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Characterizing Sensory-Mediated Changes in Human Movement: Studies on Flexibility and Joint-Related Pain

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CHARACTERIZING SENSORY-MEDIATED CHANGES IN HUMAN MOVEMENT:
STUDIES ON FLEXIBILITY AND JOINT-RELATED PAIN

by

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ABSTRACT

Capobianco, Robyn Ann (Ph.D., Integrative Physiology)

Characterizing sensory-mediated changes in human movement: studies on flexibility and joint-related pain

Thesis directed by Professor Roger M. Enoka

Human movement is controlled by the dynamic interplay between sensory input and motor output. Varying sensory input, either deliberately through manipulation, or unintentionally via pain or injury, will alter the outgoing motor command and subsequent movement patterns. My dissertation examined these interactions by evaluating sensory-mediated changes in flexibility and assessing movement in the presence of musculoskeletal disorders with associated joint-related pain.

In our first two studies, we explored changes in flexibility with sensory stimulation. First, we assessed the influence of adding transcutaneous electrical nerve stimulation (TENS) or self-massage using therapy balls to a stretching intervention of the plantar flexor muscles on ankle dorsiflexion range of motion, muscle activity, and muscle force. Although there was no influence of TENS, the addition of self-massage doubled the increase in range of motion achieved relative to stretching alone. Gains were more pronounced in less flexible individuals. Surprisingly, self-massage also increased plantar flexor maximal voluntary torque. To further explore the underlying mechanisms, we conducted a follow-up study that evaluated stretching with and without the addition of self-massage. Due to the decline in flexibility across the lifespan, we also included middle-aged adults in the study. The results were similar to our first study. The addition of self-massage increased flexibility gains achieved with stretching alone in both young and
middle-aged adults. With the addition of self-massage, middle-aged adults exhibited greater torque increases, which were associated with augmented muscle activity.

In the second two studies, we explored movement patterns as a result of joint-related pain in persons with sacroiliac joint dysfunction (SIJD) and compared them with healthy age-matched individuals. During a sit-to-stand task, individuals with SIJD had greater movement asymmetries including force loading rate when standing up, lower peak hip angle, and delayed onset of muscles key to stabilizing the joint. Furthermore, individuals with SIJD exhibited a significant reduction in the pattern of muscle activity from the painful side gluteus maximus and contralateral latissimus dorsi.

The results of this dissertation underscore the importance of afferent input in modulating range of motion and coordinating movement during activities of daily living.
DEDICATION

This dissertation is dedicated to my husband Steve and my daughter Madeline. I could not have done this without your hugs, kisses, laughter, snuggles, and support.
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CHAPTER I

INTRODUCTION
1 INTRODUCTION

Human movement is the result of a dynamic interplay between the sensory and motor components of the neuromuscular system. Afferent input from skin and muscle mechanoreceptors provide body position-sense (proprioception) and influence resting muscle tone, flexibility, and muscle coordination. Furthermore, stimulation of proprioceptors, either through deliberate manipulation or as a consequence of pain or injury, can result in a body-wide shift in the available range of motion at various joints (Dietz 2002). Two examples of this are flexibility and joint-related pain.

Flexibility, or the available range of motion at a joint, is one of the five components of physical health designated by the American College of Sports Medicine (ACSM). A certain degree of flexibility is necessary to perform activities of daily living, such as reaching, bending over, squatting, and rotating the head and torso. Changes in afferent input can modulate the body’s available range of motion. For example, foam rolling, which provides a deliberate increase in afferent input, increases flexibility at the ankle, knee, and hip (Mohr et al. 2014; Cheatham et al. 2015). In contrast, musculoskeletal disorders including plantar fasciitis, Achilles tendonitis, and non-specific low back pain are associated with decreased hamstring flexibility and ankle joint range of motion.

Similarly, the amount and type of afferent input received by the nervous system in the presence of pain or injury will result in altered motor planning and execution. For example, individuals with anterior knee pain exhibit higher internal knee abduction moments and delayed onset and duration of key stabilizing muscles (gluteus medius and vastus medialis oblique) during stair ambulation (Aminaka et al. 2011). Furthermore, motor coordination is often chronically impaired after the initial incident (Tsao et al. 2010; Hodges and Smeets 2015). To
prevent the potential for chronic maladapted motor patterns, it is crucial to identify compensatory movement patterns to develop focused rehabilitation programs.

The goals of this introduction are to provide an overview of the neuromuscular components that control movement, discuss how the nervous system modulates flexibility, and characterize the impact of joint-related pain on movement.

2 THE CONTROL OF HUMAN MOVEMENT

Neural signals are transformed into movement through the muscular system. Muscles have certain mechanical properties that modulate movement ability. Sensory feedback from somatosensory receptors and spinal reflex pathways further vary the orchestration of movement. This section will provide an overview of the neuromuscular system components that control human movement.

2.1 Tissue mechanics

From a mechanical perspective, range of motion about a joint is provided by the joint capsule (47%), muscle (41%), tendon (10%), and skin (2%) (Johns and Wright 1962). However, the primary adaptable components are muscle and connective tissue. Muscles are parallel groupings of myofibrils (fascicles) composed of sarcomeres in series and parallel. Each sarcomere comprises two contractile proteins: actin (thin filament) and myosin (thick filament). Additionally, the sarcomere contains several structural proteins, including titin. Titin is a highly elastic protein that spans more than half the length of the sarcomere, connecting the thick filament to the Z line. It acts as a molecular spring and creates passive tension within the muscle. It is this passive tension within the muscle that provides resistance to stretch (Herzog et al. 2016). Connective tissue and fascia are interwoven throughout the musculoskeletal system, making up approximately 30% of muscle mass (Alter 2004). Each myofibril is encapsulated by
endomysium, muscle fibers (fascicles) are bundled by perimysium, and the whole muscle is enveloped with epimysium. Moreover, fascia, considered the ‘soft tissue component to the connective tissue system that permeates the human body,’ binds muscles together into compartments within the body (e.g. posterior compartment of the lower leg) and mechanically connects muscles together (Schleip et al. 2012b; Wilke et al. 2018). Together with titin, connective tissue and fascia provide elasticity within the musculoskeletal system and govern resistance to passive displacement.

Muscle and connective tissue properties change with age and use. The turnover rate of collagen decreases with age, resulting in a greater deposition of type I collagen in the muscle tissue, increasing passive stiffness due to the low compliance (tolerance to undergo deformation) of collagen (Alnaqeeb et al. 1984; Schleip et al. 2006). Reduced contractile activity and muscle operation over a shortened range may adversely affect muscle composition, such as the density of connective tissue and contractile proteins. Animal models show increased connective tissue density and decreased sarcomeres in series after a period of joint immobilization in a shortened position (Williams et al. 1988). Additionally, denervated, immobilized, and dystrophic muscles show increased connective tissue density (Alnaqeeb et al. 1984). In humans, there is a significant decrease in muscle thickness, fascicle length, and pennation angle in the medial gastrocnemius and vastus lateralis after 5 weeks of immobilization, suggesting a loss of sarcomeres both in parallel and in series (de Boer et al. 2008). These changes can result in a loss of muscle force-generating capacity. As passive tension provided by connective tissue increases, the ability of the muscle to develop active tension decreases. Studies show that more strength is required to move the ankle in the presence of limited dorsiflexion and intrinsic passive resistance due to accumulation of connective tissue (Vandervoort et al. 1992).
In addition to its role in governing passive range of motion, fascia and connective tissue play an important role in force transmission between muscles and from muscles to the skeleton. In 1983, Sibyl Street found that both active and resting muscle tension were transmitted laterally between the myofibrillar components and the sarcolemma in splinted fibers from the semitendinosus muscle of a frog (Street 1983). Subsequent studies have corroborated this finding in humans. Bojsen-Møller and colleagues (2010) used ultrasound to track the displacement of the medial gastrocnemius, soleus, and flexor hallucis longus aponeuroses at origin and insertion during both a passive knee extension task (maintaining ankle angle) and during electrical stimulation of the medial gastrocnemius. Both of these tasks resulted in similar distal displacement of the medial gastrocnemius and soleus, suggesting lateral transmission of force between the two muscles. Similarly, Purslow and colleagues (2002) found that contraction of a single muscle within a fascial compartment containing multiple muscles significantly increases the intracompartmental pressure. A mathematical model estimated that the radial stress produced while contracting an upper arm muscle is equal to 50% of the longitudinal stress (Findley et al. 2015). Given this evidence, Bojsen-Møller and colleagues (2010) suggest that “it seems relevant to regard synergist muscles as a combined system of contractile and load-bearing tissues that act in concert to produce joint moments in an optimal manner” (p. 1616).

Tension can be transferred to adjacent muscles in series, and between muscles connecting into large sheets of fascia (e.g. thoracolumbar fascia). Several studies provide evidence of this in many locations of the body such as between the gastrocnemius and hamstrings (Cruz-Montecinos et al. 2015), biceps femoris and the sacrotuberous ligament (van Wingerden et al. 1993), and latissimus dorsi and the contralateral gluteus maximus (Carvalhais et al. 2013) (for a full review see Krause et al. 2016). Additionally, gluteus maximus generates force across the
sacroiliac joint, through its attachment to the thoracolumbar fascia and the aponeurosis of the erector spinae (Barker et al. 2014).

2.2 Neural components

The neural modulation of human movement occurs through the integration of sensory (afferent) input and motor (efferent) output. In addition to synaptic input from higher centers of the central nervous system (CNS), motor unit pools receive substantial input from sensory mechanoreceptors through both monosynaptic and polysynaptic projections. Mechanoreceptors relay information on internal and external stimuli both at the spinal cord level, resulting in reflex activity, and to the brain to influence the resulting motor command.

Innervation of skeletal muscle is provided by the somatic nervous system. A motor unit is the final common pathway by which the brain controls human movement, and is considered the functional unit of the neuromuscular system (Liddell and Sherrington 1925; Duchateau and Enoka 2011). It comprises a motor neuron residing in the ventral horn of the spinal cord, its axon, and all of the muscle fibers it innervates. The force produced by each motor unit depends on the innervation number (number of muscle fibers innervated by a motor neuron), muscle fiber area, and specific tension (the density of contractile proteins).

Just as synergistic muscles share space in fascial compartments, motor neurons innervating synergistic muscles share the same area in the spinal cord (Demireva et al. 2011). Studies of motor unit pools have determined that synergistic muscles receive a common input signal from higher centers (Laine et al. 2015). Motor neuron pools innervating synergistic muscles may be under the influence of a common input signal that may be modulated by afferent input (Farina and Negro 2015).
The musculoskeletal system has an extensive network of sensory receptors that provide feedback to the nervous system on both internal and external stimuli including stretch, force, touch, pressure, vibration, pain and temperature. In fact, 90% of the axons found in upper limb are sensory, underscoring the abundance and importance of afferent input (Gesslbauer et al. 2017). Sensory afferents that contain specialized encapsulated receptor cells are called mechanoreceptors and those without are known as free nerve endings. Mechanoreceptors that report on conditions generated within the musculoskeletal system reside within the muscle tendon unit while afferents in the skin and fascia detect external stimuli. Free nerve endings are responsible for transmitting information about pain.

Somatosensory information is critical for the nervous system to coordinate smooth voluntary movement. Afferent input is processed at the spinal cord level via reflex pathways and integrated within higher brain centers. Motor neurons receive thousands of inputs from homonymous muscle afferents, afferents from synergistic muscles, cutaneous receptors, and input from higher centers (Hultborn 2001; Enoka 2015). Thus, their output can be influenced by a multitude of incoming signals. Two examples of this are in response to a desired range of motion, and musculoskeletal pain. Sensory input can modulate the available range of motion at a joint as well as alter a particular motor command to avoid movements that evoke a painful stimulus (Schaible 2007).

The primary mechanoreceptors that govern range of motion are muscle spindles (length detectors) and golgi tendon organs (force detectors). Afferents arising from these receptors enter the spinal cord through the dorsal horn where they influence motor output via spinal reflex circuits, and provide proprioceptive information to the brain through ascending pathways. There are two complimentary spinal reflex pathways per mechanoreceptor. In response to muscle
lengthening, Ia afferents from muscle spindles synapse onto the homonymous motor neuron pool to increase muscle contraction (stretch reflex) and onto heteronymous motor neurons, via an inhibitory neuron, to decrease antagonist activity (reciprocal inhibition). In contrast, Ib afferents from Golgi tendon organs make disynaptic connections to reduce homonymous muscle force (autogenic inhibition) and facilitate antagonist muscle activity (reciprocal facilitation). However, both Ia and Ib afferents synapse onto the same interneuron, suggesting that muscle stretch and force are not controlled separately, but rather the signals are processed in parallel (Leonard 1998).

In addition to these simple reflex pathways, Ia afferents from muscle spindles have an extensive set of connections onto homonymous and heteronymous alpha motor neurons (Enoka 2015). For example, afferents from the soleus have been shown to make connections onto both the gastrocnemius and the quadriceps femoris. In addition, group Ib interneurons receive input from many other mechanoreceptors including Ia, Ib from other muscles, group III and IV cutaneous and joint afferents (Leonard 1998; Hultborn 2001).

Skin and connective tissue contain a rich supply of sensory afferents that report on external stimuli. These cutaneous mechanoreceptors can be classified based on their temporal response to a stimulus. Rapidly adapting afferents are responsive only at the beginning and end of a stimulus; they are quiescent in the interim. In contrast, slowly adapting mechanoreceptors provide feedback for the duration of a stimulus. There are currently four described cell types: Meissner corpuscles, Merkel discs, Pacinian corpuscles, and Ruffini endings. Meissner corpuscles and Merkel discs are primarily located in glabrous (non-hairy) skin. Meissner corpuscles are rapidly adapting receptors that lie in the dermis, and relay information about tactile sensation and light touch. Merkel discs are slow-adapting receptors located between the
epidermis and dermis that transmit information on pressure and tissue displacement. Pacinian corpuscles, the largest cutaneous receptors, are rapidly adapting and sensitive to vibration and have been found in the dermis, fascia, and joint capsules (Macefield 2005). Ruffini endings are slow adapting mechanoreceptors that lie in joint capsules as well as deep layers of the skin and fascia. They respond to skin stretch, mechanical deformation within joints, and change in joint angle (Macefield 2005). Additionally, several types of sensory afferents, including free nerve endings, Ruffini endings, and Pacinian corpuscles, have been found in human thoracolumbar fascia (Yahia et al. 1992).

Sensory information from cutaneous mechanoreceptors project to the thalamus and cerebral cortex, distributing their feedback throughout the central nervous system (Enoka 2015). Furthermore, reflex coupling between cutaneous afferents, via oligosynaptic spinal pathways and transcortical pathways, have been identified (Fallon 2005). Aniss and colleagues (1992) showed that cutaneous mechanoreceptors in the foot make multi-synaptic reflex connections to motor neurons of ankle joint muscles. When Gibbs and colleagues (1995) electrically stimulated cutaneous afferents in the second toe, they recorded cutaneomuscular reflex responses (E1 and E2) in both the ipsilateral lower limb muscles as well as the ipsilateral erector spinae. Specifically, the E1 (spinal reflex) was confined to muscles acting on the foot and toes. E2 responses, which reflect input from higher centers, were recorded in upper leg and spinal muscles in addition to lower limb muscles.

Afferent fibers that lack a specialized receptor cell are referred to as free nerve endings. Free nerve endings that transmit pain sensation are nociceptors (Purves 2008). The two primary free nerve endings are Aδ and C-fibers. Aδ fibers are thinly myelinated and respond to mechanical and thermal stimuli. C-fibers are unmyelinated and considered to be polymodal as
they respond to thermal, mechanical, and chemical stimuli. Free nerve endings and mechanoreceptors travel along different pathways to reach higher centers. However, the axons of all of these afferents enter the dorsal horn of the spinal cord where they interact via synaptic connections. One example of this is the gate control theory of pain proposed by Melzack and Wall (1965). This theory posits that afferents from different receptor types synapse onto a common interneuron that transmits pain signals to the brain. Stimulation of larger diameter myelinated afferents can mitigate signals arising from smaller diameter unmyelinated nociceptors (pain fibers) by exciting an inhibitory interneuron, thereby “closing the gate.” However, this is only one small piece of the larger issue of acute and chronic pain. Pain is known to be the result of a complex array of signal processing from multiple centers in the peripheral and central nervous system and a full discussion is beyond the scope of this paper.

