ABSTRACT


Surface electromyography (EMG), normalized to 75% maximum voluntary isometric contraction (MVIC), was analyzed in 10 male and 10 female college aged Ss during barefoot untaped and ankle taped walking and running. Using a foot switch as a time marker, peak EMG amplitude in mV and time to peak amplitude were calculated for the tibialis anterior (TA) and gastrocnemius (GN) muscles relative to time of heel strike. In walking, the time to peak TA significantly (p < .05) preceded the time to peak GN, with ankle taping having no significant (p > .05) effect on the temporal response of the triphasic burst pattern. EMG amplitude during walking presented with variability suggesting this as an area of individual difference. The temporal response of the TA and GN relative to heel strike during untaped running was characterized as a triphasic burst pattern across both genders. From this pattern there was a significant (p < .05) difference between the time peak for initial TA (TA1) and GN, as well as TA1 and a second TA burst (TA2), but not between GN and TA2. Normalized amplitude gender comparisons, as well taped and untaped running comparisons, revealed no significant (p > .05) main effects or interaction. However, the triphasic burst pattern during taped running was either significantly (p < .05) delayed by a mean of .50 sec or completely altered as indicated by lack of a TA2 in 50% of the Ss tested.
EFFECTS OF ANKLE TAPING ON THE NEUROMUSCULAR 
REGULATION OF IMPACT FORCES AT HEEL STRIKE

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We recommend acceptance of this thesis in partial fulfillment of this candidate’s requirements for the degree:

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CHAPTER I
INTRODUCTION

Background

The prevention of ligamentous ankle injuries in sport is often sought through the use of adhesive taping. Prophylactic devices are commonly advocated for protecting the foot and ankle against excessive inversion or coupled plantar flexion/inversion forces (Fumich, Ellison, Guerin, & Grace, 1981; Laughman, Carr, Chau, Youndas, & Sim, 1980). High incidences of ankle sprains undoubtedly prompted the authors of a leading athletic training textbook to include a section describing a “routine non-injury taping” technique for ankles (Arnheim & Prentice, 1997). While the ankle taping of healthy athletes is performed in the interest of injury prevention, it has not been clearly substantiated whether this practice harbors any potential adverse effects. Before the application of ankle taping can be promoted with the same conviction as regular helmet wear or the use of other protective padding, a substantial resolution to this concern is needed.

Several studies (Fumich et al., 1981; Laughman et al., 1980; Metcalfe, Schlabach, Looney, & Renehan, 1997) have suggested ankle taping may reduce ankle injuries by restricting excesses in range of motion (ROM). There has been a concern however as to whether added ankle support with taping or bracing creates a situation where disuse atrophy may occur, potentially reducing muscular girth about the ankle. Francis, Kleiner, Holcomb, and Miller (1997) examined this issue by assessing ankle size via volumetric
analysis in 39 subjects prior to and following a 7 week period in which ankle braces were worn 8 hours per day. They reported that long term bracing did not affect ankle size. With the recent growth in the body of knowledge surrounding proprioception and neuromuscular control, it is certainly conceivable that potential atrophic or girth changes is only one of several areas in need of research before the ankle taping of healthy athletes can be substantially supported or refuted.

Proprioceptors located in the muscle (i.e., muscle spindles), as well as in the articular and cutaneous structures (i.e., mechanoreceptors) supporting the joint, have been identified as major contributors to the normal human gait cycle (McClay, Lake, & Cavanagh, 1990; Mero & Komi, 1987; Osternig, Hamill, Lander, & Robertson, 1986; Scholz & Campbell, 1980). Their activation triggers sensorimotor responses vital for involuntary contributions to coordinated motion (Winter, 1983). As such, the potential of facilitating or inhibiting these important proprioceptive functions through the application of ankle taping has fueled the surge of more recent ankle taping studies. One such area of study is joint position sense. The underlying premise behind joint position sense studies has been a suggestion that the mere presence of athletic tape on cutaneous surfaces stimulates mechanoreceptors which improve limb awareness (Heit, Lephart, & Rozzi, 1996; Simoneau, Degner, Kramper, & Kittleson, 1997. This heightened somatosensory awareness has only been substantiated in nonweight bearing positions and at slower speeds of movement (Heit et al., 1996; Simoneau et al., 1997). As such, the extension of this improved joint position sense being transferred to heightened regulatory control
during higher speed locomotor functions, such as the coordination of heel strike during running, is speculative at best (Bernier & Perrin, 1998).

Another avenue examined, related to the effects of taping on proprioceptive functions, has been neuromuscular responses. One study compared premotor peroneal response time to sudden single plane inversion forces in controlled, taped, and braced conditions (Donahue, Sandrey, Kuhlman, & Edwards, 1997). Donahue and colleagues reported no differences of this neuromuscular response across any of the three conditions. Contrary to the findings of Donahue et al. (1997), Karlsson and Andreasson (1992) reported ankle taping reduced peroneus muscle reaction times during a single plane trap door inversion tilting condition in subjects with “functionally unstable ankles”. Their test methodology involving unidirectional loading forces, did not lend itself to a discussion on the effects of ankle taping on the neuromuscular regulation of impact forces.

Impact forces in walking and heel-toe running appear as a “transient peak in ground reaction force during the first .50 sec of the support phase in the cycle” (Shorten & Winslow, 1992, p. 289). Following this initial shock, a dip in vertical ground reaction force coincides with subtalar pronation (Rogers, 1995). While the calcaneal fat pad accounts for approximately 25% of the shock attenuation at heel strike (Nigg, Cole, & Bruggemann, 1995), a majority of the remaining kinetic force absorption and joint protection at this initial stage of the stance phase is afforded through coordinated neuromuscular activity (Dickenson, Cook, & Leinhardt, 1985). The timing of the muscles responding to impact forces at heel strike seems dependent upon the rate of ambulation and the inclusion or absence of a flight phase in the stride.
Beyond having a reduction in stride lengths and stride rates relative to running, the typical walking gait pattern can be characterized as having bilateral support during heel strike. Consequently, this lack of a flight phase equates to a range of 35-50% less impact force at heel strike in comparison to running (Shorten & Winslow, 1992). “Normal muscular activity” at heel strike in walking involves eccentric “pretibial group” activity (i.e., tibialis anterior) to “oppose plantar flexor moment” with the gastrosoleus functioning in eccentric fashion to “oppose dorsiflexion moment and control tibial advancement” (Norkin, 1994, p. 169). In early childhood, the respective firing of the tibialis anterior (TA) and gastrocnemius (GN) occurs as more of an inefficient cocontraction (Knutson & Soderberg, 1995). By age seven, when neuromuscular patterns in gait represent "the normal adult profile", these muscles have been described to fire in more sequential burst-like fashion at heel strike with TA activity preceding GN activity (Knutson & Soderberg, 1995, p. 308). According to Basmajian and De Luca (1985), efficient movement of this nature is the result of "learned supraspinal control and involves eliminating undesirable or useless muscle coactivation" (p. 230). It is not known if gender differences exist in the amplitude or timing of these muscles during heel strike in walking.

With the inclusion of a flight phase and posterior shear impact velocities on the calcaneus (Cailliet, 1989) occurring at rates between 1.0 to 5.0 m/s or greater dependent upon footwear (Nigg, 1985), the neuromuscular regulation of heel strike in running has received equal attention in the literature. Electromyographic (EMG) studies of running have identified the TA and GN as important “cocontractors” at heel strike (Elliott &
Blanksby, 1979; Mero & Komi, 1987). Although the TA remains active throughout the gait cycle (Reber, Perry, & Pink, 1993), heightened activity is needed in the late swing phase to pull the ankle into dorsiflexion (McClay et al., 1990). Elevations in GN activity have been reported during late swing in close proximity to heel strike (Freedman & Kent, 1987; Mero & Komi, 1987). This elevation in GN activity has been suggested to counteract excesses in ankle dorsiflexion and allow the TA to remain in a "preactivated" state (Mero & Komi, 1987). The extent of sequential TA and GN firing in quantitative values has not been reported.

Few studies have examined the effects of ankle taping on lower leg EMG activity during overground running. McKanna and Finch (1991) reported lower amplitudes in the soleus and peroneus longus for subjects running in tape. Unfortunately, temporal factors were not considered and activity of the GN was not measured. The suggestion that the practice of ankle taping has a potential of reducing EMG amplitude in muscles regulating the control of impact forces at heel strike carries with it some legitimate concerns. This information would be of interest to those regularly providing allied health care for athletes.

One phenomenon prevalent in EMG studies involving rapid movement regulation is the presence of what is termed an electrical silent period (Basmajian, 1989; Gowitske, 1984; Mita, Aoki, Tsukahara, & Yabe, 1983). While it is possible to consciously relax and suppress electrical activity in a muscle (Basmajian & De Luca, 1985), the presence of an electrical silence during rapid motions is the result of subconscious alpha motor neuron inhibition (Basmajian, 1989; Gowitzke, 1984; Mita et al., 1983). For example,
upon termination of a back swing during an upper extremity movement, an electrical period of silence in the agonist coincides with an elevation of electrical activity in the antagonist muscle(s) (Basmajian, 1989). This antagonist activity has been demonstrated to be even more marked with fast, small movements in comparison to slow, larger ones (Marsden, Obeso, & Rothwell, 1983).

