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The Effect of Anterior Tibial Translation on Quadriceps Recruitment During an Isometric Squat

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**THE EFFECT OF ANTERIOR TIBIAL TRANSLATION ON
QUADRICEPS RECRUITMENT DURING AN ISOMETRIC SQUAT**

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THESIS

Submitted to the Department of Physical Therapy
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MASTER OF SCIENCE IN PHYSICAL THERAPY

1996

The Effect of Anterior Tibial Translation on Quadriceps Recruitment During an Isometric Squat.

ABSTRACT

The purpose of this study was to determine how closed chain anterior tibial translation (ATT) affected quadriceps femoris recruitment as evidenced by electromyography.

Forty-three healthy subjects performed a unilateral squat under two conditions: anterior tibial force (ATF) and no tibial force (NTF). Two adjustable-height pulleys produced ATF by each pulling with ten percent of body weight in opposite directions on the proximal tibia and distal femur. Electromyographic (EMG) activity of the rectus femoris, vastus medialis, and vastus lateralis was monitored using surface electrodes. Percentage change in EMG area between the two squat conditions was averaged across all subjects and muscles. The squat under ATF conditions produced a significant increase in EMG activity of 14% as compared to NTF conditions ($p < .001$). Therefore, the authors concluded that ATF facilitated quadriceps femoris recruitment in a unilateral isometric squat. This facilitation technique has clinical significance in the rehabilitation of lower extremity pathologies.

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PREFACE

Definitions

ANTERIOR TIBIAL TRANSLATION (ATT): Anterior movement of the tibia relative to the femur; often used synonymously with ANTERIOR DRAWER and ANTERIOR SHEAR (Irrgang,1993).

CLOSED KINETIC CHAIN: Occurs when the distal segment is fixed, so a movement in one joint produces predictable movement in all joints of the kinetic chain (Irrgang, 1993).

CO-ACTIVATION: When the agonist and antagonist muscles contract simultaneously. Evidenced by concurrent EMG signals of antagonist muscles of greater than three percent of maximal voluntary contraction for that muscle (Grabiner, Cambell, Hawthorn, & Hawkins, 1989). Also known as CO-CONTRACTION:.

CONCENTRIC CONTRACTION: A contraction in which the muscle shortens in length while tension develops to overcome or move some resistance (Prentice, 1994).

CROSS-TALK: Unwanted EMG signal from surrounding muscles, synergists or antagonist muscles.

ECCENTRIC CONTRACTION: When the resistance is greater than the muscular force being produced, and the muscle lengthens while producing tension (Prentice, 1994).

ELECTROMYOGRAPHY (EMG): Provides an indirect indicator of muscular function. The electrical signals, which accompany the chemical stimulation of the muscle fibers, travel through the muscles and adjacent soft tissues indicating motor unit activation (Perry,1992).

FULL-WAVE RECTIFICATION: A method of processing an electromyographic signal that involves transposing all the negative signals to the positive side of the zero line (Perry, 1992).

IMPEDENCE: The opposition to the flow of alternating electrical current measured in Ohms (LeVeau & Andersson, 1992).

INTERNEURONAL MODULATION: Modification of afferent and efferent signals via small interneurons in the spinal cord and higher centers.

ISOKINETIC ACTIVITY: Accommodating variable resistance, constant speed, combination of isometric and isotonic contraction (Prentice, 1994).

ISOMETRIC CONTRACTION: Muscle contraction of constant length (Norkin & Levangie, 1992).

ISOTONIC CONTRACTION: Muscle contraction of constant tension (Norkin & Levangie, 1992).

MECHANORECEPTORS: Sensory nerve fibers that sense joint position and movement, and transmit this information back to the spinal cord.

MOTOR UNIT: An anterior horn cell, its axon, and all of the muscle fibers innervated by the axon branches (LeVeau & Andersson, 1992).

NOISE: Electrical potentials produced by electrodes, cables, amplifier or storage media and unrelated to the potentials of biologic origin (LeVeau & Andersson, 1992).

OPEN KINETIC CHAIN ACTIVITY: Occurs when the distal segment is free to move; can result in a more isolated joint movement (Irrgang, 1993).

QUADRICEPS EXTENSION MOMENT: The internal torque of the quadriceps muscles which acts to extend the knee (Haffajee, Moritz, & Svantesson, 1972).

RAW EMG SIGNAL: Unprocessed, electromyographic signal that is the basis of all methods of interpreting the myoelectrical activity from muscles (LeVeau & Andersson, 1992).

RECIPROCAL INHIBITION: Inhibition of the antagonist muscle by stimulation of the agonist.

ROOT-MEAN-SQUARE: A method of processing the electromyographic signal that represents the effective value of the quantity of an alternating current. Measures the electrical power in the signal (LeVeau & Andersson, 1992).

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CHAPTER 1
INTRODUCTION

CHAPTER 1

INTRODUCTION

Background to the Problem

Common lower extremity problems that follow certain types of knee injuries, surgical repairs, neurological pathology, and immobilization are strength, atrophy, and recruitment deficits in the quadriceps. People with anterior cruciate ligament (ACL) injuries are prime examples of patients who develop severe quadriceps weakness, although no trauma has been inflicted on this muscle group (Fu, Woo, & Irrgang, 1992; Irrgang, 1993; and Ohkoshi & Yasuda, 1989). Joint effusion has been shown to reduce quadriceps voluntary contraction (Hurley, Jones, & Newman, 1985); however, this does not explain quadriceps dysfunction that persists after effusion has resolved. Grimsby (personal communication and audiotapes, April, 1995 - March, 1996) theorizes that the injured ACL fails to activate the appropriate joint receptors that, via spinal cord interneuronal modulation, affect the alpha and gamma motor output to the quadriceps. Interneuronal modulation is also affected by knee flexion contracture and patellofemoral pain, both of which also correlate positively with quadriceps weakness (Fu et al, 1992). Pain often leads to disuse that results in more weakness and begins a vicious cycle. Weakness plus instability from also predisposes the joint to degenerative changes (Hurley et al, 1985) which further interfere with normal proprioceptive input. These signs and symptoms combine to further reduce knee extensor function which is necessary for many activities of daily living, such as ascending and descending stairs, squatting or

crouching to pick up objects, and getting in and out of an automobile (Schuldt, Ekholm, Nemeth, Arborelius, & Harms-Ringdahl, 1983).

Problem

Current literature suggests that anterior tibial translation (ATT) reflexively inhibits the quadriceps and facilitates the hamstrings via type III mechanoreceptors in the ACL and knee joint capsule (Solomonow, M., Baratta, R., Zhou, B. H., Shoji, H., Bose, W. & D'Ambrosia, R., 1987). However, unpublished research by Grimsby (1995-1996) found approximately a 38 percent increase in peak rectified EMG activity of the quadriceps and a 24 percent increase in hamstring activity when ATT was increased during squatting. Grimsby applied an anterior tibial force (ATF) equal to body weight (50 percent pulling forward on the tibia, 50 percent pulling back on the femur) via a two-pulley system. His subjects presented with varying degrees of acute and chronic ACL injury, but none had complete tears or surgical grafts. Normal subjects demonstrated a smaller percentage increase in quadriceps activity with increased ATT. Grimsby suggested that type III mechanoreceptors in the ACL or knee joint capsules facilitate quadriceps during activity.

If Grimsby's findings are correct, current ACL rehabilitation protocols that attempt to minimize ATT deprive patients of facilitation from type III mechanoreceptors that could increase quadriceps strength. Efforts to protect the injured ACL from ATT must be balanced with methods that will facilitate muscle fiber recruitment and produce tension in the collagen of the ACL. Tension along the lines of stress will direct the

orientation of new collagen fibers parallel to the direction of that stress and stimulate increased collagen production. In this way, tension optimizes healing and ligamentus strength, and increases muscular strength.

Purpose for the Study

The purpose of this study was to determine whether ATT facilitated quadriceps recruitment during an isometric squat as suggested by Grimsby, or inhibited it as current literature reports. If Grimsby's concept is correct, the resulting increased quadriceps recruitment may speed recovery of strength and enhance ligamentus healing, thus decreasing the duration of rehabilitation. If ATF increases normal proprioceptive input through ATT it could then be used as an alternative to or in conjunction with electrical stimulation of the quadriceps muscle fibers. The use of ATF instead of electric stimulation would result in less equipment, one less treatment modality, less discomfort for the patient, and less time required of the therapist during treatment sessions, thus reducing the cost of rehabilitation.

CHAPTER 2
LITERATURE REVIEW

CHAPTER 2

LITERATURE REVIEW

Mechanoreceptors

Four major types of afferent nerve fibers have been identified in human joints, three mechanoreceptors, and one nociceptor (Grabiner, 1993; Irrgang, 1993; Lehmkuhl & Smith, 1983; Wyke, 1994). Mechanoreceptors are sensory nerve fibers that sense joint position and movement, and transmit this information back to the spinal cord.

Nociceptors are pain receptors. Norkin and Levangie (1992) described five receptor types. Type I Golgi ligament endings which are located in joint ligaments, and type II Ruffini endings which are located in joint capsules, were not differentiated by other authors as they were by Norkin & Levangie. Grabiner (1993), stated that Ruffini endings exist in the medial meniscus, capsule, and ACL. Grigg and Hoffman (1982), Irrgang (1993), Lehmkuhl and Smith (1993), and Wyke (1994) recognized Ruffini or Type I receptors only in the capsule. Zimney, Schutte, and Dabezies (1986) stated they exist primarily in the ligaments. Guyton (1986) reported that Golgi and Ruffini endings have identical response properties, citing Golgi receptors in the ligaments, and Ruffini endings in the capsule. Since the behavioral characteristics are identical, these two endings will be described inclusively as Ruffini type I receptors. This will also avoid confusion between the Type III Golgi-Mazonni corpuscle and the Type I Golgi ligament ending.

As stated above, Type I receptors generally exist in superficial joint capsules and ligaments. In the knee specifically, they are found in the ACL, medial meniscus, and capsule. They demonstrate static and dynamic qualities (Grabiner, 1993; Irrgang, 1993;

Wyke, 1994) and respond to capsule stretch (Norkin & Levangie, 1992; Wyke, 1994). This stretch response allows Ruffini endings to sense fluid pressure changes (Grabiner, 1993; Norkin & Levangie, 1992), and suggests a role in arthrogenic muscle inhibition, which occurs in injured joints with effusion (Hurley et al, 1985). Grigg and Hoffman (1982) stated that Type I afferents also respond to compression. These receptors fire at rest, at the beginning of range, and at the end of the collagen stretch which contributes to kinesthetic senses of direction, velocity, amplitude and static joint position. Type I fibers also elicit tonic reflexogenic effects on muscle tone and inhibition of pain (Guyton, 1986; Irrgang, 1993; Norkin & Levangie, 1992; Wyke, 1994). Type I receptors show particular sensitivity to sudden movements and joint rotation (Grigg & Hoffman, 1982; Guyton, 1986). Because they have a very low threshold and are slow adapting they respond to very subtle movements and may fire continuously (Wyke, 1994). Generally, these receptors remain active for up to one minute following stimulation (Schwarcz, 1994).

