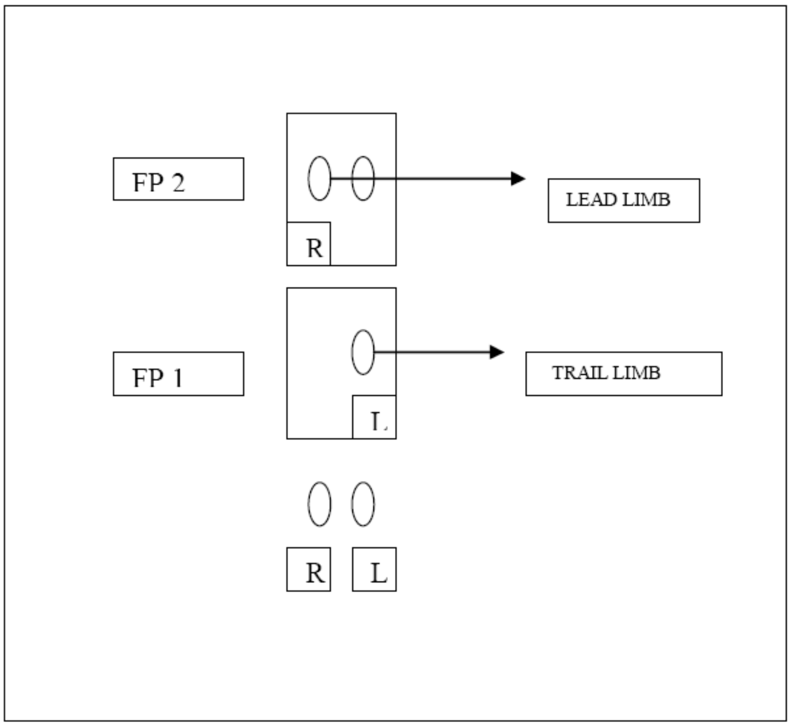


Figure 1-1 Nomenclature of the limbs during gait termination, FP1 = Force Plate 1 and FP2 = Force Plate 2

'Planned stopping' or termination of gait challenges the feedforward mechanisms of neuromuscular control. Similarly, sudden or 'Unplanned' gait termination will challenge the feedback mechanism of neuromuscular control to maintain stability and balance. Analyses of force, postural stability and muscle activation ankle joint during 'planned' and 'unplanned' gait termination may reveal the alterations in feedforward and feedback mechanisms of neuromuscular control respectively.

Hypotheses

The central hypothesis of this study is that differences between feedback and feed forward mechanisms of neuromuscular control related to ankle mechanics would be revealed by using the gait termination model. The following are the specific hypotheses for the proposed study.

Forces

Propulsive force

- The propulsive force during unplanned gait termination would be greater than the propulsive force during planned gait termination. Unplanned gait termination implies an unanticipated and sudden stop which does not provide time to reduce the propulsive force as during planned gait termination.
- The propulsive force for the AI group would be lower than that of the control group. Ligament injuries that accompany repeated ankle sprains in CAI contribute to feedback deficits in neuromuscular control (Freeman et al., 1965). Furthermore, the average amplitude of muscles responsible for generating the propulsive force at the ankle (TA and Sol) has been documented to be significantly lower in the AI group as compared to the controls during normal gait (Delahunt et al., 2006) .
- The propulsive force of the involved limb would be lower than the propulsive force of the uninvolved limb during gait termination. The evidence supporting hypothesis b) explains this hypothesis as well.

Propulsive time

Propulsive time is defined as the time from the instant of heel strike to point of maximum propulsive force generation in stance limb.

- The propulsive time for unplanned gait termination would be shorter than that observed during planned gait termination. During gait or gait termination an impulse (I = F x t) must be created within each limb. Time and force are inversely related hence the propulsive time was shorter when the propulsive force is higher.
- The propulsive time for the AI group would be longer than that of the control group. In addition to the inverse relationship between force and time, this hypothesis is based on the research findings of Nyska (2003) who demonstrated slowing down of weight transfer from heel strike to toe off during normal gait in AI.
- The propulsive time of the involved limb would be longer than that of the uninvolved limb. Similar to the hypotheses above, slowing down of weight transfer during stance phase could result in a longer propulsive time in the injured limb.

Braking force

- The braking force during unplanned gait termination would be greater than that seen during planned gait termination. Unplanned gait termination implies an abrupt termination of gait. Hence a greater braking force was generated to reduce the body's forward momentum.

- The braking force for the AI group would be higher than that of the control group. Decreased conduction velocity in the tibial and peroneal nerves has been reported in subjects having chronic ankle instability (Jazayeri Shooshtari, Didehdar, & Moghtaderi Esfahani, 2007). This finding may indicate a decrease in the braking force generated by Sol when the injured limb is the lead limb. Likewise, it may also indicate a decrease in the activity of the TA which is responsible for producing a braking force when the AI limb is a trail limb. Hence a much higher braking force of the uninjured limb will be necessary to arrest the forward momentum of the body during gait termination.

- The braking force of the involved limb would be lower than the uninvolved limb. As stated above, a decrease in conduction velocities of tibial and peroneal nerves of the injured limb may restrict force generation.

Braking time

- Braking time for the unplanned gait termination would be shorter than that observed in planned gait termination. Assuming that braking force is increased as hypothesized above and understanding that a similar impulse must be generated to terminate gait during both planned and unplanned trials, unplanned gait termination should require a shorter braking time.

- Braking time for AI group would be shorter than that seen in the control group. An Assumption based on the above hypothesis that the braking force in AI group would be higher during gait termination and understanding the impulse relationship ($I= F \times t$), braking time in AI group would be shorter.

- The braking time of the involved limb would be significantly longer than that seen for the uninvolved limb. The hypothesis that braking force generated by the injured limb would be lesser than the uninjured limb along with the understanding of the impulse relationship could mean a longer braking time in the involved limb during gait termination.

Postural stability

- The AI group would be less stable than the control group. Wikstrom and colleagues (2007) demonstrated deficits in the dynamic postural stability in the subject with history of AI using a 'single leg hop-stabilization' model.

- Unplanned gait termination would be less stable than planned gait termination. Unplanned gait termination implies an abrupt stop in gait. An associated higher braking force as hypothesized above could cause unplanned gait termination to be less stable.

- The involved limb would be less stable than the uninvolved limb. 'Jump landing' models have demonstrated a decreased dynamic postural stability in the involved limb subjects with AI. (Ross & Guskiewicz, 2004a; Wikstrom et al., 2007; Docherty, Valovich McLeod, & Shultz, 2006; Ergen & Ulkar, 2008; Evans, Hertel, & Sebastianelli, 2004; Ross et al., 2004a; Wikstrom et al., 2007). However this is the first study using a gait termination model to investigate the deficits in dynamic postural stability in subjects with AI.

Muscle Activation

Distal muscles (TA and Soleus)

- The average amplitude of muscle recruitment for the distal muscles (TA, Sol) for the AI group would be decreased as compared to the controls. A decrease in average amplitude of TA and Sol is reported in subjects with AI during normal gait (Monaghan et al., 2006) . Hence a decrease in average amplitude of distal muscles is expected even during gait termination.

- The average amplitude of the distal muscles (TA and Sol) for unplanned gait termination would be increased relative to planned gait termination. Unplanned gait termination involves an unanticipated stop with presumably increased force generation. Hence it is likely that the distal muscles show higher average amplitude than that expected in a planned stop.

- The average amplitude of the distal muscles (TA and Sol) for involved limb would be lesser than that seen in the uninvolved limb. The average amplitude of the distal muscles of the injured limb has been reported to be lower than that of the uninjured limb (Delahunt et al., 2006) and conduction velocity of the tibial and peroneal nerves is reduced.

Proximal muscle (Gluteus Medius)

- The average amplitude of muscle recruitment for the proximal muscle (GM) for the AI group would be higher as compared to the controls. A decreased reliance on the distal musculature for gait termination could imply an increased reliance on the proximal muscles for arresting the forward momentum during gait termination.

- The average amplitude of GM during unplanned gait termination would be greater than that seen in the planned gait termination. This hypothesis is based on a similar reasoning as that of the above hypothesis

- The average amplitude of GM in the involved limb would be greater than that seen in the uninvolved limb.

Significance of the Study

Many authors have investigated the alterations in either the feedback or the feed forward mechanisms in individuals with AI. However the potential difference in the feedback and feed forward mechanisms in the same set of subjects with ankle instability remains unexamined. Thus, this study is designed to examine the underlying alterations in the feed forward and feedback mechanisms in the same population of AI subjects. Specifically, I compared the ground reaction forces and dynamic postural stability during gait termination in AI subjects and controls. Potential differences in muscle activation strategies will also be evaluated in both groups during ‘planned and ‘unplanned’ gait terminations. This was useful in understanding the alteration in the neurophysiology in AI. Until the mechanism of AI is elucidated, prophylactic and rehabilitative interventions will remain ineffective for this widely prevalent and functionally debilitating musculoskeletal injury.

CHAPTER 2
LITERATURE REVIEW

Introduction

The ankle joint bears more weight per unit area than any other joint in the body (Fallat, Grimm, & Saracco, 1998) and is one of the most commonly injured joints of the body. The extreme forces that are placed on the ankle joint make it more susceptible to injury (Fallat *et al.*, 1998). Seventy-five percent of the injuries caused to the ankle are ligament injuries and 85 % of those are of the lateral ligament complex caused by inversion sprains (Baumhauer, Alosa, Renstrom, Trevino, & Beynnon, 1995). The rate of recurrence of these sprains is very high and often leads to long term symptoms like pain and instability (Baumhauer et al., 1995; Verhagen, de Keizer, & van Dijk, 1995; Yeung et al., 1994). Chronic symptoms related to ankle instability cause individuals to be less active over their lifespan (Verhagen *et al.*, 1995). Additionally the recurrent episodes of ankle injury may predispose the joint to ankle osteoarthritis (Gross & Marti, 1999; Harrington, 1979; Hertel, 2002; Verhagen et al., 1995). The following review examines the various theories present in the literature regarding instability of the ankle and the underlying neurophysiological reasons behind it. Further it evaluates gait termination as a model to study these changes in ankle instability. More specifically, insight into the mechanisms of ankle injury may be gained by examining the anatomy of the talocrural joint, differing degrees of ankle instability, neuromechanical control mechanisms, and human gait.

The Talocrural Joint.

The ankle, or talocrural joint, is a synovial hinge joint that connects the distal ends of the tibia and fibula in the lower limb with the proximal end of the talus in the foot. The distal ends of the tibia and fibula (medial and lateral malleolus) articulate with each other to form a concave surface called the ‘Ankle Mortise’. The talus articulates with this ankle mortise to form the talocrural joint (Loudon, 1998; Lynch, 2002; Nyska, 2002; Loudon, 1998).

The lateral malleolus of the fibula and the medial malleolus of the tibia along with the inferior surface of the distal tibia articulate with three facets of the talus to form the bony articulation of the joint. The mediolateral stability of the ankle joint is maintained by the medial and the lateral malleolus of the tibia and the fibula. The medial malleolus extends to the body of the talus, however the lateral malleolus fails to extend as low as the medial malleolus Thus the anatomical structure of the ankle joint is effective in preventing the eversion or medial joint sprains making the body more susceptible to lateral ankle sprains or inversion sprains (Baumhauer et al., 1995; Hertel, 2002; Meir Nyska, 2002; Smith & Reischl, 1986).

Due to this unique anatomical alignment the ankle joint is heavily dependent on its lateral ligament complex and musculature to stabilize the joint and to prevent lateral ankle sprains (Baumhauer et al., 1995; Hertel, 2002; Meir Nyska, 2002). The three major structures that stabilize the ankle are the tibiofibular ligament, the medial ligament complex and lateral ligament complex.

Tibiofibular Ligament

The tibiofibular ligament is formed by the distal and proximal extremes of the interosseous membrane that traverses the entire lower leg, connecting the tibia and fibula. The oblique arrangement of the tibiofibular ligaments aid in the distribution of force placed upon the lower leg and stabilizes the ankle from rotational forces during activity (Meir Nyska, 2002).

Medial and Lateral Ligament Complexes

The ankle joint is bound medially by the strong deltoid ligament and laterally by three ligaments, namely, the anterior talofibular ligament, the posterior talofibular ligament, and the calcaneofibular ligament (See Figure2-1,Figure2-2). The deltoid ligament supports the medial side of the joint, and is attached at the medial malleolus of the tibia and connects in four places to the sustentaculum tali of the calcaneus, calcaneonavicular ligament, the navicular tuberosity, and to the medial surface of the talus. The Deltoid ligament is stronger than any of the ligaments considered individually (Hoppenfeld S., 1976) The lateral ligament complex is made up of the anterior and posterior talofibular ligaments and the calcaneofibular ligaments.

The anterior and posterior talofibular ligaments support the lateral side of the joint from the lateral malleolus of the fibula to the dorsal and ventral ends of the talus. The calcaneofibular ligament is attached at the lateral malleolus and to the lateral surface of the calcaneus (Delahunt et al., 2006; Kapandji, 1987; Nyska, 2002; Kapandji, 1987).

During an inversion trauma the anterior talofibular ligament is the first to undergo stress and hence is most susceptible to injury. The orientation of the fibers of the calcanofibular ligament is more vertical and hence it is injured only in grade 2 sprains (partial rupture of the stabilizing ligamnets) of the ankle joint. The posterior talofibular ligament is the strongest of the lateral ligaments and is least susceptible to injury (Hoppenfeld, 1976). Each of these ligamentous structures provides passive support for the ankle joint although they do not act alone. Active restraints (muscles) also add stability.

