

THE EFFECTS OF PHYSICAL THERAPY ON INDIVIDUAL MUSCLE FORCE
PRODUCTION AND FUNCTION IN CHILDREN WITH CEREBRAL PALSY

by

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ABSTRACT

Cerebral Palsy (CP) is a chronic neurological disorder that affects the ability to perform basic motor tasks, posture, and muscle coordination. Current intervention techniques, including physical therapy, can produce vast improvements in some children, while other children do not respond. The underlying muscular improvements that are associated with physical therapy are unclear, and may be important to understanding why only some children benefit from therapeutic interventions. Musculoskeletal modeling and simulation can be used to determine changes in dynamic muscle force production and muscle contributions to body center of mass accelerations at the individual muscle level as a result of physical therapy. Changes in muscle force and function were characterized for children with CP who participated in both gait and strength training. Gait training significantly affected contributions to body support from the vasti and rectus femoris muscles, and contributions to body propulsion from the soleus, resulting in muscle functional roles more similar to able-bodied gait. Strength training significantly affected gastrocnemius contributions to support, again resulting in muscle function more consistent with able-bodied gait. Changes in muscle function as a result of gait training metrics were significantly correlated to the 6-minute walk test, which is a clinical assessment of muscle endurance. Changes in muscle force and function as a result of strength training were not consistent with measured isometric strength at the joint level, suggesting that dynamic muscle behavior during movement tasks cannot strictly be predicted by static measures. Musculoskeletal modeling and simulation are therefore needed to assess how muscles coordinate to improve mobility. Both types of therapy produced highly variable changes in muscle dynamic metrics across children, which corresponded to the highly variable neuromusculoskeletal impairments of children with CP. The use of musculoskeletal modeling and simulation can identify the muscular deficits of individual children, and have the potential to be instrumental in developing targeted physical therapy protocols that are tailored to a specific patient.

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CHAPTER 1

INTRODUCTION

Cerebral palsy (CP) is a chronic neurological disorder that affects the ability to control basic motor tasks, posture, and muscle coordination. The symptoms associated with an individual's disability can range from mild with only a slight reduction in the ability to perform daily tasks to severe motor impairment accompanied with cognitive disability, respiratory dysfunction, and sensation losses. CP may be caused by a significant trauma to the brain, genetic factors, or asphyxia resulting in a permanent brain lesion. This trauma usually occurs before, during or soon after birth. CP is one of the most common motor control disorders in children in the United States. Approximately 3 in every 1000 children are affected by CP with approximately 30% of these children having limited to no walking ability (Christensen et al., 2013a). According to the Center for Disease Control, the total lifetime costs for all children born with CP in 2000 is estimated to reach approximately 11.5 billion dollars (CDC, 2004).

CP is not a progressive disorder, meaning the primary symptoms of the disorder do not change over time. However, secondary effects can continue to worsen throughout a person's life. Some of these effects include joint degradation, reduced walking ability, muscle contractures, and bone deformities. Due to these secondary effects, the quality of life for someone with CP can diminish over time. Early interventions are needed to reduce the detrimental effects of the disorder and to improve walking ability.

Treatment options commonly used to treat the musculoskeletal symptoms associated with CP include surgical intervention, medication, and physical therapy. Surgical intervention is one of the most common treatment options, intended to improve range of motion and skeletal alignment by reducing the effects of contracted muscles or bone deformities. Medication is commonly used to temporarily paralyze a muscle or muscle group as a similar, but non-permanent method of reducing the effects of spastic muscles. Finally, physical therapy can be used as a stand-alone treatment, or in combination with surgical intervention (Chambers, 2001). Strength training, one form of physical therapy, is used to target muscle weakness (Steele et al., 2012a), one of the possible contributing factors of gait deficits in these children (Chambers, 2001; Gage, 1990). This training has been shown to improve the strength of many children

with CP; however, mobility impairments are not always consistent with strength improvements (Damiano et al., 2010; Dodd et al., 2002; Steele et al., 2012a). The results of this training are somewhat variable, with some children gaining effective range of motion at the knee during walking, while others lose significant knee range of motion (Steele et al., 2012a). Gait training, another form of physical therapy used for children with CP, is used to target muscle coordination to develop a more effective gait pattern. Gait training improves gait characteristics such as gait speed and stride length in children with CP (Begnoche and Pitetti, 2007; Damiano and DeJong, 2009; Dodd and Foley, 2007; Mutlu et al., 2009; Provost et al., 2007). These effects, however, are not always significant for individual children and more research is needed to show the effectiveness of this training for all patients (Damiano and DeJong, 2009). It is unknown why many patients do not respond, or respond negatively to treatments.

CP affects the neuromuscular system, which encompasses the complex interactions between muscle coordination, muscle mechanics and skeletal dynamics. Thus, it is important to evaluate walking mechanics at the individual muscle level. Understanding the function of individual muscles and how they contribute to altered walking kinematics before and after therapy is important for evaluating the efficacy of physical therapy in improving neuromuscular control and increasing muscle strength. This understanding is also important for developing improved physical therapy protocols targeted to individual children.

Thus, the goal of this project is to quantify the effects of physical therapy, including both gait and strength training on children with CP by characterizing muscle dynamic behavior. Musculoskeletal modeling and movement simulation will be used to determine how current generalized therapy protocols address, or do not address, underlying musculoskeletal deficits. This work will provide a basis for designing improved rehabilitation protocols for individual patients.

CHAPTER 2

REVIEW OF THE LITERATURE

A review of the literature is included to provide the reader with an understanding of current applications of musculoskeletal modeling and simulation. A review of cerebral palsy gait disorders and current treatment strategies provide background for the implementation of musculoskeletal models in this population.

2.1 Cerebral Palsy

Cerebral palsy is the most common developmental disability in the United States, and affects the independence and quality of life of many children and adults. Approximately 3 in every 1000 children are affected by CP with approximately 30% of these children having limited to no walking ability (Christensen et al., 2013a).

2.1.1 Common Gait Disorders in Cerebral Palsy

Cerebral palsy is characterized by a group of disorders that affect motion, muscle coordination, posture, cognitive function, sensation, and respiratory function. Most children with CP have some motor control difficulties that may result in a decreased ability or inability to walk or complete other basic tasks required for independence. Several possible contributing factors to gait disorders associated with CP include muscle spasticity, motor control impairment, muscle weakness, contractures, and skeletal deformities (Gage, 1990).