Type II receptors, or Pacinian corpuscles, lie in the deeper layers of distal joint capsules (Wyke, 1994) and at ligament ends (Zimney et al 1986). They are found in the capsule of the knee joint, the medial meniscus and the ACL (Grabiner, 1993). These low threshold fibers differ from type I fibers in that they are fast adapting and dynamic (Grabiner, 1993; Guyton, 1986; Irrgang, 1993; Wyke, 1994). As a result, type II receptors respond to quick changes of movement direction or speed in the beginning and mid range of collagen stretch (Grabiner, 1993; Norkin & Levangie, 1992) and remain active for only a fraction of a second (Wyke, 1994). They are inactive in immobile joints

and in joints moving at a constant velocity (Grabiner, 1993; Wyke, 1994). Type II receptors have similar pain-inhibiting effects as type I receptors, but are phasic reflexogenic rather than tonic reflexogenic (Wyke, 1994).

Type III fibers or Golgi-Mazzoni corpuscles are primarily located in the deep and superficial layers of the joint capsule (Grigg, Hoffman, & Fogarty, 1982) and ligaments (Wyke, 1994), including the ACL, collaterals and menisci of the knee (Grabiner, 1993). Type III receptors respond to both stretch and compression (Grabiner, 1993; Grigg et al 1982), but are not active until the end of joint range, secondary to their high threshold (Grabiner, 1993; Irrgang, 1993; Wyke, 1994). Golgi-Mazzoni receptors are very slow adapting, so they remain active for several minutes (Grabiner, 1993; Grigg et al, 1982; Irrgang, 1993; Wyke). When recruited, type III fibers activate wide dynamic range fibers, producing both tonic and phasic muscular responses (Grimsby, 1995-1996). Their inhibitory effect, however, can be overridden by the sympathetic nervous system (Schwarcz, 1994).

During extremes of motion, type III receptors activate protective reflex arcs such as the ACL-hamstring synergy which will be discussed in detail later. Theoretically, this synergy results in increased hamstring and decreased quadriceps muscle recruitment when the ACL is strained. Unfortunately, such reflex arcs do not respond quickly enough to effectively protect tissue from rapidly applied loads such as those associated with athletic injuries (Irrgang, 1993). Due to their dynamic nature (Wyke, 1994), Golgi-Mazzoni fibers are not active in immobile joints (Grabiner, 1993). As a result, they

would not have been recruited by the isometric procedures used in much of the literature regarding such reflex arcs as the ACL-hamstring synergy.

Type IV receptors are free nerve endings in the capsule, ligaments, and periosteum (Norkin & Levangie, 1992; Wyke, 1994). Noxious chemical or extreme mechanical stimuli (Grabiner, 1993; Norkin & Levangie, 1992; Wyke, 1994) activate these very high threshold, slow conducting and nonadapting fibers (Wyke, 1994). Three subtypes respond to specific kinds of painful stimuli. A-delta high threshold fibers responds to mechanical stimuli. A-delta heat nociceptive afferents responds to mechanical and thermal stimuli. C-polymodal afferents respond to mechanical, thermal, and chemical stimuli (Newton, 1990). Pain perception can be inhibited by appropriate stimulation of type I and II receptors and muscle spasm can be reduced via their tonic reflexogenic effects (Schwarcz, 1994).

All four types of receptors trigger afferent sensory information that enters the posterior horn lamina of the spinal cord. From there, via interneurons, the information can affect other afferent stimuli entering the dorsal lamina, proceed up to higher centers of the brain, and affect sensory perception and descending tract neuronal flow. The information can then continue on to the lateral pyramidal tract to facilitate or inhibit the sympathetic nervous system efferent flow, and/or synapse to the anterior horn and facilitate or inhibit the alpha and gamma motor units of both the ipsilateral and contralateral muscle groups. For example, when a person squats, the tibia translates, firing type I and II receptors that send sensory information to the posterior horn lamina.

The information can proceed up to higher centers of the brain to produce kinesthetic awareness, and can synapse to the anterior horn to facilitate the alpha and gamma motor neurons to the hamstrings and quadriceps muscles which stabilize the tibia and control the hip and knee flexion.

Periarticular Soft Tissue

The major components of the periarticular soft tissue of the knee joint include the cruciate ligaments, collateral ligaments, joint capsule, and menisci (Woo, Hollis, Adams, Lyon, & Takai, 1991). Kapandji (1987) described the functions of these tissues. The anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) constitute the primary restraints to anterior and posterior tibial translation, respectively. Although the PCL plays a significant role in limiting ATT, the ACL is not so significantly involved in restricting posterior translation. The medial collateral ligament (MCL) primarily restrains valgus forces. The lateral collateral ligament (LCL) restrains varus forces. Due to their similar fiber orientation (eg, posterior-superior to anterior-inferior) the ACL and the MCL work in synergy. Similarly, the posterior cruciate and the LCL are synergistic. The joint capsule is stretched by movement in all directions, except joint approximation or compression. The cruciate ligaments are often considered thickenings of the joint capsule. The menisci enhance the congruence of the articular surfaces of the tibia and femur, and transmit compressive forces between these two bones. During rotational movements, the menisci follow the femoral condyles, but during flexion and extension the menisci glide with the tibia.

The mechanical strength of some of these periarticular soft tissue structures has been assessed to determine their ability to stabilize the knee. Woo et al (1991) examined the tensile strength of the ACL from cadaveric specimens positioned at 30 degrees of knee flexion. They measured linear stiffness, ultimate load, and energy absorbed at failure. Specimen age significantly affected each of the parameters studied. Age reduced the tensile strength under all conditions. Depending upon age, the linear stiffness ranged from 180 ± 25 newtons/millimeter (n/mm) to 242 ± 28 n/mm and the ultimate load tolerance ranged from 658 ± 129 newtons (n) to 2160 ± 157 n. Energy absorbed at failure ranged from 180 ± 25 newton-meters (n-m) to 242 ± 28 n-m. The older specimens had a higher incidence of femur-ACL-tibia failure. Failure occurred predominantly by midsubstance tear for older specimens and by boney avulsion for younger specimens.

Anterior Tibial Translation

The ACL provides 85 percent of the resistance to anterior tibial translation (ATT) which is defined as anterior movement of the tibia relative to the femur (Irrgang, 1993). Anterior shear force is often used interchangeably with ATT. Because of the relationship between the ACL and ATT, ACL strain can be assumed to indicate ATT, and vice versa. Haffajee, Moritz, and Svantesson (1972) point out that translatory equilibrium of the knee joint is not achieved by the quadriceps and external moments alone. Internal forces of the ACL are required for complete equilibrium. Stress on the ACL is a function of the quadriceps' length-tension relationship and patellar ligament insertion angle (Gross, Tyson, & Burns, 1993). A reduction in the patellar ligament insertion angle as a result of

knee flexion is presumed to decrease the ATF vector produced by the quadriceps, thereby reducing ATT.

Most research agrees that ATT is inversely related to knee flexion angle, with the highest stress occurring during the last 30 degrees of knee extension, with peak translation occurring at approximately 15 degrees in open kinetic chain (Bagger, Ravn, Lavard, Blyme, & Sorensen, 1992; Jurist & Otis, 1985; More, Karras, Neiman, Fritschy, Woo, & Daniel, 1993; Smidt, 1973). Gross et al (1993) reported that two out of four subjects demonstrated peak ATT at 45 degrees. Not only was this study very small, but all subjects had insufficient ACLs. Both of these factors, sample size and ACL pathology, may account for the discrepancy between the results of Gross et al and that of the other three studies. However, Bagger et al (1992) also studied ACL deficient subjects, and the results were consistent with other research. Minimal translation exists somewhere between 60 and 75 degrees (Howell, 1990; Smidt, 1973). Since other conditions were similar, the 15 degree variation may be explained by the low intertester reliability of the KT 1000 arthrometer (Fiebert, Gresley, Hoffman, & Kunkel, 1994) used to measure ATT.

It is generally accepted that more ATT occurs in open kinetic chain movements than in closed kinetic chain movements (Fu et al, 1992; Irrgang, 1993; Voght et al, 1991; Yack, Collins, & Whieldon, 1993; Yack, Riley, & Wheildon 1994). In an open kinetic chain, the distal segment is free to move, and can result in a more isolated joint movement. In a closed kinetic chain the distal segment is fixed, so a movement in one

joint produces predictable movement in all joints of the kinetic chain (Irrgang, 1993). Compressive forces and co-contraction of the quadriceps and hamstrings are believed to be the limiting factors to ATT in closed kinetic chain. Increased load enhances co-contraction and compression, thereby decreasing ATT. Yack et al (1994) established a maximum load requirement of 25 percent body weight for this relationship to apply in their study of ACL-deficient subjects. Beyond this 25 percent, increased load did not affect the amount of ATT during the closed chain squat exercise.

Most researchers agree that open kinetic chain extension also results in more ATT than the drawer tests commonly used in evaluation (Voght et al, 1991; Yack et al, 1993; Yack et al, 1994). In contrast, Howell (1990) showed less translation with open kinetic chain extension than with a 89 newton (20 pound) drawer test. The placement of resistance in open chain exercise affects the amount of translation that occurs, so it is possible that differences in the position of the resistance pad may explain the discrepancy in Howell's findings. During resisted extension, distal placement of resistance increased ATT, while proximal placement decreased ATT, according to Jurist and Otis' (1985) study of normal knees. Their procedures varied the moment arm of resistance during open chain isometric knee extension while maintaining a constant torque of 25 foot-pounds by adjusting the force. Their findings seem to be generally accepted, based on common ACL rehabilitation protocols that will be presented later. Grimsby (1995-1996) presented a biomechanical argument that the effect of resistance location would be reversed under dynamic conditions. He theorized that overcoming the distally placed

resistance causes movement around the mechanical axis of the resistance pad. If the distal tibia moves forward around this axis, the proximal tibia would have to move backward, producing posterior displacement of the tibia relative to the femur. Therefore, Grimsby suggested that posterior tibial displacement renders open chain exercise ineffective for restoring quadriceps strength, since he found that anterior displacement facilitated this muscle group when in closed kinetic chain exercise. It does not make sense that the normal anterior displacement of the tibia could be totally reversed when the quadriceps are contracting and the knee is extending. Therefore, it is the authors' opinion that proximal placement reduced the anterior tibial translation, resulting in a relative posterior displacement compared to that with more distally placed resistance. This relative posterior displacement would reduce the quadriceps ability to extend the knee according to Draganich, Andriacchi, & Andersson (1987), resulting in increased quadriceps output.

A variety of other biomechanical factors and exercise conditions affect ATT. Ohkoshi and Yasuda (1989) measured posterior drawer forces during downward increments of a half squat. They found that trunk flexion increased the posterior drawer forces. This relationship suggests that trunk flexion would decrease the anterior drawer forces during squatting. Fiebert et al (1994) determined that internal rotation of the tibia decreased ATT in open kinetic chain isometric contraction of normal knees at 30 degrees. This makes sense anatomically, since the anterior and posterior cruciate ligaments twist around each other during internal rotation, creating tension and joint compressive forces

(Kapandji, 1987). A final factor affecting ATT is the type of contraction. Quadriceps contraction--both dynamic and isometric--increases ACL strain from 45 degrees to zero (Renstrom, Arms, Stanwyck, Johnson, & Pope, 1986). Isometric quadriceps contraction theoretically produces more ATT and ACL stress than isokinetic exercise because isometrics include less hamstring co-contraction (Howell, 1990). However, isometric quadriceps contraction was found to decrease ATT and strain beyond 75 degrees (Howell, 1990; Renstrom et al, 1986).