Although an electrical silence can commonly occur during slower motions, “there are successive activities in the agonist and antagonists including periods of common electrical silence” among closed kinetic chain motions involving impact forces (Basmajian, 1989, p. 230). To a degree there is “a partial overlapping of phasic activities in the agonist and antagonist” (Basmajian, 1989, p. 230). Electromyographic patterns of opposing muscles following an agonist-antagonist-agonist sequence of burst activation, separated by intervening periods of electrical silence, is referred to as the triphasic burst pattern (Basmajian & De Luca, 1985; Brown & Cook, 1981; Marsden et al., 1983).

There are two areas of interest when examining biphasic (i.e., the agonist-antagonist firing with complimentary electrical periods of silence during slower speed motions such as heel strike in walking) and triphasic burst patterns in closed kinetic chain activities. The first area is a simple observation and comparison of peak amplitudes in the respective muscles being measured. Because of the extraneous electrical artifacts that can enter into the EMG sample being taken, the peak amplitude measure typically represents a peak amplitude taken from raw EMG data which has been rectified and smoothed through some form of filtering and processing procedure (Lagasse, 1987; Winter, 1979). The second area of interest, far less investigated with respect to heel
strike in running, is the time referenced presentation of amplitude (Knutson & Soderberg, 1995). These two parameters represented the dependent variables examined in this study.

**Purpose of the Study**

The purpose of this study was to investigate normalized peak TA and GN EMG amplitudes and the time to peak TA and GN EMG amplitudes relative to heel strike during closed ankle taped and untaped walking and running conditions. As a secondary purpose, this study assessed gender differences in normalized peak TA and GN EMG amplitudes and the time to peak TA and GN EMG amplitudes relative to heel strike during untaped walking and running conditions.

**Statement of the Problem**

The comprehensive protection of competitive athletes often includes the routine application of ankle taping. The lack of research examining the interaction between ankle taping and the neuromuscular control of impact forces occurring at heel strike raises concerns as to the appropriateness of this practice, particularly with healthy athletes. It is not known whether mechanically restricting motions of the foot and ankle with tape has an adverse effect on the neuromuscular regulation of impact forces experienced at heel strike.

**Need for the Study**

Research in several different areas has collectively posed convincing arguments to support the use of ankle taping. Much of this work has been centered on examining forces and neuromuscular or proprioceptive capabilities during midstance of locomotor activities. The link between impact forces and the development of chronic joint problems
of the lower extremity has been an area of considerable debate for those who study the design of athletic footwear (Bates, 1985; De Wit, De Clercq, & Lenoir, 1995; Nigg et al., 1995; Stacoff, Denoth, Kaelin, & Stuessi, 1988). Certainly this same careful attention is needed with respect to the use of ankle taping.

**Null Hypotheses**

1. There are no gender differences in peak and time to peak TA or peak and time to peak GN EMG amplitude relative to heel strike during untaped walking.

2. There are no differences in peak and time to peak TA or peak and time to peak GN EMG amplitude relative to heel strike between taped and untaped walking conditions.

3. There are no gender differences in peak or time to peak TA or peak and time to peak GN EMG amplitude relative to heel strike during untaped running. There are no gender differences in peak or time to peak for a second peak TA EMG amplitude (TA2) relative to heel strike during untaped running.

4. There are no differences in peak or time to peak TA and peak and time to peak GN EMG amplitude relative to heel strike between taped and untaped running conditions. There are no differences in peak or time \( \gamma \) peak TA2 EMG amplitude relative to heel strike between taped and untaped running conditions.

**Assumptions**

1. The TA and GN muscles coordinate a biphasic burst pattern at heel strike during the walking gait cycle.

2. The TA and GN muscles coordinate a triphasic burst pattern at heel strike during higher speed heel-toe gait patterns.
3. Peak EMG amplitude and time to peak EMG amplitude relative to the time at heel strike during ambulation is a valid means of evaluating the neuromuscular response to impact forces at heel strike.

Limitations and Delimitations

1. Only healthy, volunteer subjects were used in this study.
2. The GN and TA were the only surface EMG measures obtained. No other kinetic data were collected in this study.
3. No kinematic measures were examined in this study.
4. This study did not examine beyond heel strike walking or running gait patterns.
5. This study did not examine the effects of other bracing materials.
6. This study did not examine the EMG activity of the GN and TA at heel strike following a specified training period.
7. Walking speeds for all subjects were standardized to a range of 1.45 to 1.75 m/s.
8. Running speeds for all subjects were standardized to a range of 2.61 to 2.90 m/s.
9. Upper limits of percentage maximum voluntary isometric torque were not quantitatively regulated during the dynamometer testing.
10. Only a single trial, at an indicated speed, was recorded for each of the four experimental conditions.
Definition of Terms

**Braking Force** - a culmination of vertical and anteroposterior forces regulated in the stance phase of gait.

**Closed Ankle Taping Technique** - a nonelastic adhesive taping pattern having two anchors, three stirrups, two horseshoes, two medial heel locks, two lateral heel locks, and two figure eights (Arnheim & Prentice, 1997).

**Cocontraction** - the contraction of an antagonist muscle occurring at the same time as an agonist muscle contraction (Sale, 1988).

**Impact Force** - the sum of mass and acceleration imparted on a surface transmitting equal and opposite forces through the structure or structures which are generating the motion.

**Leg Dominance** - motor preference of one leg over another typically self-defined by asking a person to state which limb they would use to kick a soccer ball the longest distance. In this situation, the nondominant limb would be considered the support leg with the kicking leg being considered the dominant limb (Tanaka et al., 1996).

**Propulsive Force** - the reversal of braking forces in the stance phase of gait.

**Triphasic Burst Pattern** - firing pattern of opposing muscles characterized by an initial burst of agonist activity in the presence of temporary antagonist electrical silence, a regulatory antagonist activity in the presence of temporary agonist silence, and a subsequent reactivation of agonist activity in the presence of temporary antagonist silence (Basmajian & De Luca, 1985).
CHAPTER II

REVIEW OF RELATED LITERATURE

Introduction

The regulation of locomotor activities is as much an involuntary process as it is a voluntary one. While the motor cortex "sends the command signals which sets the body in motion, there is no direct neuronal circuit in the brain that dictates the specific bipedal leg movements characteristic of human locomotion" (Guyton & Hall, 1996, p. 685). Winter (1983) studied "EMG profiles" in healthy subjects walking and suggested the timing of movement patterns in locomotion are under the guidance of "closed-loop feedback control requiring continuous peripheral neural input" (p. 325). When motions normally found in locomotor activities are restricted with ankle taping, there may be a potential to alter the nonvoluntary neuromuscular functions that help facilitate efficient movement and kinetic attenuation of impact forces. To examine this potential, this review focused on three primary areas. First, a survey of the research examining the effects of ankle taping on the regulation of neuromuscular functions is provided. Second, the literature on various mechanisms behind the neuromuscular regulation of dynamic joint stability to loading and impact forces is examined. Finally, the challenge behind establishing the speed of ambulation in EMG gait research is discussed.
Effects of Mechanical Support on Neuromuscular Function and Restriction of Joint Motions

In examining injury rates relative to athletic exposures, retrospective studies have supported the use of athletic taping in the prevention of ankle injuries (Garrick & Requa, 1973; Rovere, Clarke, Yates, & Burley, 1988). Several studies have evaluated the restrictive properties of ankle taping in response to physical activity. Metcalfe et al. (1997) reported shorter vertical jump heights and slower run times for the Southern Missouri agility test in subjects wearing tape. While these performance decrements were observed, a postassessment via standard goniometry revealed ankle taping reduced excesses of dorsiflexion, inversion, and eversion, but not plantar flexion (Metcalfe et al., 1997). In a similar study design using 2.5 to 3.0 hours of football practice as the independent variable, Fumich et al. (1981) reported athletic tape maintained 50% of its average residual restriction to motions of inversion, eversion, and plantarflexion/inversion. Laughman et al. (1980) collected electrogoniometric measures of three dimensional ROM for 20 subjects who engaged in conditions of untaped walking, taped walking, and taped walking following the completion of a 5.0 by 10.0 m agility course. While taping reduced multiple measures of ROM during walking, the authors reported ankle taping significantly restricted motions of plantar flexion and inversion even after subjects completed their agility course. Unfortunately, electrogoniometry was not used to examine the extent of this restriction following an agility course during running trials.