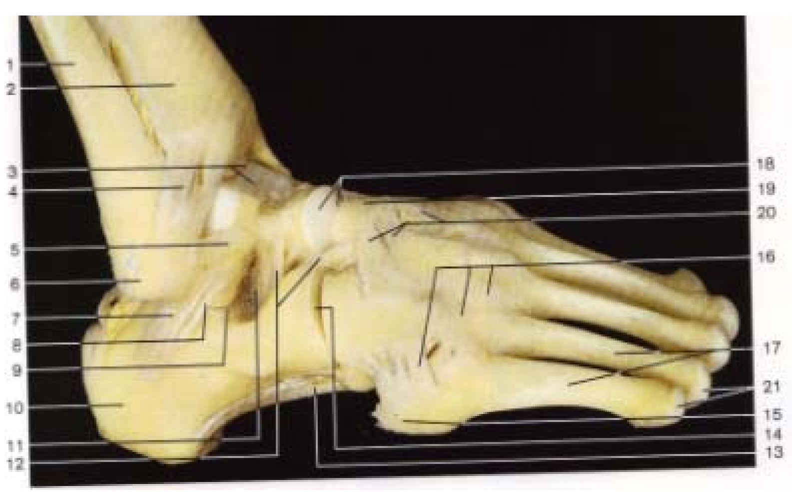
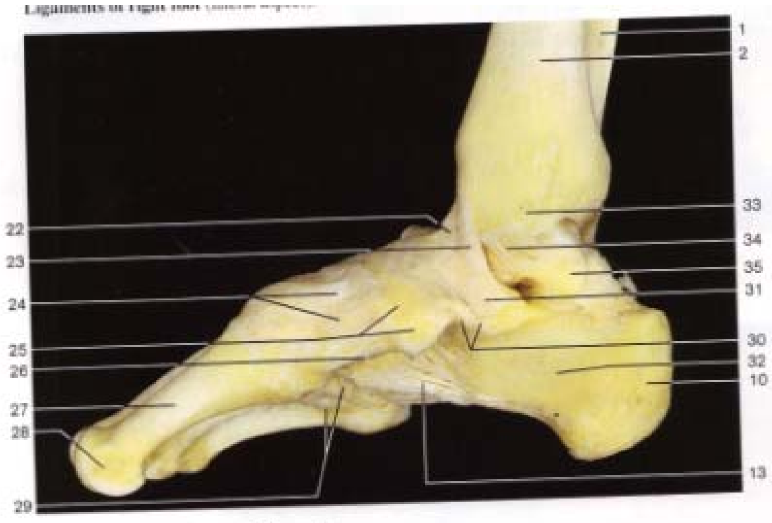


Figure 2-1 Lateral ligaments of the ankle joint



1 Fibula
2 Tibia
3 Trochlea of talus and **talocrural joint**
4 Anterior tibiofibular ligament
5 Anterior talofibular ligament
6 Lateral malleolus
7 Calcaneofibular ligament
8 Lateral talocalcaneal ligament
9 **Subtalar joint**
10 Tuber calcanei
11 Interosseous talocalcaneal ligament
12 Bifurcate ligament
13 Long plantar ligament
14 **Calcaneocuboid joint**
15 Tuberosity of fifth metatarsal bone
16 Dorsal tarsometatarsal ligaments
17 Metatarsal bones
19 Navicular bone
20 Dorsal cuneonavicular ligaments
21 Heads of metatarsal bones
22 Medial or deltoid ligament of ankle (tibionavicular part)
23 Medial or deltoid ligament of ankle (tibiocalcaneal part)
24 Dorsal cuneonavicular ligaments
25 Navicular bone
26 Plantar cuneonavicular ligament
27 First metatarsal bone
28 Head of first metatarsal bone
29 Plantar tarsometatarsal ligaments
30 Plantar calcaneonavicular ligament
31 Sustentaculum tali
32 Calcaneus
33 Medial malleolus
34 Medial or deltoid ligament of ankle (posterior part)
35 Talus

Figure 2-2 Medial ligaments of the ankle joint

Muscular Reinforcement of the Ankle Joint.

The ligamentous framework of the ankle joint is further reinforced by the surrounding muscles. These muscles are divided into four main compartments of the lower leg with each compartment serving a different function (Hoppenfeld S., 1976). The anterior compartment consists of the TA and toe extensors. These muscles serve as the primary and secondary dorsiflexors of the foot respectively. The lateral compartment holds the peroneus longus and peroneus brevis muscles, which are the main evertors of the foot and limit excessive inversion during activity. The tibialis posterior and toe flexors make up the deep posterior compartment.

These muscles are secondary invertors and plantar flexors of the foot respectively. The superficial posterior compartment is made up of the gastrocnemius and Sol muscles, the main plantar flexors of the foot and main stabilizers of ankle motion (Kapandji.I.A, 1987; Meir Nyska, 2002). While all the muscles mentioned are important, weakness of the peroneals and gastrocnemius/Sol complex would more significantly put a person at risk to injury, specifically to inversion ankle sprains (Kannus & Renstrom, 1991; Monaghan et al., 2006). An initial inversion sprain can lead to long term disability known as 'chronic ankle instability'.

Chronic Ankle Instability

As indicated, the ankle joint may not fully recover after the first episode of ankle injury. The recurrence rate of ankle sprains is as high as 70% after the initial trauma to the ankle joint (McKay et al., 2001). The symptoms that are commonly seen in the recurrent episodes are pain, swelling, tenderness and the feeling of 'giving way' or 'rolling over' of the ankle. The residual symptoms of the first episode of ankle sprain affect 55% to 72% of patients at 6 weeks to 18 months post injury (Hertel, 2002). The frequency of complications and breadth of longstanding symptoms after ankle sprain has led to the suggestion of a diagnosis of the ''sprained ankle syndrome''(Fallat *et al.*, 1998; Hertel, 2002) and to the conclusion that ''there is no such thing as a simple ankle sprain.''(Fallat *et al.*, 1998; Verhagen *et al.*, 1995). These recurrent episodes of ankle sprains after the initial trauma are termed as 'Chronic Ankle Instability' (CAI). The literature on ankle injury delineates two main theories that may lead to chronic ankle instability namely mechanical instability and functional instability.

Mechanical Instability

Mechanical instability can be caused due to the incongruence of anatomical structures at the ankle joint after the first episode of injury that may lead to abnormal joint mechanics. The contributing factors to mechanical instability are pathologic laxity, arthrokinematic restrictions and degenerative changes in the ankle joint and synovium (Hertel, 2002).

Pathologic laxity

Rasmussen and Tovborq-Jensen (1982) first described the method for graphic recording of rotatory movements in osteoligamentous ankle preparations. In this study they recorded the mobility patterns of the ankle with intact ligaments and after successively cutting the lateral collateral ligaments of the ankle in the anteroposterior direction. They concluded that talar tilt increases in the frontal plane and internal rotation of the talus increases in the horizontal plane beyond physiological limits as the degree of injury is gradually increased. Thus they suggested that there is an altered joint mechanics with injury to the anterior talofibular ligament (AFTL) and the calcanofibular ligamnet (CFL). Hollis and colleagues (1995) studied the effect of simulated ankle ligamentous injury on ankle-subtalar joint complex laxity by testing 36 intact ankles after gradually loading them with stress in the anteroposterior and mediolateral direction. The AFTL and the CFL were then gradually severed and the joints were reassessed. They found that there was a dramatic increase in talar tilt and rotation along with greater movement in the subtalar joint. Other researchers have also shown that the CFL is the second most common ligament injured along with the ATFL (Hollis, Blasier, & Flahiff, 1995). Recent studies have investigated the effect of serial sectioning of the ankle joint ligaments in cadavers. The authors have reported that the sectioning of the ATF ligament resulted in external rotation of the fibula. Furthermore injury to ATF and Deltoid ligaments resulted in greater laxity than the injury to the PTF(Beumer et al., 2006; Teramoto, Kura, Uchiyama, Suzuki, & Yamashita, 2008). Thus pathologic laxity in the talocrural and the distal tibiofibular joint is caused by the injury to the ligamentous complex. This can cause excessive accessory motions in the ankle joint leading to CAI.

Athrokinematic insufficiencies

Another factor contributing to the mechanical instability is deviation in the normal joint arthrokinematics at the talocrural and the subtalar joints(Hertel, Denegar, Monroe, & Stokes, 1999). One of the reasons for recurrent ankle sprains could be a deviance from the physiological position of the involved joint surfaces at the talocrural joint. It is suggested that there is a decrease in the posterior fibular glide observed after the initial episode of ankle injury (Hertel, 2002; Mulligan BR., 1995). Also, the lateral malleolus of the fibula may have a more posterior dislocation at the ankle joint. As seen earlier, injury to the lateral ligaments may allow excessive internal rotation of the talus (Rasmussen & Tovborg-Jensen, 1982). It is postulated that the decrease in posterior fibular glide along with the abnormal internal rotation of the talus causes the ATF to remain in a lax position predisposing the ankle to inversion sprains (Hertel, 2002). Thus changes in the normal joint arthrokinematics that result after the first episode of injury may cause mechanical instability and make the ankle joint more susceptible to injury .

Degenerative changes in the ankle joint and synovium

Repetitive ankle sprains have been known to cause degenerative changes and osteoporosis and osteophytes in the ankle joint. Gross and Marti (1999) observed that volleyball players who had unstable ankles had more osteophytes and synovial sclerosis as compared to the players without any history of ankle injury. Sugimoto et al. (1997) conducted a radiographic examination of 136 subjects with acute ankle sprains and 85 subjects with chronic ankle instability. They concluded that the varus tilt of the tibial plafond is more often seen in patients with chronic ligament instability of the ankle than in patients with acute ligament sprain (Sugimoto, Samoto, Takakura, & Tamai, 1997).

Functional Instability

Freeman et al. (1965) was the pioneer in providing the evidence that proprioceptive input is necessary for dynamic stability of the ankle. A ligament injury often results in injury to the joint mechanoreceptors. He proposed that the instability at the ankle is due to a lack of proprioceptive input from the joint mechanoreceptors. However, subsequent research revealed that insufficient input from the mechanoreceptors could not explain the instability of the ankle joint completely (Lephart, Pincivero, & Rozzi, 1998). The concept of insufficient neuromuscular control was proposed by Lephart et al. (1998) who proposed that neuromuscular control and joint stabilization are primarily mediated by the central nervous system and the spinal cord. These researchers also suggested that proprioception and the accompanying neuromuscular feedback mechanisms provide an important component for the establishing and maintaining functional joint stability. Therefore, functional stability of the ankle joint is dependent on neuromuscular mechanisms as well. Further, recent research also suggests that mechanical and functional instabilities do not totally correlate and that functional instability may exist with or without mechanical instability(Tropp, 2002). Functional ankle instability is reported to impair the performance and quality of life in sporting and non sporting population (Fallat et al., 1998; Gerber et al., 1998; Gross et al., 1999; Lynch, 2002; Osborne et al., 2003; Tropp, 2002). Hence further research on functional ankle instability is justified. The importance of sensory, proprioceptive and postural control systems as contributors towards the functional instability is described in the subsequent section.

Impairment of sensation and proprioception and postural control: Ankle injuries are known to affect afferent sensation in the lower limb (Lephart et al., 1998). Bullock and Saxton (1995) also found significant sensory deficits in the affected lower limbs of subjects who suffered ankle injuries and subsequent instability. These decreases in sensation are often associated with direct injury to the peroneal nerve which may occur during ankle sprains. Postural stability and proprioception are also known to be affected by ankle instability (Docherty et al., 2006; Ergen et al., 2008; Evans et al., 2004; Ross et al., 2004a; Wikstrom et al., 2007), but are normally evaluated in static situations. Wikstrom et al. (2006) suggested that dynamic ankle instability can be a significant factor hindering an athlete's performance and that aggressive rehabilitation or surgery may not help the ankle recover completely. These authors also suggest that the dynamic postural stability index (DPSI) can be used as an effective tool to detect the difference between the stable and functionally unstable ankle. The measurement of DPSI involves the use of a 'Jump landing model' (the participant jumps to a height 50% of his maximum vertical leap and then lands on one limb on the force plate) and 'Single leg stance model' has been used until now as a primary model to detect deficits in dynamic postural stability (Ross et al., 2004a; Wikstrom et al., 2007; Docherty et al., 2006; Ergen et al., 2008; Evans et al., 2004; Ross et al., 2004a; Wikstrom et al., 2007). The dynamic postural control can be effectively maintained by the body with the help of different feedback and feedforward mechanisms inherent to it. Hence the study of possible alterations in these mechanisms could help us understand the instability of the ankle with a better perspective.

Feedback and Feedforward Mechanisms for Postural Control

Johansson and Magnusson (1991) suggested that the stimulation of a corrective response within the corresponding system after sensory detection is often considered as feedback control. In contrast, feedforward control mechanisms have been described as anticipatory actions occurring before the sensory detection of a homeostatic disruption (Johansson et al., 1991; Riemann et al., 2002).

The ankle joint is prone to sprains and injuries when there is a sudden shift or deceleration of the body's COM (Johansson et al., 1991; Wikstrom, Tillman, Chmielewski, & Borsa, 2006).

Injury to the anatomical structures around the ankle joint (muscles, ligaments and the mechanoreceptors) is a logical consequence of an injury. Freeman et al. (1965) first put forth the pathoetiologic model for ankle instability based on the impairment of the feedback control mechanism. In this model they proposed that instability that resulted after the initial ankle injury was a result of loss of proprioceptive input from the injured ligaments. Recent studies have demonstrated that there is a decrease in proprioception and joint position sense after the first episode of injury to the ankle joint (Docherty et al., 2006; Freeman et al., 1965; Monaghan et al., 2006; Ross et al., 2004a; Wikstrom et al., 2007). These deficits can demonstrate impairments in the feedback neuromuscular controls. Empirical data also indicate that there is alteration in the neuromuscular firing patterns in unstable ankles during normal gait (Delahunt et al., 2006; Jazayeri Shooshtari, Didehdar, & Moghtaderi Esfahani, 2007; Monaghan et al., 2006; Ross et al., 2004a; Tropp, 1986). This could help explain the theory that was put forth by Freeman and support the importance of feedback neuromuscular control mechanisms and faults in feedback neuromuscular control mechanism. However, loss of proprioception can only partially explain instability at the ankle joint. Recent research suggests that the feedforward neuromuscular control is also responsible for maintaining the stability at the ankle (Delahunt et al., 2006). Subsequent to the initial trauma, the body may develop compensatory movement strategies to avoid short term effects like pain. It has been suggested that the inappropriate positioning of the ankle joint before ground contact may have important implications on the stability of the ankle joint (Monaghan et al., 2006). These compensatory movement patterns are thought to be the causal factors for the development of ankle instability. Nyska (2003) demonstrated changes in the pattern of force transfer between the foot and the floor associated with chronically sprained ankles by measuring the peak forces and their timing under several regions of the feet during level walking. The authors found that patients with ankle instability showed slowing down of weight transfer from heel strike to toe off. A reduction in foot impact and lateral shift of body weight was also noted. This was in accordance with the hypothesis that there is central reorganization of movement patterns after ankle injury causing a change in feed forward mechanisms. Change in kinetics and kinematics of the ankle joint in patients with ankle instability during overground walking at self selected velocities was demonstrated by Monaghan et al. (2006). In this study, subjects with chronic AI were found to have a more inverted position both before and immediately after heel strike compared with the control group. The subjects with AI showed a concentric evertor moment whereas the controls demonstrated an eccentric invertor moment at heel strike. The limitation of this study was that no electromyography(EMG) data were collected. Deficits in dynamic stability of the ankle joint can also be predictors of changes in the feedforward neuromuscular control. Different models are used in literature to measure dynamic stability at the ankle joint. Ross and Guskiewicz (2004a) used single leg stance and jump landing models to examine the dynamic stability of the ankle joint. They demonstrated that dynamic postural stability is affected in subjects having a chronically unstable ankle. Wikstrom and colleagues (2007) used the dynamic postural stability index (DPSI) to measure the various components of dynamic stability in a single leg hop-stabilization model. They found that there was a significant difference in the vertical as well as the anteroposterior component of dynamic stability between the control group and the subjects having a history of ankle instability. Thus, feed forward mechanisms of postural control may also contribute to AI.