Quadriceps-Hamstring Coactivation

The existence of an ACL-hamstring synergy that protects the ACL during high quadriceps loads remains a debated issue. First, one should consider that hamstring activity during coactivation is usually quite low, and that maximal quadriceps activity is significantly greater than maximal hamstring activity of normal subjects during open chain isokinetics (Baratta, Solomonow, Zhou, Letson, Chuinard, & D'Ambrosia, 1988). Grabiner, Cambell, Hawthorn, and Hawkins (1989) defined coactivation as concurrent EMG signals of hamstrings and quadriceps of greater than three percent of a pretested maximum voluntary contraction. Debate centers around the effectiveness of this synergy, and what causes it. Greater agreement exists concerning the point in the range of motion where coactivation occurs.

It is generally accepted that quadriceps-hamstring coactivation occurs during closed chain activity (Ohkoshi & Yasuda, 1989). Carlsoo, Fohlen, & Skogund (1973) examined co-contraction during gait, and demonstrated co-contraction during both open

and closed chain phases. The same amount of quadriceps activity was found when the hamstrings were activated, regardless of the knee angle, muscle length, and moment of force, during both swing and stance phases. More recent research also supports coactivation during both open and closed chain activity with high anterior shear forces (Bagger et al, 1992; Basmajian & DeLuca, 1985; Draganich, Jaeger, & Kralj, 1989; Howell, 1990; Smidt, 1973; Solomonow et al, 1987). Because hamstring activity increased during the portion of knee range when ATT is the greatest, most authors concluded that hamstring activity functioned to limit ATT and protect the ACL. This stabilizing theory of the hamstrings was further supported by More et al (1993) who found decreased ATT during the flexion phase of squatting when increased hamstring load was applied. This relationship was more pronounced with ACL deficient patients. Renstrom et al (1986) reported that co-contraction significantly increased ACL strain from 30 degrees to zero, and decreased strain from 120 to 90 degrees as compared to passive motion in cadavers. These findings suggest that this synergy is only effective within certain ranges of motion, and may explain discrepancies between studies

Draganich, Jaeger, and Kralj (1989) suggested that the golgi tendon organ, muscle spindle, ACL, and central nervous system are the possible structures responsible for coactivation. Wojts and Huston (1994) researched quadriceps, hamstring, and gastrocnemius activity in response to an ATT force in closed kinetic chain. Their results indicate the hamstrings of normal subjects are primarily activated by voluntary muscle

activity and an intermediate response that appears to be a spinal reflex with interneuronal input from higher centers.

Solomonow et al (1987) suggested that Ruffini and Golgi receptors within the ACL produce immediate response to ATT, while mechanoreceptors in the capsule and muscle spindle produce quadriceps inhibition and prolonged response time. Other studies did not indicate quadriceps inhibition, but this difference may be explained by Solomonow's inclusion of ACL-deficient patients who subluxed between 37 and 46 degrees during open chain knee extension. This subluxation may have stretched the joint capsule and other structures to end range, activating type III mechanoreceptors. If current literature on mechanoreceptors is accurate, this activation would reciprocally inhibit the quadriceps and facilitate the hamstrings. However, Grimsby (1995-1996) theorized that type III receptors are facilitatory to the quadriceps.

Basmajian and DeLuca (1985) suggest that separate controls exist for coactivation and reciprocal inhibition. They further suggest that the cerebellum plays a role in changing reciprocal inhibition into co-contraction. This triphasic pattern consists of a burst of agonist activity, followed by a decrease in agonist activity with a burst of antagonist activity, and finally co-contraction. This pattern generally occurs at higher speeds, while a low sustained antagonist activity occurs at slower speeds, and successive agonist-antagonist bursts at moderate speeds. This pattern will also vary according to the individual's physical training which can produce neurological changes that enable recruitment of more muscle fibers. Reciprocal inhibition is said to occur in rhythmic

movements, while co-contraction happens when stability is required.

Two studies strongly refute the notions of an ACL-hamstring synergy and quadriceps inhibition. Grabiner et al (1989) did not find an association between hamstring excitation, anterior drawer, extensor torque, and quadriceps excitation during isometric knee extension in open kinetic chain. The minor differences between Grabiner's study and those that support an ACL-hamstring synergy do not account for the differences in results. Baratta et al (1988) found some antagonistic quadriceps and hamstring muscle activity at all times that the knee joint was loaded in open chain isokinetic extension. They also found that antagonistic hamstring activity was nearly inversely related to its moment arm as it changed with the joint angle. Because the knee is a polycentric joint, with an instantaneous center of rotation that varies with joint angle, the moment arm of the muscles crossing the knee also varies with joint angle. This relationship maintains a near constant torque produced by the hamstrings, despite the changes in muscle activity. This would implicate the anterior capsule's sense of joint angle, not an ACL synergy, as the source of increased hamstring activity.

Quadriceps Femoris Function

Though the four heads of the quadriceps femoris are commonly thought of as acting in unison, the individual muscle heads actually have slightly differing patterns of activity (Basmajian & DeLuca, 1985). In unilateral stance, the rectus femoris (RF) demonstrates minimal activity. When a squatting motion is begun, RF activity is delayed until approximately 60 degrees, while the vasti are immediately active. Similarly, an

early cessation of RF activity occurs at approximately 40 degrees upon return from a full squat, while the vasti remain active. The vasti display a similar late onset and early cessation during open chain activity. The overall pattern of vasti and RF activity is phasic, with the vasti primarily active throughout the range of motion in closed chain and RF primarily active in open chain beyond 60 degrees of knee flexion. The RF is also described as a modulator, specifically as velocity changes during gait (Ranchos Los Amigos, 1996).

The quadriceps force required to extend the knee varies inversely with its moment arm as it changes with knee joint angle, in the same manor previously described for the hamstrings. An open kinetic chain cadaver study by Draganich, Andriacchi, and Andersson (1987) indicated that the quadriceps extension moment increased from full flexion to 20 degrees and decreased through terminal extension. Quadriceps extension moment can be defined as moment of force or torque produced by the quadriceps muscles that potentially extend the knee joint. Similarly, other researchers found that maximal quadriceps activity was required to complete the last 10 degrees of open chain knee extension (Andriacchi, Andersson, Ortengren, & Mikosz, 1984; Haffajee et al, 1972). Andriacchi et al (1984) added that increased load during this last ten degrees induced no effect on quadriceps EMG activity. Haffajee, Moritz, & Svantesson, (1972) identified a peak in quadriceps torque, requiring the least activity, at approximately 45 degrees.

The effect of knee flexion and extension on the moment arm of gravity is reversed between open kinetic chain and closed kinetic chain (Schuldt et al, 1993). With this

difference in mind, Schuldt's conclusion that the last 30 degrees of extension required the least quadriceps force does not conflict with the other studies described here.

Biomechanical analysis of closed kinetic chain knee extension illustrates that the floor reaction force crosses anterior to the knee joint, creating an extension moment at this point in the range.

A variety of other conditions can affect quadriceps activity. Isometric muscle setting prior to the active movement of straight leg raising was found by Soderberg and Cook (1983) to increase quadriceps activity. Vastus medialis was particularly affected. The vasti can also be influenced by hip flexion, which increases activity in both medial and lateral heads, and by tibial rotation. Vastus medialis is enhanced by lateral rotation of the tibia, while vastus lateralis is enhanced by medial rotation in open kinetic chain exercise. Andriacchi, Andersson, Ortengren, and Mikosz (1984) found that the addition of an adductor moment also increased quadriceps activity in open kinetic chain isometrics. This is supported by Basmajian and DeLuca (1985). Finally, EMG analysis by Habinicht and Kissinger (1994) demonstrated an increase in quadriceps femoris activity (specifically vastus medialis) in open chain isokinetics versus closed chain sit-to-stand. Recall that ATT was also reduced in a closed kinetic chain. This suggests a positive relationship between ATT and quadriceps activity, supporting Grimsby's contentions.

ACL Rehabilitation Protocols

Common qualities of current ACL rehabilitation protocols include closed kinetic

chain exercise, early full weight bearing, early restoration of full knee extension, and minimizing ACL stress induced by ATT (Haffajee et al, 1972; Irrgang, 1993; Woods & Bigland-Ritchie, 1993). Hurley, Jones, and Newman (1985) suggest a very intensive rehabilitation protocol to increase joint stability and quadriceps muscle strength, although they demonstrated only insignificant strength gains in patients with joint effusion that produced arthrogenic muscle inhibition. Blair and Wills (1991) recommend similar approaches to quickly return athletes to their sport. Fu et al (1992) reported that second surgeries were required less often when ACL patients participated in accelerated protocols. In addition to the common qualities described earlier, these aggressive rehabilitation programs were also characterized by early movement post-operatively, immediate partial weight bearing, rapid progression to light jogging, and return to sports within six months. Emphasis is placed on closed kinetic chain and proprioceptive exercise. No open chain treatment exercise is performed in the first four months post autograft surgery in order to avoid excessive ATT. The unilateral partial squat with increased proprioceptive input from ATT proposed in this study would meet many of these requirements of an aggressive protocol.

Closed kinetic chain exercise is stressed for several reasons. Less ATT occurs in closed kinetic chain than in open, therefore there is less stress placed on the injured ACL or graft (Fu et al, 1992; Howell, 1990; Irrgang, 1993; Voght et al, 1991; Yack et al, 1994). To further control translation, Irrgang (1993) recommends developing hamstring dominance with a one to one ratio of quadriceps to hamstring strength at slow speeds. A

normal ratio for slow speed contractions would be three to two, in favor of the quadriceps. Additionally, closed chain activities are considered more functional, since they are incorporated in most activities of daily living. When rising from a squat, the rectus femoris shortens over the knee joint while it lengthens across the hip joint. The reverse is true during the descent into a squat, and the result is little to no net change in length of the muscle. This pseudoisometric contraction of the rectus femoris muscle during squatting activities does not occur in most isolated open chain activities (Fu et al, 1992; Irrgang, 1993). While most exercises include concentric and an eccentric phases, closed chain activities emphasize eccentric action of the quadriceps and hamstrings. Open chain activities tend to emphasize concentric movement. Grimsby (1995-1996) found an increase in eccentric hamstring activity when the quadriceps contracted concentrically in closed kinetic chain. He concluded that eccentric training of the hamstrings was more effective in preventing ACL strain. Finally, joint reaction forces are decreased in closed kinetic chain, therefore, they are better tolerated (Fu et al, 1992; Irrgang, 1993).

Open chain activities should not be excluded, since our lower extremities also function in this capacity. The swing phase of ambulation and bicycling are two common examples of open chain activities. When rehabilitating a patient to their full functional capacity, a therapist will use a combination of closed kinetic chain, open kinetic chain, and functionally specific exercises. However, care must be taken not to place excessive stress on an acutely injured or reconstructed ACL. Even if Howell (1990) was correct in

his findings that open kinetic chain extension results in less ATT than a 20 pound drawer test, one must consider that the drawer test is not performed repetitively, as exercises are. It is well known that injury can occur from a repetitive or prolonged forces of small magnitude, as well as from isolated high forces. For safety, it is recommended that isotonic and isometric extension in open chain be limited to a range of 90 to 60 degrees, where less ATT occurs (Irrgang, 1993; Renstrom et al, 1986). Closed chain activity can be utilized earlier post-injury or post-surgery, while open kinetic chain exercise may be reserved for later stages of rehabilitation, as described by Blair and Wills (1991).