Along with the restrictive properties of ankle taping, it has been suggested that ankle taping can facilitate the neuromuscular response of key muscles about the ankle thereby improving dynamic joint stability (Karlsson & Andreasson, 1992; Konradsen &
Hojsgaard, 1993). Identified by excesses in anterior talar translation and talar tilting radiographically, Karlsson, Peterson, Andreasson, and Hogfors (1992) reported individuals with “functionally unstable ankles” had delayed peroneus longus and brevis reflex times during trap door tilting 30° from the horizontal plane. Using the same subject base, Karlsson and Andreasson (1992) later reported ankle taping helped shorten this delay of peroneus muscle reaction time in response to this same trap door tilting. Unfortunately, this provided information only pertinent to loaded, rather than impact force conditions, and the authors utilized a relatively uncommon ankle taping technique that did not include the application of heel locks (Arnheim & Prentice, 1997).

The neuromuscular response of the peroneus longus has been examined in ankle taped running conditions. Based on a finding of earlier EMG peroneal muscle activity at “pre-heel-strike” during taped running, Konradsen and Hojsgaard (1993) suggested a benefit derived from external ankle support may be one of potentially improved dynamic joint stability. McKanna and Finch (1991) however, investigated peroneal EMG activity in 10 subjects during ankle taped jogging before and after a 30 min period of supervised basketball drills. In both pretest and posttest measures, they reported ankle taping during running resulted in lower peak integrated EMG peroneal amplitudes. The authors concluded this depression of muscle amplitude could be a potential mechanism for disuse atrophy as a result of ankle taping.

Measures of joint position sense are among the more recent areas explored with respect to the efficacy of ankle taping and bracing. Position sense studies have not demonstrated heightened awareness in the weight bearing position (Heit et al., 1996;
Simoneau et al., 1997). Simoneau et al. (1997) compared joint position perception of the ankle with and without a strip of adhesive tape applied to the lower leg in standing and supine positions. They reported joint position perception was significantly improved in the taped nonweight bearing position. Based on this finding, they suggested that taping provides "added proprioceptive information that could possibly help in the proper positioning of the ankle just prior to contact during running" (p. 146). How this improved joint position perception relates to neuromuscular responses prior to or upon impact during running was not addressed.

The effects of external ankle support on postural control has also been an area receiving recent attention (Holcomb, Kleiner, & Miller, 1998; Kinzey, Ingersoll, & Knight, 1997; Kleiner & Holcomb, 1998). Kinzey et al. (1997) examined center of pressure measures for 24 healthy males in unsupported and ankle braced conditions. They concluded their comparisons "did not support or refute the concept that bracing enhances proprioception" (p. 300). More recently however, the effects of short term and long term ankle bracing on static and dynamic measures of balance have been examined (Holcomb et al., 1998; Kleiner & Holcomb, 1998). Using measures acquired on the Chattex balance system, Kleiner and Holcomb (1998) reported a 24 hr period of ankle bracing had no significant effect on static or dynamic balance in 11 healthy male subjects. Using 39 male and female subjects, Holcomb and colleagues (1998) reported these same findings following a long term ankle bracing period of 7 weeks. As with the studies on joint position sense, postural sway assessment provides information regarding
proprioception during slower movements and does not address the effects of ankle taping on the neuromuscular regulation of impact forces.

**Impact Forces and Their Ties to Injury**

Alteration of support for the foot and ankle is among the many links associated with the development of chronic musculoskeletal problems afflicting the lower extremity (Nigg et al., 1995). Support modifications, such as ankle taping, that alter rearfoot motion can potentially suppress the neuromuscular regulation of impact forces leading to a diverse spectrum of short term and long term problems. Related to the anteroposterior shear forces at heel strike, Cailliet (1989) proposed that the proximal fibers of the plantar fascia are subjected to greater tensile loading leading to the development of plantar fasciitis. As such, he recommended treatment for plantar fasciitis be “directed toward alleviating situations...where tension is placed on the calcaneus during weight bearing” (p. 139).

The literature, while lacking the mechanism of suppressed neuromuscular responses to footwear alteration, is consistent with suggestions of inappropriate heel strike attenuation with chronic wearing and degenerative changes involving weight bearing joints of the lower extremity. Several authors (Dickenson et al., 1985; Nigg et al., 1995; Rogers, 1995; Shorten & Winslow, 1992) have inferred that poor regulation of impact forces at heel strike may be a primary injury mechanism for developing osteoarthritis of the knee and/or hip joints in runners. In these cases, it is not uncommon for the individual to have an extended history of training with nonspecific articular pain (Nigg et al., 1995).
Sensorimotor Regulation in Locomotor Activities

Role of the Stretch-Shortening Cycle

The reflexive stretch-shortening cycle (SSC) is a regular part of the neuromuscular control during running (McClay et al., 1990; Mero & Komi, 1987; Osternig et al., 1986). In its most basic form, the SSC can be described as a reflexive concentric shortening of a muscle preceded by a rapid eccentric stimulus (Wilk et al., 1993). Although a similar myotatic stretch reflex is apparent in the regulation of postural sway by means of the "positive support reaction" (Spence & Mason, 1992, p. 480), use of the above definition seems more applicable in describing the role of the SSC in locomotor activities. For example, the soleus muscle during mid-stance of running, is eccentrically active when decelerating the shank over the foot, and concentrically active when lifting the heel prior to toe off (McClay et al., 1990). The greater the eccentric stimulus (i.e., greater talocrural joint moments during deceleration of the shank), the more rapid and responsive the cross sectional recruitment during the concentric shortening phase (Gowitzke & Milner, 1988). While this provides a good example of how an isolated muscle utilizes the stretch reflex during locomotion, the effects and role of the SSC in collective neuromuscular functions, particularly in regulating impact forces, is a topic requiring a more detailed description.

Coordination of the SSC in a single muscle through a myotatic stretch reflex involves concurrent neural activity with synapses servicing other muscles. Collateral synapses at the level of the spinal cord also send postsynaptic potentials to axons of alpha motor neurons innervating synergist muscles (Gowitzke & Milner, 1988), while sending inhibitory postsynaptic potentials to alpha motor neurons of the agonist muscles (Eldred,
1965; Gowitzke & Milner, 1988). This classic excitation of agonist and reciprocal inhibition of antagonist muscles, if taken out of context, can lead to the generation of misleading kinetic profiles for the activity being observed. For example, if an EMG study examining a single muscle concludes that the muscle experiences a reduced electrical amplitude, or in the extreme case, experiences a period of electrical silence when supported with a prophylactic device, it is not known whether the device simply reduces the need for that muscle to function or creates a situation which promotes a SSC of the antagonist muscle. To examine the effects of a support device or technique such as ankle taping in weight bearing situations with EMG, it seems prudent that the experimental and baseline measures should attempt to capture the true kinetic profile with respect to involuntary and voluntary functions.

The Biphasic Burst Pattern

As stated previously, the extent of excitation and reciprocal inhibition is dictated by the rate (i.e., tonic vs. phasic) and amplitude (short vs. damped motion) of the eccentric stimulus. Regarding what is known about the loading rates during walking, as well the influence of weight distribution over two bases of support at impact, the neuromuscular regulation of heel strike in this situation appears to occur through sequential activation of the TA and GN (Knutson & Soderberg, 1995). The role of the SSC in this biphasic burst pattern may occur at a rate and amplitude greater than the positive support reaction but far less than running. In considering the role of the SSC, the biphasic burst pattern described by Knutson and Soderberg (1995) can certainly fit into the model described by Norkin (1994). Restating Norkin's (1994) model at heel strike during walking, a graded
TA SSC with GN inhibition opposes plantar flexor moment. Sequentially, GN inhibition is overridden with an eccentric stimulus and responds efficiently with a braking SSC in the presence of TA silence. Continued GN activation and TA inhibition presumably occurs during tibial advancement along the same time frame as subtalar pronation and dipping of the initial vertical ground reaction force (Rogers, 1995).

**The Triphasic Burst Pattern**

The role of the SSC during ballistic limb movements takes on a whole new meaning from that of the isolated agonist excitation, antagonist inhibition phenomenon. Studies examining ballistic movements of the upper extremity describe the presence of a triphasic burst pattern (Brown & Cooke, 1981; Marsden et al., 1983). Regulation of this pattern by the muscle spindle first involves the rapid activation of an agonist which overrides reciprocal inhibition of the agonist invoking a regulatory SSC effect. This regulatory activation of the agonist overrides reciprocal inhibition of the antagonist triggering a SSC reaction of the original prime mover. While this pattern was supported to be a function of accuracy for ballistic movements (Marsden et al., 1983), Brown and Cooke (1981) reported this pattern as even move prevalent for subjects instructed to simply move “as fast as possible” (p. 97).