All of the reflex pathways described above are subject to modulation by descending pathways and segmental systems via presynaptic inhibition, neuromodulation, and task dependence (Kirsch and Stein 2000). Synaptic input from descending pathways, cutaneous, joint and muscle afferents converge on both the Ia and Ib inhibitory interneurons. Therefore increased sensory afferent input can either facilitate or inhibit nearly every motor nucleus (Hultborn 2001). Moreover, reflex gains may be modulated prior to initiating movement, suggesting that the nervous system may assign reflex gains to accomplish certain tasks (Kirsch and Stein 2000).

2.3 Integration of sensory input

Proprioception is the sense of one’s own body position in space and the effort involved during movement (Proske 2005; Goble et al. 2009). The central nervous system (CNS) coordinates positional control and skilled movement using an internal map generated from proprioceptive input received from somatosensory receptors. Muscle spindles deliver
information on position and movement, tendon organs specify force and tension, the vestibular system governs balance, and the CNS generates the overall sense of effort (Proske 2005). Additionally, cutaneous and joint receptors are important for determining the position of distal body segments and signaling end ranges of motion (Goble et al. 2009). Proprioceptive information in the presence of pain or injury can alter the internal body map, including body part representations in the sensory cortex (Tsao et al. 2010; Moseley and Flor 2012). This can ultimately influence the sequence, timing, and amplitude of muscle contractions and movement. Current research shows that sensory input provides feedback to modulate a motor command, and can influence feedforward control (Hodges et al. 2003; Röijezon et al. 2015).

Proprioception can be adjusted via stimulation of mechanoreceptors. Stimulation of plantar cutaneous afferents can influence range of motion at non-local joints as well as impact clinical balance performance. For example, Grieve and colleagues (2015) found a significant increase in range of motion at the hip joint after subjects massaged the soles of their feet for 2 min with tennis balls. In addition, healthy elderly individuals (mean age 79 yr) who received a 20 min foot massage with joint mobilization had significantly improved timed up and go and single leg balance tests (Vaillant et al. 2009). Similarly, deep massage of the plantar surface of the foot improved timed up and go, and increased range of motion and foot sensation compared to a control condition in 60 individuals with diabetic peripheral neuropathy (Yamauchi 2015). Interestingly, sensory stimulation via application of kinesiology tape has been shown to increase joint position sense in individuals with patellofemoral pain, anterior cruciate ligament tears, and in individuals with sensory deficits (Macefield et al. 2016). In a study of 25 patients with hereditary sensory and autonomic neuropathy type III, which features an absence of functional muscle spindles but preservation of large-diameter cutaneous afferents, taping the knee joint
significantly improved proprioception on a joint position matching task (Macefield et al. 2016).

In contrast, deficiencies in afferent feedback result in decrements in calibrating hand position, sustaining constant muscle force, object weight discrimination, and impaired gait and muscle onset timing during walking, and coordination of muscle contractions in multi-joint movements (Goble et al. 2009).

3 FLEXIBILITY

Flexibility, or the available range of motion at a joint, is necessary to perform activities of daily living, such as reaching, squatting, and rotating the head and torso. Limited range of motion, particularly in the ankle joint and hamstring muscles, is associated with injury as well as chronic musculoskeletal conditions including plantar fasciitis, Achilles tendonitis, and non-specific low back pain (Ryan et al. 2008; López-Miñarro et al. 2012; Han et al. 2016).

The American College of Sports Medicine recommends that adults engage in flexibility training at least twice per week to maintain adequate ranges of motion across the lifespan (Chodzko-Zajko et al. 2009). There are many benefits to flexibility training. Periodic lengthening of a muscle is required to maintain its normal resting length; a muscle held in a shortened position will adapt to this new condition (Starring et al. 1988; Hug et al. 2015). Flexibility training, including stretching and yoga, has been shown to improve range of motion, balance, and spinal posture (López-Miñarro et al. 2012; Jeter et al. 2014; Behm et al. 2016a; Gothe and McAuley 2016). In addition, increased ankle dorsiflexion enhances proprioceptive feedback to improve posture and decrease fall risk in older adults (Jeter et al. 2014, 2015).

3.1 Stretching

Stretching is a widely used modality to improve range of motion, either for aesthetics (e.g. yoga, dance) or to prevent injury. Improvements in range of motion after static stretching
have been attributed to changes in muscle, neural, and psychological mechanisms. These include increases in muscle-tendon unit length and reduced muscle stiffness, a reduction in muscle spindle excitability, and improved stretch tolerance.

Gains in flexibility appear to be modulated by changes in spinal reflex responsiveness and muscle tissue properties. Guissard and Duchateau (2004) evaluated changes in spinal reflex excitability using the Hoffman reflex (H-reflex) to assess the responsiveness of the Ia pathway and the Tendon tap reflex (T-reflex) to assess excitability of muscle spindles, as well as mechanical and contractile muscle properties after a 6wk intervention of static stretching of the plantar flexors. Ankle dorsiflexion range of motion increased by 31% (24.6° vs 32.2°) with the greatest gains appearing in the first 2 wks (56% of gain). Relative to baseline values, H-reflex decreased by 14% and T-reflex decreased by 36%. The greater change in T-reflex amplitude could be caused by reduced muscle spindle sensitivity. Furthermore, passive stiffness declined from 1.3 Nm/° at baseline to 0.76 Nm/° at the end of the intervention, whereas contractile properties, including MVC force and rate of torque development, did not change. A decrease in passive stiffness with stretching has also been shown using shear-wave elastography (Nakamura et al. 2014). Umegaki and colleagues (2015) used ultrasound shear wave elastography to measure the shear elastic modulus of the hamstring muscles in healthy young men (n = 20, 23.4 y) before and after a static stretching intervention. Final hip angle increased from 59° to 73° (p<0.01) and passive torque decreased from 32 to 28 N•m. Shear elastic modulus (kPa) significantly decreased for all three hamstring muscles when the muscle was slack (knee flexed 45°) and when fully elongated (knee at 90°) (p<0.05). Furthermore, Freitas and colleagues (2018) performed a systematic review to assess the potential of chronic stretching to alter muscle-tendon mechanical properties. The results for muscle stiffness, fascicle length, and
tendon stiffness were extremely variable, with at least half the included studies showing positive changes. The variability in results underscores the problem of study heterogeneity when evaluating the stretching literature. However, the results of these studies support a combination of neural and mechanical adaptations to stretching.

A growing body of evidence suggests that the most prominent contribution to flexibility may arise from the nervous system. A decrease in tonic reflex activity likely contributes to the observed decrease in passive torque observed with stretching (Guissard and Duchateau 2006; Abellaneda et al. 2009). Moreover, the complexity of afferent integration within the spinal cord as well as the central projections from mechanoreceptor afferents suggests that stretching may result in substantial global neural contributions. Both contralateral untrained limbs and non-local affects have been reported. For example, stretching the calf and hamstrings muscles increases cervical spine range of motion (Wilke et al. 2016). Additionally, shoulder range of motion significantly improved after a lower body stretching program (Behm et al. 2016b).

Although there is evidence for both mechanical and neural explanatory mechanisms underlying changes in flexibility with stretching, many authors conclude that stretch tolerance (Magnusson 1998; Law et al. 2009), is the most likely explanation for changes in range of motion. The phrase “stretch tolerance” was first introduced in 1998 by Magnusson and colleagues (1998). They found that while range of motion increased after a 3-week stretching intervention, the gains were not associated with changes in passive muscle properties. Consequently, the authors concluded that it was the participants’ tolerance to the discomfort of stretching that resulted in increased range of motion. Interestingly, Wyon and colleagues (2013) evaluated changes in active and passive range of motion at two levels of discomfort, defined on a 0 to 10 perception of stretching intensity scale where “0 represents no stretch and 10 is
equivalent to an aggressive stretch associated with pain, often described as a burning sensation”.

While both techniques increased passive range of motion, there was no significant difference between intensity groups. However, active range of motion was significantly higher in the low-intensity group. While perception of effort in other activities has been shown to influence performance, it is not the sole determinant factor of success (Marcora et al. 2009; Marcora and Staiano 2010).

Stretch tolerance likely plays a role in adjusting to extreme ranges of motion. However, as with most physical adjustments, flexibility is adaptive. Disuse and tissue injury result in decreased range of motion and consistent stretch practice increases range of motion. While stretch-training interventions do increase flexibility in healthy young adults, cessation of training results in a return to baseline conditions. Willy and colleagues (2001) found the increased range of motion attained after a 6-wk hamstring stretching protocol (2-30 s stretches daily x 5 d per week) dissipated after not stretching for 4 wks. However, the initial gains were quickly recovered after stretching resumed. Similarly, Guissard and Duchateau (2004) found a 25% decrease in the range of motion gain (74% retention) 30 days after termination of a plantar flexor training regimen. These findings are similar to those recorded for strength training interventions. Strength training results in increased muscle cross sectional area, as well as changes in motor unit recruitment threshold, rate coding, and increased motor neuron excitability (Enoka 2015). Just as with adaptations observed with stretch training, these changes are not permanent.

The effectiveness of stretching techniques may vary with age. Current literature suggests that in healthy, young adults, static stretching is optimal for long-term increases in range of motion whereas dynamic stretching is preferred as a warm up prior to vigorous physical activity (Behm et al. 2016a). Only one study has examined stretching as a single intervention in older
adults. Feland and colleagues (2001) evaluated range of motion after a hamstring stretching intervention in adults over the age of 65 y with ‘tight’ hamstrings (< 20° knee extension with the hip at 90°). The study compared three groups with increasing stretch durations (15 s, 30 s, 60 s) to control subjects (no stretch) and concluded that in older individuals, longer hold times are more effective than shorter hold times for improving range of motion. This may be due to the effects of stress-relaxation (decrease in passive tension under a fixed stress) on stiffer tissues.

While stretching is effective in improving flexibility, it may produce temporary deficits in strength, termed “stretch-induced force deficit” (Behm et al. 2004). Some studies suggest this is due to an alteration in length-tension relation or shortening velocity (Ryan et al. 2008). Stretching results in a more compliant muscle tendon unit, which may decrease force output by decreasing the rate of shortening (Wilson et al. 1994; Fowles et al. 2000; Behm et al. 2004). Ryan and colleagues (2008) examined the effect of 2, 4 and 8 min of passive stretching on peak torque, rate of torque development, and voluntary activation. Although each of these variables except voluntary activation (no change) declined after all stretch durations, values recovered to baseline levels at 10 min. Self-massage using an implement, such as a foam roller, roller stick, or therapy balls, often called “self-myofascial release”, has been shown to increase range of motion without performance deficit (for a full review, see Cheatham et al. 2015).

3.2 Self-massage

Self-myofascial release is a term used to describe device-assisted manual therapy techniques aimed at manipulating muscles and fascia (Schroeder and Best 2015). Current evidence suggests that the human body could not withstand the tremendous forces required to create mechanical changes to fascia with the application of these techniques (Chaudhry et al. 2008). Therefore, we will forgo the use of the term self-myofascial release and instead use more
accurate descriptions of the techniques used, namely foam rolling or self-massage. Principles underlying these techniques include increased circulation, venous and lymphatic drainage, as well as improved elasticity and flexibility of connective tissue (Karageanes 2005).

Self-massage with a device has been shown to increase range of motion without affecting muscle force (Cheatham et al. 2015). For example, knee range of motion increased by 12.7% and 10.3% at 2 and 10 min, without a deficit in maximal voluntary contraction (MVC) force, after two 2-min sessions of foam rolling the quadriceps (MacDonald et al. 2013). Similarly, Bradbury-Squires et al. (2015) reported that a 20-s intervention increased knee flexion by 10% whereas a 60-s intervention resulted in a 16% increase.

Static stretching combined with foam rolling provides a greater improvement in range of motion than either technique alone. For example, Škarabot and colleagues (2015) showed that the combination of stretching and foam rolling of the calf muscles resulted in the greatest increase in ankle dorsiflexion range of motion (9.1%) than either technique alone (6.2% stretching, 0% foam rolling). Mohr and colleagues (2014) found similar results for hip flexion; stretching plus foam rolling resulted in greater improvements (24%) than either stretching (12%) or foam rolling (7%) alone.

The explanatory mechanisms for changes observed with self-massage are currently unknown. However, improvements may be at least partially modulated by spinal reflex pathways from the excitation of muscle spindles and tendon organs. The direct application of force with a foam roller stretches the muscle, excites muscle spindles, and results in contraction of the homonymous muscle. The pressure exerted by the roller combined with the muscle-generated force may excite the tendon organs, causing relaxation of the muscle when tension on the tendon becomes extreme. Behm and colleagues (2013) evaluated spinal reflex responsiveness and
twitch contractile properties after massage at the musculotendinous junction of the soleus, tapotement massage (using percussive strokes), both massage techniques with stretching, and a control. All of the interventions resulted in decreased H-reflex compared with control subjects with tapotement plus stretch having the largest effect. Goldberg and colleagues (1992) found a greater depression of H-reflex amplitude with three minutes of deep massage (2.5 kPa) of the triceps surae compared with light massage (1.25 kPa) of the same duration. The results of these studies suggest that the observed depression in H-reflex amplitude may be due to greater Ia presynaptic inhibition or increased collection of inhibitory inputs from various cutaneous and muscle mechanoreceptors. It is possible that self-massage using a device may elicit similar changes.

3 JOINT-RELATED PAIN

Pain provides additional sensory input that may alter motor control. Changes in muscle activity may be an attempt to splint the painful area (decreasing range of motion), reduce the amplitude or velocity of a painful movement, or avoid a movement altogether (Hodges and Smeets 2015). Thus joint and muscle pain can reduce muscle strength and endurance, and disrupt muscle coordination and normative gait patterns (Graven-Nielsen et al. 1997; Fyfe et al. 2013; Falla and Hodges 2017). Consequently, this can cause chronic adaptation in both spinal and supraspinal centers (Graven-Nielsen and Arendt-Nielsen 2010; Fyfe et al. 2013). For example, a study investigating injury to and rehabilitation of the biceps femoris suggested that the muscle is chronically inhibited after injury (Fyfe et al. 2013). Furthermore, maladapted muscle length as a result of poor posture, injury, or immobilization is commonly observed in older adults with reduced mobility (Riley and Van Dyke 2012).
One example of how joint-related pain can alter human movement is sacroiliac joint dysfunction. Sacroiliac joint dysfunction is a broad term used to describe pain arising from the sacroiliac joint. In contrast to other large articulating joints in the human body, the sacroiliac joint has very little movement and does not have any muscles that cross and directly act on the joint (e.g. biceps femoris crosses the knee joint and produces knee flexion). This is due to the structure and function of the joint. It has been described as having a “keystone-like” shape as the auricular surface is billowed on the iliac side and concave on the sacral side. To describe the functional stability of this unique joint, Vleeming and colleagues (1990a) introduced the principles of form and force closure. Form closure refers to the stability provided by the complimentary convex and concave surfaces of the iliac and sacral joint surfaces. Force closure describes the function of ligaments and muscles acting across the joint to provide joint mobility and stability (Vleeming et al. 1990b).

The primary function of the sacroiliac joint is to assist with load transfer from the spine to the lower extremities through the pelvic girdle. Degeneration of the joint and ligaments can lead to changes in movement capability of the joint, stimulating nociceptive fibers within and around the joint (Szadek et al. 2008). Load transfer through the pelvis with an unstable sacroiliac joint, due to compromised form or force closure, can produce excessive loads on surrounding tissues, resulting in pain (Pool-Goudzwaard et al. 1998). Additionally, pain in the sacroiliac joint can lead to inefficient muscle recruitment, preventing necessary force closure of the joint to maintain stability (Agarwal et al. 2014). The consequence of these changes may result in abnormal joint loading, which influences the control of muscle forces and limb coordination, and is associated with joint degeneration (or worsening thereof) (McCrory et al. 2001; Herzog et al. 2003).
Walking and standing up from a chair (sit-to-stand) are predictably executed basic functional tasks. Therefore, changes in the motor performance of these tasks can be indicative of movement disorders and be assessed to further characterize compensatory patterns associated with various musculoskeletal disorders (Dietz 2002). Data acquired from biomechanical studies on walking or sit-to-stand are helpful for many reasons. First, they may help identify distinguishing motor patterns to aid in differential diagnosis of a pathological condition. Second, they may be used to design specific rehabilitation programs, which have been shown to be more effective than generalized exercise therapy (Falla and Hodges 2017; Hug and Tucker 2017). Third, they can be used to track the progress and effectiveness of an intervention (McCrory et al. 2001).