Electromyography

Definition and Characteristics

Electromyography (EMG) is commonly used to quantify, evaluate, and compare muscle activity. Perry (1992) defined EMG as an “indirect indicator of muscle function via recording of electric signals that are produced by the chemical stimulation of muscle fibers”. The chemical ion movement during depolarization and repolarization of the muscle fiber creates an electromagnetic field that is detected by EMG (Basmajian & DeLuca, 1985; Perry 1992). The EMG system records the difference in electrical potential between the recording electrode, placed over the target muscle, and the reference electrode placed over an indifferent boney site (Turker, 1993). The direction from which the depolarization approaches the recording electrode (Basmajian & DeLuca, 1985) determines whether the EMG deflection will be positive or negative.

The EMG signal is dependent on several muscle characteristics: the number of

motor units activated, fiber type, size, health, propagation velocity, state of fatigue, depth and distance of the muscle fibers in relation to the detection electrode, type and speed of contraction, and the joint angle (Basmajian & DeLuca, 1985; Perry, 1992; Winter, 1990). An increase in the number of firing motor units generally increases the amplitude of the EMG signal, but there can be cancellation of EMG signal due to superposition of motor units firing at the same time. The number of active motor units increase with increasing force except at high force levels, when the firing rate increases instead.

The number of active motor units also increases with fatigue, in order to compensate for the associated decrease in firing rate. Since slower, generally smaller, motor units are fatigue-resistant, these are the units called upon as fatigue increases. Smaller motor units have longer duration action potentials that demonstrate lower EMG frequencies. Therefore, with fatigue frequency decreases. Tissue filtering characteristics act as a low pass filter so as the frequency content of the signal shifts to the lower frequencies, more energy is transferred through the tissues to the electrodes. This results in increase of the amplitude to the recorded signal. Therefore, the result of fatigue is a decrease in frequency with an increase in amplitude.

Relationship to Muscle Force

Firing characteristics and excitability of the motor units have been studied, and a relationship has been demonstrated between integrated EMG activity and muscle tension (Basmajian & DeLuca, 1985, Haffajee et al, 1972; VanRuijven & Weijs, 1990; Woods & Bigland-Ritchie, 1993). This relationship depends on the specific muscle and possibly

the firing rate of that muscle, but is independent of training and contraction rate (Basmajian & DeLuca, 1985). For small muscles, such as in the hand, the relationship is generally quasilinear. For large muscles of the limbs, the relationship is generally nonlinear. Muscles with near uniform fiber type composition have a linear relationship between EMG and force. Muscles with a mixed fiber composition have a nonlinear relationship (LeVeau and Andersson, 1992). Various researchers have demonstrated both linear and nonlinear EMG-force relationships with the different components of the quadriceps muscle group (Basmajian & DeLuca 1985). Regardless of the specific relationship, that changes with muscle tension produce changes with EMG.

Perry (1992) best described the effects of contraction type on force and EMG output. Muscle force has been shown to vary with sarcomere length and connective tissue tensions. Eccentric and isometric contractions present similar stability of these factors, and have greater force potential than concentric contractions. Concentric contractions produce approximately 20 percent less force, and this force varies nonlinearly with the speed of the contraction. Therefore, the force-EMG relationship of concentric contractions varies with contraction speed. Dynamic concentric and eccentric contractions have shown a linear force-EMG relationship when velocity is kept constant. Similarly, the velocity-EMG relationship is linear when force is held constant. Joint position affects the muscle and moment arm lengths, thereby affecting the contraction force, and the slope of the isometric EMG-force relationship. However, the joint angle does not effect the EMG signal. Winter (1990) reported that EMG amplitude remained

constant despite muscle and moment arm length changes. When using surface electrodes, changes in muscle length resulting from changes in joint angle affect the electrode-fiber distance, which does affect the EMG signal.

Recording Methods

Several researchers have compared the use of surface electrodes to intramuscular wire electrodes (Christensen, Sogaard, Jensen, Finsen, & Sjogaard, 1995; Giroux & Lamontagne, 1990; Perry, Easterday, & Antonelli, 1981; Perry, 1992). Surface electrodes were preferred for their ease of application, and non-invasive nature. However, intramuscular wire electrodes are necessary for small, deep muscles, or when specific fascicles within one muscle need to be isolated. Surface electrodes tend to assess the activity of the whole muscle (Basmajian & DeLuca, 1985; Christensen et al, 1995; Winter, 1990), and are more appropriate for large superficial muscles and muscle groups with a common function (Christensen et al, 1995; Giroux & Lamontagne, 1990; Perry et al, 1981). Silver-silver chloride surface electrodes placed two to four centimeters apart over the midline of the muscle belly or motor point appear to be standard protocol (Baratta et al, 1988; Carlsoo et al, 1973; Christensen et al, 1995; Giroux & Lamontagne, 1990; Habinicht & Kissinger, 1994; Perry et al, 1981; Schuldt et al, 1993; Soderberg & Cook, 1983; Solomonow et al, 1987; VanRuijven & Weijs, 1990; Wojts & Huston, 1994; Woods & Bigland-Ritchie, 1993). Larger electrodes with greater spacing produce increased EMG data (Basmajian and DeLuca 1985) as a result of picking up signals from more motor units and the collection lower frequency components of the EMG signal

(Winter, 1990). Basmajian reported the most selective electrode arrangement is a bipolar configuration with a small detection surface and small spacing. However, Woods and Bigland-Ritchie (1993) found no alteration of EMG-force relationships when switching from monopolar to bipolar configurations. They maintained that this relationship depended more on physiological characteristics of the specific muscle than on the methodology.

Cross-talk from antagonist or surrounding muscles presents a problem that cannot be filtered out and may cause errors in analysis (Turker, 1993). If the EMG is quantified there will be less of a problem with cross talk (LeVeau and Andersson, 1992). However, Winter (1990) argues that surface EMG is not prone to cross-talk unless the target muscle is very small.

Good same day reliability was demonstrated by Giroux and Lamontagne (1990) for both surface and intramuscular wire electrodes and both static and dynamic contractions. No statistically significant differences in EMG recordings existed when electrodes were not repositioned. These findings were reinforced by Christensen et al (1995). Perry et al (1981) and Winter (1990) found that surface electrodes offered better reproducibility than indwelling electrodes.

In addition to specificity, indwelling electrodes may be preferred for their influence on impedance. A high input impedance of one ohm or greater and low skin impedance of 1,000 ohms or less produces the highest quality EMG signal. Input impedance varies inversely with frequency, and also depends on distance, size, and

direction of the electrode in relationship to the muscle fibers. As a result, indwelling electrodes result in a higher input impedance than surface electrodes and eliminate skin impedance (Winter, 1990). The use of active electrodes or skin preparation of shaving, abraiding, and cleaning with alcohol helps reduce the effects of skin impedance when surface EMG is employed (Basmajian & DeLuca, 1985; Winter, 1990). The amount of skin preparation will primarily depend on the instrumentation available.

Noise from the recording system can be factored out if it is determined before beginning data collection. Basmajian (1985) sites types and sources of noise as static electricity from power lines, radio, televisions, etc., thermal noise from electrodes, amplifier, power source, and filter, and artifact from motion of wires, electrode, and muscle fibers. Noise from motion is usually classified as low frequency noise. While the other sources tended to produce high frequency noise. Amplifier placement within ten centimeters and use of battery power minimizes noise from this source. Turker (1993) recommended that amplifier noise was less than five microvolts RMS as measured with a source resistance of 100 kilo-ohms and a bandwidth of 1-1,000 hertz. Other researchers expressed agreement with most of those parameters (Baratta et al, 1988; Christensen et al, 1995; Draganich et al, 1989; Giroux & Lamontagne, 1990; Habinicht & Kissinger, 1994; Schuldt et al, 1993; Soderberg & Cook, 1983; Wojts & Huston, 1994). To decrease noise or artifact associated with surface electrodes, Turker (1993) and Perry (1992) recommended the use of active electrodes in a bipolar configuration with a ground. It was also suggested that clean electrode contacts would create less thermal noise. Taping

down electrodes and wires, and minimizing the wire length of the leads have also been suggested. Noise can be identified by observing the EMG signal, since its waveform is fast or slow and undulating as opposed to the narrow frequency of muscle activity (Turker, 1993). Winter (1990) and Basmajian and DeLuca discussed the Nyquist frequency rule which states that the EMG signal should be recorded at twice the peak muscle frequency, which is approximately 350Hz when using surface electrodes. This frequency should capture the entire muscle signal while excluding higher frequency noise.

Signal Processing

The raw EMG can be analyzed directly or it can first be processed by full-wave rectification, integration, linear envelope, or root-mean-square. The raw signal is the basis of all methods of interpreting the myoelectrical activity of the muscle. The unprocessed raw signal should be monitored to detect and filter artifact. The raw signal also provides quantifiable interference patterns that represent muscle activity and can be used for statistical or higher order analysis (LeVeau and Andersson, 1992). Full-wave rectification is the absolute value of the EMG signal, such that all negative deflections are flipped to the positive. This rectified signal can be integrated to determine the area under the curve. Integration represents the trend of the signal or total muscle activity over time. The integrated EMG is related to the amplitude, duration, and frequency of motor unit action potentials (LeVeau and Andersson, 1992).

The linear envelope is similar to integration in that it represents a moving average

over time. Generally, the rectified signal is first low-pass filtered at a cutoff frequency of three to six Hz before the area is determined for linear envelope processing (Basmajian & DeLuca, 1985; Winter 1990). Similar to integration, the linear envelope represents the amount of muscle activity over time, but provides an instantaneous evaluation of the signal, rather than a continuous evaluation. It best reflects changes in state of contraction of the target muscles, and indicates onset, duration, and pattern of the muscle contraction (LeVeau and Andersson, 1992).

Root-mean-square (RMS) is an effective method of quantifying an alternating current. It measures the electrical power of the signal either as a moving average over time or as a linear voltage. As a time indicator, it provides an instantaneous measure of the power output of the myoelectric signal. RMS is related to the number of motor units firing, firing rates and durations of motor units, area of motor units, propagation velocity of electric signal, electrode configuration and instrumentation characteristics (Basmajian & DeLuca, 1985). It is independent of the effects of superposition of active motor units. RMS most completely and accurately represents motor unit behavior during a muscle contraction and is the preferred method of evaluating changes in myoelectric amplitude (Basmajian & DeLuca, 1985).

The most frequently used method to manipulate the processed signal is normalization, which involves expressing the EMG output in terms of some reference contraction. The most basic need for normalization is to accommodate the individual variation in the number and mixture of motor units sampled (Perry, 1992). This permits

valid comparison between muscles, individuals, days, or other studies. Normalization compensates for electrode re-application, thus improving between day reliability (Knutson & Soderberg, 1995; Giroux, 1990). Normalization also eliminates the dependency on muscle length and contraction rates (Baratta et al, 1988).

Summary and Implications for This Study

Four major types of joint mechanoreceptors influence muscle tone. The methods employed in this study should elicit effects of type I receptors. Type III may not have been active in the isometric methods employed in this study, since they are not active in immobile joints and are only active at the end range of collagen tension. Type II are also dynamic in nature, so isometric activity is unlikely to recruit them. However, small sways in balance and error in maintaining knee flexion may have caused enough joint movement to elicit participation of these receptors. No effects were elicited from type IV receptors, since normal exercise does not elicit pain.