Taking into consideration the ballistic nature of forces experienced at early stance during higher speed heel-toe gait patterns, as well as the research supporting the presence of early GN and TA EMG activity occurring during this event (Elliott & Blanksby, 1979; Mero & Komi, 1987), deductive reasoning would lead one to suspect that the neuromuscular regulation of impact forces at heel strike during running may occur by
A careful survey of the literature has yielded that this may actually be the case. The raw EMG figure provided in a study examining heel strike during various trials involving step downs from preselected heights, revealed this pattern (Freedman & Kent, 1987). To avoid false interpretations of heightened or reduced EMG amplitude (i.e., electrical silence) in response to ankle taping during locomotion, the assumption that sequential TA and GN burst patterns occur relative to the time at heel strike formed the basis for the walking and running EMG profiles used in this study.

Challenges in Establishing the Speed of Ambulation in Electromyographic Gait Research

During any type of gait research, it is necessary to standardize the speed of walking and running so that speed is not a confounding variable. For example, speed of ambulation will greatly affect the results of ground reaction force studies given that force is the product of mass and acceleration according to Newton's Second law. If the variable being examined is force and mass is held at a constant, then the component of acceleration requires careful standardization. Along similar lines, the collection of accurate EMG measures requires this regulation of speed. For example, ankle taped subjects moving at a rate of 5 m/s would have greater muscular activity at heel strike versus ankle taped subjects moving at a rate of 1 m/s simply because of the difference of moving rate. Conversely, requiring all subjects to walk or run at a single pace could possibly alter one's natural gait with respect to stride length or stride rate. Changing a person's natural stride length or rate creates a situation where motor unit recruitment is potentially altered leading to the generation of unnatural EMG profiles (Winter, 1983). As such, the methodology for EMG gait studies should attempt to minimize the ranges of
moving rate in ambulation as a confounding variable, while at the same time, allow
enough latitude in the ambulation rate to accommodate individual and gender differences
within the group being researched.

According to the results generated by Winter (1987), the typical cadence for walking falls between a range of 80 to 120 foot strikes per min. While it is possible to standardize overground walking pace using a metronome, such a single cadence creates the problem of potentially compromising the natural gait pattern for many individuals, therefore, an approach in standardizing the velocity of ambulation may be more appropriate. Craik and Dutterer (1995) surveyed the gait literature on walking velocities for males and females and provided a thorough table of norms in their section of a text edited by Craik and Oatis (1995). Calculating the means presented in their table, males (not instructed to walk at a specific speed) had an average walking velocity of 1.18 m/s (2.96 mph) and females (not instructed to walk at a specific speed) had an average walking velocity of 1.12 m/s (2.56 mph). As with setting a fixed cadence, it is possible to standardize the velocity of ambulation using a treadmill, however, as seen with the gender differences in walking velocity, a single treadmill speed may accommodate the natural walking velocity for one gender but not the other. This again poses the potential of disrupting natural gait patterns (i.e., males walking at the average reported walking velocity for females).

Summary

Beyond restricting extremes of joint ROM, there is some indication that the taping of ankles may improve certain proprioceptive functions including joint position sense. Much of the research investigating the effects of ankle taping on neuromuscular
responses (i.e., reaction time) has either examined EMG measures during loaded stationary conditions or confined their assessment to single muscles or synergistic muscles (i.e., peroneus longus and peroneus brevis). Based on what has been described regarding the role of the SSC in locomotion, as well the presence of electrical silence during reciprocal inhibition, it seems prudent that EMG assessments of ankle taping during impact force conditions should focus on opposing muscles. The TA and GN appear to be the primary opposing muscles functioning during heel strike. To assess the EMG of these muscles during a subject’s natural gait pattern, velocity ranges of ambulation that accommodate individual and gender differences should be employed in the research methodology.
CHAPTER III

METHODS AND PROCEDURES

Introduction

This study investigated the EMG amplitude and temporal responses of the TA and GN occurring relative to heel strike during closed ankle taped and untaped walking and running conditions. Of the various gait patterns existing, this study was limited to assessing heel-toe gait patterns for this style is representative of approximately 81% of runners and joggers (Nigg et al., 1995). Amplitude assessments of the various conditions involved comparisons of peak TA and GN EMG amplitudes normalized to EMG values obtained during a 75% maximum voluntary isometric contraction (MVIC). Temporal assessments involved comparing the time to peak TA and time to peak GN EMG amplitudes relative to the time at heel strike as indicated by a foot switch directly affixed to the subject’s heel. Gender comparisons were made using normalized amplitude and temporal data gathered for the untaped conditions.

Subject Selection

Subjects (N = 20) utilized in this study were recruited from the general student population at the University of Wisconsin-La Crosse. Prior to subject selection, the study’s procedures and data collection process underwent a preliminary departmental screening and received an expedited approval by the University of Wisconsin-La Crosse Institutional Review Board. Each subject gave their informed consent (see Appendix A) to participate in this study. All subjects had no previous history of lower extremity
surgery and had not experienced any musculoskeletal foot, ankle, knee, or hip injuries over the past year. Prior to inclusion in the study, all subjects received a brief static postural and gait examination to screen out individuals with abnormal findings such as pes planus or excessive rearfoot inversion during ambulation. Identification of the dominant lower limb was done by having each subject indicate which leg they would prefer to kick a soccer ball the furthest distance (Tanaka et al., 1996). Limb dominance, as well as other standard characteristics, were recorded for each subject and are summarized in Table 1.

Table 1. Subject Characteristics (N = 20)

<table>
<thead>
<tr>
<th>Gender</th>
<th>N</th>
<th>Age Mean ± SD</th>
<th>Ht (cm) Mean ± SD</th>
<th>Wt (kg) Mean ± SD</th>
<th>Ratio of (R): (L) Leg Dominance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>10</td>
<td>22.9 ± 5.5</td>
<td>180.2 ± 5.6</td>
<td>81.8 ± 11.5</td>
<td>3: 2</td>
</tr>
<tr>
<td>Female</td>
<td>10</td>
<td>21.7 ± 1.8</td>
<td>162.8 ± 4.7</td>
<td>62.1 ± 8.3</td>
<td>4: 1</td>
</tr>
</tbody>
</table>

Methods and Procedures

Taping Procedure

This study utilized a closed ankle taping technique that is commonly taught to and used by athletic trainers (Arnheim & Prentice, 1997). The technique included 1 ½ inch nonelastic adhesive tape (Johnson & Johnson), heel/lace pads, and underwrapping as materials. Each subject (N = 20) was taped by the same NATA certified athletic trainer.
to eliminate inter-application variations. The technique, upon application of the pads and underwrapping, began with anchors applied circumferentially distal to the belly of the GN and directly over the base of the fifth metatarsal. A pattern of three to four stirrups, dependent upon the size of the subject’s limb, were then applied. Coverage of the foot and ankle, within the borders of the anchors, included two figure eights, two medial heel locks, and two lateral heel locks. Each heel lock opposed each other on the calcaneus in the frontal plane in an attempt to have a neutral restrictive effect on rearfoot motion (Arnheim & Prentice, 1997).

Electromyographic Procedures

The muscles targeted for evaluation were the unipennate TA and the medial head of the bipennate GN. The medial GN was used in this study because it “meets the initial demands for increased force during walking…and…performs at a greater percentage of its maximum capacity” during higher loading activities (Gregor, 1993, p. 197). Several measures were taken to reduce measurement error. Data analysis including normalization, amplification, filtering, and processing of the EMG collected in this study were conducted in accordance with the guidelines established by the International Society of Electrophysiological Kinesiology (ISEK).

Electrode placement. A portable transcutaneous electrical nerve stimulator (TENS) unit (Iomed, Inc.) was used to locate the motor points for the EMG electrode placements. The procedures for locating the desired motor points first involved circulating two, 3.5 by 4.5 cm leads, lubricated with a water based gel, and gradually increasing the current’s amplitude in order to illicit an electrical paresthetic response. The pulse rate was set at
120 Hz and the pulse width was adjusted between 180 and 260 msec according to the subject's comfort to the electrical paresthesia. The amplitude was then increased while circulating the leads until finding perceptible motor response areas, or motor points, in the TA and medial GN muscles respectively. Upon location of each motor point, a marking was made with a felt tipped pen and any excess body hair was removed with an electric razor.

**Electromyographic data sampling procedures.** The Ariel Performance Analysis System (APAS) was utilized to collect and analyze all EMG measures. Preamplified (see Appendix B) rectangular bar leads (5 cm), containing two bipolar surface (silver) electrodes spaced around a reference electrode (each 1.5 cm in diameter) were placed over the marked motor points in parallel fashion to the fiber pennation of the muscles being measured. The electrodes were affixed to the skin, wires situated proximally, with Dermaclear tape for the duration of the data collection process.