Dynamic Postural Stability Index

Wikstrom et al. (2005) proposed a new force plate technology measure for the measurement of dynamic postural stability. Dynamic postural stability can be defined as the ability of the person to control balance while transitioning from a dynamic to a static state (Goldie, Bach, & Evans, 1989). Time to Stabilization (TTS) has been used as an objective measure for dynamic postural control in patients with functional instability of the ankle (McKinley & Pedotti, 1992; Ross et al., 2004b; Ross, Guskiewicz, & Yu, 2005). These authors described TTS as the time required to minimize the ground reaction force (GRF) after landing on the force plate to within a range of baseline (static) GRF. When a subject landed on a force platform after a jump, TTS was measured from the forces created in 3 directions (vertical, medial-lateral and anterior-posterior). This gave researchers 3 separate measures for dynamic postural stability. The necessity of this new measure of dynamic postural control was justified, as the authors argued that TTS has several inherent flaws. One of the major flaws as described by Wikstrom and colleagues was that this measure did not provide a common thread among the forces in 3 directions. Thus the global picture of dynamic postural stability remained unclear. Dynamic postural stability index (DPSI) was proposed by the authors as a measure that accounts for this shortcoming. DPSI was calculated by measuring stability indices in three principal directions: anterior-posterior (APSI), medial-lateral (MLSI) and vertical (VSI). Stability indices were calculated by measuring the mean square deviations assessing fluctuations around a zero point. Thus the MLSI and APSI assess the fluctuations from 0 along the frontal and sagittal axes of the force plate, respectively. The VSI assesses the fluctuation from the subject's body weight to standardize the vertical GRF along the vertical axis of the force plate. The authors proposed DPSI as a composite score of APSI, MLSI and VSI and this measure was shown to be sensitive for changes in all directions. The authors proposed the following formula for calculation of DPSI.

Formula for DPSI

MLSI = $\sqrt{} [\sum (0- x)^2$/number of data points]

APSI = $\sqrt{} [\sum (0- y)^2$ /number of data points]

$VSI = \sqrt{[\sum (\text{body weight} - z)^2 / \text{number of data points}]}$

$DPSI = \sqrt{[\sum (0- x)^2 + \sum(0- y)^2 + \sum(\text{body weight} - z)^2/\text{number of data points}]}$

Wikstrom et al. (2006) used DPSI to estimate the deficits in postural stability in subjects with self reported ankle instability. A jump protocol was used in which the participants were instructed to stand 70 cm away from the center of force plate. The participants were then instructed to jump to a height 50% of their maximum vertical leap before landing on the force plate. The authors found that there was a deficit of dynamic postural control in participants having functional instability of the ankle joint. This was demonstrated by significantly higher APSI, VSI and DPSI scores. Thus, DPSI has been used successfully as a measure to detect deficits in dynamic postural stability in subject with ankle instability. The application of this index to human gait could help unveil the deficits in the dynamic postural stability in patients with AI.

Human Gait

Introduction

Normal human locomotion is known to be a complex interaction of various mechanical and neural strategies used by the body to move from one point to a desired second point. It is a rhythmic activity in which a human body is propelled with the use of two legs alternately while maintaining the upright stability of the trunk. A typical gait cycle can be divided into three stages: gait initiation, rhythmic walking, and gait termination.

Gait Initiation

The biomechanics and the muscle activity during upright standing are well studied (Breniere & Do, 1986; Breniere & Do, 1991; Breniere & Dietrich, 1992; Crenna & Frigo, 1991). During quiet stance there is a slow, constant tonic contraction of the Sol muscle. This helps to maintain the fine balance between the locations of the COM and the COP which remain within the base of support in quiet stance. When observed in the sagittal plane, the COM is located anterior to the axis of the ankle joint. Gravity acts on the COM pushing it further away from the ankle joint. This creates a dorsiflexion torque at the ankle joint. The tonic muscle contractions of the Sol muscles help to counteract this torque by producing a plantarflexion torque of equal magnitude. The COP is moved forward beyond the location of the COM by the plantar flexion torque which pushes the COM back thus maintaining equilibrium. During gait initiation there is a slight shift of the body weight on the side of the stance foot. Pressure is created in backward and lateral direction by the foot that is about to get into the swing phase by a short burst of TA(Crenna et al., 1991). In accordance to the Newton's second law, the COM moves forward and to the side of the stance leg. This allows unloading of the swing leg making the initiation of the gait possible. Thus gait initiation involves the inhibition of the tonic postural activity of the Sol muscle along with a burst of activity in the TA which in turn causes the shift in the weight on the side of the stance limb. The abductor muscles on the side of the stance limb help in this weight shift (Brunt *et al.*, 1991)

Rhythmic Walking

Prior to taking the first step the COM is always kept within the base of support. Interestingly movement of COM anterior to the base of support marks the beginning of walking. It takes 2-3 steps to initiate rhythmic pattern of gait. The body uses energy economically by using the six determinants of gait that help in preventing excessive rise and fall of the center of mass. Once steady velocity of gait is achieved the body there is interaction of potential energy and kinetic energy based on the inverted pendulum model. The trunk is at the lowest point and has the highest kinetic energy during the phase of double limb support .During swing phase the trunk is lifted up by the supporting leg converting the kinetic energy into potential energy. During the later half of the swing phase the trunk is lowered and the COM moves in front of the supporting leg thus converting the kinetic energy to potential energy and this helps the body to slow down. This sinusoidal path of the center of mass continues till the point at which gait is terminated.

Gait Termination

The biomechanics of gait termination is less researched as compared to gait initiation and rhythmic walking. To achieve a safe and stable gait termination the forward acceleration of the body needs to be controlled to achieve a stable upright posture (Hase et al., 1998). A reduction in the forward push off force during the last step and an increase in the posterior braking force are suggested to be the possible two reasons for gait termination (Jaeger & Vanitchatchavan, 1992). During termination, there is a large burst of Sol muscle activity and a sharp reduction in TA recruitment just prior to heel strike in the leading limb. There is also a recruitment of Quadriceps and GM in the leading limb which helps in knee extension (Bishop, Brunt, Pathare, & Patel, 2002). The trail limb shows recruitment of TA, Hamstrings (mainly Biceps Femoris) and GM muscles which causes the body to move more posterior (Bishop et al., 2002; Crenna et al., 1991). This helps the COM move in a posterior direction as compared to the COP which aids the body to decelerate (Stein & Hase, 1999). The study of quantification of gait termination and the factors affecting it may help us understand the usefulness of this model for studying the deficits of postural control in AI.

Quantifying Gait Termination

Sparrow et al. (2005) discussed the need for an operational definition of gait termination The biomechanical definition of “stopping” is based on the displacement characteristics of the feet. Stopping in a gait cycle could be described as a point in time where the anterior displacement of the feet is absent and the forward progression has ceased (Sparrow & Tirosh, 2005). However a small difference exists between ‘planned’ and ‘unplanned’ patterns of gait termination. Planned gait termination is a condition in which a person has prior knowledge of the time and location where he or she needs to stop. However in unplanned gait termination the person is totally unaware of the time and location where he or she needs to stop until receiving the cue. One important difference observed between the planned and unplanned stopping is the speed of feet placement. Jaeger and Vanitchatchavan (1992) observed that the foot placement in planned gait termination takes 0.5 sec longer than that in unplanned gait termination. This additional time is required for placing both feet together in planned gait termination. Thus while quantifying gait termination by using temporal measures, the time difference between the presentation of the stopping stimulus and the final heel contact is important. Bishop et al. (2003) coincided the delivery of the stopping stimulus with significant events in gait cycle and found that there was a significant difference in the magnitude of breaking impulse which directly depended on the time point in the gait cycle at which the stopping signal was delivered. The net braking impulse that resulted when a stopping signal was given at heel strike was greater than the braking impulse when the signal was given at peak loading which was greater than that when the signal was given at midstance. They also defined a short delay at heel strike, a further delay at peak loading (braking) and the latest signal at mid-stance (the vertical force minimum). Stimulus delay and breaking force impulses can, therefore, be manipulated relative to key events of the gait cycle.

Factors Affecting Gait Termination

The termination of gait and the characteristic patterns of ground reaction forces and moments are governed by the following factors: stimulus delay, stimulus probability and velocity of gait.

Stimulus Delay

Hase and Stein (1998) described the relationship between the time at which the stopping stimulus was delivered during the gait cycle and the resulting foot placement pattern. They stated there are two main patterns of gait termination. The first pattern is when the subject stops with the right leg forward (lead limb forward) and the second with the left (trail) limb forward.

The pattern of stopping is governed by the phase of gait cycle during which the stopping stimulus was delivered. When the stimulus was applied between 35 to 70% of the gait cycle (late stance phase to mid swing phase of the right limb) the rapid stopping ended with right limb forward. Whereas, when the stimulus was delivered within the first 20% of the gait cycle, the stopping occurred with the left limb forward. When the stimulus was delivered between 20 to 35% of the gait cycle or between 70 to 85% of the gait cycle, the subject tended to stop with an extra transitional step. Very rarely did the subjects stand with feet together, except after having a particularly short step in the transition periods (Hase et al., 1998).

Stimulus Probability

Sparrow et al. (2005) proposed that the probability of stimulus delivery might have a significant effect on different variable like the stopping time, ground reaction force of the stance and the swing limb etc during gait termination. In their review of the literature, these authors stated that a majority of the studies examining different aspects of gait termination used a high probability of stimulus delivery. The authors debated that higher probability of stimulus delivery could involve a pre-planned response. Thus the responses could be faster as a result of motor planning. However Tirosh et al. (2004) examined the effect of probability of stimulus delivery on gait termination. In the experiment that was conducted they compared the effect of high stimulus probability (80%) and low stimulus probability (10%) on variables such as stopping time and stopping distance. They concluded that stimulus probability had only a weak effect on the stopping variables and stated that “Stimulus probability was found only to affect gait termination weakly and it appeared that within the laboratory, even in the low probability condition, stopping remained relatively well anticipated.”

Velocity of Gait

The maintenance of the COM within the base of support is a pre requisite for maintaining balance during gait termination. Tirosh and Sparrow (2005) developed a model to predict a safe velocity for gait. The key parameters of this model were the horizontal velocity of the center of mass normalized to body height and its position in foot lengths relative to the base of support. “The fundamentals of the model are that either a forward fall or an additional step was initiated if states exceeded the region’s upper boundary and a backward fall or step was required if the lower boundary were exceeded”. See Figure 2-3

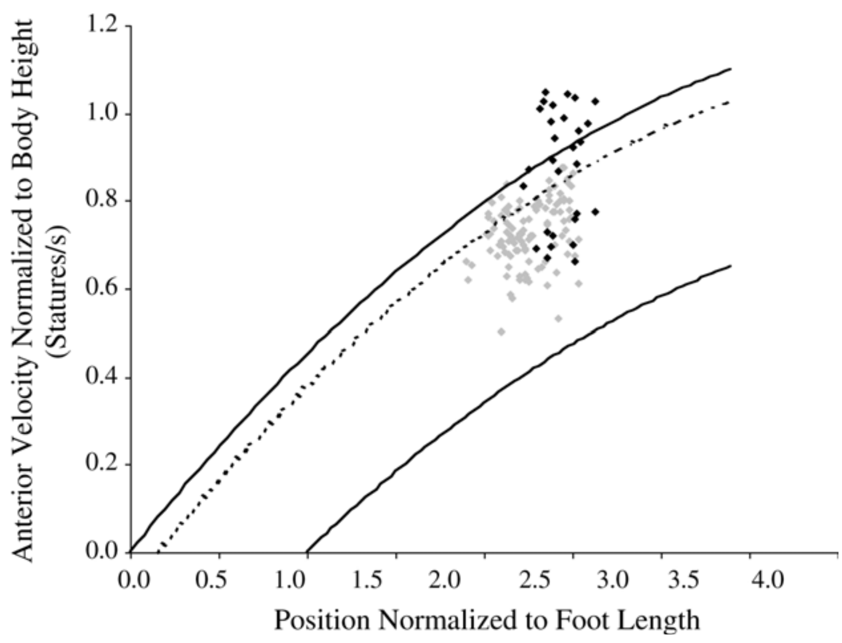


Figure 2-3 Left hip anterior velocity–position plot for young (dark symbols)and older (light symbols) adults during two-step stopping. The stability region is drawn with the upper boundary for normal conditions (upper solidline) and when recalculated to accommodate a 59% reduction in strength (middle dashed line) to simulate the capabilities of older adults, as discussedin the final section. A fall forward was anticipated if the left hiphorizontal velocity–position exceeds the upper boundary of the stabilityregion. (Reproduced from Tirosh and Sparrow [19] –

Thus, if an individual has a gait velocity above 1.1m/s at 2.4 foot lengths anterior to foot contact, he or she would terminate gait with an additional step to maintain balance. Bishop et al. (2002) examined the relationship between the cadence rate and the pattern of foot placement in gait termination. These authors measured the lower limb muscle recruitment at different cadences (50%, 100% and 150% of normal cadence rate) and found that as the cadence increased the braking forces that were produced by the lead limb during gait termination also increased. Also, there was an increased reliance on the lead limb for gait termination (Bishop et al., 2002; Sparrow et al., 2005).

Gait Termination as a Model for Elucidation of the Feedforward and Feedback Mechanisms in Ankle Injury.

The current literature on ankle instability suggests that either feedforward, or feedback control or both may be altered in individuals who have instability at the ankle joint. Unfortunately, there are no studies to date that have examined the differences in feedforward and feedback control in the same set of subjects with AI. Researchers have studied mechanisms of AI utilizing jump landing models (Ross et al., 2004b; Wikstrom, Tillman, & Borsa, 2005a). However, this experimental model may have a limited use in revealing the mechanisms behind AI. The mechanics of landing posses great variability due to multiple degrees of freedom, i.e. control of the hip, knee, ankle, and foot, which can be used to perform the task successfully. *Based on these observations, a need of an experimental model that would reveal the underlying mechanisms that lead to AI is justified. Understanding these mechanisms will be useful to device different rehabilitative interventions to prevent AI.* The present study is designed to determine if neuromuscular control alterations exist in AI subjects. To achieve this, I propose to use a gait termination model. Gait termination involves a rapid deceleration of the forward momentum of the body during steady gait. A safe termination of gait requires a complex interaction of the neuromuscular system.(Hase et al., 1998) Furthermore, it possesses a known and repeatable set of neuromuscular responses (Bishop et al., 2006; Bishop et al., 2003; O'Kane et al., 2003) and gait termination experiments can be constructed to challenge both feedforward and feedback neuromuscular control (Bishop et al., 2006). Elucidating the neuromuscular control mechanism is of fundamental interest, because it will directly impact the direction of future investigations, and lead to effective interventions for AI.