Common gait disorders associated with children with CP include true equinus, apparent equinus, and crouch gait (Chambers, 2001). True equinus is a gait disorder linked with difficulty in dorsiflexing the ankle, affecting 61% of children with CP who have altered gait (Wren et al., 2005). This type of gait can be associated with joint stiffness at the ankle or with excessive muscle spasticity in the plantarflexors. As a result of equinus, recurvatum gait can occur (Figure 2.A), in which the knee moves into hyperextension during stance, often caused by over active gastrocnemius activity during stance. Equinus can also present with increased knee flexion, known as jump gait (Figure 2.B) (Chambers, 2001).

Apparent equinus (Figure 2.C) is often characterized by increased dynamic or static knee flexion without difficulty dorsiflexing the ankle. Three dimensional gait analysis can be used to assist in the diagnosis of apparent equinus, by quantifying changes in complex gait motion at each lower limb joint. This allows observers to diagnose if the altered gait pattern is primarily associated with the knee or ankle. In the apparent equinus group of disorders, the gait deficiency is at the knee rather than the ankle. Thus, mistreating the altered gait by affecting the ankle musculature can decrease gait performance. Stiff-knee gait, a form of apparent equinus, is characterized by a reduced range of motion at the knee, specifically a lack of knee flexion during swing phase of gait (Chambers, 2001), affecting 80% of children with CP who have altered gait (Wren et al., 2005). This disorder is thought to be caused by improper firing of the rectus femoris in pre-swing (Knuppe et al., 2013).

Finally, crouch gait (Figure 2.D) is a gait disorder associated with increased hip flexion, knee flexion, and ankle dorsiflexion (Chambers, 2001), affecting 69% of children with CP who have altered gait (Wren et al., 2005). Crouch gait can be a progressive gait disorder with secondary effects adding to the musculoskeletal inefficiencies (Chambers, 2001; Gage, 1990). Crouch gait in some children is developed after improper lengthening of the Achilles tendon or preforming corrective surgeries at a very young age (Chambers, 2001).

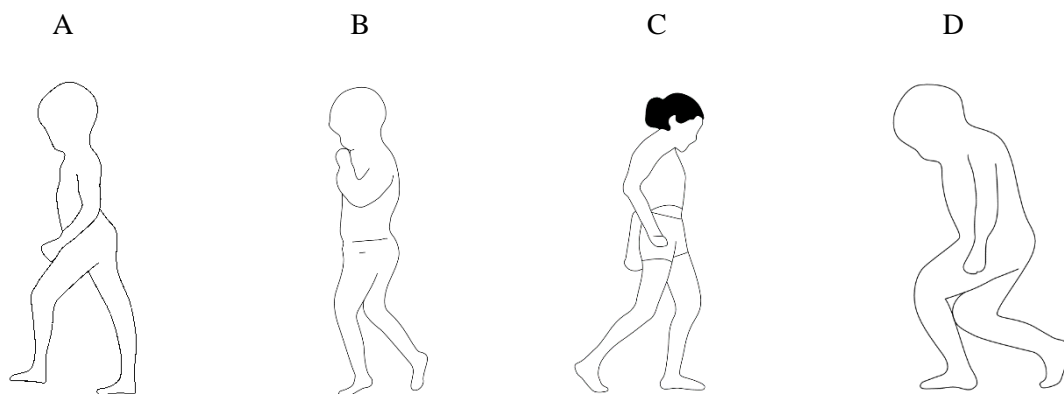


Figure 2.1 Examples of altered gait associated with CP. Each gait disorder shown is recurvatum gait (A), jump gait (B), apparent equinus gait (C) and crouch gait (D) (Chambers, 2001)

2.1.2 Clinical Diagnoses

Swift clinical diagnosis of gait disorders and effective treatment options are necessary to improve a child's overall quality of life, due to several secondary conditions associated with CP that progress throughout a child's development. For example, without intervention, bone deformities increase in severity with age (Wren et al., 2005). Several clinical tests are commonly used to diagnose spasticity, strength, independence, and overall movement, to better quantify the child's functional ability and to assist in treatment planning.

Whole body clinical tests and classifications are commonly used to develop a comprehensive assessment of each child's disability. For example, the 10-meter walk test is commonly used to evaluate the normal walking speed for a child. The 6-minute walk test is used to estimate the endurance of a child either with assisted or unassisted walking. Additional dynamic clinical assessments are used to assess the overall motor function of each child. The Gross Motor Function Classification System (GMFCS) ranges from levels I through V, and is used to differentiate the gross motor function of a child with CP based largely on the assistive devices that a child needs on a daily basis. Children classified as Level I or II are able to walk independently (for at least short distances in level II). The Gross Motor Function Measure (GMFM) is a similar assessment test for motor function; however, the ability of a child is quantified with more detail in different areas of daily activity including lying and rolling; sitting; crawling and kneeling; standing; and walking, running, and jumping.

In addition to the whole body dynamic tests, several joint level tests can be used to measure joint range of motion, spasticity, isometric joint strength, and selectivity, the ability to move an individual joint independently from other joints (Desloovere et al., 2006). Range of motion can be measured passively or actively for lower and upper limb joints. Spasticity is described as the presence of a stretch response during passive joint motion at a range of angular velocities, typically measured with the Ashworth scale, and can be verified with active muscle activity recorded with electromyography (EMG) during the stretch (Damiano et al., 2002b).

By understanding functional limitations, movement abilities, and joint level alterations, detailed goal setting and treatment planning can be completed. Improvements in these metrics indicate improved function and independence, an important goal when evaluating treatment options for each child.

2.1.3 Clinical Treatment

The treatment of different musculoskeletal impairments associated with cerebral palsy can have wide ranging results. Often the treatment of gait disorders for children with CP is unique to each child (Chambers, 2001). Treatments may include surgical intervention to increase the passive range of motion of a joint as well as skeletal deformities. In addition, forms of physical therapy, including strength and gait training show promising results in improving gait patterns and strength in children with CP.

2.1.3.1 Surgical Intervention

Surgical interventions are often used to target muscular deficiencies as well as bone deformities in children with CP. A few of the common muscular interventions include: muscle tendon lengthening surgeries used to increase the length of contracted or spastic muscles (Dreher et al., 2013, 2012; Granata et al., 2000); biarticular muscle transfer surgeries, typically performed on the rectus femoris to reduce unwanted joint moments at the knee without reducing the muscle's action at the hip (Fox et al., 2009; Knuppe et al., 2013); and gastrocnemius recessions, used to reduce the effect of contracted ankle plantarflexors (Chambers, 2001). Several common bone deformity interventions include hip varus derotational osteotomy, in which the head of the femur is adjusted to produce a better fit in the acetabulum, supra-malleolar osteotomy, which is usually used to correct tibial external rotation deformity (Hughes et al., 2012), and calcaneal, cuboid and cuneiform osteotomies (Chambers, 2001). The results of these surgeries are often positive with improvements in range of motion, (Dreher et al., 2012; Granata et al., 2000), gait kinematics (Dreher et al., 2012; Fox et al., 2009; Granata et al., 2000; Knuppe et al., 2013; Thomason et al., 2013), improved stride lengths and walking speed (Granata et al., 2000), and improved GMFM scores (Thomason et al., 2013). In addition, single event multilevel surgeries (SEMLS) that

include multiple of the surgical treatments discussed above are common in the treatment of children with spastic diplegic CP (Thomason et al., 2013).