4 CONCLUSION

Human movement is the result of a dynamic interplay between the sensory and motor components of the neuromuscular system. Alterations in the amount and type of afferent input can result in global changes in muscle activity, range of motion, and kinematics. Changes in flexibility appear to be modulated similarly to other physical endeavors; frequency, intensity, and duration of training result in neural and muscular changes. Similarly, injury can result in acute and, in some cases, chronic motor pattern adaptations. The goal of this dissertation is to examine sensory and motor interactions by evaluating sensory-mediated changes in flexibility and assessing movement in the presence of musculoskeletal disorders with associated joint-related pain.
CHAPTER II

MANIPULATION OF SENSORY INPUT CAN INFLUENCE STRETCHING OUTCOMES
ABSTRACT

**Purpose:** The primary purpose of our study was to assess the influence of modulating sensory input with either transcutaneous electrical nerve stimulation (TENS) or self-massage with therapy balls on the maximal range of motion (ROM) about the ankle joint when stretching the calf muscles. We also investigated the influence of these two conditions on the force capacity and force control of plantar flexor muscles.

**Methods:** Twenty healthy adults (25 ± 3 yr) performed 3 sessions of ankle plantar flexor stretching (3 stretches of 30 s each): stretching alone (SS), stretching with concurrent TENS (TENS), and stretching after self-massage using therapy balls (SM). TENS was applied for 60 s prior to and during each stretch, and SM was performed for 60 s prior to each of the 3 stretches. Maximal voluntary contraction (MVC) torque and force steadiness at 20% MVC were recorded before and at 15 min after the final stretch. Ankle dorsiflexion ROM was assessed before, after, and at 5, 10, 15 min after the last stretch.

**Results:** The increase in ROM was greater after SM (24%) than after SS (13%) and TENS (9%; p < 0.001). Maximal discomfort level (0-10 VAS) during stretching was similar for all conditions. MVC torque increased after SM only (p < 0.001, Cohen’s D = 1.5): SM, 16%; SS, -1%; TENS, -3%. Force steadiness did not change.

**Conclusion:** The sensory fibers that contribute to stretch tolerance were engaged by self-massage but not by TENS, resulting in greater increases in flexibility and MVC torque after self-massage.
INTRODUCTION

Flexibility, defined as the range of motion about a joint, is essential for performing activities of daily living. Stretching is often used in physical rehabilitation and athletic settings to restore or improve range of motion. Although some findings suggest that adaptations in muscular components and neural reflex pathways can contribute to the gains achieved with stretching interventions, the consensus explanation for increases in flexibility is stretch tolerance (Weppler and Magnusson 2010; Behm et al. 2016a). Stretch tolerance corresponds to the level of discomfort, pain, or feeling of tightness an individual is willing to tolerate while stretching (Folpp et al. 2006; Law et al. 2009).

The gate control theory of pain asserts that the perceived level of pain is derived from a balance between nociceptive and non-nociceptive afferent input (Melzack and Wall 1965). Non-painful input is purported to dampen or close the “gate” to the transmission of pain signals. In this pathway, common inhibitory interneurons receive excitatory inputs from unmyelinated nociceptors and inhibitory inputs arise from myelinated mechanoreceptors, namely Aß or type II fibers. In the context of stretch tolerance, input from large diameter mechanoreceptors may mitigate nociceptive signals via the pain gate theory and thereby lessen the perception of pain, allowing an individual to stretch further.

Transcutaneous electrical nerve stimulation (TENS) applied over the skin can mitigate pain via the gate control theory (Sluka and Walsh 2003; Moran et al. 2011). At low stimulus intensities, TENS elicits a tingling sensation without a muscle contraction. If TENS activates at least some of the same sensory fibers that contribute to stretch tolerance, then the application of TENS during a stretching maneuver should enable at least a transient increase in flexibility. For example, eight weeks of stretching during the concurrent application of TENS was superior to
stretching alone in increasing hamstring flexibility in adolescent soccer players with short-hamstring syndrome (Piqueras-Rodríguez et al. 2016). However, two weeks of stretch training by young women produced no difference in hamstring flexibility for the stretch group compared with stretching + TENS (Maciel and Camara 2008). The conflicting results of these two studies underscore the need for further investigation into the effects of adding TENS to a stretching protocol.

Another approach used to modulate sensory input and thereby improve range of motion is self-massage with an implement, such as a foam roller or therapy balls. Self-massage immediately prior to stretching results in greater improvements in range of motion than stretching alone (MacDonald et al. 2013; Sullivan et al. 2013; Cheatham et al. 2015). For example, three 30-s bouts of foam rolling the plantar flexor muscles combined with three 30-s calf stretches produced a greater improvement in ankle range of motion than either technique alone (Škarabot et al. 2015). Similarly, combining foam rolling with passive stretching of the hamstrings significantly improved hip flexion compared with either technique alone (Mohr et al. 2014).

Static stretching by itself, without TENS or massage, improves flexibility, but may cause a transient decrease in maximum voluntary contraction (MVC) force (Ryan et al. 2008; Behm et al. 2016a). One hypothesis to explain stretch-induced force deficit is reduced muscle activation during voluntary contraction (Trajano et al. 2017). Force steadiness, quantified as the fluctuations in force during a submaximal isometric contraction, provides one measure of the neural drive to muscle (Farina and Negro 2015). Alterations in force steadiness after an intervention imply a central treatment effect. One study, for example, found a reduction in force steadiness after a series of plantar flexor stretches (Kato et al. 2011). In contrast to static
stretching, self-massage using an implement improves flexibility without a subsequent decline in muscle performance (Bradbury-Squires et al. 2015; Cheatham et al. 2015). The effects of these instruments on force steadiness have not been investigated.

The primary purpose of our study was to assess the influence of modulating sensory input with either TENS or self-massage with therapy balls on the maximal range of motion about the ankle joint when stretching the calf muscles. The stretches were performed to the maximal tolerable level of discomfort (stretch tolerance). The secondary purpose was to determine the influence of the two conditions on MVC torque and force steadiness. Our hypothesis was that stretch tolerance would improve and enable an increase in ankle range of motion with concurrent application of TENS during stretching and when self-massage was performed prior to stretching.

MATERIALS AND METHODS

Twenty recreationally active adults (10 women, mean ± SD: age, 25.0 ± 3.7 yr; height, 172 ± 12 cm; mass, 68 ± 14 kg) with no history of neuromuscular impairment or lower limb injury volunteered for the study. All participants provided written informed consent before undergoing any study procedures. Participants were asked to refrain from exercising and stretching prior to the procedure on the day of testing. The Institutional Review Board at the University of Colorado Boulder approved the study (protocol #16-0111).

Study Design and Overview

The study comprised a counter-balanced, within-subject design with three conditions. Participants visited the laboratory on three occasions, at least five days apart, at the same time of
day. Sessions involved stretching the plantar flexors of the dominant leg, either alone without any treatment, with concurrent application of TENS, or after self-massage with therapy balls.

Upon arrival to the laboratory, participants performed an MVC task and then a force-matching task (20% MVC) with the plantar flexors of the dominant leg. Maximal ankle dorsiflexion angle was assessed in the standing stretch position prior to each intervention, immediately after, and at 5, 10, and 15 min after the final stretch. Subjects were seated between measurements. Maximal discomfort (pain) level was recorded during each range of motion assessment using a 0-10 Visual Analog Scale (VAS) with 0 representing no discomfort and 10 denoting maximal discomfort. MVC and force-matching tasks were repeated at 15 min after the final range-of-motion assessment (Figure 1).

![Figure 1: Schematic representation of the study design and test procedures.]

**Stretching Procedure**

Study participants were instructed to perform a standing calf stretch to their maximal tolerable (discomfort/pain) limit. From a standing position, participants stepped the leg to be stretched back so that both feet pointed forward, both heels were on the ground, the knee of the
stretched leg was fully extended, and the hands were placed on a wall for support (Figure 2A). Three 30-s stretches were performed with 30 s of rest between stretches.

![A) Participants were instructed to perform a calf stretch using a staggered stance with the non-dominant leg in front, and hands against a wall for support. B) Electrode pads for TENS were applied over the calf muscles. C) Participants massaged the calf muscles with two therapy balls bound together in a tote sack.](image)

**Figure 2:** A) Participants were instructed to perform a calf stretch using a staggered stance with the non-dominant leg in front, and hands against a wall for support. B) Electrode pads for TENS were applied over the calf muscles. C) Participants massaged the calf muscles with two therapy balls bound together in a tote sack.

**Stretching with TENS**

Surface electrodes were placed over the calf muscles (Figure 2B) and attached to a TENS unit (LG-Tec Elite DC6100, Current Solutions LLC, Austin, TX). Conventional biphasic TENS pulses were delivered through the electrodes using a 100 Hz pulse rate and 40 µs pulse duration to target sensory fibers (Sluka and Walsh 2003). Intensity was set at the participant’s maximal tolerable limit, but below that needed to evoke a muscle contraction. TENS was applied for 60 s before initiating the stretching intervention, remained on during the stretches, and was turned off after the last stretch. Total treatment time for TENS was 4 min.

**Stretching with therapy ball self-massage**

Participants applied self-massage to the triceps surae for 60 s using two massage therapy
balls (Yoga Tune Up® Therapy Balls, Tune Up Fitness Worldwide Inc., Sherman Oaks, CA) contained in a mesh bag prior to each of the three 30-s stretches. The rolling procedure was standardized to ensure consistency. The procedure began with the participant seated with the dominant leg extended and the contralateral knee flexed with the foot on the ground. The instructions were to push the hands into the floor to lift the hips, and move the extended leg so that the therapy balls massaged the region spanning from the Achilles tendon to just distal to the popliteal fossa at a cadence of 4 s for each stroke (Figure 2C). Participants were instructed to perform the massage procedure with a pressure level that resulted in a discomfort level of < 5 on VAS.

Range of motion measurement

A digital goniometer connected to an amplifier (Biometrics Ltd, Newport, UK) was used to measure ankle joint range of motion in the standing stretch position. With the participant in a neutral upright position, the goniometer was applied to the lateral surface of the lower leg and foot using double sided tape. The device was positioned with one end mounted on the calcaneus, parallel to the floor, and the other end perpendicular to the receiver box, over the lateral malleolus. The same investigator positioned the goniometer for all trials.

Force and EMG recordings

Muscle activity during the two force tasks was recorded using surface electrodes (Ag–AgCl, 8 mm diameter, 20 mm distance between electrodes) placed over the tibialis anterior, medial gastrocnemius, and soleus muscles. Skin surface area was shaved, abraded, and cleansed with alcohol prior to attaching the surface electrodes. Ground electrodes were placed on the
medial malleolus and patella. EMG signals were amplified (x 2,000), band-pass filtered (10 - 490 Hz), digitized at 2,000 samples/s, and recorded on a computer.

Muscle strength was quantified by measuring the peak force achieved during an isometric MVC. Participants laid supine on the testing apparatus with both legs extended and the dominant foot secured in a strap connected to a force strain-gauge transducer (Bento et al. 2010). The subject gradually increased plantar flexor force from rest to maximum during a 3-s period and then held the maximal force for 3 s. This procedure was repeated until the peak force for two MVCs differed by less than 5%. The MVC force was used to calculate the target for the force-matching task.

Force was measured with a strain-gauge transducer (MLP-300, Transducer Techniques, Temecula, CA). Force signal was digitized at 1,000 samples/s (Coulbourn Instruments, Allentown, PA) and stored on a computer where it was processed using a second-order, low-pass Butterworth filter with 100 Hz cutoff frequency (Spike2, Cambridge Electronic Design Limited, Cambridge, England).

**Force-Matching Task**

Force steadiness of the plantar flexors, quantified as the coefficient of variation for force during a steady isometric contraction, was recorded at the beginning and end of each session. After MVC force was determined, a target line representing 20% MVC force was displayed on a monitor and subjects were instructed to exert a plantar flexor force to move a cursor to the target line and to maintain a steady contraction for 20 s.
Data analysis

Torque produced during the MVC was calculated as the peak force multiplied by the moment arm (perpendicular distance from the line of action of the applied force to the center of the lateral malleolus). Muscle activity was quantified as the RMS amplitude during a 1-s period around the peak force. Coactivation of agonist and antagonist muscles during the force tasks was calculated as the ratio of EMG amplitude for tibialis anterior relative to the sum of the EMG amplitudes for the medial gastrocnemius and soleus muscles. Coefficient of variation for force (standard deviation/mean force x 100) was calculated for the middle 10 s of the steady isometric contraction.

Statistics

Data were assessed for normality using the Shapiro-Wilk test. Repeated-measures analysis of variance with post-hoc pairwise comparisons were performed to examine differences between conditions for range of motion, MVC torque, and force steadiness. Pearson’s correlation was performed to evaluate the impact of baseline flexibility on changes in range of motion and MVC torque. Effect size was calculated using either Cohen’s D or Glass’ delta, depending on the similarity of group standard deviations. Significance was set at alpha = 0.05.

RESULTS

There was no difference in range of motion at baseline for the three conditions (ANOVA p = 0.56). All three conditions significantly increased range of motion immediately after the intervention (ANOVA, p<0.001), which was retained for at least 15 min. The mean ± SD percentage change in range of motion for each condition was: stretch only, 13 ± 9%; stretch + self-massage, 24 ± 17%; stretch + TENS, 9 ± 7% (Table 1, Figure 3A). Post-hoc pairwise
comparisons indicated no statistically significant difference between stretching alone and stretch + TENS (p = 0.18). However, stretch + self-massage significantly increased range of motion compared with stretch only (p < 0.001, Glass’ delta = 1.38). A moderately negative correlation (r = -0.51; p = 0.02) was found between baseline range of motion and its change after the stretch + self-massage condition. The negative sign indicates that less flexible individuals experienced greater changes in range of motion. There was no such correlation for the other two conditions.

MVC torque values were consistent at baseline across conditions (ANOVA, p = 0.93). Mean ± SD change for each condition was: stretch, -1 ± 13%; stretch + TENS, -3 ± 10%, stretch + self-massage, +16 ± 11% (Table 1, Figure 3B). The changes in MVC torque after stretch + TENS were not statistically different from those observed after stretching alone (p = 0.62). Stretching after self-massage with therapy balls significantly increased MVC torque compared with stretching alone (p < 0.001, Cohen’s D = 1.47). The change in MVC torque from before to after each condition was not correlated with baseline or change in range of motion (p > 0.5).

The influence of the three conditions on muscle activation was assessed by comparing EMG amplitude during the MVC and force steadiness tasks. The increase in MVC torque observed with stretch + self-massage was accompanied by an increase in plantar flexor EMG amplitude (p = 0.04; Table 1). However, EMG amplitude also increased for the stretch + TENS condition (p = 0.01), despite a decrease in MVC torque. No significant difference was observed after stretching only (p = 0.77). There were no significant changes in EMG amplitude for an antagonist muscle (tibialis anterior) for any condition. The coactivation ratio (tibialis anterior/(medial gastrocnemius + soleus)) during the MVC did not significantly change for any condition. There were no statistically significant differences in force steadiness at 20% MVC force between the three conditions, both before and after the intervention (mean ± SD): stretch, 2.3 ± 1.1% to
2.1 ± 1.1%; stretch + TENS, 1.7 ± 0.5% to 1.8 ± 0.7%; stretch + self-massage, 1.5 ± 0.8% to 1.5 ± 0.4% (p = 0.60).

Participant-reported discomfort (pain) was recorded on VAS during each condition. Maximal VAS during the stretches was similar across conditions: stretch, 3 ± 2; stretch + TENS, 4 ± 2; stretch + self-massage, 4 ± 2. TENS intensity averaged 35 ± 13 mA. Average VAS during self-massage was 2 ± 1.