All raw EMG was sampled within a band pass filter of 8 Hz to 28 kHz. Data were filtered at a low pass cutoff frequency of 500.00 Hz. The sampling time, initiated by either EMG lead at a pretrigger percentage of 25 (trigger set at 1.000 mV), was set at 3.00 sec. Following a full wave rectification, the EMG was smoothed at a moving average of .100 sec. These procedures for treating the raw EMG were utilized for the measures taken in the four experimental conditions as well as for the EMG amplitude measures used in the normalization process.
**Normalization procedures.** According to a recent review article of surface EMG by De Luca (1997), recordings of EMG amplitude measures at a full percentage of maximum voluntary contractions are "exceptionally unstable and do not provide a suitable reference point" (p. 154). As such, he recommended EMG reference amplitudes used for normalization be sampled during a percentage of maximal contraction below 80% of maximum. To follow such procedures using a manual muscle test would only introduce a new source of error. This error primarily extends from the subjectivity involved in this method of assessment. In general, the grading of muscle strength manually requires a high level of skill and is limited to the experience of the person administering the test (Kendall, McCreary, & Provance, 1993). Therefore, this study established the percentage of maximum contraction using an isokinetic dynamometer (Cybex 340).

Several factors were considered when establishing the test position for determining the MVIC and percent MVIC torque values. While a test position reflecting similar angles of the hip, knee, and ankle flexion at heel strike would be ideal from a specificity standpoint, this position lends itself to the potential of accessory muscles influencing torque values (i.e., less isolation of the muscle being tested). Additionally, this position because of the placement of cushions, strapping, etc. on the universal body exercise table (UBXT) of the Cybex System, could place compressive forces on the EMG leads during sampling which may introduce artifacts that could contaminate the raw EMG being collected. The position selected in this study involved placing the subject supine on the UBXT with the lower extremity rested at 55° knee flexion over the popliteal pad and the
ankle positioned neutral (0°) resting the plantar surface of the foot against the foot plate of the dynamometer. These relative measures of joint ROM were confirmed with a goniometer prior to strapping the foot and ankle firmly to the dynamometer and the thigh and waist to the UBXT.

To gather the MVIC for the GN muscle, the subjects were given three trials in which they were instructed to plantar flex their ankle as hard as possible. A test session of 7.0 sec with a rest period of 5.0 sec was used on the dynamometer. These time parameters were also used for the TA. In the three test trials for the TA, the subjects were instructed to dorsiflex their ankle as hard as possible. The highest torque value among each of the three trials, for each muscle, was taken and multiplied by .75 to establish the 75% MVIC values that were used for normalizing the EMG measures taken during the four experimental trials. The subjects were given a 15 sec period to reach a sustained corresponding percentage of maximum torque (75%) as indicated on the dynamometer's visual display monitor. Once the subjects maintained the desired level of 75% maximum torque, a three sec EMG sample was taken and stored on the APAS computer.

**Synchronization of heel strike with the EMG activity during the four test conditions.** To serve as a time marker for EMG activity occurring prior to and following heel strike, a foot switch (Motion Analysis Systems) was placed over the plantar surface of the heel covering an area 3.0 cm in diameter. The foot switch was situated proximally and its wiring was modified to enable interfacing with an open channel on the A/D board of the APAS via a BNC cable. Upon heel strike, 10 V were recorded concurrently with the EMG record.
Data generated for analysis. Following processing of the raw EMG, an amplitude measure was recorded for each of the reference samples taken during the 75% MVIC tests. In each of the walking conditions, peak EMG amplitudes from the processed data (Y values), were recorded for the TA and GN. Also, the temporal values (X values) for the start of heel strike, peak TA (same point identified in the Y value), and peak GN (same point identified in the Y value) were determined (see Appendix C) and recorded on a subject data sheet (see Appendix D). The peak amplitude (expressed in mV) for the TA was normalized to the mV recorded in the 75% MVIC test by dividing its value over the 75% MVIC value and multiplying by 100. The peak GN was normalized to its 75% MVIC measure in similar fashion. For temporal measures, the time to peak TA and the time to peak GN were calculated by subtracting the heel strike time. All values for running were recorded and calculated in similar fashion. The exception for running was a second peak TA value appearing after a period of TA electrical depression or silence and in sequence to the peak GN burst which occurred on a separate channel, when applicable.

Testing Conditions

The four test conditions of untaped walking, taped walking, untaped running, and taped running used in this study were conducted on a wooden runway measuring 180 cm by 580 cm. Two strips of white tape were used to mark off a 3.05 m (10 ft) distance. For the walking trials, the subjects were required to cover the specified distance within a time of 2.27 to 2.73 sec. For the running trials, the time was between 1.36 and 1.51 sec (see Table 2 for time frame calculations). These times for walking and running were
calculated from walking (1.45 to 1.47 m/s) and running (2.61 to 2.90 m/s) velocity ranges that attempted to accommodate for individual and gender differences. Once the subject was able to cover the marked distance within the specified time frame, the next test condition was collected. A total of four test conditions with one trial per condition were collected. No athletic shoes or other form of footwear were worn in any of the four conditions.

Table 2. Formula for Calculating Time Frames

1. If mean velocity \( v = \frac{\text{displacement (d)}}{\text{time (t)}} \) or \( t = \frac{d}{v} \)

2. Then time can be solved for by:

\[
t = \frac{10 \text{ ft}}{(X \cdot 5280 \text{ ft mi}^{-1})/3600 \text{ sec hr}^{-1}} \quad \text{or} \quad t = 10 \text{ ft} / X \cdot 1.467
\]

Randomization of test order. In an attempt to remove within subject variability regarding athletic taping (i.e., removing and reapplying tape with subtle changes in tape pattern and tension), only the order of test speeds were randomized. Essentially, each subject following the collection of MVIC EMG measures on the dynamometer, selected one of eight cards which determined whether they started with or without tape and in what order of speeds they would engage in the experimental trials (see Table 3 for condition orders).

Regulation of walking and running speeds. A photoelectric timing device was used to verify all walking and running times. Two photocells (Cutler Hammer, Inc.), each aimed at reflectors across the width of the runway, were clipped to poles set at the start and finish of the marked off 3.05 m walkway. Wires from the photocells were interfaced
with a digital clock (KEPtron). One photocell started the digital clock while a second stopped the clock expressing values to the nearest .01 sec. To prevent premature triggering of the photocells by the subject’s arms, the photocells were set to the trochanteric height of each subject. Additionally, the EMG and foot switch wires were fastened with Velcro straps to the subject’s waist and thigh to minimize this means of artificially setting on or off the digital clock. During testing, subjects were instructed to continue walking or running beyond the photocell at the end of the 3.05 m distance.

Table 3. Matrix of Speed Condition Order

<table>
<thead>
<tr>
<th>1. WUT, RUT, WT, RT</th>
<th>5. WUT, RUT, RT, WT</th>
</tr>
</thead>
<tbody>
<tr>
<td>2. RUT, WUT, WT, RT</td>
<td>6. RUT, WUT, RT, WT</td>
</tr>
<tr>
<td>3. WT, RT, WUT, RUT</td>
<td>7. WT, RT, RUT, WUT</td>
</tr>
<tr>
<td>4. RT, WT, WUT, RUT</td>
<td>8. RT, WT, WUT, RUT</td>
</tr>
</tbody>
</table>

WUT = walk untaped  RUT = run untaped  WT = walk taped  RT = run taped
Statistical Treatment

Untaped Walking

A gender time to peak TA and time to peak GN (referenced to heel strike) comparison in the walking without tape condition was made using a two-way ANOVA. Significant main effects or interaction were explored using Tukey’s post hoc test. A gender peak TA and peak GN (normalized to 75% MVIC EMG mV values) comparison in the walking without tape condition was made using a two-way ANOVA. Significant main effects or interaction were explored using Tukey’s post hoc test. All alpha levels were set at .05.

Untaped Running

A gender time to peak for an initial TA burst (TA1), time to peak GN, and time to peak for a second TA burst (TA2) comparison in the running without tape condition was made using a two-way ANOVA. Significant main effects or interaction were explored using Tukey’s post hoc test. A gender peak TA1, GN, and peak TA2 comparison in untaped running was made using a two-way ANOVA. Significant main effects or interaction were explored using Tukey’s post hoc test. All alpha levels were set at .05.

Taped Walking

The time to peak TA and time to peak GN was compared for untaped and taped walking using a two-way ANOVA with repeated measures. Significant main effects or interaction were explored using Tukey’s post hoc test. The peak TA and the peak GN was compared for untaped and taped walking using a two-way ANOVA with repeated measures. Significant main effects or interaction were explored using Tukey’s post hoc test. All alpha levels were set at .05.
**Taped Running**

The time to peak TA1, time to peak GN, and time to peak TA2 was compared for untaped and taped running using a two way ANOVA with repeated measures. Significant main effects or interaction were explored using Tukey's post hoc test. The peak TA1, GN, and peak TA2 for untaped and taped running was compared using a two-way ANOVA with repeated measures. Significant main effects or interaction were explored using Tukey's post hoc test. All alpha levels were set at .05.
CHAPTER IV
RESULTS AND DISCUSSION

Introduction

This study’s procedures generated TA and GN EMG data for 10 male and 10 female volunteers who engaged in trials randomized by speed (see Table 2) of barefoot walking, barefoot closed ankle tape walking, barefoot running, and barefoot closed ankle tape running. Following full wave rectification and moving average processing of the EMG data, the time at heel strike (calculated by the APAS to the nearest .001 sec) along with peak and time to peak TA, as well as peak and time to peak GN EMG amplitude, were recorded on a data sheet (see Appendix D) for each subject. In cases where a triphasic burst pattern was present (see Appendix C), values were recorded for peak and time to peak TA1, GN, and peak and time to peak TA2 EMG amplitudes.