Many of these surgeries, however, may be accompanied by negative side effects including a decrease in the efficacy of the treatment in the years following initial intervention (Dreher et al., 2012), and can create undesired results on gait if the surgery is performed without addressing other deficiencies (Figure 2.) (Chambers, 2001).

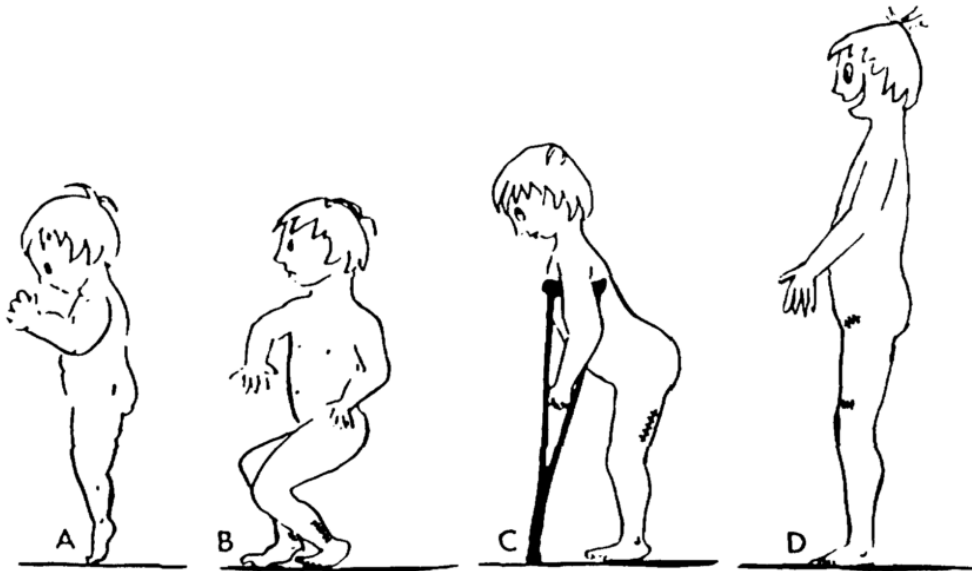


Figure 2.2 A sequence of surgical interventions represented for a child with spastic CP. First a child with equinus gait (A) has Achilles tendon lengthening surgery. This surgery causes a crouch gait (B) due to untreated hamstring and hip flexor spasticity. Surgery to lengthen the hamstrings results in forward collapse of the child's upper body (C) and finally with hip flexor lengthening and a rectus femoris transfer the patient can walk normally (D) (Gage, 1990).

Medication and therapeutic interventions are commonly used in combination with surgery, or as independent treatment options. Medications commonly used to paralyze a spastic or mis-firing muscle include baclofen, botulinum toxin type A (Hughes et al., 2012), phenol, and alcohol (Chambers, 2001). Therapy interventions are commonly used in combination with surgical interventions however therapy such as gait and strength training can also be used independently.

As many of these interventions can produce negative results with improper diagnosis, musculoskeletal modeling and simulation can be used to improve the understanding of individual muscle mechanics and assess the effects of treatments on muscle function.

2.1.3.2 Strength Training

Improvements in walking speed (Dodd et al., 2002) and muscle strength (Steele et al., 2012a) have been observed after children with CP have participated in strength training. One study investigated the correlation between strength training, knee extensor strength, and minimum knee flexion angle during gait. In this study, knee extensor strength was increased with strength training by an average of 23% with a maximum increase of 215%. However, changes in minimum knee flexion angle measured during gait were limited with an average change across all subjects of 0°, indicating no change after training. The maximum changes after training in minimum knee flexion angle during gait was 9° towards full knee extension, and 15° toward increased flexion (Steele et al., 2012a). These results indicate that although strength training can improve muscular strength, a direct correlation to improved knee angle kinematics does not exist. Other studies reported no significant changes in walking speed or cadence; however significant changes in knee extension and hip extension angle and changes in peak isokinetic joint torques were observed after strength training (Damiano et al., 2010). Results from strength training programs are often substantially variable with a wide range of changes after training (Damiano et al., 2010). As each child's deficits are unique, generalized strength training programs may not be as effective as programs targeted to specific muscle groups based on the needs of the individual child. One previous study has evaluated changes in dynamic muscle force production as a result of strength training using a musculoskeletal modeling approach. Changes for a single child, who showed substantial improvement after strength training, were found in the gluteus maximus during first double support, with an overall increase in knee and hip extensor moments. In addition, this study found increased potential for the gluteus maximus and vasti muscles to extend the hip and knee (Damiano et al., 2010).

Several concerns related to strength training have limited the use of this therapy. However, negative results, including increased spasticity, reduced range of motion, and reduced activity related to strength

training, have not been supported (Dodd et al., 2002; Steele et al., 2012a). Although there has been no indication that strength training worsens a child's muscle spasticity, there was a correlation between children with significant spasticity and negative results in minimum knee flexion angles during gait resulting from the strength training program (Steele et al., 2012a). These results indicate that although strength training may not cause increased spasticity, those children with spasticity may not be good candidates for this therapy. Spasticity, rather than muscle weakness, may be a major contributing factor for their altered gait patterns.

Based on these previous results, strength training can increase strength and may improve motor activity without adverse effects (Dodd et al., 2002). Strength training programs that focus on specific areas of muscle weakness for each child may be more effective than those that target general lower-extremity strength (Steele et al., 2012a).

2.1.3.3 Gait Training

Gait training or partial body weight support treadmill training (PBWSTT) is a type of physical therapy focused on improving walking motion through repetitive walking tasks. This therapy helps the child walk with normal kinematics without requiring the full strength and coordination to support their body weight during locomotion. The ability to conduct the training on a treadmill with body weight support makes this repetitive training method feasible for a broad range of children with different gait abnormalities and ambulatory levels (Damiano and DeJong, 2009).

Results from gait training studies have shown promising results related to overall activity levels and walking kinematics. Changes in gross motor function measures (GMFM-66 or 88) have shown improvements in dimension E, the category evaluating walking and running ability after PBWSTT (Begnoche and Pitetti, 2007; Kurz et al., 2011b; Provost et al., 2007). There have also been changes reported in functional balance post PBWSTT therapy (Provost et al., 2007) and compared to overground training (Grecco et al., 2013).