CHAPTER 3
METHODS

This experiment was a single session, single subject mixed model design. We analyzed the activation patterns of the muscles of lower extremity along with the propulsive force, propulsive time (time after heel strike during which maximum propulsion occurs), braking time and the braking force (ground reaction force) that was generated during gait termination. These data were compared between limb (involved, uninvolved) between the groups (healthy, AI), within task (planned and unplanned gait termination) and across different sub-phases of stance using EMG. Twenty young adults in the age group of 18 to 30 yrs and with a history of recurrent ankle sprains were recruited to participate in the experimental group. Twenty healthy young adults were recruited in the control group. This sample represented both genders.

Recruitment

Participants were recruited for this study from the campus of the University of Florida. The participants were informed about this study during various lectures and laboratory classes. The interested participants were further screened for the inclusion and exclusion criteria as discussed below. The subjects satisfying the age, inclusion and exclusion criteria were contacted and the interested participants were provided with additional details regarding the study. Initially, the participants were screened using a ankle injury questionnaire and Ankle Injury Questionnaire tool (Ross & Guskiewicz, 2004c; Ross, Guskiewicz, Gross, & Yu, 2008; Rozzi, Lephart, Sterner, & Kuligowski, 1999). This questionnaire contains 6 items. The responses to each of the items were helpful in gaining insight into the injury status. They indicated which ankle was injured, the nature of treatment or immobilization, the duration of immobilization, the rate of recurrence of ankle sprains, whether they have returned to their initial level of activity and if they attribute the symptoms of instability to the first ankle injury. The eligible participants also signed the informed consent forms. Each testing session took about two hours including the initial set up.

Inclusion Criteria

- At least one grade 2 injury (partial loss of integrity of the lateral ligament complex of the ankle) which required immobilization in the past.
- History of recurrent ankle sprains (at least 1 in the past 6 months) which the patient attributes to the previous ankle injury.
- The responses of the participant to specific parts of Ankle Injury Questionnaire.(See appendix A)

Exclusion Criteria

- Fracture or any other acute or chronic orthopedic complication for which weight bearing is contraindicated.
- Systemic illnesses and conditions which could interfere with balance and gait (cataracts, glaucoma etc causing problems with clarity of vision, head injury, vestibular involvement, hypertension, diabetes).
- Subjects with a history of bilateral ankle instability were excluded from the study.

Instrumentation

Force Platform

Gait termination trials were performed along an 8-m walkway containing force plates. Ground reaction forces were collected using two force plates (Type 4060-10 Bertec Corporation, Columbus, Ohio). The force plates (0.4m x 0.6m) were mounted flush with the surface of the walkway in the laboratory so that full foot contact occurred on both force plates during normal walking. The force plates were oriented so that the laboratory coordinate system coincides with the right posterior corner of the force plates, with the X-axis aligned in the direction of forward progression, Y-axis aligned in the mediolateral direction and Z-axis in the vertical direction (Figure 3-1). Forces and moments along the 3 principle axes were sampled at 1200Hz. The force plate amplifiers were turned on at least one hour before the actual data collection.

The Motion Capture System

The motion analysis for this study was performed with a VICON Nexus 1.0 motion capture system (VICON motion systems, California). Gait trials were digitally recorded with 8 MX 20+ cameras mounted around the force platforms along the perimeter of the lab (see Figure 3-1). The orientation of the cameras in the room was such that at any time instance within the calibrated volume each reflective marker was in the field of view of at least 2 cameras. The data were captured at 120Hz. Prior to the data collection the 'capture volume' was calibrated using a T-shaped calibration wand (with 5 retroreflective markers). The same wand was used to set the volume origin of the capture volume which oriented the cameras with the three axes. Within the calibrated volume, the motion capture system is accurate to < 2mm. The video cameras and force platform recordings were synchronized using the VICON Nexus motion capture system.

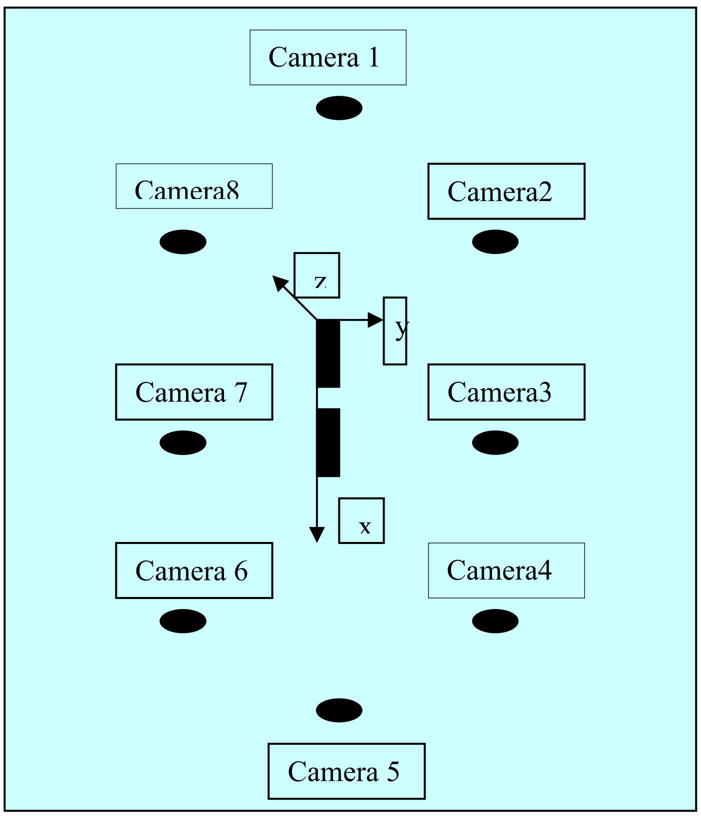


Figure 3-1. Placement of the cameras and the force plates

Electromyography (EMG)

A KONIGSBERG T-42AL-8T (Konigsberg Instruments Inc, California) telemetric electromyography unit was used to record muscle activity for all trials. Bipolar 1-mm x 10-mm Ag/AgCl surface electrodes with an inter detection surface distance of 1.5-cm was used to detect the muscle activity. The KONIGSBERG T-42AL-8T telemetric EMG system uses a transmitter which is attached to the subject during the walking trials. This transmitter broadcasts the muscle activity of the subject to the VICON Nexus 1.0 motion capture system. The signals were band pass filtered (20-4Khz) and full wave rectified. The processed EMG and amplified force plate signals was sampled at a rate of 1200Hz. A sliding average with a window frame of 25 milliseconds was calculated to smooth the data. Stance phase of gait for the trail limb was divided into four sub-phases [1-heel strike (HS) to peak loading (P1), 2-peak loading (P1) to mid-stance (MS), 3-mid-stance (MS) to second peak load (P2), and 4- second peak load (P2) to toe-off (TO)]. See Figure 3-2. The stance phase of gait for the lead limb was divided into two sub phases (heel strike (HS) to peak loading (P1), peak loading of the lead limb (P1) to mid stance of the stance limb (MS)) based on the vertical ground reaction force (Figure 3-3) (Bishop et al., 2006; Bishop et al., 2003).

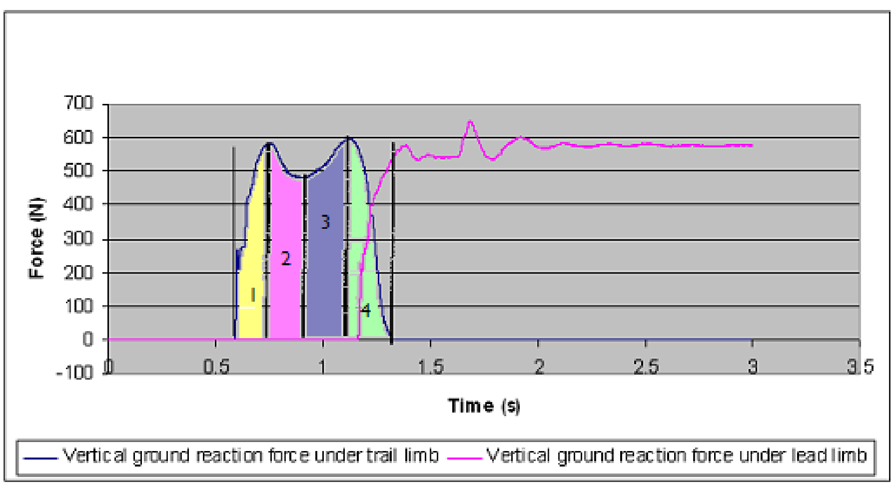


Figure 3-2. Force curve with sub phases of the trail limb.

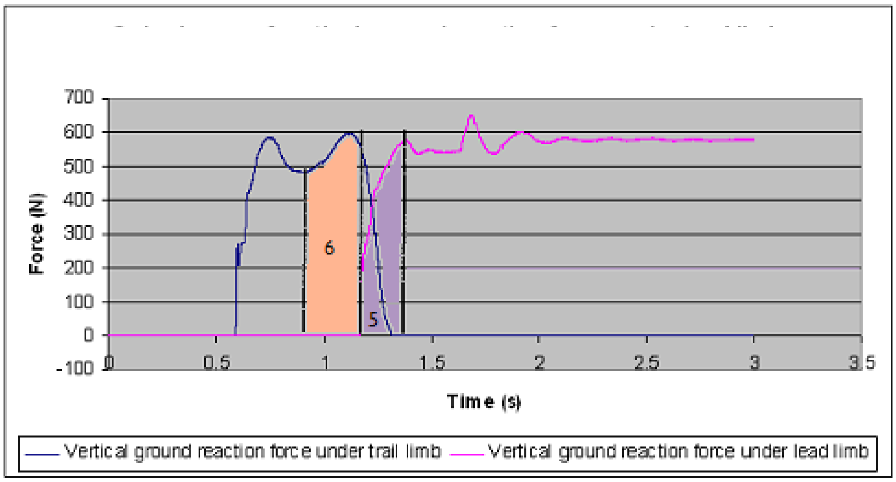
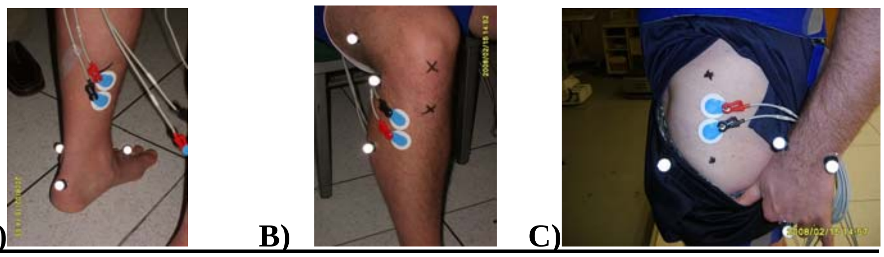


Figure 3-3. Force curve with sub phases of the of lead limb

Further, average amplitude of EMG for each of the six sub phases of gait was calculated. This amplitude was normalized with the average amplitude of the corresponding phase for the 'normal walking trials'

Testing and Subject Preparation

Data were collected in a single testing session (an approximate duration of 2hrs) during which participants wore dark-colored tight-fitting shirts, shorts and walked bare foot along the 8m long walk way. Thrity-six passive retroreflective markers were placed over anatomical landmarks according to the Plug in Gait marker system (Figure 3-5). The 36 markers were used to construct a simple 15-segment model using the VICON motion capture system. The spatial orientation of these markers was used by the VICON motion capture system to estimate the walking velocity during each trial. EMG data was collected (Konigsburg instruments T-42AL-ST) from the TA, Sol, and GM as representative muscle groups in both lower extremities. Prior to the electrode application the skin was shaved and cleaned with alcohol to reduce skin impedance which allowed for a clearer signal (Brask, Lueke, & Soderberg, 1984; Hung & Gross, 1999; Isear, Jr., Erickson, & Worrell, 1997; Ninos, Irrgang, Burdett, & Weiss, 1997). The muscles were palpated for their exact anatomical location and two bipolar surface electrodes with an inter detection surface distance of 1.5cm placed on the muscle bellies bilaterally. The exact anatomical location of the electrodes is described in Table 3-1 and shown in Figure 3-4 (Brask *et al.*, 1984; Perotto A, 1994).



A) **B)** **C)**

Figure 3-4**.** Electrode placement a) Soleus b) Tibialis Anetrior c) Gluteus Medius

Table-3-1. Electrode placement:

Muscle	Electrode location
Soleus	Placed one inch lateral and 3 inch inferior to the lower edge of the gastrocnemius belly on either side
Gluteus Medius	Spaced approximately at the midpoint of the perpendicular line joining the iliac crest and the greater trochanter.
Tibialis Anterior	Placed 1.5 inches lateral and 1 inch inferior to the tibial tuberosity .

Static Trial

Once the participant was fitted with reflective markers and EMG electrodes, one 5 second static trial was collected. In this trial the participant stood as still as possible in the capture volume of the cameras in order to reconstruct the live model of the participant in the VICON software. All 36 markers had to be visible to the cameras and labeled in the system.

Procedure for Testing Unplanned Gait Termination

Each participant was instructed to walk along the 8 m long walk way at normal gait velocity and was instructed to stop only when cued by a buzzer. The buzzer was triggered on random trials just before the heel strike of the lead limb so that the participant was able to stop within one foot strike after the trigger. Each limb was the lead limb in 5 trials. Unplanned gait termination velocity was also compared to self-selected gait velocity to ensure that gait speed was kept constant. Trials were repeated until we recorded 10 valid trials**.** The participant also performed normal walking trials in which no signal was given to prevent counting or anticipation of gait termination. These trials were interspersed within the sequence of unplanned stopping trials and constituted at least 60% of total number of trials recorded. During data collection the unplanned and normal walking trials were recorded before the planned gait termination trials. This eliminated the learning effect.

Procedure for Testing Planned Gait Termination

In this task the participant was instructed to perform a planned gait termination. The participant performed 10 trials. Each limb served as the lead limb in 5 trials of planned gait termination. In each trial, the participant walked down the 8 meter long walkway and stopped on the second force plate. The participants were instructed to walk with his or her usual gait velocity.

Procedure for calculation of the DPSI

The ground reaction force data were used to calculate the Dynamic Postural Stabiltiy Index (DPSI) for a duration of 1 sec post heel strike in the lead limb. DPSI was calculated for both involved and uninvolved limbs, AI and control groups and during unplanned and planned gait termination trials.

Figure 3-5. Placement of the retro reflective markers and the wireless EMG transmitter

Statistics

Statistics for timing of stimulus delivery and gait velocity: Comparison of the kinetic and kinematics and muscle recruitment between the control and the AI groups during gait termination was reasonable only if the gait velocity remains same for both the groups and the relative timing of stimulus delivery is same for both groups (AI and control), both limbs (involved/dominant and uninvolved/non dominant) and for each condition (planned, unplanned and catch) (Sparrow et al., 2005). A three way factorial ANOVA (2:group x 2:limb x 3:condition) was computed for velocity. A two way factorial ANOVA (2:group x 2:limb) was performed for the timing of stimulus delivery. Bonferonni post hoc analysis was performed when necessary. A traditional level of significance (α =0.05) was used..