Table 1. Results for range of motion (ROM)(°) and maximal voluntary contraction (MVC) torque (N·m) before and at 15 min after the intervention, EMG amplitude (mV) during MVC, and coactivation ratio (TA/MG+SOL), mean (± SD) values. SM= self-massage with therapy balls; TA= tibialis anterior; MG = medial gastrocnemius; SOL = soleus.

<table>
<thead>
<tr>
<th></th>
<th>ROM</th>
<th>MVC Torque</th>
<th>MG + SOL</th>
<th>Coactivation Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Stretch only</strong></td>
<td>Before</td>
<td>25.9 ± 6.2</td>
<td>29.5 ± 14</td>
<td>0.465 ± 0.274</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>29 ± 6.8</td>
<td>29.0 ± 13.7</td>
<td>0.422 ± 0.057</td>
</tr>
<tr>
<td></td>
<td>% Change</td>
<td>12.5 ± 8.5</td>
<td>-0.7 ± 12.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>p-value</td>
<td>0.00</td>
<td>0.52</td>
<td>0.11</td>
</tr>
<tr>
<td><strong>Stretch + TENS</strong></td>
<td>Before</td>
<td>28.1 ± 6.3</td>
<td>31.0 ± 15.2</td>
<td>0.394 ± 0.172</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>30.5 ± 6.2</td>
<td>29.8 ± 14.2</td>
<td>0.426 ± 0.182</td>
</tr>
<tr>
<td></td>
<td>% Change</td>
<td>9.3 ± 7.4</td>
<td>-2.5 ± 10.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>p-value</td>
<td>0.00</td>
<td>0.09</td>
<td>0.01</td>
</tr>
<tr>
<td><strong>Stretch + SM</strong></td>
<td>Before</td>
<td>26.7 ± 6.9</td>
<td>31.0 ± 14.9</td>
<td>0.404 ± 0.194</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>32.6 ± 7.0</td>
<td>35.6 ± 16</td>
<td>0.449 ± 0.184</td>
</tr>
<tr>
<td></td>
<td>% Change</td>
<td>24.3 ± 16.9</td>
<td>16.3 ± 11.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td>p-value</td>
<td>0.00</td>
<td>0.00</td>
<td>0.04</td>
</tr>
</tbody>
</table>
DISCUSSION

The goal of our study was to compare the effects of combining either TENS or self-massage with stretching on stretch tolerance, range of motion about the ankle joint, force capacity, and force steadiness. Contrary to our hypothesis, the concurrent application of TENS during stretching did not influence either stretch tolerance or range of motion compared with stretching alone. In contrast, the increase in range of motion after self-massage with therapy balls...
prior to stretching was nearly twice that for the other two conditions. Notably, study participants reported similar discomfort levels during the stretches across all three conditions. Whereas stretching with and without TENS was associated with a small average decrease in maximal voluntary torque, the addition of self-massage with therapy balls significantly increased MVC torque in all participants.

The evidence to support the use of TENS in conjunction with stretching to improve flexibility is varied. A 10-min application of TENS (100 Hz, 40 µs) prior to stretching did not increase hamstring flexibility or stretch tolerance in young women (Maciel and Camara 2008). However, a 1-min application of TENS (50 Hz, with muscle contraction) concurrent with stretching resulted in greater improvements in straight-leg raise compared with stretching alone in adolescent male soccer players with short hamstring syndrome (Piqueras-Rodríguez et al. 2016). The most comprehensive study of TENS and stretching evaluated the effects of the treatment on pressure pain threshold, muscle hardness (transverse muscle stiffness), and range of motion in 15 young men (Karasuno et al. 2016). The protocol comprised 20 min of TENS followed by ten 30-s stretches of the hamstrings with 30 s of rest after each stretch (10 min of static stretching). TENS current (100 Hz, 200 µs) was set at 80% of pain threshold and evoked muscle contractions. The improvements observed on all tested measures decayed by ~50% after 10 min. In our study, TENS provided no added benefit to the increase in the range of motion after stretching alone and resulted in a small decline in MVC torque at 15 min. Therefore it is likely that the discomfort experienced during stretching is not attenuated via gate control.

In contrast to TENS, the addition of self-massage amplified the effects of stretching on ankle flexibility. This finding is consistent with literature on the addition of foam rolling to a stretch program. For example, one study found a greater increase in ankle range of motion in
athletes when foam rolling was combined with stretching (9.1%) than stretching alone (6.2%) (Škarabot et al. 2015). Similarly, the increase in hip-flexion range of motion was greater when stretching was combined with foam rolling (24%) than after either stretching (12%) or foam rolling (7%) alone (Mohr et al. 2014).

The significant increase in range of motion observed after self-massage may be due to modulation of neural pathways. For example, Behm et al. (2013) evaluated the responsiveness of spinal reflex pathways and twitch contractile properties after massage at the musculotendinous junction of the soleus, tapotement massage (using percussive strokes) of the triceps surae, both massage techniques with stretching, and a control condition. The amplitude of the Hoffmann (H) reflex decreased after all three conditions, but the effect was greatest for the tapotement plus stretch condition. The decrease in H-reflex amplitude was evident immediately after the intervention and up to 2 min later. Twitch contractile properties, including peak torque, time to peak torque, half relaxation time, and electromechanical delay did not change across conditions. Similarly, Goldberg et al. (1992) found a greater depression of H-reflex amplitude with 3 min of deep massage of the triceps surae compared with light massage of the same duration. It is possible that the self-massage technique used in the current study had a similar effect, and the reduced responsiveness of spinal reflex pathways allowed for a greater range of motion.

The improvements observed after self-massage with an implement may be influenced by an increase in the temperature of skin, muscle, and fascia, which would reduce connective tissue stiffness and lessen passive resistance to stretch (Taylor et al. 1990). Several authors have suggested that increased heat from foam rolling may reduce muscle and connective tissue viscoelasticity, thereby augmenting range of motion (MacDonald et al. 2013; Bradbury-Squires et al. 2015; Cheatham et al. 2015). A systematic review on the effect of heat from hot packs,
ultrasound, or diathermy, applied for 7 - 20 min before or during stretch, concluded that heat application potentiated the acute and sustained effects of stretching, perhaps due to changes in viscoelastic properties or altered sensations to stretch (Nakano et al. 2012). However, 60 s of foam rolling applied to quadriceps did not alter skin temperature (Murray et al. 2016).

Static stretching has been shown to produce a temporary decrease in the maximal force an individual can produce (stretching-induced force deficit) (Behm et al. 2004). Some studies suggest an alteration in the length-tension relation and a more compliant muscle tendon unit (Fowles et al. 2000; Behm et al. 2004; Behm and Chaouachi 2011). However, the declines in peak torque and rate of torque development after several minutes of passive stretching return to baseline values within 10 min of recovery (Ryan et al. 2008). In contrast to static stretching, studies of self-massage report flexibility gains at the hip, knee, and ankle without a decrement in muscle force (Cheatham et al. 2015). For example, Halperin et al. (2014) found self-massage of the plantar flexors with a roller stick increased MVC force by 8.2% at 10 min after the intervention, whereas static stretching decreased it by 4% despite an increase in ankle range of motion that lasted for at least 10 min. If stretch-induced force deficit is attributable to disfacilitation of synaptic inputs to motor neurons (Trajano et al. 2017a), then perhaps the self-massage procedure is able to reverse this effect and augment the inputs to motor neurons.

The fluctuations in force during submaximal isometric contractions, termed force steadiness, provides an index of the variability in common synaptic input received by the motor neurons that innervate a muscle (Negro et al. 2009; Farina and Negro 2015). The influence of stretching on force steadiness appears to depend on the magnitude of the target force. For example, Kato et al. (2011) reported significantly worse force steadiness at 20% MVC, but not at 5, 10, or 15% MVC, after a series of five passive static stretches of the plantar flexors. In
contrast, none of the conditions in our study influenced force steadiness (20% MVC), despite substantial change in range of motion and MVC torque. However, Kato et al. measured force steadiness at 2 min after stretching, whereas we measured it at 15 min after the stretch. The changes in MVC torque observed in our study at 15 min, therefore, were unrelated to the mechanisms responsible for force steadiness.

The current study has limitations. The self-massage condition added an additional three minutes to the duration of the protocol, and TENS added one minute. This time was not accounted for in the stretching-only condition. However, total time under stretch was identical for all three conditions and subsequent measurement time points after each final stretch did not differ. Due to the current delivered during the TENS condition, EMG signals could not be recorded during the stretching sessions.

**CONCLUSION**

The application of TENS did not improve stretch tolerance and was associated with a minor decrease in range of motion and force capacity when compared with stretching alone. In contrast, the addition of self-massage significantly improved stretch tolerance, resulting in greater flexibility compared with stretching alone. Furthermore, participants were able to produce greater MVC torque after the self-massage + stretch condition.
CHAPTER III

SELF-MASSAGE WITH THERAPY BALLS IMPROVES FLEXIBILITY IN YOUNG
AND MIDDLE-AGED ADULTS
Abstract

We examined the influence of stretching alone or combined with self-massage on maximal ankle dorsiflexion angle, maximal voluntary contraction (MVC) torque and muscle activity of the calf muscles, and subcutaneous tissue thickness in 15 young (25 ± 3 yrs) and 15 middle-aged (45 ± 5 yrs) adults. Participants performed two sessions of calf muscle stretches (3x 30-s stretches, 30-s rest): stretch after a 60-s control condition and stretch after 60 s of self-massage with therapy balls. Evaluations were performed before and 5 min after the intervention. Linear mixed effects model revealed no main effect for age on range of motion or MVC torque and significant main effects for treatment and time. Change in ankle dorsiflexion angle was significantly greater after stretch + self-massage: stretch = 12 ± 9 %, + self-massage = 24 ± 15 % (Hedges’ g = 0.98, p < 0.001). Similar results were observed for MVC torque: stretch = -4 ± 16 %, + self-massage = 12 ± 16 % (Hedges’ g = 0.97, p = 0.0001). Changes in MVC torque and absolute EMG amplitude were correlated. Subcutaneous tissue thickness was not altered by treatment. These benefits were more pronounced in less flexible middle-aged adults.
INTRODUCTION

Sufficient ankle dorsiflexion is necessary for performing activities of daily living, such as walking, climbing stairs, sitting and rising from a chair, and squatting. Limited range of motion at the ankle joint, for example, is associated with increased hip adduction during a step-down task, dynamic knee valgus during a squat, increased vertical ground reaction forces during landing, anterior cruciate ligament injury, and patellofemoral pain (Fong et al. 2011; Bell-Jenje et al. 2016). Restricted ankle dorsiflexion also alters the kinematics of walking by causing heel off to occur earlier at the end of the stance phase and increasing knee extension, which together increase stress and pressure on the forefoot (Yoon et al. 2014). These compensatory kinematics are associated with increased injury risk and may lead to or exacerbate osteoarthritis in the lower extremity (Ota et al. 2014).

Motor function and flexibility decline with advancing age, even in healthy adults (Gajdosik 2002; Justice et al. 2014), with some changes evident in middle age. For example, middle-aged (46 ± 6 yr) men were 23% less flexible on the sit-and-reach test and 18% less flexible during a prone hip extension task than young men (22 ± 1 yr) (Behm et al. 2011). Other studies report a decline in ankle joint flexibility of up to 50% after the age of 55 yr (Chiacchio et al. 2010). This decline is clinically significant as ankle dorsiflexion range of motion is correlated with calf muscle strength in women; less range of motion is associated with weaker muscles (Gajdosik 2002). Moreover, loss of flexibility about the ankle joint may contribute to the proximal shift in joint power during walking that is evident in older adults and appears to begin during middle-age (Kulmala et al. 2014). Reductions in dorsiflexion range of motion are also associated with an increased risk of ACL injury, which is becoming more prevalent in middle-aged adults (Legnani et al. 2011). Therefore, flexibility training during middle-age may attenuate
both the age-related loss of flexibility and the increased risk of knee injuries.

Static stretching is effective at increasing range of motion, but it can temporarily alter strength and performance measures, at least in young adults. Several studies, for example, have reported decreases in peak isometric torque, rate of force development, and power production after static stretching (Behm et al. 2016a). Shortening contractions performed prior to stretching can attenuate, but do not eliminate, the force deficit observed after stretching (Kay and Blazevich 2010). However, a dynamic warm-up performed in conjunction with static or dynamic stretching can prevent stretch-induced deficits in sprinting and jumping (Blazevich et al. 2018). Less is known about the influence of stretching on range of motion and athletic performance in middle-aged adults. Both static and dynamic stretching improved sit-and-reach scores without reducing performance in young or middle-aged adults (Behm et al. 2011). Similarly, 10 minutes of static stretching improved dynamic balance and did not alter jumping performance in 10 middle-aged adults (age 40 – 60 yr) (Handrakis et al. 2010). Neither of these studies, however, evaluated the influence of stretching on maximal force production.

In contrast, self-massage with various devices, such as foam rollers, roller-sticks, and therapy balls, improves flexibility without reducing muscle strength or performance in young adults (Cheatham et al. 2015). Furthermore, self-massage prior to stretching results in greater improvements in flexibility than stretching alone (MacDonald et al. 2013; Cheatham et al. 2015; Capobianco et al. 2018). For example, three 60-s repetitions of self-massage on the calf muscles prior to a three 30-s standing calf stretches nearly doubled the increase in range of motion compared with stretching alone (Capobianco et al. 2018). Similarly, three 30-s bouts of foam rolling combined with three 30-s calf stretches produced a greater improvement in ankle dorsiflexion than either technique alone (Škarabot et al. 2015). One potential explanation for this
Effect is that self-massage temporarily decreases fluid content in the local area, thereby reducing the resistance to changes in length (Sleboda and Roberts 2017). Ultrasound imaging may be used to detect these changes in fluid dynamics through measurements of muscle thickness (Mayans et al. 2012). Importantly, the influence of self-massage with a device on flexibility and muscle force capacity has not been evaluated in middle-aged adults.

Little is known about the influence of stretching, with or without the addition of self-massage, on range of motion and force production in middle-aged adults. The purpose of our study was to compare the influence of stretching with and without self-massage using therapy balls on range of motion about the ankle joint, MVC torque and muscle activity of the plantar flexors, and subcutaneous tissue thickness over the distal medial gastrocnemius muscle in young and middle-aged adults. We had three hypotheses. First, baseline dorsiflexion range of motion would be less in middle-aged adults compared with young adults. Second, self-massage would increase range of motion more than stretch alone for both young and middle-aged adults. Third, the greater range of motion after self-massage would be accompanied by a reduction in subcutaneous tissue thickness and an increase in MVC torque.

MATERIALS AND METHODS

Experimental Design

The study comprised a counter-balanced, within-subject design with two treatment arms (Figure 1). Participants visited the laboratory at the same time of day on two occasions with at least two days between visits. Each session began with the participant walking for five minutes as a warm-up activity. The individual then lay prone on an examination table with the right foot secured in a foot plate that was connected to a force transducer. Subcutaneous tissue thickness over the distal medial gastrocnemius at the level of the musculotendinous junction was assessed.
with ultrasound imaging prior to participants performing isometric MVCs. Participants subsequently performed one of the two treatments with the right plantar flexors: stretching only with a control condition or stretching after self-massage with therapy balls. The other treatment arm was performed on the second day. The influence of each treatment on range of motion was characterized by measuring maximal ankle dorsiflexion angle prior to beginning one of the treatments, immediately after the last stretch, and five minutes later. Ultrasound imaging and the isometric force task were repeated after the final range-of-motion assessment.

**Figure 1.** Schematic diagram of the study protocol.

**Participants**

Fifteen young adults (9 women; age, 25 ± 3 yr; height, 172 ± 10 cm; body mass, 54 ± 12 kg; BMI, 22 ± 2) and fifteen middle-aged adults (9 women; age, 45 ± 5 yr; height, 173 ± 9 cm; body mass, 73 ± 14 kg; BMI, 24 ± 3) with no history of neuromuscular impairment or lower limb
injury volunteered for the study. Participants were asked to refrain from performing strenuous lower body exercise or stretching for 24 hours prior to testing. We estimated that a sample of 12 participants per group would be needed to achieve a power of 0.8 ($\alpha = 0.5$), based on range of motion changes from a previous study. All participants provided written informed consent before undergoing any study procedures. The University Institutional Review Board approved the study (protocol #17-0111).