Improvements have also been seen in walking kinematics and gait characteristics after gait training. Significant changes have been seen in stride length (Begnoche and Pitetti, 2007; Mutlu et al., 2009) as

well as a reduction in stride length differential between the right and left leg indicating improved gait symmetry (Begnoche and Pitetti, 2007). In addition, decreases in stepping harmonics were seen after training indicating more consistent stepping pattern and an increase in the frequency of peak power also indicating more control of the stepping kinematics after training (Kurz et al., 2011b). Increased walking speed was also noted after training (Damiano and DeJong, 2009; Dodd and Foley, 2007; Kurz et al., 2011b; Mutlu et al., 2009; Provost et al., 2007). Ankle and hip moments also were reported to appear closer to normal gait moments (Mutlu et al., 2009), as well as improved range of motion at the ankle with sloped walking treadmill training (Willerslev-Olsen et al., 2014). Finally, increases in walking endurance have been reported (Dodd and Foley, 2007; Provost et al., 2007). Despite evidence that gait training provides improvements for children with CP, many results are not significant, as some children do not respond to treatment (Damiano and DeJong, 2009). Importantly, no adverse effects have been seen from this treatment (Damiano and DeJong, 2009; Dodd and Foley, 2007; Mutlu et al., 2009).

Although these changes have been seen at the whole body level, the alterations to individual muscle coordination and strength are still not well understood. By identifying the changes at a muscular level as a result of gait and strength training, a targeted treatment protocol may be established to identify specific abnormal muscle recruitment strategies or areas of muscle weakness, thereby improving the effectiveness of therapy.

2.2 Musculoskeletal Modeling and Simulation

The use of musculoskeletal modeling and simulation has several advantages including its cost-effectiveness compared to experimental studies, the ability to access metrics not experimentally accessible, and asking “what if?” questions. Modeling and simulation can be used in design prototyping, theoretical studies, and the identification of cause and effect relationships. The application of dynamic rigid body modeling to assist in the understanding of the human musculoskeletal system has led to a number of important discoveries regarding movement control. It also allows for a safe environment to investigate injury mechanics, and it allows for evaluating the effectiveness of clinical interventions prior to the implementation of clinical treatment (Piazza, 2006).

Inverse dynamics is commonly used to estimate joint moments required to recreate experimentally collected motion data. Inverse dynamics uses a rigid body model with experimentally collected ground reaction forces and motion capture data to determine joint moments and forces based on the Newton-Euler equations of motion (Zajac et al., 2002). Advanced techniques are used to reduce the error produced by differentiating noisy experimental position data to determine segment accelerations as well as error created from modeling assumptions (Cahouët et al., 2002). Inverse dynamics has several advantages compared to other computational techniques. Only the acceleration of distal segments is required to determine joint acceleration, meaning that modeling errors in the upper body do not affect the results in the lower limbs (Zajac et al., 2002). Unfortunately, inverse dynamics methods are also highly sensitive to uncertainties in acceleration data (Cahouët et al., 2002; Zajac et al., 2002) and skin mounted marker movement that violates the rigid body modeling assumption, producing additional model error (Zajac et al., 2002).

Based on the joint moments determined from inverse dynamics, muscle forces are estimated using static optimization techniques. Because muscle forces cannot currently be measured experimentally in humans, musculoskeletal modeling and simulation are used to understand the dynamic muscle force production before and after therapy. As the skeletal system has more muscle actuators than degrees of freedom (i.e., muscle redundancy problem), optimization techniques are required to estimate muscle forces at each time step of the analysis. Although there are limitations to these analyses, such as low confidence in the optimization criterion (based on inconsistencies with human neural control schemes and underestimated co-activation within muscles), many studies have used static optimization to successfully predict motion parameters such as hip contact forces in prosthetic implant design (Zajac et al., 2002). A hybrid approach combining static optimization and proportional-derivative (PD) control, “Computed Muscle Control” (CMC), is used to determine muscle forces in musculoskeletal models accounting for muscle activation and deactivation dynamics (Thelen et al., 2003). CMC enables a forward dynamic simulation to be generated in a computationally efficient framework.

With computational tools used to estimate muscle forces, the ability to understand muscle coordination principles in healthy and pathological gait has greatly advanced. However, muscle forces alone do not describe how muscles drive the body to achieve a specific motion. To understand muscle function, first an understanding of dynamic coupling is required. In a musculoskeletal model, a muscle can produce instantaneous linear and angular accelerations at all of the segments and joints within the linkage, even the joints it does not cross (Zajac et al., 2002). An example of dynamic coupling is the function of the soleus muscle in mid-stance. The activation of this muscle during stance not only acts to accelerate the ankle into plantarflexion, but it also acts to accelerate the knee and hip into extension, even though the muscle only physically crosses the ankle joint (Zajac et al., 2002). Evaluating dynamic coupling mathematically, the relationship of a single muscle to the acceleration of every joint arises from the non-diagonal mass matrix, $I(q)$. The equation used to describe motion at each joint in the body can be seen in equation 2.1.

$$I(q)\ddot{q} = R(q)F^{mus} + G(q)g + V(q, \dot{q}) + F^{non}(q, \dot{q}) \quad (2.1)$$

Where: q, \dot{q}, \ddot{q} are position, velocity, and acceleration of each generalized coordinate, respectively

$I(q)$ is the system mass matrix

$R(q)$ is the matrix of muscle moment arms

F^{mus} is the vector of muscle forces

$G(q)g$ is a gravity vector

$V(q, \dot{q})$ is a Coriolis and centripetal vector

$F^{non}(q, \dot{q})$ is a vector of external force terms (Zajac et al., 2002)

In addition to the action of a muscle to accelerate joints throughout the body, bi-articular muscles can act to accelerate joints they cross in the opposite direction from their anatomical orientation. For example, the rectus femoris produces a hip flexion and knee extension moment when the muscle is activated. During walking however, the rectus femoris acts to accelerate the knee and hip into extension despite producing a flexion moment at the hip (Hernández et al., 2008; Zajac et al., 2002). Consideration of dynamic coupling is necessary to understanding the contribution of muscles to the altered gait of children with CP. Through simulation, the underlying muscular causes of altered whole-body kinematics can be determined, and used for guiding treatment decisions to improve walking ability.