Statistics for Kinetic and Kinematics

Force: A 3-way MANOVA (2:group x 2:limb x 3:condition) with repeated measures on the last factor was computed for propulsive force, propulsive time, braking force and braking time. Bonferonni post hoc analysis was performed when necessary. A traditional level of significance (α =0.05) was used. See Appendix B for a detailed list of dependent and independent variables.

Statistics for Stability

DPSI was calculated for each of the independent variables. The effect of the independent variables on the subcomponents of DPSI i.e. APSI (Antero posterior stability index), VSI (Vertical stability index) and MLSI (Mediolateral stability index) was also studied. The 'involved' or 'uninvolved' status of the limb is important in the AI group. Hence it was included as one of the independent variables. The empirical results from the prior studies have shown that for the control group the DPSI is independent of the dominance of the limb (Wikstrom et al., 2005a; Wikstrom, Tillman, Smith, & Borsa, 2005b). However for the control group the dominant and non dominant status of the limb were included as an independent variable to maintain statistical balance. A 3-way MANOVA (2:group x 2:limb x 2:condition) was computed for APSI,MLSI, VSI and DPSI . Bonferonni post hoc analyses were performed when necessary. A traditional level of significance (α =0.05) was used. See Appendix 2 for a detailed list of dependent and independent variables.

Statistics for EMG

MANOVA (2:group x 2:limb x 6:phase 2:condition) with repeated measures on the last factor were computed for 3 muscles on each limb across 6 phases. Bonferonni post hoc analysis was performed when necessary. A traditional level of significance (α =0.05) was used. See Appendix B for a detailed list of dependent and independent variables.

CHAPTER4
RESULTS

Demographics

The sample size consisted of 40 participants. Twenty participants were recruited in the AI group (age 20.2±1.2years, height 169.8±9.7cm and mass 74.2 ±20.2kg) and 20 participants were recruited in the control group (age 20.4± 1.6years, height 164.3 ±7.9cm and mass 64.2 ±10.6kg). Amongst the participants recruited in the AI group, six participants had their left limb injured whereas 14 had their right limb injured. All the participants in the control group identified the right limb as their dominant limb. The data collected from all 20 participants in each group were used to compute the force and stability statistics. However, the EMG statistics were computed using the data from 20 participants form the control group and 18 participants from the experimental group. These data were considered outliers created by technical problems. The probable reason for this was that the data were collected when the battery of the EMG telemetric unit was almost exhausted. Independent sample t-tests were performed to compare anthropometric data (age, height, weight) between both groups. No significant statistical differences were observed between groups ($p>0.05$).

Velocity

A 3-way ANOVA (2:group x 3:condition x 2:limb) was computed to compare the velocity across each of the independent variables. No significant group [$F (1,199) = 1.12$, $p = 0.29$] or limb [$F (1,199) = 0.03$, $p = 0.84$] main effects were noted. However, a significant condition [$F (2,199) = 4.22$, $p = 0.01$] main effect was observed. Bonforroni post hocs revealed that the gait velocity during unplanned gait termination was significantly higher than the gait velocity during planned gait termination ($p < 0.01$). No significant two-way interactions were revealed for condition x group [$F (1,199) = 0.13$, $p = 0.84$], condition x limb [$F (1,199) = 0.08$, $p = 0.77$] or group x limb [F (1,199) = 0.48, p = 0.46]. Similarly, no significant three-way interaction was detected for condition x group x limb [F (1,199) = 0.48, p = 0.48]. Gait velocity data appear in Table 4-1.

Table 4-1 Velocity (m/s) (Mean ± SD). * Indicates a statistically significant difference between unplanned and planned trial (p≤0.05).

Variable	Level	M ±SD
1) Condition	a)catch	1.23±0.01
	b)planned*	1.19±0.11
	c)unplanned*	1.24±0.48
2) Group	a)healthy	1.21±0.01
	b)AI	1.23±0.11
3) Limb	a)right	1.22±0.11
	b)left	1.22±0.11

Relative Time of Stimulus Delivery

A two-way ANOVA (2:group x 2:limb) was computed to determine the relative difference in time between stimulus delivery and heel strike of the trail limb. There was no significant difference between groups [F (1,38) = 1.9, p = 0.17] or limbs [F (1,38) = 0.061, p = 0.80]. Also no group x limb [F (1,38) = 0.05, p = 0.14] interaction was detected. Time of stimulus delivery data appear in Table 4-2.

Table 4-2 Relative timing of stimulus delivery(s) (Mean ± SD)

Variable	level	M ±SD
1)Group	a)healthy	0.07±0.07
	b)AI	0.05±0.08
2) Limb	a)right	0.06±0.08
	b)left	0.05±0.09

Force

A 3-way MANOVA (2:group x 2:limb x 3:condition) with repeated measures on the last factor was computed for propulsive force, propulsive time, braking force and braking time.

Significant condition [F (8,374) = 96.28, $p < 0.001$] and group [F (8,187) = 6.32, $p < 0.001$] main effects were observed. However no significant main effect for limb [F (4,187) = 0.075, p = 0.99] was noted. There were no significant two way interactions: condition x group [F (8,374) = 1.12, p = 0.34], condition x limb [F (4,187) = 0.45, p = 0.76] or group x limb [F (4,187) = 0.230, p = 0.921]. Also, no significant condition x group x limb [F (4,187) = 0.41, p = 0.79] interaction was observed. Follow-up univariate ANOVA were performed for propulsive force, propulsive time, braking force and braking time. The results for each of these dependent variables are provided below.

Propulsive Force

Subsequent ANOVA revealed that there was a significant condition main effect [F (2,199) = 68.96, $p < 0.001$]. Bonferroni post hocs revealed that the propulsive force during both planned and unplanned gait termination was significantly lower than that in catch trials ($p < 0.001$). It was also observed that the AI group had higher propulsive force than the control group [F (1,199) = 8.40, p = 0.004]. Propulsive force data appear in Table 4-3.

Table 4-3. Propulsive Force (N) (Mean ± SD). *Indicates a statistically significant difference between planned and catch trial ($p \leq 0.01$). †Indicates a statistical significant difference unplanned and catch trial ($p \leq 0.01$). Δ Indicates statistically significant difference between AI and healthy ($p \leq 0.01$).

Variable	level		M ±SD
Propulsive Force	1) Condition	a) catch* †	144.61± 38.16
		b) planned*	83.54± 20.71
		c) unplanned†	79.68± 28.63
	2) Group	a) healthy$^{\Delta}$	88.60±.33.64
		b) AI$^{\Delta}$	99.82±.40.79
	3) Limb	a) involved/dominant1	02.38± 42.23
		b) uninvolved/healthy	81.96± 25.37

Propulsive Time

The ANOVA did not reveal significant condition [F (2,199) = 1.64, p = 0.20] or group [F (1,199) = 0.694, p = 0.40] main effects. Propulsive Time data are shown in the subsequent table Table 4-4.

Table 4-4 Propulsive Time (s) (Mean ± SD)

Variable	level		M ±SD
Propulsive Time	1) Condition	a) catch	0.62± 0.14
		b) planned	0.59± 0.10
		c) unplanned	0.57± 0.14
	2) Group	a) healthy	0.59± 0.12
		b) AI	0.58± 0.13
	3) Limb	a) involved/dominant	0.59± 0.13
		b) uninvolved/ healthy	0.58± 0.12

Braking Force

A significant condition main effect [F (2,199) = 31.40, p < 0.001] was observed. Further, Bonferroni post hocs revealed that braking force during unplanned gait termination was significantly higher than braking force during planned gait termination (p < 0.01) Braking force in the AI group was higher than in the control group [F (1,199) = 24.17, p < 0.001]. Braking force data are shown in Table 4-5.

Table 4-5 Braking Force (N) (Mean ± SD).† Indicates a statistically significant difference between planned and unplanned (p ≤ 0.01). $^{\Delta}$ Indicates statistically significant difference between healthy and AI (p ≤ 0.01).

Variable	level		M ±SD
Braking Force	1) Condition	a) planned†	177.42± 56.21
		b) unplanned†	222.89± 84.39
	2) Group	a) healthy $^{\Delta}$	161.61± 62.2
		b) AI $^{\Delta}$	207.11± 80.89
	3) Limb	a) involved/dominant	173.18± 72.33
		b) uninvolved/ healthy	201.13± 77.54

Braking Time

No significant group main effect [F (1,199) = 0.007, p = 0.93] was noted. However, a significant condition main effect [F (2,199) = 214.34, p < 0.001] was observed. Bonferroni post hocs revealed no significant differences with the braking time during planned and unplanned gait termination. Data for Braking time are shown in Table 4-6.

Table 4-6 Braking Time (s) (Mean ± SD)

Variable	level		M ±SD
Braking Time	1) Condition	a) planned	0.70± 0.11
		b) unplanned	0.67± 0.17
	2) Group	a) healthy	0.58± 0.23
		b) AI	0.58± 0.26
	3) Limb	a) involved/dominant	0.51± 0.28
		b) uninvolved/ healthy	0.68± 0.12

Stability

A 3-way MANOVA (2:group x 2:limb x 2:condition) was computed for APSI, MLSI, VSI and DPSI. Significant main effects were noted for condition [F (4,149) = 8.39, p < 0.001] and group [F (4,149) = 4.68, p < 0.001]. No significant limb main effect was detected [F (4,149) = 1.82, p = 0.12]. No significant condition x group [F (4,149) = 0.77, p = 0.54], condition x limb [F (4,149) = 0.27, p = 0.89], group x limb [F (4,149) = 0.89, p = 0.46] or condition x group x limb [F (4,149) = 0.87, p = 0.48] interactions occurred. Subsequent ANOVA were computed for each dependent variable, the results of which are discussed below.

Anteroposterior Stability Index (APSI)

The APSI score for the AI group was higher than the control group [F (1,159) = 8.33, p = 0.004]. The APSI score for unplanned gait termination was higher than planned gait termination [F (1,159) = 26.88, p < 0.001]. APSI scores can be found in Table 4-7.

Table 4-7 Anteroposterior Stability Index (APSI) (Mean ± SD). * Indicates a statistically significant difference between unplanned and planned ($p \leq 0.01$). † Indicates statistically significant difference between healthy and AI ($p \leq 0.01$).

Variable	level		M ±SD
APSI	1) Condition	a) planned*	0.13± 0.01
		b) unplanned*	0.15± 0.02
	2) Group	a) healthy †	0.13± 0.019
		b) AI †	0.14± 0.02
	3) Limb	a) involved/dominant	0.14± 0.02
		b) uninvolved/ healthy	0.14± 0.02

Mediolateral Stability Index (MLSI)

The ANOVA revealed no significant group [F (1,159) = 0.93, p = 0.33], condition [F (1,159) = 0.25, p = 0.30] or limb [F (1,159) = 1.13, p = 0.28] main effects. MLSI data appear in Table 4-8.

Table 4-8. Mediolateral Stability Index(MLSI) (Mean ± SD)

Variable	level		M ±SD
MLSI	1) Condition	a) planned	0.11± 0.69
		b) unplanned	0.03± 0.01
	2) Group	a) healthy	0.10± 0.69
		b) AI	0.02± 0.01
	3) Limb	a) involved/dominant	0.11± 0.69
		b) uninvolved/ healthy	0.14± 0.02

Vertical Stability Index(VSI)

No significant condition [F (1,159) = 0.07, p = 0.79] or group [F (1,159) = 2.82, p = 0.09] main effects were noted for VSI. VSI data are shown in Table 4-9.

Table 4-9. Vertical Stability Index (VSI) (Mean ± SD)

Variable	level		M ±SD
VSI	1) Condition	a) planned	0.17± 0.03
		b) unplanned	0.17± 0.03
	2) Group	a) healthy	0.17± 0.03
		b) AI	0.18± 0.04
	3) Limb	a) involved/dominant	0.17± 0.04
		b) uninvolved/ healthy	0.17± 0.03

Dynamic Postural Stability Index (DPSI)

A higher DPSI score was detected during unplanned gait termination trials than during planned gait termination trials [F (1,159) = 7.22, p = 0.008]. A higher DPSI score was observed in AI group than in the control group [F (1,159) = 9.87, p = 0.002]. Table 4-10 shows the DPSI data.

Table 4-10 Dynamic Postural Stability Index(DPSI)(Mean ± SD) * Indicates a statistically significant difference between unplanned and planned (p≤0.01) † Indicates statistically significant difference between AI and healthy (p≤0.01).

Variable	level		M ±SD
DPSI	1) Condition	a) planned*	0.22± 0.02
		b) unplanned*	0.23± 0.03
	2) Group	a) healthy †	0.22± 0.02
		b) AI †	0.24± 0.04
	3) Limb	a) involved/dominant	0.23± 0.03
		b) uninvolved/ healthy	0.23± 0.03

EMG

Tibialis Anterior

A four-way MANOVA (2: group x 2: limb x 6: phase x 2: condition) with repeated measures on the last factor was computed for TA. Mauchly's test indicated that the assumptions of sphericity were violated (p < 0.001). Hence, the Greenhouse Geisser adjustment was made. Significant condition [F (1, 36) = 49.90, p < 0.001], limb [F (1, 36) = 4.47, p = 0.04] and phase [F(5,32) = 8.19, p < 0.001] main effects were observed. No significant main effect was identified for group [F (1, 32) = 1.03, p = 0.41]. A significant limb x phase [F (5,32) = 3.12, p = 0.02] interaction was noted. No significant condition x group [F (1,36) = 2.19, p = 0.14], condition x limb [F (1,36) = 1.84, p = 0.18], condition x phase [F (5,32) = 1.52, p = 0.20], group x limb [F (1,36) = 0.97, p = 0.33], or phase x group [F (5,32) = 1.03, p = 0.41] interactions were detected. No significant three or four way interactions were noted: condition x group x limb [F (1,36) = 0.71 p = 0.40], condition x group x phase [F (5,32) = 0.98 p = 0.44], limb x group x phase [F

(5,32) = 0.95 p = 0.46], condition x limb x phase [F (5,32) = 1.81 p = 0.13], and condition x group x limb x phase [F (5,32) =1.29 p = 0.29].