**Procedures**

Each session involved the participants performing three repetitions a 30-s standing calf stretch with 30 s of rest between repetitions. The self-massage procedure required participants to massage the triceps surae for 60 s using two massage therapy balls contained in a mesh bag prior to performing each of the three 30-s stretches. In the stretch-only session, participants rested for the same duration (60 s) with the legs in the same position as the self-massage session (Figure 2).

![Figure 2](image_url)

**Figure 2.** A) Participants were instructed to perform a calf stretch using a staggered stance with the left leg in front, and hands against a wall for support. B) Participants massaged the calf muscles with two therapy balls bound together in a tote sack.
Stretching

Study participants were instructed to perform a standing calf stretch of the right leg with both hands flat against a wall while maintaining an almost vertical torso. Participants used a staggered stance with the left foot approximately 30 cm from the wall, the left knee bent, and the right foot behind the body, and the heel remaining on the ground at a distance that elicited a maximal tolerable stretch of the calf muscles.

Self-massage with therapy balls

Participants applied 60 s of self-massage to the calf muscles of the right leg using massage therapy balls (Yoga Tune Up® Therapy Balls, Tune Up Fitness Worldwide Inc., Sherman Oaks, CA). The procedure began with the participant seated, the right leg extended, and the left knee flexed with the foot on the ground. The instructions were to push down with the hands to lift the hips off the ground, and actively slide the extended (right) leg back and forth across the therapy balls to massage the region spanning from the Achilles tendon to the popliteal fossa. The medial, mid-sagittal, and lateral portions of the calf muscles were massaged for 20 s each. Participants were instructed to perform the massage procedure with a pressure that resulted in a discomfort level of up to 5 on a numerical rating scale with 0 corresponding to no pain or discomfort and 10 representing maximal discomfort.

Range of motion

A digital goniometer connected to an amplifier (Biometrics Ltd, Newport, UK) was used to measure ankle dorsiflexion angle. With the participant standing in a neutral position, the goniometer was attached to the lateral surface of the right ankle using double-sided tape. One end was mounted over the calcaneus, parallel to the floor, and the other end was attached to the lateral side of the lower leg so that the goniometer passed over the middle of the lateral
malleolus. Participants were monitored during the maximal range-of-motion assessments to ensure appropriate body position, which required an erect torso, right knee fully extended, and right heel on the ground.

*Ultrasound imaging*

Subcutaneous tissue thickness was measured with a portable musculoskeletal ultrasonic imaging system (DP-30, Mindray North America, Mahwah, NJ, USA) with a 7.5 MHz linear transducer (38 mm footprint). The ultrasound probe was positioned over the muscle-tendon junction of the medial gastrocnemius, defined as the convergence of the deep and superficial aponeuroses. The perpendicular distance between the epidermis of the skin and insertion of the gastrocnemius into the Achilles tendon (subcutaneous tissue thickness) was measured using the native software on the device (Figure 3). A skin marker was used to record the placement of the transducer to ensure consistency across the two testing sessions.

![Figure 3](image.png)

*Figure 3.* The musculotendinous junction of the medial gastrocnemius and the Achilles tendon (1 on the image) was visualized on ultrasound imaging. The distance from the epidermis to the specified location was measured before and after each condition.
**Force measurements**

The strength of the calf muscles was quantified as the peak force achieved during an isometric MVC with the ankle joint positioned at 90 degrees. Participants lay prone on the testing apparatus with both legs extended and the right foot secured in a foot plate connected to a force transducer. The task involved gradually increasing plantar flexor force from rest to maximum in 3 s and then maintaining maximal force for 3 s. This procedure was repeated until the peak force for two MVCs differed by less than 5%.

Force was measured with a strain-gauge transducer (MLP-300, Transducer Techniques, Temecula, CA). The force signal was digitized at 1,000 samples/s (Coulbourn Instruments, Allentown, PA) and stored on a computer for subsequent processing with a second-order, low-pass Butterworth filter with 100 Hz cutoff frequency (Spike2, Cambridge Electronic Design Limited, Cambridge, England).

**Electromyography**

Muscle activity during the force tasks was recorded from medial gastrocnemius and soleus using high-density surface EMG electrodes (Sessantaquattro, OT Bioelettronica, Turin, Italy; 2 x 8 grid array, inter-electrode distance 10 mm). A single ground electrode was placed over the medial malleolus. To reduce impedance, the skin surface area was shaved, abraded, and cleansed with alcohol prior to attaching the electrodes. Electrodes were secured on the skin with Omnifix retention tape (Hartmann USA, Inc., Rock Hill, SC). Electrode placement was marked on the skin to ensure consistent placement across testing sessions. EMG signals were band-pass filtered (10 - 500 Hz), sampled at 2048 Hz, and recorded on a computer.

Prior to initiating the study, the reliability of high-density EMG recordings was assessed. We recorded absolute EMG amplitude for the medial gastrocnemius and soleus at 50, 75, 85, and
100% MVC force in five individuals on two occasions. The correlations between absolute EMG amplitude (µV) and normalized muscle force (% MVC) across force levels were between 0.95 and 0.99 across individuals and testing sessions.

Data analysis

The peak force recorded during the MVC was converted to a torque by calculating the product of the peak force and the moment arm, which was measured as the perpendicular distance from the line of action of the applied force to the center of the ankle joint. Muscle activity was recorded using a 2 x 8 electrode grid array, which resulted in 16 single-differential signals from adjacent sites on the array. The sum of the rectified and averaged signals during a 1-s interval that included the peak force was used to calculate the amplitude of the muscle activity.

Statistics

The data are presented as means ± SD. A linear mixed effect model was performed to examine the effects of age, treatment, time, and interactions on outcome measures. Post hoc pairwise comparisons were used to evaluate significant effects. Effect size was calculated using Hedge’s g. Correlation coefficients (Pearson’s product-moment correlation or Spearman’s rank correlation) were calculated between baseline flexibility and changes in range of motion and MVC torque, and change in MVC torque and EMG amplitude. All statistics were performed in R version 3.4.

RESULTS

A linear effects mixed model was performed controlling for age, time, treatment, and the interaction between time and treatment on range of motion and MVC torque. There was no main effect for age on range of motion (p = 0.42) or MVC torque (p = 0.84). However, a main effect
was observed for time and treatment for both range of motion ($p < 0.001$) and MVC torque ($p = 0.011$). Post-hoc pairwise comparison revealed that the increase in maximal ankle dorsiflexion angle was significantly greater after the stretch + self-massage treatment ($p = 0.0002$, 95% CI = 6.22 - 17.64) (stretch, 12 ± 9 %; stretch + self-massage, 24 ± 15 %; Hedge’s g = 0.98) (Figure 4A). Values for each age group and treatments are presented in Table 1. Similarly, a main effect was observed for the interaction of treatment and time on MVC torque ($p = 0.011$). Pairwise comparison revealed the change in MVC torque was significantly different between treatments ($p = 0.001$, 95% CI = 6.8 – 24.5) with a mean ± SD change of -4 ± 16 % for stretch alone and 12 ± 16% after stretch + self-massage (Hedge’s g = 0.97) (Figure 4B). Changes in MVC torque were correlated with changes in absolute EMG amplitude for both stretch alone ($r = 0.61$, $p = 0.028$) and stretch + self-massage ($r = 0.85$, $p = 0.0005$) (Table 2).

**Table 1.** Mean ± SD for range of motion (°).

<table>
<thead>
<tr>
<th></th>
<th>Young</th>
<th>Middle-aged</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Stretch only</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before</td>
<td>28.1 ± 5.5</td>
<td>26.4 ± 4.5</td>
</tr>
<tr>
<td>After</td>
<td>30.8 ± 6.2</td>
<td>29.3 ± 3.8</td>
</tr>
<tr>
<td>Final</td>
<td>30.7 ± 6.1</td>
<td>29.9 ± 3.8</td>
</tr>
<tr>
<td>% Change</td>
<td>9.6 ± 7.8</td>
<td>14.4 ± 9.9</td>
</tr>
<tr>
<td><strong>Stretch + self-massage</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before</td>
<td>27.9 ± 4.7</td>
<td>25.9 ± 4.4</td>
</tr>
<tr>
<td>After</td>
<td>32.3 ± 6.3</td>
<td>30.5 ± 4.1</td>
</tr>
<tr>
<td>Final</td>
<td>33.7 ± 6.7</td>
<td>32.6 ± 4.4</td>
</tr>
<tr>
<td>% Change</td>
<td>20.7 ± 13.2</td>
<td>27.2 ± 16</td>
</tr>
</tbody>
</table>
Table 2. Mean ± SD for torque (N·m) and sum of the EMG amplitude (µV) for medial gastrocnemius (MG) and soleus (SOL) muscles during the plantar flexor MVC.

<table>
<thead>
<tr>
<th></th>
<th>Young Adults</th>
<th></th>
<th>Middle-aged Adults</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MVC torque</td>
<td>MG+SOL</td>
<td>MVC torque</td>
<td>MG+SOL</td>
</tr>
<tr>
<td>Stretch</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before</td>
<td>92 ± 31</td>
<td>16.0 ± 6.8</td>
<td>88 ± 39</td>
<td>11.6 ± 5.6</td>
</tr>
<tr>
<td>After</td>
<td>88 ± 27</td>
<td>14.4 ± 5.7</td>
<td>86 ± 38</td>
<td>11.7 ± 5.9</td>
</tr>
<tr>
<td>% Change</td>
<td>-2.4 ± 14.5</td>
<td>-5.05 ± 23.9</td>
<td>-5 ± 18</td>
<td>3.4 ± 24.7</td>
</tr>
<tr>
<td>Stretch + self-massage</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before</td>
<td>95 ± 31</td>
<td>14.8 ± 6.5</td>
<td>91 ± 37</td>
<td>11.5 ± 5.3</td>
</tr>
<tr>
<td>After</td>
<td>101 ± 26</td>
<td>17.5 ± 7.4</td>
<td>103 ± 42</td>
<td>14.1 ± 7.6</td>
</tr>
<tr>
<td>% Change</td>
<td>11 ± 20</td>
<td>25.2 ± 33.8</td>
<td>13 ± 12</td>
<td>22.4 ± 21.8</td>
</tr>
</tbody>
</table>
**Figure 4.** Percentage changes in range of motion for ankle dorsiflexion (A) and MVC torque (B) after each treatment (stretch alone and stretch + self-massage).
Baseline range of motion was correlated with change in range of motion for both stretch + self-massage ($r = -0.7, p = 0.005$) and stretch alone ($r = -0.6, p = 0.02$) in middle-aged adults, but not young adults. The negative correlations indicate that less flexible middle-aged adults experienced greater improvements in ankle dorsiflexion range of motion. The change in plantar flexor MVC torque after each treatment was not correlated with baseline or change in range of motion across age groups or treatments. There were no main effects of age ($p = 0.2$) or treatment ($p = 0.1$) on subcutaneous thickness, as measured with ultrasound.

**DISCUSSION**

The goal of our study was to assess the influence of adding self-massage using therapy balls to a stretching protocol on ankle flexibility, plantar flexor muscle force and EMG amplitude, and changes in tissue thickness over the plantar flexors in young and middle-aged adults. Consistent with two of our hypotheses, the addition of self-massage increased the range of motion and MVC torque more than stretch alone for both age groups. Contrary to our third hypothesis, there were no statistically significant changes in subcutaneous thickness by age or treatment.

In contrast to our first hypothesis, the baseline dorsiflexion angle was not statistically different between our young and middle-aged participants. This result differs from the reduced flexibility observed in middle-aged men compared with young men on the sit-and-reach (-23%) and prone hip-extension (-17.5%) tests (Behm et al. 2011). Nonetheless, we found that less flexible middle-aged adults experienced greater increases in range of motion after both treatments, especially after the self-massage treatment. Furthermore, the self-massage treatment increased MVC torque and was correlated with an increase in the amplitude of muscle activation
for both age groups. As declines in ankle dorsiflexor and plantar flexor strength are related to increased injury risk and a proximal shift in joint power during walking (Fong et al. 2011; Kulmala et al. 2014; Bell-Jenje et al. 2016), incorporating self-massage within stretching protocols may be an effective approach to slowing these adaptations.

The possible explanations for the increased flexibility and muscle force after the self-massage treatment include changes in local fluid dynamics, reduced reflex responsiveness, improved stretch tolerance, and increased blood flow. The potential influence of local fluid dynamics is based on the observation that higher fluid pressure within skeletal muscle increases passive tension (Sleboda and Roberts 2017). This same effect has been hypothesized for fascia; higher water content results in increased fascial stiffness (Schleip et al. 2012a). Self-massage with an implement may temporarily decrease fluid content and pressure in the local area, decreasing passive tension, and allowing for greater tissue mobilization before the tissue rehydrates (Schleip et al. 2012a; Beardsley and Škarabot 2015). We hypothesized that we would be able to detect this change by measuring subcutaneous tissue thickness on ultrasound imaging, but our method was not sensitive enough to detect such changes. Doppler ultrasound may be able to provide greater resolution to detect tissue hydration changes in future research studies.

Increases in range of motion after a self-massage treatment may be due to the modulation of neural pathways through activation of mechanoreceptors. The musculoskeletal system is richly innervated by sensory receptors in the skin, muscle, superficial fascia, and deep fascia (Simmonds et al. 2012; Gesslbauer et al. 2017) that may prime the somatosensory cortex by reporting changes in tissue length and pressure prior to stretching (Abraira and Ginty 2013). For example, a short bout of self-massage using tennis balls on the soles of the feet increased performance on the sit-and-reach test, indicating improved hamstring and lumbar spine
flexibility (Grieve et al. 2015). Similarly, studies that evaluated changes in spinal reflex responsiveness after various massage techniques report a reduction in H-reflex amplitude after treatment (Behm et al. 2013). For example, soleus H-reflex amplitude declined after painful (-58%) and uncomfortable (-43%) massage using a roller stick (Young et al. 2018). The reduction in reflex responsiveness may be due to increased la presynaptic inhibition or increased inhibitory synaptic input from cutaneous mechanoreceptors onto spinal neurons (Hultborn 2001). It seems possible, therefore, that our self-massage treatment may have similarly reduced reflex responsiveness to allow for a greater range of motion. However, such an effect cannot explain the observed increase in MVC torque.

Changes in flexibility have been attributed to adaptation in muscular components and reflex pathways. However, an increased tolerance to stretch discomfort (stretch tolerance) is often the primary explanatory mechanism for improvements in flexibility (Law et al. 2009). Increased stretch tolerance by overloading cutaneous afferents, effectively dulling pain, is a plausible explanation for increased flexibility after adding self-massage. The pressure applied to the skin stimulates low-threshold mechanoreceptors, which can temporarily reduce pain sensation (Habig et al. 2017). For example, pain pressure threshold, measured with a pressure algometer, increased (decrease in pain sensation) after a 3-min bout of foam rolling over the iliotibial band in asymptomatic individuals (Vaughan and McLaughlin 2014). This effect was also observed in symptomatic individuals. Similarly, pain pressure threshold increased after a therapist applied light or deep massage over tender spots on the lower leg, but not after a sham treatment (Aboolardada et al. 2015). Interestingly, the effect was observed when the treatment was applied on either the affected or unaffected leg. These findings suggest the involvement of a central pain-modulating mechanism, such as the gate control theory. It is likely that the addition
of self-massage prior to stretching reduces the discomfort associated with stretching to the maximal tolerable limit as well as performing an MVC, thereby increasing the range of motion and peak torque during an MVC.

As observed previously, the addition of self-massage prevents the force deficit observed with the stretch alone. Even short bouts of static stretching may result in a temporary reduction in maximal muscle force (Ryan et al. 2008; Trajano et al. 2017b), which is not observed after self-massage (Cheatham et al. 2015; Capobianco et al. 2018). Self-massage of the plantar flexors with a stick increased MVC force by 8.2% at 10 min after the intervention, whereas static stretching decreased it by 4% (Halperin et al. 2014). Similarly, MVC torque increased by 16% after stretch + self-massage using therapy balls, but decreased by 1% after stretch alone (Capobianco et al. 2018). Outcomes from our study were similar with force decreasing, on average, after stretching and increasing after adding self-massage in both age groups.