The complex relationship of dynamic coupling in the human body can be examined using an induced acceleration analysis. Using the equations of motion (Eq. 2.1) the instantaneous acceleration caused by a muscle can be determined (Zajac et al., 2003). To determine the acceleration of the model at each joint due to one muscle, the instantaneous position of the model, muscle force, and component of the ground reaction force produced by the muscle are required. The decomposition of the ground reaction force into individual muscle contributions, especially at a single instant in time, is difficult (Zajac et al., 2003). Several different methods have been established to determine the component of the ground reaction force attributed to a single muscle, including a rigid contact method (Anderson and Pandy, 2003), an integration method (Neptune et al., 2001), a perturbation method (Liu et al., 2006), and a modified hard constraint model method (Hamner et al., 2010).

Although musculoskeletal modeling is a powerful tool used to evaluate muscle function and coordination during human motion, there are limitations. Kinetic quantities computed using musculoskeletal simulation are difficult or impossible to measure experimentally and therefore, validation of the results of a simulation is extremely difficult. Methods of validation currently are limited to kinematics, kinetics, and electromyographic experimental data that can be compared with model outputs to confirm experimental consistency (Piazza, 2006). The optimization criteria used to estimate muscle forces are not validated in the real human system, meaning that what the human body “optimizes” for during gait or any other motion is unknown, and therefore accurate optimization criteria are unknown. Finally, identification of muscle coordination in pathological gait is challenging as a full understanding has yet to be reached for normal gait. Therefore the extension to pathological gait only establishes more unknowns and modeling challenges (Zajac et al., 2003). Despite these challenges, the use of this technology has led to many important discoveries regarding human motion, and with continued validation, the application of musculoskeletal modeling can become even more impactful.

2.2.1 Modeling of Children with Cerebral Palsy

Previous studies have used modeling techniques to investigate pathological gait conditions associated with CP, although many of these studies have not included changes in musculature, bone deformities or

other musculoskeletal characteristics associated with common CP gait disorders (Fox et al., 2009; Steele et al., 2013, 2012b, 2011, 2010). In addition to modeling considerations and assumptions required for healthy, able-bodied adult populations, modeling considerations specific to children with CP are necessary to accurately represent this population.

One consideration is the effects of using generic adult musculoskeletal geometry scaled by subject height and mass to fit the geometry of a child. Previous studies have shown that mass and length scaling of muscle strength in generic musculoskeletal models is valid for modeling the strength of children (Correa and Pandy, 2011). More data are needed to understand effects of other scaling practices on the model of children, including inertial effects, mass distributions, muscle moment arms and others. In addition to most CP research being completed on children, most children with CP who have gait difficulties also have bone deformities and muscular changes (e.g., spasticity, contracture, weakness, surgical lengthening).

Tibial torsion is a common bone deformity in children with CP, which occurs as abnormal muscle forces and loading cause the bones to twist as they grow, developing an altered angle between the flexion axis at the knee and ankle. Excessive tibial torsion has been shown to shift the induced knee joint acceleration, and diminishes the contribution of the soleus to support and progression (Schwartz and Lakin, 2003). Hicks et al. found that not only the capacity of the soleus, but also the capacity of the posterior gluteus medius and gluteus maximus to extend the hip and knee were significantly reduced with excessive tibial torsion with angles of at least 30° due to changes in the skeletal platform on which the muscle acted (Hicks et al., 2007). In addition, external tibial torsion can disrupt the stability of the foot during mid to late stance (Schwartz and Lakin, 2003). These previous findings indicate that including a model of tibial torsion may be important in identifying the functional roles of several important muscles that contribute to support and propulsion of the body.

Muscle spasticity is another common characteristic associated with CP that is rarely modeled in musculoskeletal simulations. One model has recently been developed and tested against single joint motion dynamics with success in modeling spasticity. The model with spasticity produced lower peak

knee angles, slower fiber lengthening velocities and lower peak fiber length, consistent with spastic muscles in children with CP (Van der Krogt et al., 2013). Although this model has been proven to effectively model the spastic catch of a child with CP, the implementation of this spasticity model has not yet been included in a full body model or used to simulate walking simulations.

Finally, modeling and simulation are rarely used to assess assistive devices used by children with CP. Ankle foot orthoses (AFOs) are commonly prescribed to children with CP (Knutson and Clark, 1991) to improve ankle dynamics. The purpose of AFOs for each child can range from reducing the risk of falling for children who exhibit drop foot, to assisting body support during stance in children with crouch gait (Bregman et al., 2009). Although these AFOs have been shown to affect walking speed, cadence, ankle kinematics, and ankle power, the same studies also show the duration and timing of muscle activation in lower limb muscles are unchanged with the application of these devices (Lam et al., 2005; Radtka et al., 2005, 1997). One study modeled the effects of an AFO on walking using a non-linear logarithmic model based on mechanical testing of an individual AFO. Crabtree *et al.* found the contribution of the soleus to ankle acceleration during gait was reduced, while the contribution from the gastrocnemius was increased (Crabtree and Higginson, 2009). Many studies have evaluated the stiffness of these devices, reporting stiffness ranges from 0.2-2.667 Nm/deg with the evaluation of both polypropylene and carbon fiber orthoses with different design configurations (Cappa et al., 2005; Crabtree and Higginson, 2009; Sumiya et al., 1996). The impact of stiffness on the child's kinematics, however, has not been fully investigated. As many children with CP use some type of AFO to assist in walking or improve bony alignment, a comprehensive understanding of the effect of these devices on muscle function and muscle coordination is important. In addition, accurately incorporating these devices in musculoskeletal models allows expansion of modeling and simulation studies to many additional children with CP using with their own prescribed AFOs.

2.3 Lower-Limb Muscle Function

Muscles in the lower limbs are the primary contributors to support and forward propulsion of the body while maintaining balance during gait (Anderson and Pandy, 2003; Liu et al., 2006; Neptune et al.,

2004; Pandy et al., 2010). Modeling studies of CP gait, specifically crouch gait, show an increased average total muscle force required to support and propel the body (Hicks et al., 2008). Also, investigation of crouched walking pattern indicate an increased ability to produce transverse plane forces, which increases the resistance to balance perturbations (Hoang and Reinbolt, 2012).

At the beginning of stance, muscles that support the body concurrently brake the center of mass (COM), whereas in the second half of stance, the muscles support and propel the body COM forward (Liu et al., 2006). Several studies have used musculoskeletal modeling to evaluate the muscles that contribute most significantly to the support of the body during gait. In normal gait, support of the COM is achieved through precisely timed actions of a few key muscle groups (Figure 2.3) (Anderson and Pandy, 2003), whereas in crouch gait, most muscles are active throughout single limb stance, generating relatively constant COM accelerations (Steele et al., 2010). The major muscle groups responsible for coordination of able-bodied gait include the plantarflexors and dorsiflexors, hip and knee extensors, and hip abductors (Anderson and Pandy, 2003). These muscles also contribute to COM motion in crouch gait (Steele et al., 2010). In normal gait, the ankle dorsiflexors, hip abductors (Anderson and Pandy, 2003), uniaxial hip extensors, and uniaxial knee extensors (Anderson and Pandy, 2003; Neptune et al., 2004) support the body during early stance. During mid-stance when the body is supported by only one leg, skeletal support is largely provided by the gluteus medius and minimus (Anderson and Pandy, 2003; Liu et al., 2006). In late stance when the leg of interest is now the trailing limb the soleus and gastrocnemius provide significant support (Anderson and Pandy, 2003; Liu et al., 2006; Neptune et al., 2001).