No research was found that identified the specific point in knee range of motion where peak tibial translation occurs in closed kinetic chain. Without additional research, peak translation in closed kinetic chain is best estimated to occur at 15 degrees flexion, based on open kinetic chain findings. However, compressive forces involved in closed kinetic chain activity may alter this joint angle-ATT relationship. Trunk inclination and tibial rotation have also been shown to affect ATT and quadriceps activity and thus needed to be controlled. A smaller angle of knee flexion should have improved subjects' balance and maintenance of a vertical trunk position. Since quadriceps recruitment in

closed kinetic chain is lowest from 20 degrees to full extension, a slightly larger knee angle was chosen for this study. The closed chain activity under investigation should have produced hamstring co-contraction, and theoretically, this would limit ATT. However, research shows that the hamstrings cannot limit ATT from 30 degrees to full extension, where ATT may be at its highest. Since the forces applied in this investigation would only affect ATT, not resist knee flexion and extension, this study would not be influenced by the effects of location or amount of resistance on ATT and hamstring co-contraction. In summary, a knee angle of 30 degrees was chosen in an attempt to elicit the most ATT without compromising quadriceps activity.

The safety of the translatory forces applied in this study was demonstrated by Kapandji (1987), Kannus et al (1991), Race and Amis (1994), and Woo et al (1991). When age of the subjects included in this investigation is considered, at least 319 pounds of tensile force should be tolerated by the ACL, and at 30 degrees of knee flexion, the PCL of young adults should tolerate 900 pounds. Furthermore, the ATF of 20 percent of each subject's body weight was dispersed throughout the periarticular soft tissue structures, including the menisci, capsule, and collateral ligaments.

As stated previously, the individual muscle heads of the quadriceps group present slightly different patterns of activity (Basmajian & DeLuca, 1985). In unilateral stance, the rectus demonstrates minimal activity, and has been shown to be delayed until 40-60 degrees of knee flexion has occurred. The vasti are primarily active throughout the range of motion in closed chain. As a result, the rectus was not expected to demonstrate

significant EMG activity at the 30 degree position chosen for this study. The vasti should have displayed significant EMG activity.

Common qualities of current ACL rehabilitation protocols include closed kinetic chain exercise, early full weight bearing, early restoration of full knee extension, and minimizing ACL stress induced by ATT (Haffajee et al, 1972; Irrgang, 1993; Woods & Bigland-Ritchie, 1993). According to the stated aggressive ACL protocols, a squat activity would provide optimal healing, facilitate early functional strength gains by providing tension along the lines of stress, and increase motor unit recruitment. Although active squatting would have been more functional and aggressive, an isometric movement provided more control of variables for experimental conditions. Also, isometric EMG-force relationships appeared to be better understood and more valid than dynamic relationships.

Surface electrodes have been shown to have less associated noise for large superficial muscle groups, and to be more reliable than indwelling electrodes. Therefore, surface electrodes were most appropriate to measure the quadriceps muscle EMG activity. Normalization of the EMG output is required for intersubject, intermuscle, and between day comparisons. Since the subjects of this study provided their own control for intramuscle comparison, normalization was not required in this study. Since total recruitment, would best reflect a facilitation or inhibition of the quadriceps, area was the variable chosen to answer the research question under investigation. While the peak amplitude might have reflected total recruitment, it could have varied more with minor

perturbations of balance. RMS processing appeared to offer the most valid measurement of the area, since it is an indicator of power output of the EMG signal. Also, it is not dependent on cancellation due to superposition of active motor units.

Hypothesis

The authors hypothesized that application of an ATF during quadriceps contraction would significantly increase quadriceps EMG activity, in accordance with Grimsby's findings. Grimsby suggested that ATF activated the type I, II, and possibly type III receptors. He theorized that type III mechanoreceptors had a facilitatory affect on the quadriceps femoris muscle.

CHAPTER 3

METHODS

CHAPTER 3

METHODS

Study Site

The study was conducted in the therapeutic exercise laboratory of the Physical Therapy Department at Grand Valley State University. Approval to use this facility was obtained from the Director of the Physical Therapy Department. Subjects reported to the laboratory during January and February 1996 for a one-visit test.

Subjects

Forty-three volunteer subjects, 20 male and 23 female, were selected from among GVSU students and residents of the West Michigan demographic area. The participants ages were 19 to 44 years old, with a mean age of 28 years. Exclusion criteria of pregnancy within the last 6 months, pre-existing knee pathologies or surgeries, current low back, hip, knee, foot or ankle pain, joint laxity, and neurological disorders were screened via a questionnaire completed by each subject (See Appendix A). Each subject read and signed a consent form approved by the Human Subjects Review Board of Grand Valley State University (See Appendix B). Test sequence was assigned by drawing from a 43-card deck. Each card was labeled with a sequence, with each sequence equally represented. Right or left extremity was assigned in an identical manner. This method ensured random assignment with equal representation of sequences and extremities. Prior to data collection, ten additional subjects participated in a pilot study in order to identify and correct problems in data collection and processing.

Apparatus

Pulleys were used to create an anterior tibial transition. Two Vigor¹ adjustable height pulleys were mounted three feet apart facing toward each other. A colored foot print with demarcations for the malleoli and forefoot was positioned equidistant between the two pulleys, such that one pulley was in front and the other was behind the subject. A standard household scale with an analog reading was used to measure body weight, which determined the pulley load. A standard goniometer with clear plastic arms was used to measure the amount of knee flexion, and a plumb line provided a reference for trunk inclination. A multiple channel Myosoft/Myosystem 1200² EMG system and disposable silver-silver chloride surface electrodes were used to measure muscle activity. This system offered an input impedance of 20 mega-ohm to 1 giga-ohm, sensitivity of one microvolt, and recording bandwidths of 16 to 500 hertz. The surface electrodes were 1.0 centimeter diameter discs mounted on flexible adhesive and covered with a saline gel. The silver chloride provided a stable skin-electrode interface by diminishing the polarization, while the saline gel provided quality signal transmission (Perry, 1992).

Testing Procedures

EMG activity was recorded during all conditions using surface electrodes. The electrodes were placed on the vastus medialis, vastus lateralis, and rectus femoris muscles in a bipolar set-up. The vastus intermedius was omitted because it is a deep muscle and the surface electrodes would not be able to pick up activity from that particular muscle.

¹ Vigor Equipment Inc., 4915 Advance Way, Stevensville, Michigan 49126

² Noraxon USA, Inc., 13430 N. Scottsdale RD., Suite 104, Scottsdale AZ, 85254

For optimal signal reception, accepted skin preparation methods of shaving and vigorous cleaning with alcohol was employed. Since neither preamplification nor active electrodes were employed in this study, abraiding of the skin seemed appropriate for reduction of skin impedance. However, characteristics of the EMG system used eliminated this need. Noise from the recording system was automatically cancelled out by the recording system at the beginning of each test. A bipolar configuration and clean electrode contacts was also used to reduce the noise or artifact associated with surface electrodes.

Two electrodes were placed over each muscle belly with a center-to-center distance of 2.8 centimeters. Appropriate balance between selectivity and quantity of data required electrode spacing within two to four centimeters. Specific electrode placement was taken from Ericson, Nisell, Arborelius, and Ekholm (1985) and is presented in Table 3.1. The motor points used by other authors appeared to coincide with these locations, but Ericson et al (1985) were the only authors to provide specific measurements for location. The ground electrode was placed on the head of the fibula.

Table 3.1

Electrode Placement, Modified from Ericson, Nisell, Arborelius, and Ekholm (1985)

Muscle	Electrode position
Rectus femoris	50% of distance from ASIS to apex of patella
Vastus medialis	20% of distance from ASIS to medial knee joint line
Vastus lateralis	25% of distance from ASIS to lateral knee joint line

After being weighed, each subject was instructed to stand with their test foot directly over the colored foot print with demarcations bisecting the malleoli transversely and the second toe longitudinally. Athletic shoes were worn for support. One pulley cuff was then placed below the knee joint line, pulling forward on the tibia, and a second cuff was placed above the patella, pulling backward on the femur. Pulley height was adjusted so that the cables were perpendicular to their respective limb at 30 degrees of flexion. Each pulley was loaded with 10 percent of the subject's body weight, so that total ATF was 20 percent of body weight (See figure 3.1). This pulley load was chosen based on patient comfort and manageability for the testers. The load was applied slowly, in order to monitor the subject's response. All subjects were able to tolerate this load. Equal weight was maintained on each pulley so that only translational force was added, not resistance to knee flexion or extension. The addition of resistance could have created several confounding variables, such as increase contraction in the quadriceps, hamstrings, gluteal, or adductors muscles, which in turn may have affected ATT.

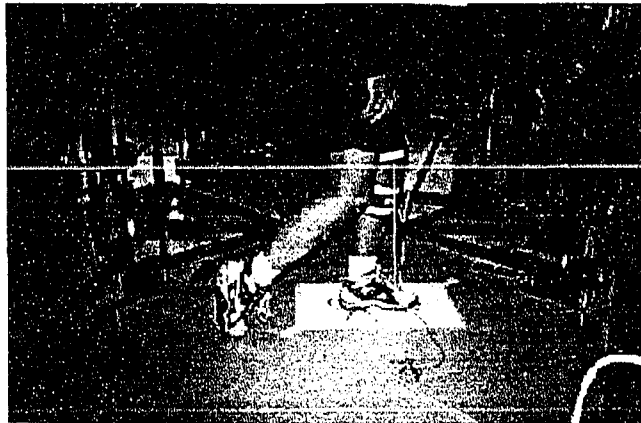


Figure 3.1: Pulley set-up for anterior force conditions with subject positioned in a 30 degree squat.

Each subject was instructed to maintain a unilateral squat on the randomly selected lower extremity while maintaining an erect trunk and without purposefully contracting the thigh muscles. The non-tested leg was extended back in a posterior toe-touch placement to help ensure the subjects balance. Subjects were instructed not to bear weight on this extremity (See figure 3.1). Each subject was then instructed to squat on their front (tested) leg to 30 degrees knee flexion as measured by a goniometer aligned with the bony landmarks of the lateral midline of the femur, using the greater trochanter for reference, and lateral midline of the fibula, using the lateral malleolus for reference. While each subject maintained the 30 degree squat position, EMG measurements were recorded for a three second interval for both the ATF and NTF conditions. The ATF EMG measurement was taken with the tension of the pulleys engaged (See figure 3.1), and the NTF EMG measurement was taken with the tension of the pulleys released (See figure 3.2). These tests were performed consecutively in the preselected random sequence, without rest in between.

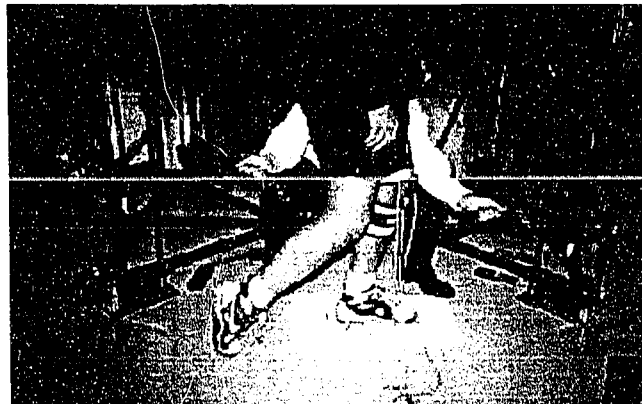


Figure 3. 2: Removal of pulley tension for no tibial force conditions with subject positioned in a 30 degree squat.