All time to peak EMG amplitude measures generated in the descriptive statistics were calculated by taking the recorded time of peak EMG (mV) and subtracting the time recorded for heel strike. These time referenced data were used when examining the temporal patterns with inferential statistics. All amplitude measures used in statistical comparisons were normalized to reference EMG amplitudes (mV) collected during a 75% MVIC. Gender and ankle taping represented the study’s independent variables. Normalized EMG amplitudes (i.e., a percentage of mV measured during a 75% MVIC) and time to peak amplitudes referenced to the time at heel strike served as the independent variables.
Results

Temporal Patterns in the Untaped Conditions

Walking. As depicted in Table 4, the mean time to peak TA following heel strike in untaped walking was .004 sec and the mean time to peak GN was .160 sec. A two-way ANOVA was used to compare the time to peak TA and the time to peak GN EMG amplitude during the untaped walking condition across genders. There was a significant (p < .05) main effect for the time to peak values. Tukey’s post hoc test was used to compare pairwise differences. In both genders, the time to peak TA was significantly (p < .05) faster than the time to peak GN.

Table 4. Temporal Activation of the Tibialis Anterior (TA) and Gastrocnemius (GN) During Untaped Walking

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Time To Peak TA(^a) (sec) Mean ± SD</th>
<th>Time To Peak GN(^b) (sec) Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>10</td>
<td>.004 ± .004*</td>
<td>.160 ± .192</td>
</tr>
<tr>
<td>Female</td>
<td>10</td>
<td>.005 ± .005*</td>
<td>.296 ± .154</td>
</tr>
<tr>
<td>Total</td>
<td>20</td>
<td>.004 ± .005*</td>
<td>.228 ± .183</td>
</tr>
</tbody>
</table>

* Significance at the .05 level

\(^a\)Time referenced to peak amplitude by subtracting heel strike from peak TA.

\(^b\)Time referenced to peak amplitude by subtracting heel strike from peak GN.
Running. The raw EMG during untaped running in all subjects revealed the presence of a triphasic burst pattern at heel strike (see Appendix E). As with the untaped walking, all temporal data were recorded on a data sheet (see Appendix D) and their respective descriptive statistics were calculated (see Table 5).

Table 5. Temporal Activation of an Initial Tibialis Anterior Burst (TA1), a Gastrocnemius Burst (GN), and a Second Tibialis Anterior Burst (TA2) During Untaped Running

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Time to Peak TA1 (sec) Mean ± SD</th>
<th>Time to Peak GN (sec) Mean ± SD</th>
<th>Time to Peak TA2 (sec) Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>10</td>
<td>-.033 ± .093*</td>
<td>.086 ± .034ψ</td>
<td>.191 ± .099</td>
</tr>
<tr>
<td>Female</td>
<td>10</td>
<td>-.022 ± .047*</td>
<td>.089 ± .049ψ</td>
<td>.270 ± .281</td>
</tr>
<tr>
<td>Total</td>
<td>20</td>
<td>-.027 ± .074*</td>
<td>.088 ± .041ψ</td>
<td>.230 ± .209</td>
</tr>
</tbody>
</table>

* Significant (p < .05) from the time to peak TA2.
ψ Significant (p < .05) from the time to peak TA only.

Consistent with previously reported findings, initiation and time to peak TA EMG activity occurred prior to heel strike as indicated by a negative mean value for TA1 in Table 5. A two-way ANOVA was used to compare time to peak TA1, time to peak GN, and time to peak TA2 EMG amplitude during the untaped running condition across genders. There was a significant (p < .05) main effect for the time to peak values.
Tukey's post hoc test was used to compare pairwise differences. Across gender, there was a significant ($p < .05$) difference between the time to peak TA1 and time to peak GN, and between the time to peak TA1 and time to peak TA2. However, there was no significant ($p > .05$) difference between the time to peak GN and the time to peak TA2 (see Figure 1).

Figure 1. Time to Peak EMG Amplitude for an Initial Burst of the Tibialis Anterior (TA1), the Gastrocnemius (GN), and a Subsequent Burst of the Tibialis Anterior (TA2) During Untaped Running.

**Normalized Amplitudes in the Untaped Conditions**

All TA and GN amplitudes were normalized to a 75% MVIC. The normalized values represent percentage values expressed to the nearest whole number. As with the time
referenced data, a sequential TA1, GN, and TA2 indicates a triphasic burst pattern that has been normalized in amplitude.

**Walking.** In the walking without tape condition, the mV percentage of 75% MVIC for the TA ranged from 14 to 7100% ($M = 874, SD \pm 1761$) where the GN ranged from 39 to 829% ($M = 309, SD \pm 240$). Using the entire data set ($N = 20$), a gender TA and separate gender GN comparison failed Levene’s Test of Equality of Variances. It must be noted this variability did not exist between the EMG values collected during the MVIC tests. The removal of seven outliers (see Table 6 for the adjusted walking amplitude data) was necessary to restore homogeneity among the amplitude data (i.e., a p value for Levene’s

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Peak TA$^a$ (%) mV Mean ± SD</th>
<th>Peak GN$^b$ (%) mV Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>7</td>
<td>58 ± 45</td>
<td>260 ± 237</td>
</tr>
<tr>
<td>Female</td>
<td>6</td>
<td>241 ± 155*</td>
<td>265 ± 222</td>
</tr>
<tr>
<td>Total (untaped)</td>
<td>13</td>
<td>142 ± 146</td>
<td>262 ± 195</td>
</tr>
<tr>
<td>Total (taped)</td>
<td>13</td>
<td>152 ± 161</td>
<td>335 ± 239$^\psi$</td>
</tr>
</tbody>
</table>

* Significant ($p < .05$) from male peak TA.
$^\psi$Significant ($p < .05$) from the peak GN during untaped walking.

$^a$EMG during walking normalized to EMG sampled during a 75% TA MVIC.
$^b$EMG during walking normalized to EMG sampled during a 75% GN MVIC.
Test of Equality of Variances exceeding .05). These adjusted data were used to assess the gender amplitude differences as well as used for the taping statistical comparisons in walking conditions with respect to TA and GN EMG amplitudes (see Table 6). A two-way ANOVA was used to compare peak TA and peak GN EMG amplitude during the untaped walking conditions across genders (N = 13). There was a significant (p < .05) gender by muscle interaction. Tukey’s post hoc test was used to explore this interaction. Females had significantly (p < .05) greater normalized peak TA amplitudes than males during untaped walking.

**Running.** Normalized amplitudes for untaped running consisted of percentage values, calculated to the nearest whole number, generated for TA1, GN, and TA2 (N = 20). A two-way ANOVA was used to compare peak TA1, peak GN, and peak TA2 EMG amplitudes during the untaped running condition across gender. There were no significant (p < .05) main effects or interaction.

**Effects of Ankle Taping on EMG Measures During Walking**

Using the data summarized in Table 7, the time to peak TA and time to peak GN in taped and untaped walking trials were compared using a two-way ANOVA with repeated measures (N = 20). There were no significant (p > .05) main effects or interaction.

Using the data from Table 6, a two-way ANOVA with repeated measures was used to compare TA and GN amplitudes during taped and untaped walking conditions (N = 13). A significant (p < .05) muscle by condition interaction was found. Tukey’s post hoc test was used to explore this interaction. A significant (p < .05) increase in amplitude was found for the GN in the taped walking condition.
Table 7. Comparison of Tibialis Anterior (TA) and Gastrocnemius (GN) Temporal Activation During Taped and Untaped Walking (N = 20)

<table>
<thead>
<tr>
<th>Condition</th>
<th>Time to Peak $\text{TA}^a$ (sec) Mean ± SD</th>
<th>Time to Peak $\text{GN}^b$ (sec) Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Untaped</td>
<td>.004 ± .005</td>
<td>.228 ± .183</td>
</tr>
<tr>
<td>Taped</td>
<td>.004 ± .001</td>
<td>.244 ± .176</td>
</tr>
</tbody>
</table>

$^a$Time referenced to peak amplitude by subtracting heel strike from peak TA.
$^b$Time referenced to peak amplitude by subtracting heel strike from peak GN.