In crouch gait, previous work shows an increased contribution to support during stance from the ankle plantarflexors and a decreased contribution in the gluteus medius compared to unimpaired gait (Steele et al., 2010). Increased strength requirements in the knee extensors, specifically the vasti group, are required to produce the crouched gait and increased with crouch severity (Steele et al., 2012b), potentially due to a loss of skeletal support contributions in mid-stance seen in normal gait. Interestingly, the soleus contributes substantially to support during stance (Steele et al., 2010), however the motion of crouch gait can be reproduced with greatly reduced ankle plantarflexor strength (Steele et al., 2012b). As previously

seen, crouch gait can result from prior surgeries (Wren et al., 2005), such as over-lengthening of the gastrocnemius-soleus complex, effectively reducing the strength of the muscle without addressing other muscular discrepancies (Chambers, 2001; Gage, 1990). Although surgery may reduce isometric muscle strength, there is insufficient evidence to correlate changes in dynamic muscle force production.

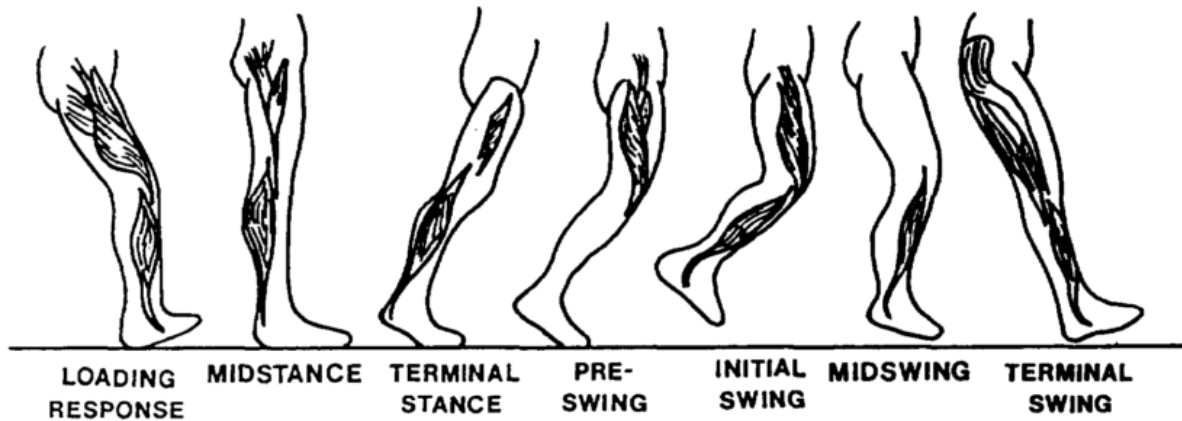


Figure 2.3 Illustration of muscles active during normal gait (Gage, 1990)

While supporting the body, muscles must also provide body propulsion to move the COM forward. Most significantly, the soleus and gastrocnemius provide support and maintain forward propulsion (Neptune et al., 2001). The soleus provides propulsion of the trunk, while the gastrocnemius generates energy to the leg for swing initiation during late stance (Neptune et al., 2001). In crouch gait, the propulsion contributions from the quadriceps and the plantarflexors applied opposing accelerations, with the quadriceps producing braking contributions and the plantarflexors providing propulsion. These contributions were matched in timing and magnitude, preventing a net forward propulsion acceleration (Steele et al., 2013).

The function of individual muscles to accelerate specific joints is also useful in understanding normal and pathological gait. In normal gait, the gluteus maximus and vasti muscles accelerate the hip and knee into extension in early stance while the gluteus medius and soleus extend these joints in mid stance (Arnold et al., 2005). Even without alterations to muscle force production, the overall posture of crouch gait reduces the ability of muscles to effectively produce hip and knee extension (Hicks et al., 2008). The

effectiveness of muscles known to produce extension moments in able-bodied gait are reduced by more than 50% in crouch gait postures (Hicks et al., 2008). The effect of tibial torsion also reduced the ability of the soleus and gluteal muscles to produce extension moments (Hicks et al., 2007). In addition, pelvic anterior tilt, common with hamstring musculotendon lengthening (Dreher et al., 2013), substantially reduces the extension capacity of muscles at the hip (Hicks et al., 2008). Because of the flexed posture, increased muscle strength is required to support the body (Hicks et al., 2008).

The hamstrings and rectus femoris, which are biarticular muscles that cross both the hip and knee, contribute little to support during normal gait (Anderson and Pandy, 2003; Arnold et al., 2005; Liu et al., 2006). Arnold *et al.* show that motion at the knee was not affected by the hamstrings during terminal swing although the muscle was active during this time in normal gait. This result implies that muscle tendon lengthening surgeries conducted on the hamstrings in children with CP do not directly affect terminal knee angles during swing for children with crouch gait. Instead, by weakening the biarticular hamstrings through surgical intervention, the uniarticular hip extensors must produce more force and thereby increase knee extension indirectly (Arnold et al., 2007). Modeling studies have found that there are no significant changes in hamstring strength required for crouch gait patterns (Steele et al., 2012b), and that the action of the hamstrings during crouch gait accelerated the hip into extension, with small changes at the knee (Hicks et al., 2008).

Knee kinematics during gait are important in understanding potential changes associated with altered gait patterns in CP. Both crouch and stiff knee gait have significant knee kinematic alterations. Many modeling studies investigated areas of muscle weakness and altered function for children with crouch gait. Musculoskeletal simulations of children with CP walking with a stiff-knee gait have been used to evaluate the effect of rectus femoris transfers, showing improved knee flexion moments in swing with the transferred muscle (Delp et al., 2007; Fox et al., 2009). In addition, studies conducted on able-bodied individuals have established that the most influential muscles in producing knee flexion velocity during pre-swing and swing include iliopsoas and gastrocnemius. The vasti, rectus femoris and soleus contributed most significantly to reduced knee flexion velocity in pre-swing (Goldberg et al., 2004a).