Immediately after the EMG recording of the second squat condition, a resting EMG measurement was taken without tension on the pulleys. The subject transferred all their body weight to the nontested limb while maintaining trunk and test-foot position. The resting measure was taken last for convenience, to reduce total testing time, and to eliminate repositioning the knee for the two different squat conditions. This helped control variations in knee flexion and internal and external rotation of the tibia relative to the femur. The total testing time was approximately 30 seconds, including adding or removing pulley tension, and repositioning the subject for the resting measure. All EMG data was recorded at a frequency of 1,000 hertz. This frequency met the specifications of the Nyquist frequency rule, described earlier, and was determined based on the frequency demonstrated in the pilot study. Collection at this frequency also maintained the option of analyzing the frequency spectrum for identification of fatigue, if necessary.

CHAPTER 4

RESULTS

CHAPTER 4

RESULTS

Data Analysis

EMG data analysis was conducted on only the middle one-second of each three-second test in order to decrease the influence of timing errors. This procedure also allowed time for the subjects to stabilize their balance, but not enough time to produce fatigue. Since visual examination of the raw EMG signal did not reveal amplitude changes that would suggest fatigue and testing time was short, frequency analysis was deemed unnecessary.

The raw EMG signal was converted to root mean square (RMS) using a bandwidth of 15 data points, and the area under this curve was determined for further analysis. This bandwidth was chosen based on the small time period under analysis, and visual comparison of the RMS conversion at a variety of bandwidths. Outliers were then identified as EMG areas falling beyond a distance of 1.5 times the interquartile range from the upper and lower boundaries of the interquartile range. These outliers were reset at a value equal to the extreme end of the interquartile range plus or minus one unit. A unit was defined at one decimal place. This process is known as Winsorizing, and it preserved the trend of the outliers without skewing the data. The resting EMG measurements demonstrated many obvious outliers, and it appeared that many subjects had difficulty relaxing the leg for a true resting measure. In fact, the values in three of the

resting measures were so high that using them would have resulted in a negative EMG value for the no force condition. Since many resting EMG measurements were invalid, they were not subtracted from the EMG values of the two squat conditions, nor were they considered in the final analysis of data.

Many subjects did demonstrate a resting EMG near or equal to zero, so resting EMG levels were assumed to be zero microvolts. Thus, the EMG area produced by the two force conditions was analyzed at the measured value, with no adjustment for rest. Differences between the two force conditions were therefore analyzed immediately after management of outliers in the processed EMG signals. The percentage change from no tibial force (NTF) to anterior tibial force (ATF) conditions was calculated, using NTF values as the reference value. Winsorizing was also used to manage the outliers in percent change. Winsorizing procedures were the same as previously described, except that “one unit” was defined as two decimal places. Finally, repeated measures analysis of variance (ANOVA) of the mean percentage change from NTF to ATF conditions over all muscles and all subjects (“grand mean” percentage change) was performed to assess any difference in the area between these two conditions. The level of significance was set at $p < 0.05$. A significant increase in EMG activity during the ATF conditions would support the hypothesis of this study and support Grimsby’s findings. A decrease under these conditions would support the current literature that suggests ATT produces quadriceps inhibition.

Results

The raw data showed a wide variation in recruitment levels between individuals

which was evidenced by large standard deviations. This variation created many outliers. Application of the Winsorizing technique permitted the inclusion of these outliers without skewing the data. The mean EMG areas across all subjects under NTF conditions was 13.056 microvolts-squared for the rectus femoris, 28.521 microvolts-squared for the vastus medialis, and 45.565 microvolts-squared for the vastus lateralis (See Table 4.1). Standard deviations were 07.022, 16.041, and 26.094 respectively. The grand mean EMG area across all three quadriceps muscles and all subjects under these conditions was 29.047 microvolts-squared.

Table 4. 1

Mean EMG Area (microvolts-squared) During an Isometric Squat Under No Tibal Force Conditions, after Winsorizing

Muscle	Mean	Standard Deviation
Rectus Femoris, n=43	13.056	07.022
Vastus Medialis, n=43	28.521	16.041
Vastus Lateralis, n=43	45.565	26.094
Average for all Muscles	29.047	14.990

Under ATF conditions, the mean EMG area across all subjects was 14.388 microvolts-squared for the rectus femoris, 33.453 microvolts-squared for the vastus medialis, and 49.205 microvolts-squared for the vastus lateralis (See Table 4.2). The

standard deviations were 07.104, 18.884, and 27.330 respectively. The grand mean EMG area under ATF conditions was 32.349 microvolts-squared.

Table 4. 2

Mean EMG Area (microvolts-squared) During an Isometric Squat Under Anterior Tibial Force Conditions, after Winsorizing

Muscle	Mean	Standard Deviation
Rectus Femoris, n=43	14.388	07.104
Vastus Medialis, n=43	33.453	18.884
Vastus Lateralis, n=43	49.205	27.330
Average of all Muscles	32.349	16.080

Analysis of Variance

Analysis of variance indicated a significant effect of anterior tibial force on EMG activity, $F(1,42) = 13.32$, $P = 0.001$. Similarly, EMG activity varied significantly with the component quadriceps muscle being tested, $F(2,84) = 76.84$, $P < 0.001$. However, there was no significant difference in the influence of anterior tibial force on each of the component quadriceps muscles, $F(2, 84) = 2.50$, $P = 0.088$. Although the calculated EMG area and microvolt change between conditions varied among the tested muscles, the *percentage* difference between conditions was fairly consistent across muscles. As a result, the percent change in EMG area between the two conditions was not significantly

different between muscles, $F(2,84) = 2.42$, $p = 0.095$.

The trend in EMG activity by muscle and by condition was also consistent.

Under both NTF and ATF conditions, the rectus femoris demonstrated the least activity, the vastus medialis demonstrated moderate activity, and the vastus lateralis demonstrated the most activity. The individual activity in each of these muscles was higher under ATF conditions than NTF conditions. The increase in mean EMG area from NTF to ATF conditions across all subjects (See Figure 4.1) was 01.332 microvolts-squared for the rectus femoris, 04.932 microvolts-squared for the vastus medialis, and 03.640 microvolts-squared for the vastus lateralis. These figures represent an average increase of 03.299 microvolts-squared from NTF to ATF conditions.

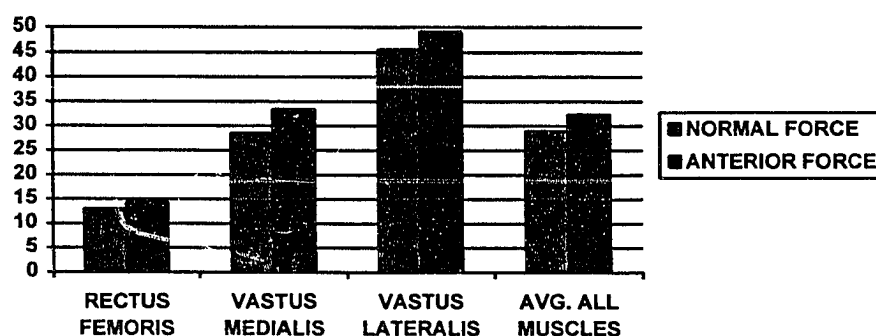


Figure 4. 1: Mean EMG area (microvolts-squared) for no tibial force conditions compared to anterior tibial force conditions across all subjects.

Although the particular muscle tested did not influence the percentage change in EMG area between test conditions, force condition continued to demonstrate a significant effect, $F(1,42) = 33.48$, $P < 0.001$. These results demonstrated an increase in quadriceps

activity from NTF conditions to ATF conditions. Rectus femoris showed a mean 12.1 percent increase in EMG area, while vastus medialis and lateralis increased 18.1 percent and 10.5 percent, respectively (See Table 4.3). The standard deviations were 0.154 for rectus femoris, 0.216 for vastus medialis, and 0.239 for vastus lateralis. These findings represented a mean increase in EMG area of 14 percent from NTF to ATF conditions.

Table 4.3

Mean Percentage Increase in EMG Area from No Tibial Force Conditions to Anterior Tibial Force Conditions, after Winsorizing

Muscle	Mean % Increase	Standard Deviation
Rectus Femoris, n=43	12.1	0.154
Vastus Medialis, n=43	18.1	0.216
Vastus Lateralis, n=43	10.5	0.239
Average of All Muscles	14.0	0.150

CHAPTER 5
DISCUSSION

CHAPTER 5

DISCUSSION

Purpose

The purpose of this study was to determine whether anterior tibial translation (ATT) facilitated quadriceps recruitment during closed kinetic chain exercise as suggested by Grimsby, or inhibited it as current literature reported. Recruitment was evidenced by EMG activity. Although the subjects of this study demonstrated a wide variety of EMG output, EMG activity was consistently higher under anterior tibial force (ATF) conditions. The grand mean EMG area for all subjects and all muscles under ATF conditions was 32.394 microvolts-squared. Under no tibial force (NTF) the grand mean was 29.047 microvolts-squared. These results reflect an average 14 percent increase in EMG activity from ATF conditions as compared to NTF conditions. Thus, the results of this study clearly supported Grimsby's findings that ATT facilitated quadriceps recruitment as measured by EMG. However, Grimsby's theory that the type III receptors were responsible for the facilitation of the quadriceps was not demonstrated. The isometric conditions and the relatively low ATF of 20 percent body weight applied in this study was insufficient to recruit type III mechanoreceptors which are activated by dynamic movement and end range collagen tension (Grabiner, 1993; Wyke, 1994; Irrgang, 1993). Type II receptors are also dynamic (Grabiner, 1993; Guyton, 1986; Irrgang, 1993; & Wyke, 1994), so they would not have been recruited during the isometric conditions.

Therefore, these results implicate type I receptors, which respond to beginning and end range stretch and compression during static and dynamic activities.

Discussion of Results

The facilitation of the quadriceps femoris through ATT seems to contradict the concept of an ACL-hamstring synergy that inhibits the quadriceps when ATT increases, at least in closed kinetic chain. However, this does not refute the theory that ATT may also facilitate or inhibit the hamstrings under certain conditions. Both inhibition through the ACL-hamstring synergy and facilitation through ATT may depend on the amount of ATF or actual amount of ATT. Changes in quadriceps and hamstring moment arms produced by changes in knee angle may also be responsible for changes in muscle activity, as suggested by Barratta et al (1988). However, this study utilized isometric muscle activity which eliminated moment arm changes. As a result, moment arm changes could not have produced the EMG changes that were demonstrated.

The raw data showed a wide variation in recruitment levels and patterns between individuals and a wide variety of outliers. Under both NTF and ATF conditions, the rectus femoris demonstrated the least activity, the vastus medialis demonstrated moderate activity, and the vastus lateralis demonstrated the most activity. The low rectus femoris activity may be explained by looking at the pattern of quadriceps recruitment during a squat, as described by Basmajian and DeLuca (1985). They reported that the rectus femoris is active when knee flexion is greater than 40 to 60 degrees. In the presented study, the subjects were instructed to maintain 30 degrees of knee flexion.

Unfortunately, this amount of knee flexion fell short of the amount required to facilitate the rectus femoris muscle. Also, the rectus femoris muscle is a two joint muscle. It functions as a hip flexor as well as a knee extensor, so this muscle may not be the first one recruited during knee extension. Since the vastus medialis and lateralis are primary knee extensors, they would be recruited first, creating a higher EMG output. Also, the rectus femoris is known as a modulator, rather than a prime mover. It is recruited primarily by changes in speed (Rancho Los Amigos, 1996) and would not have been optimally recruited by the isometric conditions used in this study. Considering that the knee angle was less than 40 degrees and the subjects performed an isometric task, the question arises as to why the rectus femoris was recruited at all. One possibility is that Basmajian and DeLuca's (1985) description of rectus femoris activity may not apply to isometric conditions. A second possibility is that balance perturbations may have called upon the rectus as a modulator. The most likely explanation is found in Lombard's paradox which describes simultaneous recruitment of the rectus femoris and hamstring muscles during rising.