One hypothesis to explain stretch-induced force deficit is a decrease in muscle activation during voluntary contraction (Trajano et al. 2017b). High-density surface EMG recordings allow for a more global measurement of muscle activity than bipolar recordings. In our study, changes in MVC torque and absolute EMG amplitude were strongly correlated, supporting the notion that changes in force are associated with changes in muscle activation. However, the correlations were lower than our reliability measurements, leaving the possibility that factors other than muscle activation may have contributed to the changes in MVC torque.

The greater gains achieved with the combination of stretching and self-massage with an instrument may improve peripheral vascular function. For example, 3 months of static stretching increased vascular endothelial function and decreased arterial stiffness (Shinno et al. 2017). Moreover, there was a 74% increase in blood flow immediately after foam rolling, and a 53%
increase from baseline at 30 min later (Hotfiel et al. 2017). The mechanical stress on endothelial cells from stretching and self-massage results in the release of nitric oxide, which increases vasodilation that may augment performance during a maximal contraction through increased phosphocreatine replenishment or speeding the return to baseline pH between contractions.

There are some limitations to our study. It is difficult to standardize the pressure of self-massage using the therapy balls, which may have contributed to variability in spinal reflex depression among our participants (Young et al. 2018) and thereby influenced the subsequent changes in range of motion. Although we asked subjects to maintain a moderate pressure during rolling, it is possible that some participants used a lighter pressure. Additionally, our ultrasound method likely did not have sufficient resolution to detect relatively small changes in subcutaneous thickness and the measurement should be repeated with more precise technology.

**Conclusions**

Self-massage with therapy balls prior to stretching increases ankle dorsiflexion range of motion more than stretching alone in both young and middle-aged adults. Furthermore, the addition of self-massage significantly increased plantar flexor MVC torque and was correlated with an increase in muscle activation of the calf muscles during an MVC. Both stretching treatments conferred a greater benefit in less flexible middle-aged adults, suggesting it may be more important for improving range of motion in individuals already exhibiting an age-related decline in range of motion.
CHAPTER IV

SACROILIAC JOINT DYSFUNCTION PATIENTS EXHIBIT ALTERED MOVEMENT STRATEGIES WHEN PERFORMING A SIT-TO-STAND TASK
ABSTRACT

**Background:** The ability to rise from a chair is a basic functional task that is frequently compromised in individuals diagnosed with orthopedic disorders in the low back and hip. There is no published literature that describes how this task is altered by sacroiliac joint dysfunction (SIJD).

**Purpose:** To compare lower extremity biomechanics and the onset of muscle activity when rising from a chair in individuals with SIJD and healthy persons.

**Study design:** Six women with unilateral SIJD and six age-matched healthy controls performed a sit-to-stand task while we measured kinematics, kinetics, and muscle activity.

**Methods:** Subjects stood up at a preferred speed from a seated position on an armless and backless adjustable stool. We measured kinematics with a 10-camera motion capture system, ground reaction forces for each leg with force plates, and muscle activity with surface electromyography. Joint angles and torques were calculated using inverse dynamics. Leg loading rate was quantified as the average slope of vertical ground reaction force during the 500-ms interval preceding maximal knee extension.

**Results:** Between-leg differences in loading rates and peak vertical ground reaction forces were significantly greater for the SIJD group than the control group. Maximal hip angles were significantly less for the SIJD group (p = 0.001). Peak hip moment in the SIJD group was significantly greater in the unaffected leg (0.75 ± 0.22 N•m/kg) than the affected leg (0.47 ± 0.29 N•m/kg, p = 0.005). There were no between-leg or between-group differences for peak knee or ankle moments. The onset of activity in the latissimus dorsi muscle on the affected side in the SIJD group was delayed and the erector spinae muscles were activated earlier than in Controls.
Conclusions: Individuals with SIJD have a greater vertical ground reaction force on the unaffected leg, generate a greater peak hip moment in the unaffected leg, use a smaller range of motion at the hip joint of the affected leg, and delay the onset of a key muscle on the affected side when rising from a seated position.
INTRODUCTION

Sacroiliac (SI) joint dysfunction (SIJD) is a significant cause of low back pain, accounting for up to 40% of nonspecific low back pain cases (Bernard and Kirkaldy-Willis 1987; Sembrano and Polly 2009). The impact of SIJD on quality of life is comparable to that observed with other debilitating orthopedic conditions, such as hip osteoarthritis and spinal stenosis, and is higher than many cardiovascular-related medical conditions (Cher et al. 2014). Indirect health-care expenditures associated with low back pain range from $7 - $28 billion per year and individuals lose an estimated 5.2 hours of work time per week (Stewart et al. 2003; Koenig et al. 2016). In a recent clinical trial on SI joint fusion, 19% of subjects were not working due to back pain (Polly et al. 2016).

The SI joint assists in transferring load from the spine to the lower extremities by providing the stability required to support this force transfer and permitting the mobility required to facilitate bipedal locomotion, postural changes, and expansion during parturition (Sturesson et al. 1989). The anatomical capabilities of the SI joint that enable these movements were characterized by Vleeming and colleagues with the principles of form and force closure (1990a, 1990b). Form closure refers to the stability provided by the complimentary convex and concave surfaces of the iliac and sacral joint surfaces. In women, the joint surface is somewhat smaller, less curved, and more backward tilted than in men, possibly contributing to the greater incidence of SIJD in women (Vleeming et al. 2012). Force closure describes the function of ligaments and muscles acting across the joint to provide joint mobility and stability. The primary muscles responsible for force closure are the gluteus maximus and latissimus dorsi as tension is transmitted from one side of the trunk to the other through their connection into the superficial...
layer of the thoracolumbar fascia (Vleeming et al. 1995; Pool-Goudzwaard et al. 1998; Carvalhais et al. 2013).

Diagnosing SIJD can be challenging as associated symptoms include pain at the low back and hip, in addition to pain at the joint itself (Polly et al. 2016). The etiology of SIJD is unknown, but one hypothesis is that it results from inadequate force closure needed to stabilize the joint during movement. Possible explanatory mechanisms include inadequate muscle coordination, reduced muscle strength, and alterations in ligamentous tension (Pool-Goudzwaard et al. 1998).

Joint and muscle pain are known to alter muscle activation, muscle coordination, normative gait patterns, and basic motor function (Graven-Nielsen et al. 1997; Falla and Hodges 2017). Joint loading, the control of joint movements through coordinated muscular contractions, and joint integrity and stability are critical factors that influence degenerative responses (Herzog et al. 2003). Load transfer through the pelvis with an unstable SI joint can produce excessive loads on surrounding tissues, resulting in pain (Pool-Goudzwaard et al. 1998). An individual may attempt to splint the painful area or avoid movements that exacerbate pain by changing the distribution of muscle activation (Hodges and Smeets 2015). The consequence of these changes may result in abnormal joint loading, which influences the control of muscle forces and limb coordination, and is associated with joint degeneration (or worsening thereof) (McCrory et al. 2001; Herzog et al. 2003). Moreover, leg-loading asymmetry that results from pain or injury on one side has been linked to the pathogenesis of osteoarthritis in the knee or hip of the healthy leg (McCrory et al. 2001).

The ability to rise from a chair is a basic functional task that requires approximately 60% of the available lumbar flexion range of motion (Hsieh and Pringle 1994). Torque at the hip joint
during this movement can exceed that observed during walking or climbing stairs, which may exacerbate symptoms in individuals suffering from SI joint pain (Gross et al. 1998).

Patients with SIJD present similarly to individuals with hip and spine disorders. Some studies have investigated sit-to-stand mechanics in individuals with end-stage hip osteoarthritis and low back pain, but there are no published studies that have examined this task in individuals with SIJD (Arendt-Nielsen et al. 1996; Al-Obaidi et al. 2003; Shum et al. 2005; Abujaber et al. 2015). The purpose of our study was to compare lower extremity biomechanics and the onset of muscle activity when rising from a chair in individuals with SIJD and healthy persons. We hypothesized that the unilateral discomfort experienced by this patient population would result in asymmetrical leg loading and alterations in the onset of muscle activity compared with healthy controls.

MATERIALS AND METHODS

Twelve women volunteered to participate in the study. Six individuals were diagnosed with unilateral SIJD by an orthopedic surgeon specializing in diagnosing and treating SIJD. Diagnostic criteria included a positive result on at least three out of five physical provocative maneuvers designed to target the SI joint and pain relief after localized steroid injections (Szadek et al. 2009). These criteria were used in a recent randomized controlled trial of SI joint fusion (Whang et al. 2015). Six healthy, age-matched (±5 yrs) women without back, SI joint, or hip pain, served as the control group (Control). Study exclusion criteria included: bilateral symptoms (SIJD group), severe back or hip pain from other causes, prior spinal fusion at any level, major lower extremity surgery, difficulty walking due to conditions other than SIJD, severe arthritis, neurological disorders, vestibular or visual disturbance affecting balance, fibromyalgia, chronic rheumatologic condition, drug or alcohol abuse, or uncontrolled psychiatric disease. All study
participants were asked to rate pain severity on a visual analog scale (VAS), measured from 0 to 10 cm, and complete the Oswestry Disability Index (ODI) to assess overall disability (Fairbank and Pynsent 2000). The study was approved by the University of Colorado Boulder Institutional Review Board (protocol #15-0586). Written informed consent was obtained from all participants before beginning study-related procedures.

**Procedures**

*Sit-to-stand task*

An armless, backless piano stool was placed on two force plates (Bertec Corp., Columbus, OH) with each foot on a separate force plate. The stool was adjusted so that the top was even with the back of the participant’s knee joint line (popliteal fossa) (Abujaber et al. 2015). With arms placed across the chest, the participant stood up until fully upright, and then sat back down. This task was performed at a self-selected speed and repeated three times. We concurrently recorded kinematic, kinetic, and electromyographic data throughout each task.

*Motion analysis*

Kinematic data were obtained with a 10-camera motion capture system (VICON, Centennial, CO). The data were recorded at 100 Hz and low-pass filtered (2nd order bidirectional Butterworth, 10-Hz cutoff). Spherical reflective markers were placed on the sternal notch, C7 spinous process, and bilaterally on the acromioclavicular joint, scapula, fifth lumbar vertebrae, iliac crest, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), greater trochanter, medial femoral condyle, lateral femoral condyle, medial malleolus, lateral malleolus, head of the first metatarsal, head of the fifth metatarsal, and posterior heel. A rigid thermoplastic shell with four reflective markers was placed bilaterally on the thighs and shanks to track the motion of these segments.
Ground reaction forces were measured for each leg at 1000 Hz and then low-pass filtered (Bidirectional 2\textsuperscript{nd} order Butterworth, 30-Hz cutoff).

Wireless surface EMG signals (Noraxon USA, Scottsdale, AZ) were sampled at 1000 Hz and used to monitor activity in seven muscles bilaterally. Electrodes were placed according to SENIAM guidelines over the following muscles: latissimus dorsi, erector spinae, gluteus maximus, biceps femoris, medial gastrocnemius, rectus femoris, and tibialis anterior (Hermens et al. 2000).

**Data Analysis**

Joint angles and moments of force (torque) were calculated using inverse dynamics with Visual 3D (C-Motion, Germantown, MD). Leg loading rate was calculated by taking the average slope of the vertical ground reaction force during the 500-ms window preceding peak knee extension. Initiation of the sit-to-stand task (movement onset) was defined as the time at which the total vertical ground reaction force first exceeded body weight by 5%.

EMG signals were band-pass filtered (10-490 Hz) and full-wave rectified. Two investigators independently determined muscle onset using visual inspection (Jesunathadas et al. 2012). Muscle onset is reported as the percentage of time from movement onset to peak knee extension.

All data were processed and analyzed using a custom Matlab script (MathWorks, Inc., Natick, MA). Data were assessed for normality using the Shapiro-Wilk test. Normally distributed data were compared by group using an independent groups two-tailed Student’s t-test. Within-subject data were compared using paired t-tests. Group-by-leg interaction for muscle onset was assessed using a two-way ANOVA. Significance was set at \( \alpha = 0.05 \). Effect size was calculated using Hedge’s \( g \) with small, medium, and large effect sizes are defined as 0.2, 0.5 and 0.8,
respectively. Statistics were performed in R and the results are reported as mean ± SD (R Core Team (2013) date unknown).

**RESULTS**

Mean (± SD) participant age was 39 ± 7 yrs for the SIJD group and 37 ± 8 yrs for Controls. There were no statistically significant differences in height (168 ± 14 cm, 162 ± 6 cm, p > 0.05) or body mass (72 ± 14 kg, 59 ± 6 kg, p = 0.07), but BMI was greater for the SIJD group (27.7 ± 5.0) than for Controls (21.2 ± 2.0) (p=0.02). SIJD symptom duration was 24 ± 11 months and ODI score was 41 ± 18% (Table 1). The VAS pain scores for the SIJD group were: low back, 6.3 ± 1.5 cm; affected leg, 6.7 ± 1.9 cm; unaffected leg, 2.7 ± 2.3 cm. Control subjects reported 0 cm on VAS and 0% on ODI.

**Table 1: Oswestry Disability Index (ODI) and pain scores measured on a visual analog scale (VAS)**

<table>
<thead>
<tr>
<th>Subject</th>
<th>ODI score (%)</th>
<th>VAS: Low Back</th>
<th>VAS: Affected leg</th>
<th>VAS: Unaffected leg</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>44</td>
<td>7</td>
<td>5</td>
<td>3</td>
</tr>
<tr>
<td>2</td>
<td>16</td>
<td>3</td>
<td>4</td>
<td>0</td>
</tr>
<tr>
<td>3</td>
<td>54</td>
<td>7</td>
<td>7</td>
<td>3</td>
</tr>
<tr>
<td>4</td>
<td>52</td>
<td>8</td>
<td>8</td>
<td>6</td>
</tr>
<tr>
<td>5</td>
<td>58</td>
<td>8</td>
<td>7</td>
<td>4</td>
</tr>
<tr>
<td>6</td>
<td>22</td>
<td>7</td>
<td>7</td>
<td>0</td>
</tr>
</tbody>
</table>

VAS, 0 – 10 cm: 0 = no pain, 10 = maximal pain. ODI score: 0 - 20% minimal, 21 - 40% moderate, 41 - 60% severe
Figure 1: Vertical ground reaction forces (VGRF) for each side (above) and knee angle (dashed line below) for a subject in the SIJD group (A) and a control subject (B). The solid trace indicates the affected leg (A) and the right leg (B). The vertical line indicates movement onset when the summed VGRFs first exceeded body weight by 5%. Average loading rates for each leg were calculated from the 500-ms window preceding peak knee extension (shaded area).

Average leg loading rate was symmetric for the Control group (13 ± 11 N/s difference between sides), but was significantly different between legs for the SIJD group (112 ± 48 N/s).
difference) \((p = 0.008, \text{Hedge's } g = 1.07)\) (Table 2, Figure 1). Although the difference in peak vertical ground reaction force between legs was negligible for the Control group \((3 \text{ N})\), there was a significant difference \((103 \text{ N})\) between legs for the SIJD cohort \((p < 0.0001, \text{Hedge's } g = 1.17)\).

**Table 2:** Results for the difference between legs in loading rate, peak vertical ground reaction force, and minimal and maximal hip flexion angles.

<table>
<thead>
<tr>
<th>Difference between sides</th>
<th>Hip angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Leg loading rate (N/s)</td>
</tr>
<tr>
<td>SIJD</td>
<td>112 ± 48*</td>
</tr>
<tr>
<td>Control</td>
<td>13 ± 11</td>
</tr>
</tbody>
</table>

* \(p < 0.05\) between groups. VGRF = vertical ground reaction force. Maximal hip angle occurred during the seated position, whereas minimal hip angle corresponded to full extension when standing.

Minimal and maximal hip angles were symmetric between legs for both groups, with no significant between-group differences for minimal hip angle (Table 2). However, maximal hip angle was significantly less in the SIJD group \((59.9 ± 9.9°)\) than the Controls \((85.2 ± 9.9°; p = 0.001)\).