Currently, only the rectus femoris is treated in connection with stiff knee gait, although there are several other muscles significant in normal gait that could contribute to a change in knee flexion during swing for those with stiff knee gait.

Strength programs have targeted knee and hip extensors; however, weak quadriceps may not be the cause of crouch gait in all subjects (Steele et al., 2012a, 2012b). Instead, as hip abductors and ankle plantarflexors have shown to be weak in crouch gait (Steele et al., 2012b), and contribute substantially to support during normal gait (Anderson and Pandy, 2003), these muscles may be more effective targets for strength training, as well as the gluteus maximus to improve hip and knee extension (Arnold et al., 2005). Through the use of modeling and simulation, a detailed understanding of changes in muscle function associated with altered gait mechanics have been established (Fox et al., 2009; Hicks et al., 2008, 2007; Knappe et al., 2013; Steele et al., 2013, 2012b, 2011, 2010). However, modeling and simulation have not been used to evaluate the changes in muscular force and function associated with specific treatment interventions, including strength and gait training for a group of children. Performing such an evaluation may contribute to improved efficacy of treatment interventions.

2.4 Summary

Cerebral palsy is a disorder that affects many children and can be detrimental to mobility, independence, and quality of life. The gait deficits experienced for each child vary widely and can be dramatically altered with surgical intervention and over time. Physical therapy is a common intervention for these children, both in combination with or independent of surgical intervention. Different physical therapy methods, including gait and strength training, are proven to be safe and effective. However, the improvements in individual muscle function as a result of physical therapy are unclear, may be used to improve therapy protocols by identifying muscular deficits for individual children.

Whole body musculoskeletal modeling and simulation provides insight into muscle force and function, which are not accessible using experimental techniques alone. Modeling has previously been used to identify muscle function during able-bodied gait and to develop an understanding of possible areas of muscle weakness associated with crouch gait. However, no studies have used modeling and

simulation to understand the changes in the musculoskeletal system associated with current physical therapy treatments for children with cerebral palsy. By understanding the changes in muscle force and function associated with current therapy techniques, improved physical therapy protocols can be developed. The ultimate goal of this work is to ensure that the success of physical therapy is maximized for each individual child.

CHAPTER 3

EFFECTS OF GAIT TRAINING ON MUSCLE FORCE AND FUNCTION

Cerebral palsy (CP) is a perinatal brain disorder that often results in mobility impairments.

Approximately 1 in every 323 children in the United States are affected by CP, with only about 58% of those having full walking ability (Christensen et al., 2013b). Gait training is a form of physical therapy used for children with CP with the goal of improving muscle coordination to develop more effective gait patterns. The ability to conduct the training on a treadmill with body weight support makes this repetitive training method feasible for a broad range of children with different gait patterns and ambulatory levels (Damiano and DeJong, 2009). Results from gait training have included improvements in gait speed, stride length, stepping harmonics, normalized ankle and hip moments, increased ankle range of motion, and increased isometric joint strength (Begnoche and Pitetti, 2007; Damiano and DeJong, 2009; Dodd and Foley, 2007; Kurz et al., 2011a; Mutlu et al., 2009; Provost et al., 2007). These positive effects, however, are not always significant for each individual child (Damiano and DeJong, 2009) and little is known about how gait training targets the function of individual muscles to produce mobility improvements.

Altered gait patterns are often attributed to changes in the neuromuscular and musculoskeletal systems, which encompass the complex interactions between muscle coordination, muscle mechanics, and skeletal dynamics. Thus, it is important to evaluate walking mechanics at the individual muscle level. Understanding how individual muscles contribute to altered walking kinematics before and after therapy is important for evaluating the efficacy of physical therapy in improving neuromuscular control and increasing muscle strength. Musculoskeletal modeling and simulation provide a framework to quantify the function of individual muscles. In previous work, musculoskeletal models and movement simulations assess altered muscle control and mechanics associated with CP (eg. Hicks et al., 2008; Steele et al., 2013, 2010). In addition, musculoskeletal models have been used to investigate the potential effectiveness of different surgical interventions for children with CP (Fox et al., 2009). Further, investigation into strength requirements for altered kinematics identified weak hip abductors and ankle plantarflexors as potentially significant factors in the development of crouch gait (Steele et al., 2012b). However, modeling

and simulation approaches have not been used to evaluate the changes in muscular force and function to provide body propulsion, support and mediolateral balance that are associated with gait training.

Thus, the purpose of the study was to characterize changes of individual muscle forces and functional roles for multiple children with CP who have undergone gait training. Through a modeling and simulation approach, the muscles responsible for specific changes to a child's gait pattern can be identified. The methodology presented in this work can be applied in the future to develop improved physical therapy protocols that are targeted to individual children.

3.1 Methods

Seven children with spastic diplegic CP (13.4 ± 3.0 years, 152 ± 14.8 cm, 46.6 ± 13.4 kg) were selected from a group of children who underwent a six-week gait training protocol. All children were classified with Gross Motor Function Classification System (GMFCS) level I or II, indicating that the child could walk unassisted at least for short distances. Three of these children underwent a gait analysis and gait training protocol wearing their own prescribed ankle foot orthoses (AFO). Prior to and after the gait training protocol, a gait analysis session and clinical outcome assessments were completed.

3.1.1 Experimental Design

Gait training was completed over a period of six weeks with training sessions three times per week and each session lasting for 30 minutes. An overhead harness system provided support at the torso and waist. The level of support was reduced throughout the training protocol until it was 10% or less of body weight by the end of the training. The support was used to allow the child to focus on improving their stepping kinematics. The therapist provided subtle hand cues and verbal instruction to assist the child in improving their lower limb posture, ankle push off, and toe clearance during swing. A 6-minute walk test was conducted before and after gait training for each child.

3.1.2 Gait Analysis

Each gait analysis session consisted of collecting 3D kinematic data at 120 Hz using a six-camera motion capture system (Vicon Motion Systems, Ltd., Oxford, UK) and ground reaction forces (GRF) at

960 Hz using four force plates (AMTI, Inc., Watertown, MA). Sixteen lower body markers were used to track leg and pelvic kinematics. Kinematic markers were placed on the pelvis (left and right anterior superior iliac spine, left and right posterior superior iliac spine), thigh (lateral mid-thigh and lateral femoral epicondyle), shank (lateral mid shank and lateral ankle malleolus), and foot (heel and toe). Each child walked back and forth along an instrumented walkway. Three trials were included for simulation analysis for each child before and after the six-week gait training. The average walking speed for all walking trials in this study was 1.13 ± 0.15 m/s prior to training, and 1.18 ± 0.14 m/s after training, with no significant difference in walking speed between training conditions ($\alpha=0.05$).