Subsequent ANOVA revealed that the average amplitude of TA during unplanned gait termination was higher than that during planned gait termination [F (1, 36) = 49.90, p < 0.001]. Additionally, the average amplitude of TA for the uninvolved limb was higher than the involved limb [F (1, 36) = 4.47, p = 0.04]. Also, Bonferroni post hocs revealed that the average amplitude of TA was maximal in phase 4 [F (2.11, 180) = 12.23, p < 0.001]. A significant interaction of limb x phase [F (3.84, 38.33) = 5.40, p = 0.001] was observed. However, no significant condition x group [F (1,36) = 2.19, p = 0.14], condition x limb [F (1,36) = 1.84, p = 0.18], condition x phase [F (3.29,118.74) = 2.23, p = 0.08], group x limb [F (1,36) = 0.97, p = 0.33] and phase x group [F (5,32) = 1.03, p = 0.41] interactions were observed. Also, No significant condition x group x limb [F (1,36) = 0.71, p = 0.40], condition x group x phase [F (3.28,32) = 0.36, p = 0.095], limb x group x phase [F (3.84,32) = 1.79, p = 0.32], condition x limb x phase [F (2.47,89.14) = 2.63, p = 0.063] and condition x group x limb x phase [F (2.47,32) = 1.24, p = 0.29] interactions were detected. Average amplitude (TA) data are shown in Table 4-11.

Table 4-11 Average Amplitude (TA) (Mean ± SD) * Indicates a statistically significant difference between unplanned and planned (p≤0.01) † Indicates statistically significant difference between the involved/dominant and uninvolved/ non-dominant limb (p≤0.05).

Variable	level		M ±SD
Average Amplitude	1) Condition	a) planned*	0.99± 0.03
		b) unplanned*	1.20± 0.05
	2) Group	a) healthy	1.17± 0.05
		b) AI	1.11± 0.05
	3) Limb	a) involved/dominant†	1.05± 0.03
		b) uninvolved / non dominant †	1.18± 0.04
	4) Phase	a) one	0.99± 0.04
		b) two	1.03± 0.05
		c) three	0.97± 0.05
		d) four	1.58± 0.12
		e) five	1.14± 0.07
		f) six	1.13± 0.05

Soleus

A MANOVA (2: group x 2: limb x 6: phase x 2: condition) with repeated measures on the last factor was computed for average amplitude of Sol. Mauchly's test indicated that the assumptions of sphericity were violated ($p < 0.001$). Hence, the Greenhouse Geisser adjustment was made. Significant condition [$F(1, 36) = 34.99$, $p < 0.001$], limb [$F(1, 36) = 8.65$, $p = 0.006$] and phase [$F(5, 32) = 10.50$, $p < 0.001$] main effects were revealed for average amplitude. Significant limb x phase [$F(5, 32) = 2.72$, $p = 0.03$], condition x phase [$F(5, 32) = 4.82$, $p = 0.002$] and condition x group x phase [$F(5, 32) = 3.35$ $p = 0.06$] interactions were also observed. However, no significant two way interactions were detected for condition x limb [$F(1,36) = 1.92$, $p = 0.17$] , group x limb [$F(1,36) = 1.86$, $p = 0.18$] or phase x group [$F(5,32) = 0.93$, $p = 0.47$]. No significant three way interactions were identified for condition x group x limb [$F(1, 36) = 2.18$ $p = 0.14$], limb x group x phase [$F(5,32) = 0.91$ $p= 0.48$], condition x limb x phase [$F(5,32) = 0.99$, $p = 0.43$] or condition x group x limb x phase [$F(5,32) = 0.29$ $p = 0.91$].

Subsequently, ANOVA were computed which revealed that the average amplitude of Sol during unplanned gait termination was greater than planned gait termination [$F(1, 36) = 34.99$, $p < 0.001$]. Average amplitude of uninvolved limb was greater than the involved limb [$F(1, 36) = 8.65$, $p = 0.006$]. There was a significant main effect for phase [$F(2.93, 105.94) = 9.2$, $p < 0.001$]. Bonferroni post hocs revealed that average amplitude of Sol was highest in phase 5 Also the AI group demonstrated higher average amplitude of Sol than the control group [$F(1, 36) = 4.18$, $p = 0.048$]. A significant limb x phase [$F(3.57, 128.72) = 4.58$, $p = 0.003$] interaction was also observed. A significant condition x phase [$F(3.45, 124.46) = 3.37$, $p = 0.023$] interaction was noted. The average amplitude of Sol was higher for unplanned gait termination during all six sub phases.

No significant two way interactions were identified for condition x limb [F (1, 36) = 2.52, p = 0.12], group x limb [F (1, 36) = 1.06, p = 0.30] or phase x group [F (5, 32) = 0.74, p = 0.52], condition x group [F (1, 36) = 3.27, p = 0.08]. Similarly, no significant condition x group x limb [F (1,36) = 2.82 p = 0.10], condition x group x phase [F (3.45, 32) = 0.99 p = 0.42], limb x group x phase [F (3.38,32) = 0.30 p = 0.84], condition x limb x phase [F (3.43, 32) = 1.48 p = 0.19], condition x group x limb x phase [F (3.43, 32) = 0.26 p = 0.86] interactions were observed. Data for average amplitude Sol for each of the independent variable appear Table 4-12.

Table 4-12 Average Amplitude (Soleus) (Mean ± SD). * Indicates a statistically significant difference from planned (p≤0.01). † Indicates a statistically significant difference from healthy (p≤0.05). $^{\Delta}$ Indicates statistically significant difference from the uninvolved/ non-dominant (p≤0.01)

Variable	level		M ±SD
Average Amplitude	1) Condition	a) planned*	1.13± 0.05
		b) unplanned*	1.42± 0.06
	2) Group	a) healthy †	1.13± 0.07
		b) AI†	1.16± 0.08
	3) Limb	a) involved/dominant $^{\Delta}$	1.19± 0.05
		b) uninvolved/ non dominant $^{\Delta}$	1.36± 0.06
	4) Phase	a) one	1.31± 0.06
		b) two	1.24± 0.09
		c) three	0.88± 0.08
		d) four	1.25± 0.07
		e) five	1.61± 0.11
		f) six	1.36± 0.18

Gluteus Medius

A MANOVA (2:group x 2:limb x 6:phase x 2:condition) with repeated measures on the last factor were computed for average amplitude of GM. Mauchly's test indicated that the assumptions of sphericity were violated (p < 0.001). Hence, the Greenhouse Geisser adjustment was made. Significant condition [F (1, 36) = 11.49, p = 0.0002] and phase [F(5,32) = 14.77, p < 0.001] main effects were identified. However, no significant limb [F (1, 36) = 0.03, p=0.86] or group [F (1, 36) = 1.75, p=0.19] main effects were revealed. No significant condition x limb [F(1, 36) = 2.13, p = 0.15], limb x phase [F (5, 32) = 1.17 p = 0.34], condition x phase [F(5, 32) = 1.88, p = 0.12], condition x group [F (1, 36) = 1.46, p = 0.23], group x limb [F(1, 36) =1.67, p = 0.20] or phase x group [F (5, 32) = 1.56, p = 0.19] interactions were detected. Also, no significant condition x group x limb [F (1, 36) = 0.001 p = 0.97], condition x group x phase [F (5, 32) = 1.77 p = 0.14], limb x group x phase [F (5, 32) = 0.818, p = 0.54], condition x limb x phase [F (5, 32) = 2.26, p = 0.07], condition x group x limb x phase [F (5, 32) = 0.96, p = 0.45] interactions were detected.

Subsequent ANOVA revealed that the average amplitude of GM was higher during unplanned gait termination than during planned gait termination [F (1, 36) = 11.49, p = 0.002]. Statistical analysis also revealed that the muscle had the highest average amplitude during phase six [F (2.52, 90.75) = 12.19, p < 0.001]. A significant condition x group x phase [F (5, 32) = 2.4, p = 0.03] interaction was detected for average amplitude of GM. No significant condition x limb [F (2.17,36) = 2.13, p = 0.15], limb x phase [F (2.67,96.30) = 1.05 p = 0.35], condition x phase [F (5,74.6) = 1.63, p = 0.20], condition x group [F (1,36) = 1.46, p = 0.23], group x limb [F (1,36) =1.67, p = 0.20] or phase x group [F (2.5,32) = 1.68, p = 0.14] interactions were noted. No significant three way condition x group x limb [F (1, 36) = 0.001 p = 0.97], limb x group x phase [F (2.70, 32) = 1.45, p = 0.20], condition x limb x phase [F (2.28, 82.31) = 2.66, p = 0.07], condition x group x limb x phase [F (2.28,32) = 0.78, p = 0.47] interaction were observed. Table 4-13 shows the data for the average amplitude of GM.

Table 4-13 Average Amplitude (GM) (Mean ± SD). * Indicates a statistically significant difference between unplanned and planned ($p \leq 0.01$).

Variable	level		M ±SD
Average Amplitude	1) Condition	a) planned*	1.19± 0.05
		b) unplanned*	1.40± 0.07
	2) Group	a) healthy	1.36± 0.07
		b) AI	1.22± 0.07
	3) Limb	a) involved/dominant	1.29± 0.07
		b) uninvolved/ healthy	1.30± 0.06
	4) Phase	a) one	1.05± 0.06
		b) two	1.07± 0.05
		c) three	0.12± 0.06
		d) four	1.54± 0.13
		e) five	1.35± 0.09
		f) six	1.62± 0.07

CHAPTER 5
DISCUSSION

The purpose of this study was to determine if differences exist between feedback and feedforward neuromuscular control mechanisms in participants with a history of AI during gait termination. We tested this by means of three major outcome measures namely force, stability and EMG. Initially gait velocity and the relative timing of delivery of the auditory cue with respect to the gait cycle were measured to confirm whether comparison of groups was valid. We attempted to reveal the underlying differences in force by comparing the propulsive force, propulsive time, braking force and braking time. The force results were partially supportive of the hypotheses. Dynamic postural stability in both the groups was compared on the basis of four different stability measures: APSI, VSI, MLSI and DPSI. The data supported the hypothesis that the AI group would be less stable as compared to control group. The EMG findings supported the differences in the propulsive and braking force thus contributing to the global picture of differences in feedback and feedforward neuromuscular controls in the AI group. The EMG results supported the central hypothesis however only partially supported the individual hypotheses. The results and hypotheses for individual outcome measures are examined relative to the few previously published studies and discussed below.

Gait Velocity

It has been well documented in the literature that the pattern of muscle recruitment during gait termination changes with a change in gait velocity. For example, Crenna et al. (2001) suggested that with an increase in gait velocity the pattern of muscle recruitment used for gait termination shifts from distal to proximal. Hence in this study the prerequisite for a valid comparison between groups was to ensure that gait velocity was the same. The results indicated that both the groups had very similar gait velocities. There was no statistically significant difference in the gait velocity between groups. However, results indicated that the participants in both the AI and control groups walked faster during unplanned gait termination trials. The difference in the mean velocities during planned (1.23±0.01 m/s), unplanned (1.24±0.48 m/s) and catch (1.19±0.11 m/s) was practically negligible (~3%). Sparrow et al. (2005) proposed that the probability of stimulus delivery and an anticipation of a stop might have an effect on different variables like the stopping time and the ground reaction force of the stance and the swing limb during gait termination. The main goal during data collection was to avoid anticipation of a stop during unplanned gait termination trials. In fact, the finding that the gait velocity during unplanned gait termination was greater indicates a minimal anticipation of a stop.

Relative Timing of Stimulus Delivery

It is well documented in literature that the pattern of foot placement at the instant when the person terminates gait depends on the timing of stimulus delivery relative to the phase of gait cycle. Hase and Stein (1998) documented that when the stimulus was applied between 35 to 70% of the gait cycle (late stance phase to mid swing phase of the right limb) the rapid stopping ended with right limb forward. Whereas, when the stimulus was delivered within the first 20% of the gait cycle, the stopping occurred with the left limb forward. When the stimulus was delivered between 20 to 35% of the gait cycle or between 70 to 85% of the gait cycle, the subject tended to stop with an extra transitional step. Very rarely did the subjects stand with feet together, except after having a particularly short step in the transition periods (Hase et al., 1998). Similar findings were reported by Tirosh and Sparrow (2004) who studied the effect of stimulus delay on gait termination. A visual stopping signal was presented 10ms before the heel strike of the trail limb and 450ms post heel strike (just before heel off of the trail limb). The authors concluded that the pattern of final foot placement is directly related to the relative timing of stimulus delivery (Bishop et al., 2003; Delahunt et al., 2006; Tirosh & Sparrow, 2004). Hence, we chose to keep the relative timing of stimulus delivery constant in both groups across all conditions. The results indicate that there was no difference in timing delivery between groups or across conditions.

Forces

Propulsive Force

Hase and Stein (1998) analyzed the changes in forces that occurred during gait termination. They documented that propulsive force in the trail limb is specifically reduced during gait termination. Crenna and colleagues (2001) described the different motor programs used during gait termination. These authors have also identified a reduction in the propulsive force as a strategy used in the trail limb during gait termination. Interestingly, recent studies have reported that there are significant force sense deficits in functionally unstable ankles (Bishop et al., 2003; Docherty & Arnold, 2008; Hase et al., 1998; Jaeger et al., 1992). Docherty and Arnold (2008) measured the ability of subjects with history of AI to produce 10 %, 20% and 30% of maximum voluntary isometric and isokinetic torques and compared it with controls. The authors noted that the AI group lacked the ability to accurately reproduce a given force. On the other hand it has been documented that in the AI population the ankle has an increased joint velocity at heel strike during normal gait (Bishop et al., 2003; Monaghan et al., 2006). This is indirect evidence of inability of the AI group to modulate propulsive forces even during normal walking. In the current study we tried to compare the propulsive forces produced by the AI population with the propulsive forces produced by healthy adults during planned and unplanned gait termination and during catch trials. The major finding was that the propulsive force during both planned and unplanned gait termination was significantly less than the catch trials for both AI and the control group. The first hypothesis made was that propulsive force during unplanned gait termination would be higher than during planned gait termination. This hypothesis was based on the assumption that unplanned gait termination was an unanticipated stop which may not provide enough time to reduce the propulsive force. Interestingly the results did not support this hypothesis: the propulsive force during unplanned gait termination was in fact less than during planned gait termination, although the differences were statistically not significant. Unplanned gait termination involves a sudden stop and hence would involve a greater reliance on feedback neuromuscular control for gait termination. It could be argued that the feedback mechanisms were equally effective in reducing the propulsive force as the feedforward mechanisms.

The second hypothesis was that the propulsive force in the AI group would be lower than in the control group. This hypothesis was based on findings of previous studies which report lower average amplitude of the muscles responsible for producing propulsion during normal gait in AI (Delahunt et al., 2006). The results of the current study failed to support the hypothesis. Interestingly propulsive force during gait termination in AI group was greater than in controls. In our experiment the AI group was unable to reduce propulsive force both during unplanned and planned gait termination when compared to the propulsive force during catch trials. Bishop and colleagues (2003) have demonstrated that in healthy adults during unplanned gait termination, the trail limb strategy is used to reduce propulsive force when the stimulus is given at heel strike. Empirical data suggesting sensory proprioceptive deficits and altered neuromuscular feedback in AI could explain the finding for unplanned gait termination (Delahunt et al., 2006; Docherty et al., 2006; Freeman et al., 1965; Monaghan et al., 2006; Ross & Guskiewicz, 2004d; Wikstrom et al., 2007). The propulsive force during unplanned gait termination in AI group could be larger due to decreased feedback from the ankle joint. The inability of the AI group to reduce the propulsive force during planned gait termination when compared to the propulsive force during catch trials suggests a potential reorganization of feedforward neuromuscular control.