Effects of Ankle Taping on EMG Measures During Running

Examination of the raw EMG for the taped running trials revealed 50% of the subjects had observable triphasic burst pattern as indicated by the lack of TA2 (see Appendix F). For those having triphasic burst patterns in both taped and untaped running conditions (see Table 8), the time to peak TA1, time to peak GN, and time to peak TA2 EMG amplitude were compared using a two-way ANOVA with repeated measures (N = 10). There was a significant (p < .05) time to peak amplitude by condition interaction. Tukey’s post host test was used to explore this interaction. The time to peak TA2 EMG amplitude was significantly (p < .05) greater during the taped running condition. A two-way ANOVA with repeated measures was used to compare the peak GN and the highest peak TA amplitudes in the taped and untaped running conditions (N = 20). There were no significant (p > .05) main effects or interaction.
Table 8. Temporal Comparison of the Triphasic Burst Pattern During Taped and Untaped Running (N = 10)

<table>
<thead>
<tr>
<th>Condition</th>
<th>Time to Peak TA1(^a) (sec)</th>
<th>Time to Peak GN(^b) (sec)</th>
<th>Time to Peak TA2(^c) (sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>Untaped</td>
<td>-.004 ± .008</td>
<td>.186 ± .006</td>
<td>.204 ± .116</td>
</tr>
<tr>
<td>Taped</td>
<td>-.005 ± .007</td>
<td>.110 ± .005</td>
<td>.250 ± .124*</td>
</tr>
</tbody>
</table>

* Significantly (p < .05) greater time to peak EMG amplitude in the taped condition.

\(^a\) Time referenced to peak amplitude by subtracting heel strike from peak TA1.
\(^b\) Time referenced to peak amplitude by subtracting heel strike from peak GN.
\(^c\) Time referenced to peak amplitude by subtracting heel strike from peak TA2.

**Discussion**

**Neuromuscular Regulation of Impact Forces at Heel Strike During Walking**

The results of this study support the idea that the regulation of heel strike during walking occurs by way of a biphasic burst pattern between the TA and GN as previously described by Knutson and Soderberg (1995). Rapidly following heel strike, the time to peak amplitude for the TA preceded the time to peak amplitude for the GN. The temporal firing of this sequential TA and GN activation at heel strike was similar among genders. With respect to EMG amplitude, there also seems to be a trend for individual differences across gender as indicated by the lack of homogeneity within these data. Following adjustment of these data with the removal of several outliers, there is some indication that females tend to rely more heavily on the TA at heel strike during walking.
This finding was certainly unexpected taking into consideration that there were no gender differences relative to the EMG amplitude during 75% MVIC. Further, males generated greater lb-ft measures than females when establishing the maximum isometric torque values.

Closed ankle taping does not appear to have an adverse effect on the neuromuscular regulation of impact forces at heel strike during walking. Although there was a delay in the mean time to peak GN during the taped walking condition (see Table 7), this delay in time to peak value was not supported by statistical comparison. Based on the adjusted data set for amplitude however, ankle taping may generate an increase in amplitude for the GN at heel strike during walking. This increased amplitude did not occur with simultaneous amplitude depression of the antagonistic TA. Again, with the variability existing in the entire data set as indicated by Levene's Test for Equal Variances, this finding was only present among 13 subjects.

Neuromuscular Regulation of Impact Forces at Heel Strike During Running

The results of this study indicate that the neuromuscular regulation of impact forces at heel strike do not occur as a true cocontraction between the TA and GN. This finding contradicts statements made in other EMG studies investigating the role of these muscles during heel strike (Elliott & Blanksby, 1979; Freedman & Kent, 1987; Mero & Komi, 1987). In the present study, firing of the TA and GN relative to the time at heel strike (see Table 5 for mean time to peak values) was more consistent with a triphasic burst pattern response as in the raw EMG appearing in the study by Freedman and Kent (1987). The presence of the time to peak TA occurring at a mean of -.027 sec prior to heel strike
during running remains consistent with previously reported findings (Elliott & Blanksby, 1979; Mero & Komi, 1987; Reber et al., 1993). As seen with the biphasic burst pattern at heel strike in walking, there was no gender difference in the rate at which this triphasic burst pattern occurs relative to the time at heel strike in running.

**Characteristics of the triphasic burst pattern at heel strike during untaped running.** As stated, initiation of the triphasic burst pattern occurs prior to heel strike with no differences in the triphasic burst pattern duration between gender. Relative to the time at heel strike, there is an initial burst of the TA in the presence of GN electrical depression or silence that precedes a burst of the GN in the presence of TA electrical depression or silence. Following this rapid sequential TA and GN firing, there is an overlapping second burst of TA activity either during heel strike or shortly after heel rise. There is no difference in TA and GN amplitude and the amplitude of the second TA burst is no different from that of the initial one.

**Effects of ankle taping on the neuromuscular regulation of impact forces at heel strike during running.** The results of this study suggest that closed ankle taping alters the timing of a triphasic burst pattern at heel strike in running. As reported in 50% of the subjects during the taped running condition, the triphasic burst pattern was completely absorbed as indicated by the absence of a second TA burst. For those having triphasic burst patterns in both taped and untaped conditions, there was a significant ($p < .05$) mean delay of .50 sec of this neuromuscular pattern in the condition where subjects ran with the support of ankle tape. While the application of ankle tape did not appear to elevate or depress amplitudes of the TA or GN relative to the time at heel strike in
running, perturbations of elevated electrical activity during this condition were observed for both muscles between heel strikes in a substantial number of the subjects tested. Unfortunately, attempts to describe this observation statistically were unsuccessful.

**Implications**

According to Guyton and Hall (1996), conditions in which antagonistic muscles collectively fire to maintain joint stiffness, “the spindles on both sides of the joints are activated at the same time” (p. 690). Although this simultaneous activation of antagonistic muscle spindles represents the mechanism by which muscle stiffness around a joint is regulated in a stationary position, the presence of this mediated cocontraction during locomotor activities is certainly undesirable. Heightened neuromuscular activity in an antagonist during locomotion creates a situation where mechanical efficiency is reduced. Specifically, this heightened activity “impairs, through reciprocal inhibition, the ability to fully and efficiently activate agonist motor units” (Sale, 1988, p. S140). As such, we learn rather early in life, to regulate dynamic joint stability at heel strike during walking through sequential TA and GN firing (Knutson & Soderberg, 1995).

While the role of the muscle spindle in coordinating the myotatic stretch reflex is well understood, its interactive role with the proprioceptive regulation of impact forces is an area of complexity that is less defined. The largest area of deficit in literature for this concern is the relationship and function of antagonistic muscles during impact forces. Hutton and Atwater (1992) tied this deficiency in the literature together nicely by stating:

Taken together, the above adaptations seen in both muscle spindles and golgi tendon units would render stretch reflex pathways more excitable to impending elongation following a conditioning contraction (e.g. as in reciprocal activation patterns of antagonistic muscles such as those seen in locomotion). Surprisingly,
little attention has been given to this body of literature either in the neurosciences, exercise sciences or rehabilitation medicine (p. 410).

Examining amplitude and temporal responses in the antagonistic TA and GN muscles in taped and untaped conditions, this study investigated this area related to impact forces at heel strike.

At slower speeds of locomotion, closed ankle taping does not appear to affect the way males and females regulate impact forces at heel strike. Likewise during higher speed locomotor activities, the time to peak TA is not affected by the use of closed ankle taping. The means by which impact forces at heel strike are sequentially regulated in both genders however is either significantly (p < .05) delayed in total duration by .50 sec or completely altered with the removal of TA2. Taking into account what Shorten and Winslow (1992) described regarding the degree of impact during the first .50 sec of the support phase, as well the importance these muscles play in the regulation of impact forces, these findings suggest that ankle taping creates a situation where impact forces experienced by the weight bearing joints of the lower extremity may be altered. As Reber et al. (1993) reported in reference to strayed bouts of TA activity during the entire gait cycle, the observed electrical perturbations between heel strikes in the present study may indicate a mechanism for increased muscular fatigue during running as a result of taping.

Summary

The results of this study suggest the neuromuscular regulation of impact forces at heel strike may be dependent upon the speed of ambulation. Maintenance of dynamic joint stability even at slower speeds of ambulation where no flight phase exists during stride,
does not appear as a true cocontraction. During the untaped walking trials, the TA and GN EMG profile in this study appeared as a biphasic burst pattern with the time to peak TA discretely preceding the time to peak GN. At faster speeds, during the running untaped trials, TA and GN EMG activity relative to the time at heel strike appeared as a triphasic burst pattern response. All TA and GN amplitudes collected in this study were normalized to EMG collected during a 75% MVIC as indicated via dynamometry. The normalized amplitudes in the walking trials demonstrated excessive variability. With an adjustment of the walking amplitude data, gender comparison indicated that females during untaped walking relied more heavily on the TA at heel strike. Of further note, there was an elevation of the GN at heel strike during taped walking across genders, however, these amplitude changes could not be shown using the entire data set.