Peak hip-flexion moment in the sagittal plane was not statistically different between legs for Controls \((p = 0.4)\) (Table 3), but was significantly greater on the affected side \((0.47 ± 0.29 \text{ N} \cdot \text{m/kg})\) than the unaffected side \((0.75 ± 0.22 \text{ N} \cdot \text{m/kg})\) in the SIJD group \((p = 0.005)\). The difference in peak hip moment between legs was significantly different between groups \((p = 0.005, \text{Hedges } g = 0.87)\). There were no between-legs or between-group differences for peak knee or ankle moments.
Table 3: Normalized peak hip moment (torque) and peak vertical ground reaction force (VGRF) results by side and group.

<table>
<thead>
<tr>
<th></th>
<th>Control</th>
<th></th>
<th>SIJD</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left</td>
<td>Right</td>
<td>Unaffected</td>
<td>Affected</td>
</tr>
<tr>
<td>Peak Hip Moment</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Normalized to body mass (N•m/kg)</td>
<td>0.83 ± 0.24</td>
<td>0.87 ± 0.19</td>
<td>0.75 ± 0.22</td>
<td>0.47 ± 0.29*</td>
</tr>
<tr>
<td>Normalized to body mass and height (N•m/kg•m)</td>
<td>0.49 ± 0.13</td>
<td>0.52 ± 0.10</td>
<td>0.47 ± 0.13</td>
<td>0.29 ± 0.18*</td>
</tr>
<tr>
<td>Peak VGRF (N/BW)</td>
<td>0.65 ± 0.02</td>
<td>0.65 ± 0.05</td>
<td>0.65 ± 0.3</td>
<td>0.58 ± 0.02*</td>
</tr>
</tbody>
</table>

* p < 0.05 between legs. BW = body weight

Table 4: Muscle onset timing as a percentage of the time from movement onset to peak knee extension.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Control (%)</th>
<th></th>
<th>SIJD (%)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left leg</td>
<td>Right leg</td>
<td>Unaffected</td>
<td>Affected</td>
</tr>
<tr>
<td>Latissimus dorsi</td>
<td>12 ± 2</td>
<td>13 ± 2</td>
<td>9 ± 3</td>
<td>12 ± 5</td>
</tr>
<tr>
<td>Erector spine</td>
<td>12 ± 3</td>
<td>11 ± 2</td>
<td>9 ± 3</td>
<td>8 ± 4</td>
</tr>
<tr>
<td>Gluteus maximus</td>
<td>18 ± 3</td>
<td>17 ± 5</td>
<td>14 ± 4</td>
<td>15 ± 5</td>
</tr>
<tr>
<td>Biceps femoris</td>
<td>21 ± 5</td>
<td>21 ± 6</td>
<td>18 ± 6</td>
<td>18 ± 3</td>
</tr>
<tr>
<td>Medial gastrocnemius</td>
<td>14 ± 4</td>
<td>14 ± 4</td>
<td>15 ± 9</td>
<td>15 ± 7</td>
</tr>
<tr>
<td>Rectus femoris</td>
<td>18 ± 7</td>
<td>19 ± 5</td>
<td>15 ± 3</td>
<td>14 ± 2</td>
</tr>
<tr>
<td>Tibialis anterior</td>
<td>3 ± 6</td>
<td>2 ± 6</td>
<td>3 ± 5</td>
<td>2 ± 4</td>
</tr>
</tbody>
</table>
Figure 2: Representative EMG recordings (mV) for an SIJD subject for the right side. Solid vertical line indicates movement onset when the summed vertical ground reaction forces first exceeded 5% of body weight. Dashed vertical line indicates time of peak knee extension. LD = latissimus dorsi; ES = erector spinae; GMx = gluteus maximus; BF = biceps femoris; MG = medial gastrocnemius; RF = rectus femoris; TA = tibialis anterior.

Average onset times for the recorded muscles occurred after the vertical ground reaction force first exceeded 5% of body weight, which indicates that other muscles contributed to the initial vertical acceleration of the center of mass (Figure 2). The onset of EMG activity for all muscles was symmetrical between sides for the Control group (Table 4). Latissimus dorsi was activated at the same time in both legs for the Control group, but the onset was delayed on the affected side (12 ± 6 %) relative to the unaffected side (9 ± 3 %) in the SIJD group (p=0.05). However, variability of latissimus dorsi onset, as indicated by the coefficient of variation ((standard deviation/mean) x 100) was higher for the SIJD group (40%) than for the Control
group (20%). Additionally, SIJD subjects activated erector spinae earlier (unaffected: 9 ± 3%; affected: 8 ± 4%) than Controls (right: 12 ± 3%; left: 11 ± 2%) (p=0.05).

DISCUSSION

This is the first study to evaluate sit-to-stand kinematics, kinetics, and the onset of muscle activity in individuals with SIJD. Consistent with our hypothesis, individuals with SIJD preferentially loaded the unaffected leg when rising from a chair and when fully erect, thereby reducing the load on the affected leg. Furthermore, they used a smaller range of motion at the hip joint of the affected leg and produced a greater peak hip moment on the unaffected side. Additionally, individuals with SIJD activated the erector spinae muscles (bilaterally) earlier than Controls and exhibited a delayed onset of the latissimus dorsi muscle on the affected side, which suggests disruption in the force-closure mechanism between sides.

Our results illustrate that the compensatory movement strategies adopted by persons with SIJD are similar to those observed in individuals suffering from low back pain and severe hip osteoarthritis. Shum et al. (2005) evaluated hip flexion in 60 individuals with low back pain (LBP) during a sit-to-stand task and found values similar to those for our SIJD cohort: 64 ± 11° for LBP; 60 ± 10° for SIJD. In a study evaluating sit-to-stand performance in individuals with severe hip osteoarthritis, Abujaber et al. (2015) reported differences in peak hip moments (normalized to body mass and height) between the unaffected (0.48 N•m/kg•m) and affected sides (0.36 N•m/kg•m). Our results were more asymmetric (unaffected: 0.47 N•m/kg•m; affected: 0.29 N•m/kg•m,) indicating a greater adjustment in our SIJD group. Similarly, peak vertical ground reaction forces were greater in the unaffected side in both our study and one on individuals prior to total hip arthroplasty (McCrory et al. 2001). As abnormal joint loading has been shown to result in joint degeneration on the healthy limb, it is possible that individuals with
SI joint dysfunction may develop early osteoarthritis in the hip of the unaffected side if left untreated (McCrory et al. 2001).

The onset of muscle activity was variable between individuals and between our two groups. SIJD subjects activated their erector spinae muscles earlier than the Control subjects, which is consistent with our finding that individuals with SIJD have smaller hip-flexion angles and thus extend at the hip earlier than Controls. However, variability within each group was high. In the presence of pain, individuals may adopt different motor strategies to accomplish the same task (Hug and Tucker 2017; van Dieën et al. 2017). The delay in activation of the latissimus dorsi on the affected side of SIJD participants suggests a possible disturbance of the force-closure mechanism. Previous research has shown that individuals with low back pain exhibit decreased anticipatory responses in preparation for an impending perturbation (Hodges and Richardson 1996; Radebold et al. 2000; Nelson-Wong et al. 2013). Latissimus dorsi provides an essential component of the force-closure mechanism for the SI joint, important for stability during weight-bearing. The delay in latissimus dorsi activation suggests deficient movement preparation for rising from a chair in the SIJD group. Further studies with larger sample sizes are needed to explore differences in muscle activation in people with SIJD.

Our study has limitations. Our rigorous inclusion criteria made subject recruitment challenging. However, this was necessary to ensure the study population had SIJD and not any other back- or hip-related condition. Nonetheless, peak hip flexion moments (Abujaber et al. 2015) and maximal hip flexion angles (Shum et al. 2005) for the Control group were similar to those reported for larger sample sizes. Despite the small sample size, the effect sizes were typically large. Our SIJD cohort, however, was overweight (BMI: SIJD, 27, vs Control, 22), which may have accounted for the smaller peak hip moment on the unaffected side (SIJD, 0.79...
N•m/kg; Control, 0.83 N•m/kg). For example, obese individuals (BMI = 35) generate a lower torque around the hip joint during a sit-to-stand task (peak hip torque: obese, 0.59 N•m/kg, control 0.98 N•m/kg) (Sibella et al. 2003). Furthermore, muscle activation timing was variable across individuals. This may have been attributable to the sit-to-stand task being performed at a self-selected speed.

The data from our study represent the first step in understanding the kinematics and kinetics in individuals with SIJD during an activity of daily living. Many cases of SIJD are misdiagnosed as low back pain, perhaps due to poor diagnostic criteria or lack of understanding of the disease presentation (Bernard and Kirkaldy-Willis 1987; Sembrano and Polly 2009). Individuals with SIJD can present with pain in one or more regions including the sacroiliac joint, gluteal region, low back, and hip joint. Our study shows that persons with SIJD stand up from sitting using a similar strategy to those with chronic low back pain and severe hip osteoarthritis. Misdiagnosis of SIJD may lead to unnecessary lumbar spinal surgery (Polly and Cher 2016). Katz et al. (2003) found that SIJD was the etiology of low back pain in 32% of patients treated with lumbar fusion surgery. The similarities in symptomology and physical presentation between low back pain and SIJD, and the possible consequence of misdiagnosis, underscore the need for a careful clinical examination to include the SIJ.

Motor coordination has been shown to change with various musculoskeletal disorders, such as low back pain and anterior knee pain (Tsao et al. 2010). Studies in individuals with low back pain show that non-specific conservative management results in small-to-moderate effects that are not maintained over time (Falla and Hodges 2017). In a randomized trial, manual therapy aimed at mobilizing the SIJ was found to be more effective than intraarticular joint injections or physiotherapy in individuals with SIJ-related back and leg pain (Visser et al. 2013). Moreover,
skilled training, but not general exercise, results in cortical reorganization and significant long-term improvements in individuals with low back pain (Tsao et al. 2010). Taken together, these observations suggest that changes in the strategy used to perform the sit-to-stand task could be used to further refine diagnostic criteria and contribute to the development of targeted physiotherapy treatment.

CONCLUSION

We found that individuals diagnosed with SIJD adopt similar movement patterns when rising from a chair as individuals with low back pain and severe hip osteoarthritis. They preferentially loaded the unaffected leg, maintained this asymmetry when fully upright, and experienced a larger peak hip moment on the affected side. Furthermore, the onset of muscle activity was disrupted in key muscles providing force closure of the joint. Thus, it is imperative that we seek to further understand this disease and how to differentiate it from other lumbopelvic disorders.
CHAPTER V

INDIVIDUALS WITH SACROILIAC JOINT DYSFUNCTION DISPLAY A DEPRESSED MUSCLE SYNERGY BETWEEN THE GLUTEUS MAXIMUS AND CONTRALATERAL LATISSIMUS DORSI WHEN WALKING
ABSTRACT

Introduction: Sacroiliac joint dysfunction (SIJD) is a debilitating condition that involves pain in the lower back. In healthy individuals, the combined actions of gluteus maximus and contralateral latissimus dorsi likely stabilize the joint and minimize displacement of the involved joint surfaces. The purpose of our study was to compare the muscle weights of identified synergies between individuals with SIJD and control subjects when they walked at the same speed. We hypothesized that individuals with SIJD would lack a significant muscle synergy involving gluteus maximus on the affected side and contralateral latissimus dorsi when walking and that the timing of this muscle synergy would be more variable.

Methods: Electromyographic (EMG) signals were recorded from 16 muscles when individuals with SIJD and healthy control participants walked at 1 m/s on a force-measuring treadmill. Non-negative matrix factorization (NMF) was used to identify patterns of EMG activity (muscle synergies). The output from the NMF analysis comprised information about the timing of each synergy and the relative contribution (weight) of each muscle.

Results: Individuals with SIJD lacked a muscle synergy between gluteus maximus on the affected side and the contralateral latissimus dorsi, which was present in control subjects. Moreover, the timing of this muscle synergy was more variable for individuals with SIJD than healthy participants.

Conclusion: The results are consistent with the hypothesis that individuals with SIJD exhibit reduced coactivation of gluteus maximus and contralateral latissimus dorsi to help stabilize the sacrum and ilium during walking.
INTRODUCTION

Sacroiliac joint dysfunction (SIJD) is the etiology for 15-30% of lower back pain (Bernard and Kirkaldy-Willis 1987; Sembrano and Polly 2009) and causes debilitating pain that impairs mobility (Cher et al. 2014). Symptoms of SIJD often include pain in the low back and hip, in addition to pain at the actual joint, all of which can alter gait patterns and muscle activation strategies.

The sacroiliac joint assists in load transfer from the spine to the lower extremities (Sturesson et al. 1989). As such, it must provide the necessary stability for this force transfer while maintaining a degree of mobility that does not constrain activities of daily living. The joint experiences up to 4800 N of shear force, translates 1.6 mm in the anterior-posterior plane, and may rotate up to 4 degrees while ambulating (Sturesson et al. 1989). Moreover, when laying supine and performing a straight leg raise, the ilium of the rested leg rotates 0.8 degrees backwards and 0.3 degrees inward (Kibsgård et al. 2017).

The function of this joint has been described by Vleeming and colleagues with the principles of form and force closure (2012). Form closure refers to the alignment of the concave surface of the sacrum with the convex surface of the ilium, whereas force closure is provided by surrounding muscles, ligaments, and fasciae. A notable component of force closure is the connection from the gluteus maximus to the contralateral latissimus dorsi through the thoracolumbar fascia. Although the etiology of SIJD is unknown, one potential cause is the disruption of coactivation between these muscles to support force closure of the joint (Vleeming et al. 2012; Barker et al. 2014; Kibsgård et al. 2017).

Our study is the first to apply a muscle synergy analysis to the electromyographic (EMG) signals recorded during walking to quantify the relative activities of the gluteus maximus and
latissimus dorsi muscles in individuals with SIJD. A muscle synergy comprises a group of muscles that perform complimentary actions and can be activated with simple control strategies (Lee 1984; Olree and Vaughan 1995; Lee and Seung 2001; Tresch and Jarc 2009; Clark et al. 2010). Muscle synergies can be quantified by applying a factorization analysis, such as non-negative matrix factorization (NMF) (Lee and Seung 1999, 2001), to EMG data recorded during a cyclic task and identifying muscles with similar activation patterns (Ting and Macpherson 2005; Cappellini 2006; Tresch and Jarc 2009; Clark et al. 2010; Oliveira et al. 2014). The NMF algorithm factors the original EMG signals into two vectors: timing and muscle weighting. Muscles that are activated with similar spatiotemporal patterns are grouped into a synergy with the relative contribution of each muscle indicated by its weight.

Locomotion studies have found that three to five muscle synergies can account for 90% of the variance in EMG signals from many limb and trunk muscles during walking and running (Olree and Vaughan 1995; Cappellini 2006). For example, Olree and Vaughan (1995) characterized the muscle synergies required to reconstruct the original EMG signals as contributing to either a propulsion or loading function during walking. The propulsion synergy, which occurred in late stance, involved the gastrocnemius, adductor magnus, and hamstring muscles. The loading synergy, which appeared in early stance, involved gluteus maximus, gluteus minimus, and erector spinae muscles. Although multiple studies suggest low back pain alters muscle activation patterns (Hanada et al. 2011; van den Hoorn et al. 2015; Falla and Hodges 2017; van Dieën et al. 2017), no studies to date have examined muscle activation strategies of individuals with SIJD during gait. This information could provide clinicians with insight into how individuals with SIJD adapt to the pain and could lead to improved treatment plans.
The purpose of our study was to compare the muscle weights of identified synergies between individuals with SIJD and control subjects when they walked at the same speed. We hypothesized that individuals with SIJD would lack a significant muscle synergy involving gluteus maximus on the affected side and contralateral latissimus dorsi when walking and that the timing of this muscle synergy would be more variable.

METHODS

Participants

Six women (age = 37.2 ± 5.9 yrs, BMI = 29.5 ± 5.0) who were diagnosed with unilateral SIJD (2 left side, 4 right side), free of neurological disorders, without surgery to the lower limbs or back, who were not currently pregnant or pregnant within the past two years, and otherwise in good health and six healthy age-matched women (age: 38.8 ± 7.0 yrs, BMI = 21.3 ± 5.9) agreed to participate in the study. Inclusion criteria for the SIJD cohort included a positive result on at least three of five physical examination maneuvers specific to the diagnosis of SIJ disorders, relief on intra-articular joint injection, and pain for at least six months (Szadek et al. 2009). None of the participants indicated that their pain began in the peri-partum period. All individuals were diagnosed by a single physician to ensure diagnostic consistency. The protocol was approved by the IRB at the University of Colorado Boulder (approval number 15-0586). Written informed consent was obtained from all participants.