The third hypothesis was that propulsive force in the involved limb would be lower than the propulsive force in the uninvolved limb. This hypothesis was again based on the finding of decreased EMG amplitude in the muscles responsible for producing propulsion during normal gait in AI (Delahunt et al., 2006). In addition, recent EMG studies regarding H:M ratios (H reflex is the electrical equivalent of a monosynaptic stretch reflex, M wave signifies a maximal response that can be obtained on stimulation of the muscle. M wave values are used of normalization of H reflex values and the H/M ratios are used commonly in reflex studies) report that post ankle injury there is a significant arthrogenic inhibition in muscles like the Sol which are responsible for producing propulsive force (Bishop et al., 2003; Hase et al., 1998; Jaeger et al., 1992; McVey, Palmieri, Docherty, Zinder, & Ingersoll, 2005). However the present results failed to support this hypothesis. There was no significant statistical difference between the involved/dominant and the uninvolved/nondominant limb. A possible reason for this finding is the method of statistical analysis. In our analysis we grouped the dominant limb in the control group with the injured limb in the AI group and the nondominant limb in the control group with the uninjured limb of the AI group. Possibly this comparison failed to reveal the potential main effect of limb on the propulsive force as the differences between the involved and the uninvolved limbs of the AI group were masked by the inclusion of the control group (Fig 5-1).

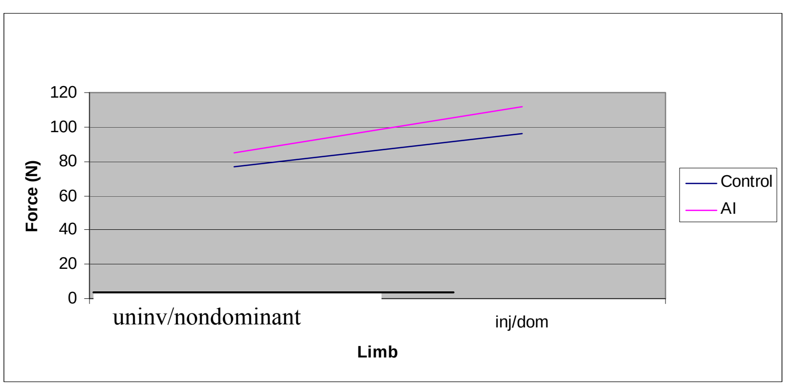


Figure 5-1. Comparison of propulsive force between limbs and across groups.

Braking Force

The first hypothesis stated was that the braking force during unplanned gait termination would be higher than that seen during planned gait termination. The results of this study supported this hypothesis. Jaeger and colleagues (1992) were the first to document that a reduction in propulsive and an increase in braking force were the two main characteristic findings during gait termination. During normal gait termination there is both reduction in propulsive force in the trail limb along with an increase in braking force in the lead limb (Bishop et al., 2003; Crenna, Cuong, & Breniere, 2001; Hase et al., 1998; Jaeger et al., 1992). Hence it could be argued that during unplanned gait termination an increased braking force was required to arrest the forward momentum of the body generated by a higher propulsive force for a safe gait termination.

The second hypothesis stated that the braking force in AI group would be higher than that seen in the control group. This hypothesis was supported by the results of this study. It is well documented in the literature that both reduction in propulsive force of the trail limb and an increase in the braking force of the lead limb are required for a safe gait termination (Hase et al., 1998; Jaeger et al., 1992). The requirement of a higher braking force for a safe gait termination in the AI group could be explained as a consequence of a higher propulsive force. A higher propulsive force in AI group would make it necessary for the muscles to produce a higher braking force for gait termination. In contrast, the controls demonstrated less braking force. It was interesting to note that in control group there was a greater reduction in propulsive force in the trail limb. Thus the control group was able to effectively modulate the trail limb and the lead limb strategies during gait termination. Alternatively, the AI group depended heavily on the lead limb strategy, possibly due to feedbackback deficits during unplanned gait termination and feed forward deficits during planned gait termination.

The third hypothesis stated that the braking force in the involved limb would be lower than that in the uninvolved limb. Recent studies report that there are definite deficits in force production in the injured limb (Docherty et al., 2008; Fox, Docherty, Schrader, & Applegate, 2008). Docherty et al. (2008) have also reported force sense deficits and the inability of the injured limb in the AI group to consistently reproduce a specific level of force. In another study, Fox and colleagues (2008) have reported that the injured limb shows deficits in eccentric plantarflexion torque when compared to the normal limb. However, in spite of the obvious group differences in braking force the results of this study failed to support the first hypothesis. In our analysis for comparing the differences between limbs we grouped the dominant limb in the control group with the injured limb in the AI group and the nondominant limb in the control group with the uninjured limb of the AI group. This comparison may have failed to reveal the statistical main effect of limb on the braking force as the differences between the involved and the uninvolved limbs of the AI group were masked by the inclusion of the control group.

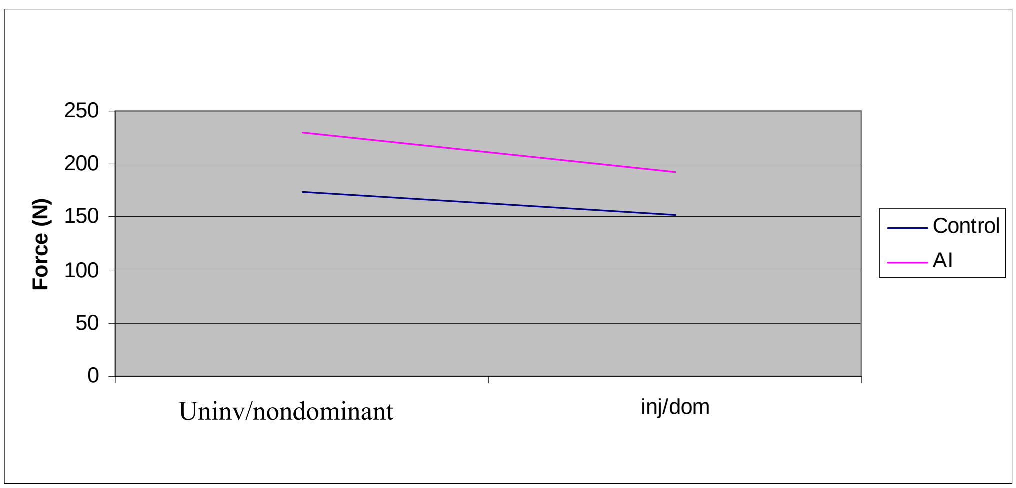


Figure 5-2 Comparison of braking force between limbs and across groups

Propulsive Time and Braking Time

Crenna and colleagues (2001) described the motor programs that are used during gait termination. In their study they described that the lead limb motor program is quite robust. The onset latencies of the individual muscle recruited in this strategy (Quadriceps and Sol) showed a close correlation, and the spatio-temporal parameters were always scaled in parallel. However they reported that trail limb motor program is flexible and velocity dependent. The trail limb showed a delay of 150ms in the muscle recruitment after the braking stimulus was applied. In contrast, the limb showed a delay of 330 ms after the stimulus was applied. Even the lead limb strategy showed a distal to proximal muscle recruitment mainly in Sol and Quadriceps. Increase in walking speed shifted the pattern of muscle recruitment in the trail limb from distal to proximal. However, in our study there was no practically significant difference in the velocity across groups, conditions and limb. Hence it could be argued that the same lead limb and trail limb motor programs were executed while terminating gait in both the conditions, for both limbs and in both groups. Thus if the basic sequence of the pattern of muscle recruitment remained the same it could be argued that the propulsive time and the braking time remained the same across groups, condition and limb.

Stability

The first hypothesis stated that the AI group would be less stable than the control group. In this study, dynamic postural stability was evaluated based on scores for APSI, MLSI, VSI and DPSI. The results supported this hypothesis. Ross et al. (2004) theorized that people with FAI take longer to decelerate their center-of-mass oscillations because they allow their center of mass to approach the limits of stability which lead to large external moments that act on the joint to destabilize the body. Previous researchers have documented differences in dynamic postural stability between AI and control groups based on the APSI, MLSI, VSI and DPSI scores

(Wikstrom et al., 2005b; Wikstrom, Tillman, Kline, & Borsa, 2006; Wikstrom et al., 2007). Specifically dynamic postural stability deficits in the AI group were observed. Wiksstrom et al (2005) reported that the DPSI (0.85±0.17), APSI (0.36±0.09) and VSI (0.73±0.17), scores in their study were higher/worse in the AI group as compared to DPSI (0.73±.17), APSI (0.30±0.06) and VSI (0.61±0.13) scores in controls indicating deficits in dynamic postural stability in AI. In the present study the APSI and the DPSI scores for the AI group were higher than the control group indicating that the AI group was less stable. These findings correlate to those documented in literature and in turn reinforce the finding of feedback neuromuscular deficits that exist in the AI group. However, in the present study the values of DPSI and all its subcomponents are lower than those reported by Wikstrom and colleagues (2006) in the studies done on jump landing. The likely reason for this finding is that the present study used the gait termination model for measuring the DPSI and its subcomponents. The task of gait termination involves a lesser impact than the task of jump landing hence the values of all subcomponents are lower. It was also interesting to note that in this study we did not find any significant differences in the VSI sub component of DPSI. The possible explanation for this finding is that the study used the gait termination model. Previous studies which have documented significant differences in VSI between the AI and the control groups have used the 'jump landing model'. The vertical destabilizing component in this model is expected to be greater than the gait termination model. On similar lines the APSI and the DPSI scores in unplanned gait termination were higher than the score in planned gait termination. Unplanned gait termination involved an unanticipated and sudden stop. Such gait termination is known to pose a greater challenge to maintenance of balance and postural stability (Bishop et al., 2003). The findings of this study reinforce this finding and thus indicate that the unplanned gait termination posed a greater challenge to maintenance of the postural stability. Planned gait termination involved an anticipated stop. Studies have documented that during planned gait termination subjects decrease propulsive force and increase braking force at least two steps prior to stopping (Wearing, Urry, Smeathers, & Battistutta, 1999) with as much as 90% of the deceleration occurring in the final step (Jian, 1993). Thus it could be argued that this type of gait termination involves feedforward planning.

The stability indices failed to show any significant differences between dominant/involved and non dominant/uninvolved limbs. Thus the results did not support the hypothesis that the involved limb would be less stable than the uninvolved limb. This finding seems to contradict the earlier finding of group differences in the stability indices. However this finding could be explained on the basis of findings of other studies. Wikstrom et al. (2007) reported that for healthy individuals the DPSI scores in the dominant limb were not significantly different than those for the non dominant limb. In this study the dominant and the involved limb were together statistically compared to the non dominant and uninvolved limb. Since we had 20 healthy subjects the actual differences in the involved and the uninvolved limb could be masked due to the design of the statistical analysis of this study.

EMG

Distal Muscles (TA and Soleus)

The first hypothesis stated was that the average amplitude of the TA and Sol in AI group would be less than the control group. This hypothesis was supported by the results of this study. The average amplitude of Sol was significantly less in the AI group as compared to the control group across all six phases. This finding is however contradictory to that reported in the literature. Delahunt and colleagues (2006) compared the integral EMG of TA and Sol between the AI and controls. They indicated that the Sol muscle in the AI group had a higher average EMG as compared to the controls. However the average integral EMG of TA was reported to be less than the controls. The likely reasons for the results of our study to be different from those reported are two. First the Delahunt study investigated the muscle activity of TA and Sol during normal walking. Conversely, our study measured the average amplitude of these muscles during the task of planned and unplanned gait termination. Second the results reported by the study are calculated for the time frame of 200ms before heel strike to 80 milliseconds after heel strike. In our study we measured the average amplitude of the muscles during the terminal swing phase and whole of the stance phase of each limb. The group data were not statistically different for the TA. However the mean of the average amplitude of TA was AI group was lower than that of the control group. This finding supports the data indicating that the propulsive force for AI group was higher than in the control group. During gait termination, the TA is responsible for reducing the propulsive force of the trail limb (Bishop et al., 2002; Crenna et al., 1991). During gait termination, TA is known to be most efficient at reducing the anterior tibial advancement in the trail limb between heel strike to midstance phases (Crenna et al., 2001). The lower average amplitude of TA in the AI group may explain the finding of increased propulsive force in the AI group. The significant limb x phase interaction supports this finding more specifically. The average amplitude of TA in the involved limb was less than the uninvolved limb during the first four phases of gait termination (Fig 5-3).

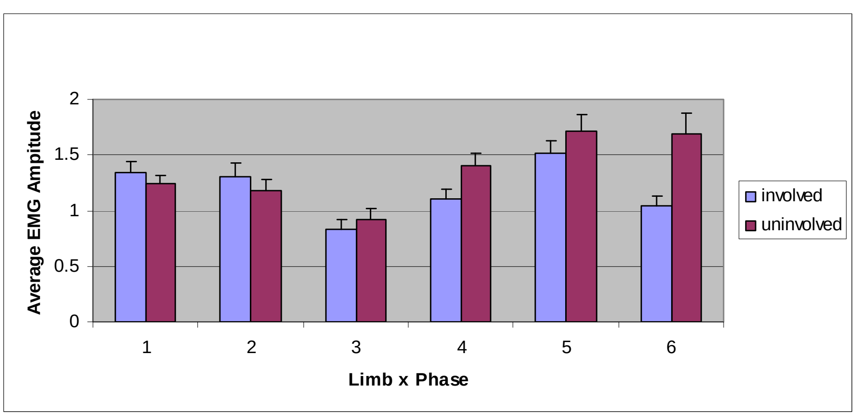


Figure 5-3 Average amplitude of TA in limb x phase interaction

Also other plantar flexors like the gastrocnemius could possibly be contributing toward propulsion. However this justification is only a surmise as our study did not measure the muscle activity of either gastrocnenius or peroneal muscle group. A burst of Sol muscle activity in the lead limb is responsible for increasing the braking force during gait termination (Crenna et al., 1991). The results of this study indicate that the braking force in the AI group was more than the control group. However the average amplitude of the Sol muscle in the AI group was lower than the control group. These contradictory findings may indicate that the AI group relied on muscles other than Sol for generating the braking force.