The higher activity demonstrated by the vastus lateralis compared to the vastus medialis was also an unexpected finding. Basmajian and DeLuca (1985) reported that the vastus lateralis and medialis displayed similar activity during a squat. Again, isometric conditions may recruit the quadriceps muscles in different ratios than active squatting. Foot position may also have affected the output of the vasti, since internal tibial rotation and external tibial rotation increase the activity in the vastus lateralis and medialis,

respectively (Soderberg & Cook, 1983). Alignment of the foot in the saggital plane was a relative internal rotation compared to the normal anatomical resting position of slight external rotation. This relative internal rotation would also increase the stretch on the ACL and may have increased the level of vastus lateralis muscle activity in this study. It may also explain the discrepancy between these results and the reports of Basmajian and DeLuca (1985), if anatomical positioning was used in their research.

Several other factors may have also contributed to the variations in EMG output between muscles and the variety of outliers found in this study. Variations between subjects recruitment patterns and the variety of outliers may have resulted from variations in subjects' muscle fiber composition, skeletal structure, or muscle recruitment pattern. Since the quadriceps muscles are mixed in composition, individual proportions of slow twitch, fast twitch, and intermediate fibers, are known to vary somewhat between individuals. Skeletal variations may have included genu varum or valgum, excessive pronation or supination of the ankle, or muscle imbalances between hip and knee internal and external rotators. These structural variations could have altered stresses to the knee and hip joints and influenced muscle recruitment which would have resulted in the differences in the EMG readings. Trunk inclination, balance, and amount of weight bearing on the untested lower extremity may have also influenced EMG output of the quadriceps femoris muscles. These influences will be discussed in detail, under the "limitations" section.

Reliability and Validity

Overall, the reliability and validity of the methods employed in this study are

supported by the literature regarding surface EMG, and the acceptance of goniometric reproducibility within five to ten degrees. The high statistical power indicated by the low p-values and high f-statistics also infers reliability and validity. However, a large number of outliers and a wide variation in EMG recruitment patterns suggest a lack of control which will be discussed later. Unfortunately, this study did not include a test of reliability of the procedures used, making proof of reliability in the presence of such lack of control difficult. A repeated measurement test of reliability would have verified the reliability of our measurement tools, but a stable variable with good control was required for validity with such a test. Fortunately, the lack of reliable data was compensated for by the large number of subjects that participated in the study. The use of means from a considerable number of subjects compensated for the lack of control by averaging out the effects of random variation. Although this averaging over a large subject pool could not alter the reliability of the research methods, it enhances the statistical power, despite questionable reliability of the data collection methods. Further, the extremely high F-statistic and low P-value indicated that the changes produced in this study are unlikely to have resulted from chance.

Reliability of the measurement tools used in this study was considered high, because comparisons were made within subjects, not between subjects, electrodes were not repositioned at any time during the testing, and the testers did not change jobs during the testing of any one subject. For these three reasons, this study required only intratester reliability of the equipment and methods involved. Two testers were utilized with each

subject. One tester prepared the pulley apparatus and instructed the subject. The second tester applied the electrodes, attached the leads to the machine, and collected the EMG data. The intratester reliability of goniometric measurements has been generally accepted within approximately five to ten degrees. Good same day reliability was also demonstrated for surface electrode under both static and dynamic conditions by Giroux and Lamontagne (1990) and Christensen et al (1995). Giroux and Lamontagne (1990) determined that no statistically significant differences existed when electrodes were not repositioned between measurements. Based on these facts, good reliability would be expected from this study. Any lack of reproducibility of the EMG measures taken in this study, is more likely contributable to random variation rather than reliability of measurement tools or methods.

Construct validity was obtained through control of bias. Subject and tester bias was reduced through consistency in instructions. Verbal and tactile instructions (Appendix C) were given in a consistent manner during all conditions in order to prevent differences in encouragement or discouragement that could affect the subject effort. Subject bias was further reduced by blinding the subject to the EMG readings and to the hypothesized relationship between ATF and EMG activity. Experimenter bias was further reduced by the use of an EMG machine that produced objective numerical data. Operational definitions also increased the construct validity of this study since both ATF and EMG activity are both unidimensional constructs. Construct validity of this research project should be considered very high.

Internal validity was achieved through the controls described in chapter three for biomechanical influences on quadriceps activity and ATF such as trunk inclination, tibial rotation and degree of knee flexion. These controls attempted to limit the independent variable to one: the pull/load on the tibia. Based on the wide variation in EMG output demonstrated by the subjects and the number of identified outliers, these controls may not have been sufficient. Major areas that lacked control will be discussed in the proceeding “limitations” section. Less powerful variables that may have threatened control, as well as several successful methods of eliminating variables, will be discussed here. Single-session testing for all sessions helped to eliminate maturation and attrition as confounding variables. Testing effects such as learning could still have influenced data, especially with respect to unilateral balance. Randomization of the order in which the two conditions were applied and randomization of the assignment of the leg to be tested controlled for leg dominance, learning effect, and fatigue. Individual knee laxity may have been another confounding variable. A higher degree of laxity in the collateral ligaments could have allowed a greater amount of translation to occur, while reduced laxity or hypomobility could have restricted translation. This could have affected the amount of proprioceptive output which in turn would have affected the EMG output. Comparison within subjects and screening for pathology and other conditions that may have effected laxity may have helped to prevent this variable from influencing the data. Finally, subject selection was not truly randomized, since most subjects were Grand Valley State University students. Therefore, influences of age and activity level may

have introduced bias and additional variables. However, these factors were not expected to strongly influence the two major variables of joint translation or EMG output. Overall, many potential variables were well controlled. As stated previously, it was assumed that averaging data from a large subject pool would account for any remaining random variation. Therefore, the results of this study should be valid, despite some apparent lack of control and unlikely reproducibility of individual EMG measurements.

Limitations

It became apparent throughout the testing procedures that some variables lacked control. The primary problems were the amount of weight bearing on the untested leg, maintenance of unilateral balance, and the ability to relax during the resting measure phase. The amount of weight bearing would affect ATT. Weight bearing may have been controlled by the use of a scale which could have monitored the amount of force applied through toe touch weight bearing. Fluctuations in weight bearing could have affected the EMG output of the various muscles tested, creating a confounding variable when analyzing the cause and effect relationship regarding EMG changes. Full weight bearing on the tested lower extremity was the optimal situation. However, decreased weight bearing could have decreased EMG output and increased weight bearing could have increased EMG output. Subjects who had difficulty with balance in a unilateral stance may have increased their weight bearing on the nontested leg, or simple lack of kinesthetic awareness may have influenced weight bearing.

Many subjects demonstrated and reported difficulty with unilateral balance. Also,

most subjects stated that they felt more secure when the pulley load was applied. The pulleys may have supplied some kinesthetic awareness or tactile cues that helped maintain erect posture, and they may have also provided anteroposterior stability for balance. The body naturally sways when standing to maintain its center of gravity. Even minor losses of balance will activate some of the muscles in the lower kinetic chain in order for the body to readjust and maintain an erect posture. Sway may have resulted in subtle changes in lower extremity and trunk alignment that in turn may have influenced the firing of the quadriceps muscle or hamstring muscles or altered tibial translation.

Other influential factors on EMG activity might include variations in knee angles, tibial rotation, and trunk inclination. As previously discussed, tibial translation, quadriceps muscle activity, and hamstring muscle activity are related (Bagger et al, 1992; Basmajian & DeLuca, 1985; Draganich et al, 1989; Howell, 1990; Smidt, 1973; Solomonow et al 1987). The degree of trunk inclination was only loosely controlled to maintain an up right posture. Trunk forward inclination may have displaced the center of gravity, altering the flexion and extension moments and torque on the quadriceps. A decrease in the flexion moment of gravity would have required less quadriceps force, resulting in reduced EMG activity. The resultant changes in muscle firing could have been one reason for the changes in EMG output of the quadriceps muscle groups. Elimination of balance issues as variables was attempted by analyzing only the middle one second of data for each condition. This procedure allowed the subjects time to find a balanced position, but not enough time to fatigue. Analysis of the middle second also

accounted for timing errors, since the EMG was not switch activated.

The resting measures were also a problem. Three subjects "resting" EMG were higher than the EMG values for the NTF conditions. A more accurate resting measure could have been obtained in supine or sitting. Unfortunately, the equipment available to produce translational forces could not accommodate supine or sitting. The authors compensated for this problem by having the subject shift all of their weight to the nontested leg to "unweight" the tested leg. The subject was still instructed to maintain foot position and erect posture. Again, a scale may have been one way to verify unweighting of this leg. Another option may have been to step onto a small step and allow the tested leg to dangle during resting measurements. Without an accurate way to measure the resting state of the tested lower extremity, resting was assumed to be zero. However, a slight amount of EMG activity often occurs at rest. The inclusion of this resting EMG in the values for the test conditions would not have effected the difference between the two test conditions, since the same resting value was present in both conditions. The resting value may have affected the percent change, by inflating the NTF values used as the reference for calculating percent change. Dividing by a potentially larger number may have resulted in a smaller percentage. However, this smaller percentage was statistically significant, so a larger percentage certainly would be significant as well.

Another random variable may have been the subjects foot position. Variation in this position was decreased by the use of floor markings that were consistent between

subjects and conditions. However, accurate foot position without consideration of trunk alignment with respect to the frontal plane did not control tibial rotation alignment. Further, fixing the distal end of an extremity does not control the proximal end. Although subjects were instructed and observed to align their knee over their foot, balance and kinesthetic awareness could have permitted internal rotation of the tibia. Small variations in rotation may not have been visible to the testor or obvious to the subject. Internal rotation would have require more recruitment from the vastus lateralis muscle group or might have decreased ATT and increased stretch on the ACL.

This lack of control resulted in many outliers in the data collected. These outliers may have resulted from normal neuromusculoskeletal variations between subjects, as discussed previously. Outliers may also have resulted from methodological problems. Even though they were accounted for through Winsorizing, the outliers may still have affected data analysis. Lack of normalization of EMG data also presented some limitations to this study. Lack of normalization limited the capability to compare the results to other studies and limited generalizability and reproducibility. A comparison cannot be made unless the units are the same. The basic unit in this study was microvolts, but normalized data utilizes percent as the basic unit, most commonly percent MVC. Based on the results of this study, it would be accurate to state that ATF facilitated the quadriceps muscle in a unilateral stance during a 30 degree isometric squat. This study estimated the microvolt increase at 14 percent above that for normal conditions. Unfortunately, this 14 percent cannot be directly compared to percent MVC

increases found in Grimsby's unpublished study. However, the result of this study support the trend of Grimby's findings that showed a 38 percent increase in peak EMG output during the performance of an anterior tibial force facilitated partial squatting exercise.

The amount of anterior tibial force, if any, was not measured in this study due to the inability of placing both an arthrometer and the pulleys around the knee joint simultaneously. However, visible and palpable ATT was identified when ATF was applied using the pulley system described in this study. The amount of translation varied among individuals even though 20 percent body weight was consistently used for the ATF. Variations in joint laxity between subjects may have resulted in different amounts of translation under body weight loads. This study did not attempt to determine the amount of anterior tibial force required to affect the quadriceps EMG, since the applied force was not varied, nor was the amount of ATT measured. The necessary amount may vary between normal and pathological joints.