Experimental procedure

Participants walked at 1 m/s for 30 s on a dual-belt, force-measuring treadmill (Bertec Corp, Columbus, OH). Muscle activity was recorded with bipolar wireless (Noraxon, USA) electrodes from the following muscles, bilaterally: tibialis anterior (TA), medial gastrocnemius
(MG), vastus lateralis (VL), biceps femoris (BF), gluteus maximus (GM), erector spinae (ES), and latissimus dorsi (LD). Electrodes were attached to the skin using SENIAM guidelines (Hermens et al. 2000). Force and muscle activation signals were recorded synchronously at 1 kHz. The positions of 32 reflective markers were recorded at 100 Hz using a 3D 10-camera system (VICON, Centennial, CO). All data were collected in Vicon (Centennial, CO) and exported to MATLAB for processing (Mathworks version 2015a, Natik, MA). Statistical analyses were performed using R (version 2.14.0).

**Data analysis**

EMG signals were recorded during 15 strides (left heel strike to left heel strike) for each subject. Left heel strike was defined as the index (i) with a local minimum for the position of the left heel marker, vertical ground reaction force equal to 0 N, and the vertical ground reaction force at the index (i + 10) being at least 5 N greater than the force at index i. The EMG activity from each muscle was then resampled to 0.1% of stride duration to obtain 1000 samples for each stride. EMG signals were band-pass filtered (10-490 Hz) and rectified. A linear envelope was created by low-pass filtering (Butterworth, 4th order, cutoff frequency 10 Hz) and normalizing the signal to the peak value during the stride for each subject (Clark et al. 2010). Data were averaged across the 15 strides for each muscle to produce the original EMG (EMG_o) matrix, an \( m \times t \) matrix (16 muscles by 1000 samples) for each subject.

**Non-negative matrix factorization**

Figure 1 depicts the NMF approach used to decompose the EMG signals (left panel) into timing and weighting matrices for each muscle synergy (middle panel), and the subsequent estimation of the original EMG signals by cross multiplication of the two vectors (right panel). The filtered, rectified, and normalized EMG signals were entered into a 16 x 1000 matrix (16
muscles at 1000 time points) as shown in the left panel of Figure 1. An iterative NMF algorithm (Lee and Seung 1999, 2001) was then applied with a progressive increase in the number of factors until a criterion was met. The NMF algorithm decomposed the original $m \times t$ EMG matrix into an $m \times n$ matrix encoding the weights of each muscle on a factor (synergy) and an $n \times t$ matrix encoding the activation timing of that factor during the stride. The product of the timing and weighting matrices should approximate the original signals (Lee and Seung 1999, 2001; Ting and Macpherson 2005; Clark et al. 2010).

$$EMG_o \approx W \times T$$

An NMF may be performed with the number of factors ranging from 1 to the number of input signals (16 in our study). The number of factors was deemed sufficient when three criteria were met (Tresch and Jarc 2009): first, the variability accounted for (VAF) in the reconstructed EMG signals ($EMG_r$) exceeded 90% of the original EMG recordings ($EMG_o$); second, the plot of the VAF as a function of number of synergies (scree plot, bottom of middle panel in Figure 1) was concave down for the number of factors accepted; and third, subsequent synergies did not explain more than 5% of the variance in the original EMG signals.

$$VAF = 1 - \frac{(EMG_o - EMG_r)^2}{EMG_o^2}$$

The weights of each muscle in the $m \times n$ matrix were normalized to the maximal weight within that factor (synergy) so each muscle received a weight between 0 and 1. NMF requires this normalization procedure so that weights across different factors can be compared (Lee and Seung 1999, 2001; Clark et al. 2010; Lawrence et al. 2015).
Figure 1: Schematic diagram of the NMF procedure. A. The rectified and filtered EMG signals averaged across 15 strides were concatenated and a non-negative matrix factorization was performed for each subject with between 1-8 factors (muscle synergies). B. Each factor comprised a weighting vector and a timing vector for the 16 muscles (middle panel). The criterion for the number of synergies required for each subject was the value when $\geq 90\%$ of the variance in the original EMG signals was explained and no additional synergies accounted for $> 5\%$ of the unexplained variance. C. The cross product of the total weighting and timing matrices approximate the original 16 EMG signals (right panel).

Once the number of factors had been determined, they were organized into functional groups (Kristiansen et al. 2016). This was accomplished by aligning the timing matrix of each factor with a functionally relevant event in the gait cycle, such as heel strike and toe off. The approach used in our study was to identify the factors with a peak in their timing vectors closest to heel strike and toe off bilaterally for each individual, and then ordering the factors from left heel strike to right toe off. Subsequently, the synergies for each participant were compared by finding the peak correlation in the timing vectors.
Statistics

Data were tested for normality using the Shapiro-Wilk test. Because the number of factors was not normally distributed, the Mann-Whitney U test was used to compare the number of factors required to reconstruct the EMG\textsubscript{0} signals between groups with alpha = 0.05. Similarly, the muscle weights within each muscle synergy were non-normally distributed. Due to non-normality and our \textit{a priori} hypothesis of a difference in the muscle weights of gluteus maximus and latissimus dorsi, a Mann-Whitney U test was used to evaluate differences in muscle weights for gluteus maximus and contralateral latissimus dorsi during the synergies for a loading response (early stance). Effect size was calculated as $Z/\sqrt{N}$. The time series of the muscle synergies were compared within each group using Pearson correlation coefficients for each participant. The mean Pearson r for each synergy and group were calculated (21 comparisons within each group and synergy). Differences between groups were examined with correlation coefficients for the all synergies with a Mann-Whitney U test and the effect size was calculated with Hedge’s g.
Figure 2: Rectified and filtered EMG signals recorded with bipolar surface electrodes from control subjects (A) and individuals with SI joint dysfunction (B). Black lines represent the average activation patterns across all subjects within each group with grey lines denoting the standard deviation. The EMG amplitudes for each muscle were normalized to its maximum amplitude during a stride.
RESULTS

Figure 2 shows ensemble averaged EMG data for the six participants in each group, with the values for each person representing the average for 15 strides. The grey bars indicate the standard deviation. The averaged EMG from each participant was decomposed using the NMF approach (Lee and Seung 1999, 2001).

Number of synergies

The number of muscle synergies required to reconstruct the original EMG signals with the stated criteria (Tresch and Jarc 2009) ranged from 3-5. There was no significant difference in the number of synergies required to reconstruct the EMG patterns between the control and SIJD groups (p = 0.64, r = 0.008). A scree plot was calculated for each participant and then averages were calculated for the two groups. Figure 3 depicts scree plots of the average variance accounted for by each synergy for SIJD individuals (Figure 3A) and controls (Figure 3B). The average number of synergies was 4.2 ± 0.4 for controls and 4.0 ± 0.7 for SIJD.
Figure 3: Scree plots for control subjects (A) and individuals with SI joint dysfunction (B) with the variance explained by the reconstructed EMG (from multiplying the W and T matrices) on the y-axis and the number of synergies on the x-axis.

Muscle activation timing

Figure 4 shows the average (dark lines) and individual (grey lines) timing data for each muscle synergy in control subjects and individuals with SIJD. The correlation coefficient (21 comparisons within each group) for the timing of the third muscle synergy was significantly less (p = 0.002, Hedge’s g = 1.7) for individuals with SIJD (r = 0.54 ± 0.09) than control subjects (r = 0.83 ± 0.05). These results indicate that the timing of this synergy was more variable for SIJD individuals.
Figure 4: Averaged timing vectors for SI joint dysfunction (A) and controls (B). Black lines represent the average and grey lines are individual subject data. The synergies are ordered chronologically based on the time of peak activation beginning with left heel-strike at time point 0.

Muscle synergies

Figure 5 displays the muscles that contributed to each synergy for the two groups of participants. Each muscle synergy comprises muscles that are activated at similar times. The muscles shown in the plots had confidence intervals that did not contain 0 after 1000 bootstraps of the NMF procedure (Sawers et al. 2017), which indicates these muscles were significantly
active for a given synergy. The muscles involved in the synergies were functionally related; for example, Synergies 2 and 4 for control subjects (Figure 4) were activated from late stance to toe off and included gastrocnemius and biceps femoris of the stance leg as well as the contralateral tibialis anterior. Synergies 1 and 3 comprised muscles involved with trunk and hip stabilization (erector spinae, latissimus dorsi) as well as hip extension (gluteus maximus). These synergies are consistent with the two synergies for propulsion and two for the loading response described by Olree and Vaughan (1995).

Table 1 shows mean ± SD muscle weights for the muscle synergies 1 and 3 (loading response) for the two groups of participants. The affected side is defined as the side diagnosed with SI joint dysfunction and includes the ipsilateral gluteus maximus. The muscle weights for gluteus maximus on the affected side (p = 0.014, effect size = 0.70) and contralateral latissimus dorsi (p = 0.009, effect size = 0.74) in the loading response were significantly less than those for control subjects. There was no difference between muscle weights (gluteus maximus, p = 0.82; latissimus dorsi, p = 0.76) for the loading on the unaffected side.

Table 1: Mean ± SD normalized muscle weights for the synergies associated with the loading response. The average values in the first two rows for the individuals with SI joint dysfunction were derived from the gluteus maximus (GM) on the affected side and the contralateral latissimus dorsi (LD). Data from the control subjects are reported as right or left only. †Indicates a significant difference in muscle weight (p < 0.05).

<table>
<thead>
<tr>
<th></th>
<th>Control</th>
<th>SI</th>
</tr>
</thead>
<tbody>
<tr>
<td>Affected (SI) or right (control) GM</td>
<td>0.85 ± 0.13</td>
<td>0.29 ± 0.10†</td>
</tr>
<tr>
<td>Contralateral (SI) or left (control) LD</td>
<td>0.75 ± 0.13</td>
<td>0.38 ± 0.08†</td>
</tr>
<tr>
<td>Unaffected (SI) or left (control) GM</td>
<td>0.83 ± 0.08</td>
<td>0.95 ± 0.01</td>
</tr>
<tr>
<td>Contralateral (SI) or (control) right LD</td>
<td>0.71 ± 0.06</td>
<td>0.71 ± 0.12</td>
</tr>
</tbody>
</table>
Figure 5: Mean ± SD normalized muscle weights for each muscle synergy for SI joint dysfunction and control subjects. Muscles are plotted if the confidence interval of their muscle weight did not contain 0 after Bootstrapping the NMF procedure 1000 times. LBF: left biceps femoris, LES = left erector spinae, LGM = left gluteus maximus, LIO = left internal oblique, LLD = left latissimus dorsi, LMG = left medial gastrocnemius, LRF = left rectus femoris, LTA = left tibialis anterior, RBF = right biceps femoris, RES = right erector spinae, RGM = right gluteus maximus, RIO = right internal oblique, RLD = right latissimus dorsi, RMG = right medial gastrocnemius, RRF = right rectus femoris, RTA = right tibialis anterior)
DISCUSSION

Individuals with SIJD exhibit significantly reduced muscle weights between gluteus maximus on the affected side with contralateral latissimus dorsi in a synergy during walking at 1 m/s (Figures 4 and 5). This muscle synergy was present on both sides (1\textsuperscript{st} and 3\textsuperscript{rd}) in control subjects during the loading response, similar to previous findings (Olree and Vaughan 1995). Moreover, the timing of this synergy was more variable in individuals with SIJD than controls, which likely indicates less stability in the timing of this synergy across subjects and steps. Nonetheless, there were no differences in the number of synergies needed to account for 90% of the original EMG data between SIJD and controls (Figure 3) using established criteria (Tresch and Jarc 2009).

The absence of a synergy involving gluteus maximus on the affected side and contralateral latissimus dorsi may compromise pelvic stability and contribute to SIJD by limiting force closure of the joint (Vleeming 2012, Kibsgård et al. 2017). Several studies support the muscular actions of force closure. For example, Barker et al. (2014) found that 70% of the muscular force generated by the gluteus maximus could act perpendicular to the plane of the SIJ, 14% of which is through its attachment into the thoracolumbar fascia. Carvalhais et al. (2013) determined that activation of the latissimus dorsi resulted in a shift of the contralateral hip. Moreover, van Wingerden and colleagues (2004) demonstrated that the gluteus maximus significantly contributes to muscular actions that cause force closure by compressing the SIJ and preventing excessive shear forces. Furthermore, Kibsgård et al. (2014) found that there is a 0.5 degree rotation between the ilium and sacrum during single leg stance in individuals with SIJD. The lack of synergist activity on the affected side of individuals with SIJD likely enables slight
displacements between the two joint surfaces and contributes to the pain associated with the disorder.

Pain in the lower back or SIJ alters muscle activation patterns during gait (Arendt-Nielsen et al. 1996; Hanada et al. 2011; van Dieën et al. 2017; Falla and Hodges 2017). Consistent with the result of our study, Hanada and colleagues (2011) found significant differences in trunk EMG activity during walking in individuals with chronic low back pain and suggested the muscles were activated in different subsets, which indicate different muscle synergies. Moreover, Arendt-Nelson and colleagues (1996) demonstrated that when injected with a hypertonic saline injection, individuals with chronic low back pain had significantly different EMG activity during walking when compared with healthy controls and suggested that noxious stimuli may alter reflex pathways causing different muscle activation patterns. Additionally, van den Hoorn and colleagues (2015) demonstrated differences in trunk—but not calf—muscle synergies during gait after a hypertonic saline injection. This finding led to the conclusion that muscle synergies involved in propulsion may be relatively stable in the presence of pain, whereas secondary synergies may be altered (Falla and Hodges 2017). Taken together with the results of our study and a recent review by van Dieën and colleagues (2017), it appears that pain can modulate EMG activity during gait.

Consistent with previous studies that have examined muscle synergies during walking, our study found three to five muscle synergies were needed to recreate the original EMG waveforms (Olree and Vaughan 1995; Cappellini 2006; Clark et al. 2010). Additionally, the muscle synergies were associated with key events the gait cycle, as determined by functional sorting (Kristiansen et al. 2016). Based on the approach developed by Olree and Vaughan (1995), the identified muscle synergies contributed to either propulsion (gastrocnemius, tibialis
anterior, biceps femoris) or the loading response (gluteus maximus, erector spinae, latissimus dorsi). These results suggest that the motor plan used for walking (Lee 1984; Cappellini 2006) may differ between individuals with SI joint dysfunction and healthy controls.

Our study has some limitations. Due to stringent study criteria, enrollment was difficult. However, this was necessary to ensure the results of the study represented changes associated with SIJD and were not due to other low-back disorders. Nonetheless, the effect sizes were large. Additionally, all of our study subjects were women, which is consistent with the prevalence of the disorder among women (>70% of individuals treated for SIJD) (Whang et al. 2015). In contrast, Hoffman and colleagues (2011) found that men with low back pain exhibit greater lumbopelvic motion than women, which suggests that our findings may not be generalizable to men. Furthermore, the SIJD cohort was overweight (BMI: SIJD, 27; Control, 22), which may have influenced the distribution of joint torques in the lower extremity and altered muscle activation patterns (Sibella et al. 2003). This study investigated the dynamics of muscular activation and the associated stability transmitted through the fascia. Additionally, there is a significant influence of ligaments on joint stability, which we could not quantify in vivo (Hammer et al. 2013). Therefore, we expect that our results explain some of the differences in muscular activation patterns during walking in individuals with SIJD, but there is likely a strong influence of mechanical structures as well (Hammer et al. 2013).

**CONCLUSION**

Although women with SIJD exhibit the same number of synergies as healthy women when walking at the same speed, the results of our study are consistent with the hypothesis that the synergy between gluteus maximus on the affected side and the contralateral latissimus dorsi is compromised in individuals with SIJD. This information provides clinicians with evidence
that women with SIJD may present with altered patterns of EMG activity and may suggest strategies for future rehabilitation approaches.
REFERENCES