The application of closed ankle taping did not affect the temporal firing of the biphasic burst pattern during walking. During running, where impact at heel strike is much greater, the regulation of dynamic joint stability may be compromised. Specifically, ankle taping creates a situation where either the duration of the entire triphasic burst pattern is delayed or completely altered by the removal of TA2.
CHAPTER V

CONCLUSIONS AND RECOMMENDATIONS

Conclusions

When athletes return to sport following ankle injury, the use of ankle taping has been shown to prevent excessive range of motion (Fumich et al., 1981; Laughman et al., 1980; Metcalfe et al., 1997). Other research has suggested ankle taping can prevent injury, or re-injury, by improving proprioceptive functions (Heit et al., 1996; Karlsson & Andreasson, 1992; Simoneau et al., 1997). The bulk of ankle taping research however has not addressed the role of muscular activity in the regulation of impact forces.

The results of this study demonstrated that ankle taping alters the regulation of sequential muscle activity relative to the time of heel strike during running. During strides between successive heel strikes, perturbations of antagonistic electrical activity between the TA and GN were observed. This coactivation of opposing muscles resembled the EMG profile of the TA and GN in adolescents below the age of seven described by Knutson and Soderberg (1995), and may indicate a possible mechanism for less mechanical efficiency and earlier muscle fatigue as a result of ankle taping.

While there is a concern for atrophic changes as a result of ankle taping, the potential alteration of the neuromuscular profile at heel strike represents a problem of greater concern. During running, the lower extremity relies heavily on subconscious neuromuscular mechanisms to attenuate shock forces (Winter, 1983) and prevent chronic articular trauma (Dickenson et al., 1985). Examining the temporal findings of this study,
it seems apparent that ankle taping alters the neuromuscular regulation of impact forces at heel strike. Whether this change has a beneficial or adverse effect on the joint reaction forces experienced in the lower extremity during running is not known. As such, ankle taping should be reserved for athletes needing protection during the proliferative/maturation phase of healing and should not be thought of as a replacement for therapeutic exercise. Furthermore, based on these findings, the taping of healthy athletes should be undertaken with caution due to the uncertainty of whether or not taping will increase or decrease impact forces.

**Recommendations**

The attenuation of impact forces is an area of deficiency in the literature on ankle taping. While this study demonstrated that the neuromuscular profile of the TA and GN at heel strike in running is altered as a result of taping, it is not known whether this results in increased or decreased impact forces. Clearly, ground reaction force platform analysis would be helpful in the examination of this issue.

The variability in the normalized amplitude data presented problems in the gender and taping amplitude comparisons. Having homogeneity within the reference EMG samples taken during the 75% MVIC suggested this intensity provided a suitable means of normalization as suggested by De Luca (1997). Future research should determine whether individual differences in the amplitude of biphasic activity at heel strike in walking exists based on stride rate and stride length. The variation within this study may have been due to the lack of repeated trials taken at the same speeds for each condition.
It is important to point out the necessity of using accurate motor points when determining EMG electrode placement. In this study, it was necessary to retest two subjects due to an error made with regard to motor points. The problem that existed was that these subjects were tested using a motor point in a more proximal area of the medial GN. Upon retesting with the EMG lead placed at a motor point in the distal belly of the medial GN, the normalized EMG was more consistent with the other subjects. An interesting area to examine would be the role of the GN in knee extension during toe-off versus its role serving as an intervening burst in the triphasic burst pattern at heel strike. Inquiries into what role the lateral GN and soleus, as well as the relationship between the medial and lateral heads of the GN during these events of the gait cycle are also areas worthy of investigation.

In this study, periods of excitation and inhibition, as indicated by electrical activity and electrical silence were observed relative to the time at heel strike. It is not known what role the golgi tendon organ (GTO) has in regard to these sequential muscle functions. Thus it is not known if the phasic inverse stretch reflex can be lowered to enhance regulation via sequential firing by surpressing the muscle spindle in anticipation of a loading force (i.e., cushioning effect of antagonistic muscles). Along similar lines, it is not known if injury or disuse influences the sequential neuromuscular regulation closed kinetic chain impact forces (i.e., sequential TA and GN firing at heel strike).

Accommodation to the application of athletic taping was not examined in this study. It is not known if the neuromuscular system will accommodate to the routine use of athletic taping enabling individuals to coordinate normal sequential muscular activity at
heel strike during running. Further, this study did not examine whether the loosening of tape generated a restoration of normal sequential EMG activities relative to heel strike. Future researchers may wish to examine these areas of inquiry.

One last area of study that might help expand the knowledge of athletic taping which is related to its effect on the regulation of impact forces would be to examine the kinematics of the lower extremity. As concluded, ankle taping introduces a situation where neuromuscular regulation of impact forces at heel strike during running is altered. Taking into account what is known about the properties of ankle taping regarding the restriction of motions about the foot and ankle, curiosity exists as to whether the body responds to this altered neuromuscular control at heel strike through some form of proximal kinematic adjustment.
REFERENCES


APPENDIX A

CONSENT FORM
INFORMED CONSENT FOR PARTICIPATION
IN A RESEARCH PROJECT ENTITLED “EFFECTS OF
ANKLE TAPEING ON THE NEUROMUSCULAR REGULATION
OF IMPACT FORCES AT HEEL STRIKE”

I ______________________, give my informed consent to participate in a study examining the effects of ankle taping on muscle function during walking and jogging. I have been informed my identity will be referred to by number throughout the study and that any measure collected on me will be pooled into a group of measures taken from other subjects for the purpose of analysis. I authorize the presentation of test procedures and any results, data, and photos for dissemination as long as my identity is protected.

I have been informed the measurement conditions and test procedures carry with them minimal risk to my health. Further, it has been explained to me that the measurement preparations used in this study involve the use of motor point electrical stimulation and may produce temporary minor discomfort but in no way poses any real danger to my well being. I am also aware that any excess body hair on small areas on my dominant side lower leg will be removed via an electric razor in the interest of collecting accurate measures. If at any time I decide this removal of body hair would be a cosmetic burden, I am free to withdraw myself from this study without any penalty.

I attest to having no known physical or orthopaedic problems that would preclude my participation in this study. I have been informed that I am free to remove myself from this study at any time without penalty. Any questions I have regarding any aspect of this study can be answered by the principal investigator or the study’s faculty thesis chair. I am free to contact Robert Pettitt at 784-3672 or the thesis chair, Dr. Marilyn Miller, by phone at 785-6527 or in person at 0134 Mitchell Hall. Having read this statement in the presence of the investigator whose signature appears below, I give my full informed consent as signed and dated in the space labeled participant signature.

______________________________ / ___ / 1998
(Participant Signature) month day

______________________________ / ___ / 1998
(Investigator) month day
APPENDIX B

EMG TREATMENTS
EMG Treatments

Amplification

Gain: 349
Common Mode Rejection Ratio: double differential of 104
DC input impedance: 100,000 Megaohms
Quiescent Current: 12 mAmps

Sampling Range: 8 Hz – 28 KHz

Low Pass Cutoff: 500.00 Hz

Sampling Time: 3.00 sec.

Trigger: 1,000 mV

Pre-trigger percentage: 25

Rectification: full wave

Processing: moving average of .100 sec (linear envelope)
APPENDIX C

RECTIFIED AND PROCESSED EMG FOR UNTAPED RUNNING

(AMPLITUDES AND TIMES LABELED)
Rectified/Linear Envelope:
Running Without Tape

Mvols
25
20
15
10
5
0
Sec
0.6
0.8
1.0
1.2

H= Tibant
Tc( H)= .180

I= Gastrocn
Tc( I)= .180

X=.782

X=.726
Y=.159

X=1.054
Y=.173

X=.879
Y=.111
APPENDIX D

SUBJECT DATA SHEET
Subject Data Sheet

Name: __________________________ M  F
(Circle)
Assigned ID Number __________________
Age: _____
Ht: _____
Wt: _____
Trochanteric Ht: ______ Limb Dominance: R  L
Peak GN Torque (MVIC): ______ _____ _____ Max Torque Value _____
Peak TA Torque (MVIC): ______ _____ _____ Max Torque Value _____
75% GN Torque Value ______
75% TA Torque Value ______

List of Condition Order:
_____________________________________________________________________

DYNAMOMETRY AMPLITUDE VALUES (mV)

TA mV @ 75% MVIC: ______
GN mV @ 75% MVIC: ______

EXPERIMENTAL CONDITIONS

<table>
<thead>
<tr>
<th>Condition</th>
<th>Y (Mv)</th>
<th>X (msec)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>TA</td>
<td>GN</td>
</tr>
<tr>
<td>Walk no tape</td>
<td>______</td>
<td>______</td>
</tr>
<tr>
<td>Walk with tape</td>
<td>______</td>
<td>______</td>
</tr>
<tr>
<td>Run no tape</td>
<td>______</td>
<td>______</td>
</tr>
<tr>
<td>Run with tape</td>
<td>______</td>
<td>______</td>
</tr>
</tbody>
</table>
APPENDIX E

RAW EMG: RUNNING WITHOUT TAPE
Raw EMG: Running Without Tape

G = footswitch

H = tibant

I = gastroc
APPENDIX F

RAW EMG: RUNNING WITH TAPE
Raw EMG: Running with Tape

H = tibant

I = gastroc

G = foot switch