Clinical Implications

This study clearly demonstrated greater recruitment of the rectus femoris, vastus medialis, and vastus lateralis through ATT under isometric squat conditions. The use of ATT to facilitate quadriceps muscle recruitment can be used on patients with neurological pathology, after release from immobilization, following surgical repairs and with other conditions that produce quadriceps strength and recruitment deficits. Application of ATF can provide proprioceptive input that helps to retrain neurologic control and normal

recruitment patterns, tension that optimizes collagen production for ligamentus strength, and muscle activity that increases contractile strength, endurance, blood flow, and nutrition to all tissues. The increased proprioceptive input caused by the induced ATT could be used as an alternative to or in conjunction with electrical stimulation of the quadriceps muscle fiber. The use of ATT through application of an ATF instead of electrical stimulation would result in less equipment, one less treatment modality, faster functional return, less discomfort for patient, and less time required of the therapist during a treatment session, thus reducing the cost of rehabilitation. The use of ATT to facilitate muscle strengthening, muscle re-education, and collagen production along the lines of stress would also lead to decreased duration of rehabilitation.

This facilitory effect of ATT also challenges current ACL rehabilitation protocols that attempt to minimize ATT which then deprives the patient of this facilitation. Protection of the injured ACL or other tissues from excessive ATF must be balanced with facilitation of muscle fiber recruitment and production of tension along the lines of stress in order for optimal healing to occur. The approach used in this study and by Grimsby would accomplish this goal. Although an ATF of 20 percent should be safely tolerated, the therapist should consider the relative tissue or graft strength during each stage of healing before choosing to utilize ATT as a method of quadriceps facilitation. An ACL graft is known to be weakest at approximately weeks seven through nine, following reconstruction. Further study regarding the degree of translation produced by these methods of applying ATF should be done to determine the safety during periods of graft

weakness. Also the attempted prevention of ATT during open chain knee extension exercise by ACL deficient patients should be reconsidered at least during the early and later stages of rehabilitation when graft strength is high.

While the methods employed in this study may not be readily available in many facilities due to the need for an adjustable height, two-pulley system, the use of theraband or theratubing may be a viable alternative. Translatory forces could also be altered through other methods such as open versus closed chain activities, distal versus proximal placement of resistance, amount of resistance to knee extension, and type of muscle contraction. Isometric knee extension in open kinetic chain with distal placement of resistance has demonstrated the greatest amount of ATT and ACL stress when compared to other common exercise conditions (Fu et al, 1992; Howell, 1990; Irrgang, 1993; Jurist & Otis, 1985; Voght, Bell, & Rhodes, 1991; Yack et al, 1994). Factors influencing ATT, described above, can be varied in a sequence that will gradually increase translation to the maximal level, according to the patients tolerance and stage of healing. For example, the isometric methods employed in this study may be useful in the earlier stages of quadriceps rehabilitation, when conscious control over quadriceps recruitment, stability, and coordination are just beginning to develop. This adaptability increases the safety of ATT as a method of facilitation and makes it applicable to more patients. Both orthopedic and neurologic patients may gain control, coordination, and strength by incorporating ATF into functional exercise regimes. Furthermore, ATF can be applied by a variety of methods, many of which are accessible to the patient for their home program.

Recommendations for Further Research

The conclusions drawn from this study can only be generalized to closed kinetic chain isometric squatting activity of healthy individuals without pre-existing knee pathology. Further study should test the assumptions made here on subjects with neurological pathology, status post immobilization, after surgical repairs and other conditions that produce quadriceps weakness. An additional study should also compare the effects of ATT under different exercise conditions, such as open chain activities and dynamic activities. Other muscles and joints should be tested to determine the effect on antagonist and synergistic muscles, and explore the application of Grimsby's theories to other joints. Since all joints have proprioceptors, it follows that facilitation of the normal joint translation, and thus the normal proprioceptive input, may facilitate muscular recruitment at any joint. Testing the effects of posterior tibial translation under all of the above mentioned conditions is also recommended. Attempts should also be made to quantify the amount of anterior or posterior translation required to produce a response and the amount of ATT produced by various amounts of ATF. This study incorporated submaximal muscular contractions. It would be interesting to assess the effects of ATT on maximum EMG during isometric and isotonic contractions. If ATT could increase maximum muscular forces, applications could be made to strength training programs and athletic populations.

Replication of this study using normalized EMG could be compared to similar studies, such as Grimsby's study on subjects with ACL pathology in dynamic activity.

This comparison would increase the relevance to other populations and conditions if the outcomes are similar. Such replication would also reinforce the validity of the hypothesis that ATT facilitates quadriceps femoris recruitment.

Conclusion and Summary

Under both NTF and ATF conditions the vastus lateralis presented the most EMG activity, followed by the vastus medialis, and finally the rectus femoris. The individual activity in each quadriceps muscle was higher under ATF conditions than NTF conditions. Although the calculated EMG area and microvolt change from NTF to ATF conditions varied between muscles, the *percentage* change between conditions was fairly consistent across all muscles. Despite the identification of many limitations and variables that lacked control, the P-value and F-statistic indicated a highly significant relationship between tibial force and EMG output. ATF conditions demonstrated palpable and visible ATT, therefore, it was valid to conclude that the ATT was directly responsible for the increase in EMG activity demonstrated in this study. This conclusion supports the authors' hypothesis that ATT increases quadriceps femoris recruitment. These results have direct clinical implications in the rehabilitation of patients that have quadriceps femoris recruitment or strength deficits.

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APPENDIX

APPENDIX A

GRAND VALLEY STATE UNIVERSITY
DEPARTMENT OF PHYSICAL THERAPY
HEALTH SCREENING QUESTIONNAIRE

FOR

MASTERS THESIS: THE EFFECT OF ANTERIOR TIBIAL TRANSLATION
ON QUADRICEPS RECRUITMENT DURING AN ISOMETRIC SQUAT.

Date: ___/___/___

Name (please print): _____

First 5 Digits of SS#: _____ Age: _____ Sex: Male () Female () (check)

CIRCLE THE APPROPRIATE ANSWER TO EACH QUESTION BELOW:

1. Do you currently have any pain, discomfort, or dysfunction in your

low back? Yes / No
hips? Yes / No
knees? Yes / No
ankles? Yes / No
feet? Yes / No

2. Have you ever had any injuries or surgeries to your knees? Yes / No

If yes, please explain. _____

Have you fully recovered from your injury or surgery? Yes / No / NA

3. Has your doctor ever recommended orthotics? Yes / No

Are you wearing them? Yes / No / NA

4. Have you ever had any neurological problems? Yes / No

If yes, please explain. _____

5. Are you pregnant now or have you been pregnant in the last 6 months? Yes / No

6. Do you consider yourself unusually flexible or double-jointed? Yes / No

Subjects signature Date

Witness Date

FOR RESEARCHERS USE ONLY:**Pulley Load:**

Body Weight: _____ X .20 = _____

Load Used: _____

Electrodes:

ASIS to Lateral Joint Line: _____ 25%: _____

ASIS to Patellar Apex: _____ 50%: _____

ASIS to Medial Joint Line: _____ 20%: _____

Pulley Placement:

Floor to Joint Line: _____ + 6 = _____ Femur

- 12 = _____ Tibia

Sequence: No - Ant.

Ant - No

APPENDIX B

GRAND VALLEY STATE UNIVERSITY DEPARTMENT OF PHYSICAL THERAPY INFORMED CONSENT

PROJECT TITLE: THE EFFECT OF ANTERIOR TIBIAL TRANSLATION ON QUADRICEPS RECRUITMENT DURING AN ISOMETRIC SQUAT.

I, _____, freely and voluntarily agree to participate in the research project under the direction of Natalie Howard, SPT, and Penny P. Tussing, SPT, to be conducted at Grand Valley State University therapeutic exercise laboratory. I understand the following to be true:

1. The study is being conducted in order to examine the changes in muscle contraction produced by forward movement of the shin bone. The conclusions drawn from this study may be used to improve physical therapy treatment of the knee joint and thigh muscles.
2. I have been selected for this study because I do not have any current or previous injury, pain, surgeries or other difficulties involving my lower back, hips, knees, ankles, or feet, nor have I recently been pregnant.
3. I will be required to hold a partial squat position for approximately 15 seconds. One-half of this time will include the use of a two-pulley system, with one strap placed above my knee and one placed below my knee. The weight on each pulley will be equal to 20 percent of my total body weight, but will be decreased if it causes significant discomfort.
4. The muscle activity will be recorded by seven electrodes that will be placed on the skin of the front of my thighs and outside of my calf. The skin surface under the electrodes will be shaved and vigorously rubbed with alcohol to assure good contact. Wires will attach the electrodes to a monitor, where current will be received and measured, NOT delivered to the subjects.
5. Total testing time is estimated at 45 minutes or less.
6. Testing procedures are not expected present any risk of injury; however, mild knee joint or muscle soreness lasting one or two days may occur.
7. I will report to the tester any feelings of pain or discomfort that may develop during the test.
8. I have the right to discontinue my testing or inclusion in this study at any time, for any reason, and without penalty.

9. The test results will be used in a Master's thesis for students in the Grand Valley State physical therapy program, but all subjects names will be strictly confidential.

10. I will have the opportunity to ask questions or contact either tester regarding the study at anytime, and to have these questions answered to my satisfaction. The phone numbers at which the testers maybe contacted are: Penny Tussing (616) 538-3441, Natalie Howard (616) 785-1560. I may also contact the physical therapy department at Grand Valley State University at (616) 895-3365.

12. Grand Valley State University is in no way responsible for the administration of research involved with this project. Grand Valley State University is in no way liable for any compensation for any time volunteered or evolving from my participation in this project.

I acknowledge that I have read and understood the above page, and based upon this information, I am voluntarily agreeing to participate in the study.

Signature of participant

Date

Signature of witness

Date

Please check here _____ if you are interested in receiving a summary of the study results. Address (please print): _____

Phone: () _____

I have explained and defined in detail the research procedures to which the subject has consented to participate.

Signature of researcher

Date

Signature of Witness

Date

APPENDIX C

SUBJECT INSTRUCTIONS

1. One tester will be operating the EMG machine/computer and applying the electrodes. This tester will be concerned with only the EMG set-up and operation and will not offer any instruction to the subject regarding the pulley or the squat. Pulley set-up, subject positioning, and subject instruction will be the job of the second tester.
2. Stand in the middle of the two pulleys with your right (left) foot on the colored foot print. Your ankle bones should line up with black (blue) lines. Your second toe should line up with the top line.
3. (Set up the pulley apparatus.) Are the pulleys tolerable? Are they causing discomfort or pain?
4. Listen to these instructions: You will place your left (right) nontested leg behind you, with toe-touch for balance. Put as little weight as possible on this foot. Try to put all of your weight on your right (left) foot. Keeping your back erect, you will squat down slowly until I tell you to stop. I will be measuring the angle at your knee with this goniometer. You will then hold the position until I say stand up. To avoid confusion, do not listen to the other tester. I will let you know when the test is complete. You may then stand up and put your weight on both feet, but **DO NOT** move your right (left) foot. Any questions?
5. (Using a goniometer and plumb line, position subject in a partial squat.)
6. (Signal the EMG operator when the subject is ready.)
7. When the EMG operator indicates the test is complete tell the subject to stand up but do not move your right (left) foot.
8. You have completed the tests, thank you for your participation. Do you have any questions or concerns?
9. Remove the apparatus from the subject, answer any questions, and escort the subject from the room.