The second hypothesis stated that the average amplitude of TA and Sol would be higher for unplanned gait termination as compared to the planned gait termination. This hypothesis was supported. The average amplitudes of TA and Sol were significantly more during unplanned gait termination as compared to planned gait termination. These results support the finding that the propulsive force during unplanned gait termination was lower (though not statistically significant) than during planned gait termination. As stated earlier, an increased TA activity in the trail limb is responsible for reduction of propulsive force of the trail limb. The results of our study demonstrated higher average amplitude of TA during unplanned gait termination than during planned gait termination. The average amplitude of Sol was higher during unplanned gait termination than during planned gait termination. As mentioned earlier, the Sol is the primary muscle responsible for producing braking force in the lead limb during gait termination (Crenna et al., 1991; Delahunt et al., 2006; Hase et al., 1998). Hence the higher average amplitude of Sol during unplanned gait termination justified the higher braking force that was evident during unplanned gait termination. More specifically a significant condition x phase interaction indicated that the Sloeus muscle had higher average amplitude during all six sub phases of the limb during gait termination (Fig 5-4).

The third hypothesis stated that the average amplitude of TA and Sol for the involved limb would be lower than the uninvolved limb. This hypothesis was supported. Lower average amplitude of TA in the uninvolved limb and corresponds to the higher propulsive force in the uninvolved limb.

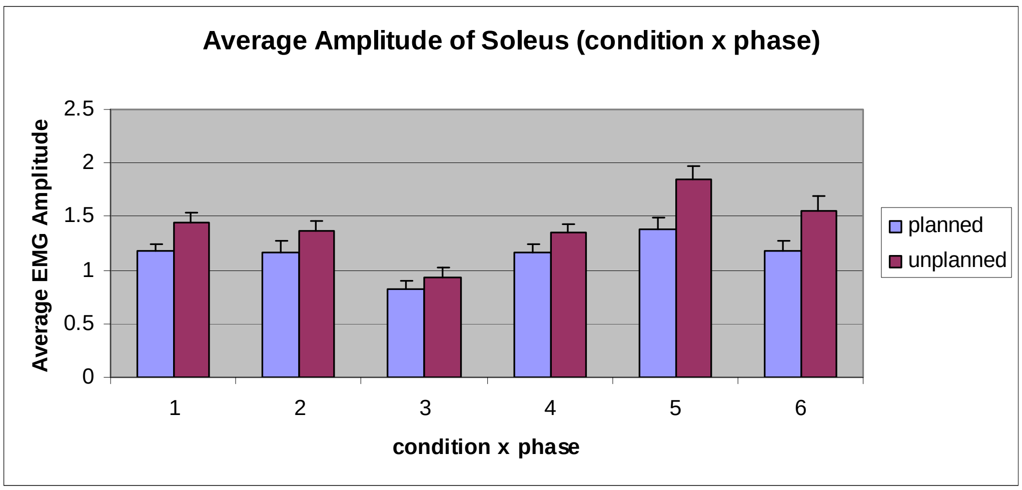


Figure 5-4 Average amplitude of Soleus in condition x phase interaction

Likewise lower TA activity in the injured limb justifies the higher propulsive force in the AI group. The finding of lower average amplitude of the Sol in the involved limb does not support the finding that the braking force in the AI group was higher than the control group. These two contradictory findings indicate that the involved limb may rely on a muscle other than Sol for producing the braking force for planned gait termination

Proximal Muscle (Gluteus Medius)

To date researchers have focused on the contribution of GM during gait termination in healthy adults (Delahunt et al., 2006; Hase et al., 1998; Crenna et al., 2001). The current study was the first to measure the average amplitude of GM in AI group during gait termination. The

GM along with Quadriceps and Sol is responsible for the producing the braking force in the lead limb (Crenna et al., 2001). Three hypotheses were made. The first hypothesis stated that the average amplitude of GM during unplanned gait termination would be greater than during planned gait termination. The second hypothesis stated that the average amplitude of GM for involved limb would be higher than that in the uninvolved limb. The third hypothesis stated was that the average amplitude for the AI group would be higher than that of the control group. The average amplitude of GM for unplanned gait termination was higher than that seen in planned gait termination which supported the first hypothesis made for GM. This finding also supported the result of higher braking force during unplanned gait termination that was discussed earlier. Bullock et al (1994) investigated the activation patterns of proximal muscles like Errector spinae, Gluteus Maximus and Hamstrings. The authors reported there is a definite delay in the onset time of the proximal muscles on the injured side (Bullock-Saxton, Janda, & Bullock, 1994) .The present study however failed to show any difference between the involved and the uninvolved limb or between the AI and the control group thus failing to support the second and the third hypotheses. Poor signal to noise ratio due to the deep anatomical location of the muscle could be a possible cause.

Synopsis of the Results

It has been well documented in the literature that ankle joint is more susceptible to injury when there is a sudden shift or deceleration of COM of the body (Johansson et al., 1991; Wikstrom et al., 2006). Gait termination involves a rapid deceleration of the forward momentum of the body during steady gait. A safe termination of gait requires a complex interaction of the neuromuscular system (Hase et al., 1998) and involves a known and repeatable set of neuromuscular responses (Bishop et al., 2006; Bishop et al., 2003; O'Kane et al., 2003). Accordingly, gait termination experiments can be constructed to challenge both feedforward and feedback neuromuscular control (Bishop et al., 2006). Crenna et al. (2001) investigated the strategies and patterns of muscle recruitment during gait termination in healthy individuals. Other researchers have also investigated the deficits of force and changes in the muscle recruitment patterns in AI group during normal gait (Delahunt et al., 2006; Monaghan et al., 2006). Interestingly, the current study is the first to examine the forces, stability and the patterns of muscle recruitment in AI population during gait termination. In this study we attempted to unveil the feedback and feed forward deficits by analyzing forces, stability and the average amplitude of the muscle recruitment during unplanned gait termination and feedback deficits during planned gait termination. The results suggest that both feedback and feedforward deficits coexist in AI. This finding is similar to the recent work of Hertel and colleagues (2008). In this study we found significant feedback deficits in the AI group when we analyzed both the propulsive force and braking force. Both propulsive force and braking force in the AI group were higher than that seen for the control group during unplanned gait termination. More specifically the results demonstrated that the AI group relied more on lead limb strategy than the trail limb strategy during gait termination. This statement can be supported by the higher propulsive force and the braking force that can be seen in the AI group as compared to controls. The EMG findings for TA and Sol also revealed some interesting differences between groups. The average amplitude of TA for the involved limb was less than the uninvolved limb. Further a significant limb x phase interaction revealed that the TA in the involved fired less than the uninvolved limb during all four subphases of stance (Fig 5-3).

These two findings suggest that the involved limb was less effective in executing the trail limb strategy of reducing the propulsive force and also support the observation that the AI group relied less on the trail limb strategy of reducing the propulsive force during gait termination. The average amplitude of TA during unplanned gait termination was larger than the planned gait termination which helps explain the decreased propulsive force during unplanned gait termination. This finding also suggests that the feedback mechanisms of neuromuscular control are equally effective in reducing the propulsive force when the stimulus to stop is given just before heel strike and is consistent with the findings reported by Bishop and colleagues that the trail limb strategy is effective during unplanned gait termination even when the trigger is given at heel strike (Bishop et al., 2003; Delahunt et al., 2006). The increased average amplitude of Sol during unplanned gait termination supported the result that braking force during unplanned gait termination was higher than during planned gait termination. Interestingly, it was also noted that the average amplitude of Sol in the AI group was less than that in the control group. This finding failed to explain the result that the AI group generates a higher braking force than the control group during gait termination. However another muscle might be responsible for producing the braking force in the AI group during gait termination. Greater average amplitude of the GM was evident during unplanned gait termination as compared to planned gait termination which supports the higher braking force that is evident during unplanned gait termination. However, the present study failed to demonstrate any group or limb differences in the average amplitude of GM. The result of increased propulsive forces in AI group as compared to controls during planned gait termination suggests that the AI group was not able to effectively reduce the propulsive force even during planned gait termination. Thus feedforward deficits coexist with the feedback deficits in AI.

This study was the first designed to specifically investigate dynamic postural stability deficits in an AI group by using DPSI and its components during gait termination. Compromised dynamic postural stability was noted in the AI group as compared to control and during unplanned gait termination than during planned gait termination. An interesting finding of this study was that there were significant deficits in the anteroposterior stability indicated by a higher APSI score in AI population and in the involved limb. Also the DPSI score for the involved limb was higher than the uninvolved limb. All these finding suggest that there are significant dynamic postural stability deficits in the AI population even during a simple and functional task like unplanned gait termination.

Limitations

This study included an AI population across only a very small age range. Although the inclusion criterion allowed the age range of 18 to 30 years, we were only able to collect data on participants between 20- 22 years for both the groups. Presumably, AI affects individuals of all ages. Further there were no specific criteria based on the functional deficits that determined the inclusion criterion as no standardized screening questionnaire (AJFAT) was used. In addition, the basic screening of the participants was done on the basis of the information provided that was subjective and retrospective. Another significant limitation was that this study used surface EMG to measure the average amplitude of the muscles during gait termination. Although practically difficult, indwelling EMG would be able to give more accurate results especially for muscles like GM which have an anatomically deeper location. In this study we investigated only three representative muscles for studying the strategies of gait termination. Furthermore only average amplitude of the muscle was measured. Further investigation of muscle recruitment based on the variables like onset time, onset duration, and peak amplitude could give us valuable information regarding the pattern of muscle recruitment during gait termination. Also investigation of muscles (peroneal group and the Hamstrings) could help us further understand the strategies used by the subjects with AI during gait termination.

Conclusion

From the results of this investigation it appears that the control strategies used by the AI group differ from that of the control group. In the current study we investigated force, stability and the muscle recruitment in AI group and made comparisons to the control group both during planned and unplanned gait termination. The results for the force and the stability measures revealed that there are alterations in both feedback and feed forward neuromuscular control mechanisms in persons who suffer from AI. Higher DPSI and APSI scores in the AI group along with the finding of a definite difference in the average amplitude of TA and Sol between the AI and control group and between planned and unplanned gait termination supported the central hypothesis that feedback and feedforward deficits in neuromuscular control exist for AI individuals and that the gait termination model was effective in revealing both the those deficits.

Future research should be directed at understanding the underlying reasons behind feedforward deficits and could be beneficial for clinicians who evaluate and develop rehabilitation protocols for individuals with AI. More specifically, the time of onset of feed forward deficits may be important. Also, a possible relationship between feedback and feedforward neuromuscular deficits and whether they are reversible is of concern. Study of variables like the onset time, onset duration and RMS of Sol and TA would be helpful to broaden our understanding of the problem of AI. Further research targeting the peroneal and the hamstring groups in the AI population could help us understand the global picture of the changes in the pattern of muscle recruitment during gait termination in this population. Elucidation of these changes could help us device specific rehabilitation strategies to treat and prevent one of the most commonly occurring and functionally debilitating musculoskeletal injury.

APPENDIX A
ANKLE INJURY QUESTIONNAIRE

Ankle Injury Questionnaire

1. Which ankle have you sprained in the past?

If neither, please turn in this questionnaire R L

Neither

2. Did the initial injury to your ankle require crutches, immobilization, or both, of any form (cast, braces, etc.)? Y N

How long were you on crutches or immobilized?

______ days

3. How many times have you sprained your ankle since the initial injury?

______

4. How many times have you had episodes of your ankle "giving way" or "rolling over" during daily activity (athletic or otherwise)? ______

5. Do you have pain, instability, or weakness in your involved ankle?

Y N

6. Do you attribute these signs and symptoms to the previous injuries associated you're your involved ankle? Y N

To qualify as an AI individual their responses will have to match a specific format :

Question 1- R or L but no history of bilateral trauma

Question 2- Yes but no requirements were made on the number of days

Question 3- must be >1

Question 4- must be >1

Question 5- must answer Yes

Question 6- Must answer Yes

APPENDIX B
STATISTICAL TABLE

Table B-1. Statistics for force

Type of statistic	Independent variable(IV)	Dependent variable(DV)
MANOVA	1)Condition(planned,unplanned, catch) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) Propulsive Force 2) Propulsive time 3) Braking force 4) Braking Time
3 way factorial ANOVA (2x2x3)	1)Condition(planned,unplanned, catch) 2) Group (AI, control) 3) Limb (involved, uninvolved)	1) Propulsive Force
3 way factorial ANOVA (2x2x3)	1)Condition(planned,unplanned, catch) 2) Group (AI, control) 3) Limb (involved, uninvolved)	1) Propulsive time
3 way factorial ANOVA (2x2x3)	1)Condition(planned,unplanned, catch) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) Braking force
3 way factorial ANOVA (2x2x2)	1)Condition(planned,unplanned, catch) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) Breaking time

Table B-2. Statistics for stability

Type of statistic	Independent Variable(IV)	Dependent Variable(DV)
MANOVA	1)Condition-(planned,unplanned) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) DPSI 2) APSI 3) MLSI 4) VSI
3 way factorial ANOVA (2x2x2)	1)Condition-(planned ,unplanned) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) DPSI
3 way factorial ANOVA (2x2x2)	1)Condition-(planned,unplanned) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) APSI
3 way factorial ANOVA (2x2x2)	1)Condition-(planned,unplanned) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) VSI
3 way factorial ANOVA (2x2x2)	1)Condition-(planned,unplanned) 2) Group (AI, control) 3) Limb (involved,uninvolved)	2) MLSI

Table B-3. Statistics for EMG(Soleus)

Type of statistic	Independent Variable(IV)	Dependent Variable(DV)
3 way factorial ANOVA (2x2x2)	1)Condition(planned,unplanned) 2) Group (AI, control) 3)Limb(involved, uninvolved)	1) Average amplitude (Soleus)

Table B-4. Statistics for EMG (TA)

Type of statistic	Independent Variable(IV)	Dependent Variable(DV)
3 way factorial ANOVA (2x2x2)	1)Condition-(planned ,unplanned) 2) Group (AI, control) 3) Limb (involved, uninvolved)	1) Average amplitude (TA)

Table B-5. Statistics for EMG(GM)

Type of statistic	Independent Variable(IV)	Dependent Variable(DV)
3 way factorial ANOVA (2x2x2)	1)Condition-(planned ,unplanned) 2) Group (AI, control) 3) Limb (involved,uninvolved)	1) GM Average Amplitude

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BIOGRAPHICAL SKETCH

I was born in 1981 in a city called Pune, in India, to Mr. Sanjay Suresh Hatttangadi and Mrs. Geetanjali Sanjay Hattangadi. I completed my primary and high school education in Pune and moved to the city of Mumbai to complete my professional education. I completed by bachelor’s degree in physical therapy from Lokmanya Tilak Municipal Medical College in Mumbai. These four years gave me the opportunity to receive the highest standard of education in my field from one of the prestigious and highly regarded colleges in the India for physical therapy education. These four years were the most memorable years of my life. Becoming one of the best clinicians has been my goal ever since. I decided to pursue my graduate education here in the United States of America. I was lucky to be able to complete my master’s education in the field of biomechanics at the University of Florida. This unique experience as helped me grow at an individual and professional level. It was indeed a great academic and cultural experience.