3.1.3 Model Development

The kinematic and GRF data were used to develop musculoskeletal models for each child and simulate their movement before and after gait training in OpenSim 3.2 (Delp et al. 2007). Each model had 14 rigid body segments, 21 degrees of freedom, and 92 Hill-type musculotendon actuators with musculoskeletal geometry (Delp et al., 1990) and muscle force-length and force-velocity properties (Zajac, 1989). Passive torques were added to the major lower limb joints, including hip flexion/extension, hip adduction/abduction, knee flexion/extension, ankle dorsiflexion/plantarflexion, and subtalar inversion/eversion, to represent the effects of ligaments and soft tissues (Anderson, 1999). The generic model was scaled to match the child's height and weight. Model scaling for each child was based on anatomical measurements, such as knee width and ankle width, as well as 3D marker position data from a static standing calibration trial.

3.1.4 AFO Characterization

Three children in this study wore their own prescribed ankle foot orthoses during gait analysis sessions. An AFO model was developed in Opensim 3.2 by applying additional passive torques at the ankle and subtalar joints (Table 3.1). A passive actuator with damping ("CoordinateLimitForce" actuator in OpenSim 3.2) was applied to the ankle and subtalar joints to model an AFO for the three children who used these braces. The stiffness was modeled as a torsional spring with a single stiffness factor (k_{Θ}). The

stiffness in each direction of motion at the joint was altered independently (i.e., dorsiflexion stiffness was independent of plantarflexion stiffness). First-order torsional damping was also included in each joint, and was considered constant across the entire joint range of motion. Finally, neutral regions where the AFO did not apply any resistive torque were included over a small range of joint angles to account for the fit of the AFO. As the stiffness and damping parameters of each child's AFO were not directly measured, parameters were estimated for each type of AFO based on a comprehensive literature study of reported stiffness values (Table 3.1). The torque profile (Figure 3.1) developed for the AFO (Eq. 3.1) was constrained to not exceed the total joint torque determined using an inverse dynamics approach.

$$\begin{aligned} \tau_{AFO} &= k_1 * |\theta| - d * |\dot{\theta}| \\ &\text{when } \theta > \theta_{upper\ limit} \\ \tau_{AFO} &= -(k_2 * |\theta| - d * |\dot{\theta}|) \\ &\text{when } \theta < \theta_{lower\ limit} \end{aligned} \quad (3.1)$$

Where: τ_{AFO} is the passive joint torque
 θ is the joint angle (ankle and subtalar)
 $\dot{\theta}$ is the joint velocity
 k_1 is a torsional stiffness factor applied above the upper limit joint angle threshold
 k_2 is a torsional stiffness factor applied below the lower limit joint angle threshold
 d is the damping coefficient
 $\theta_{lower\ limit}$ and $\theta_{upper\ limit}$ are bounds on the neutral region

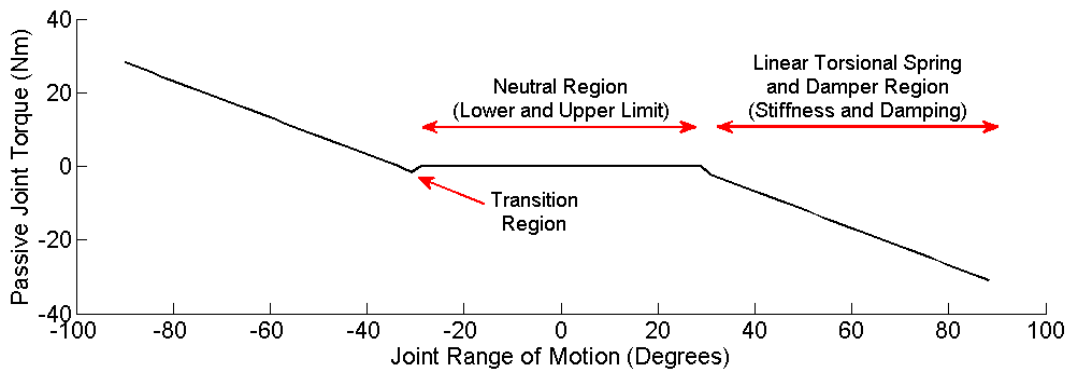


Figure 3.1 AFO passive joint torque profile for specified model parameters.

In addition, a probabilistic analysis was completed to determine the sensitivity of AFO estimated parameters on model results for this study. A 200-trial Monte Carlo simulation was completed to determine the effects of AFO model parameters on dynamic muscle force production for all major lower limb muscle groups. AFO model parameters that were investigated included the ankle and subtalar

stiffness coefficient approximations, ankle and subtalar damping coefficients, and the neutral angle. The neutral angle was a simplified method where the upper and lower limits were set to a single value, eliminating the neutral region. Similarly, a 50-trial Monte Carlo simulation, using the same design described above, was also used to evaluate the effects of AFO parameters on muscle contributions to the center of mass (COM) for all major lower limb muscle groups (CHAPTER 5).

Table 3.1 Parameters for AFO Models for three children with CP.

Child	AFO Type	AFO Model Parameters								References
		Ankle Stiffness (N/deg)		Subtalar Stiffness (N/deg)		Damping (N/deg/s)		Neutral Angle (deg)		
		DF	PF	IN	EV	Ank	Subtalar	Ank	Subtalar	
5	Hinged	0.05	2	0.75	0.75	0.01	0.01	-3	0	[5][3]
6	Posterior Leaf Spring	0.18	0.18	1.2	1.2	0.01	0.01	0	0	[1][2][3][4][5]
7	Solid	0.5	0.5	1.2	1.2	0.01	0.01	0	0	[1][3]

1. Bregman et al., 2009; 2. Crabtree and Higginson, 2009; 3. Kobayashi et al., 2011; 4. Sumiya et al., 1996; 5. Yamamoto et al., 1993

3.1.5 Gait Simulations

Three walking trials for each child were selected for analysis from the data collected before and after gait training, totaling 42 full gait cycle walking trials. For the walking trials, marker trajectories were filtered with a second order low-pass bidirectional Butterworth filter with a cutoff frequency of 6 Hz. The GRF data were filtered with the same filter at a low pass cut off of 20 Hz. A least-squares optimization algorithm (Lu and O'Connor, 1999) was used to determine the inverse kinematics solution that minimized the error between experimental marker positions and the rigid body model position, subject to kinematic constraints of the musculoskeletal model, in Visual3D (C-Motion Inc, Germantown, MD).

Simulations were generated in OpenSim 3.2 to reproduce measured kinematics for each child. First, a residual reduction algorithm (RRA) was used to improve the dynamic consistency between the ground reaction forces and the inverse kinematics solution by making slight adjustments to the walking kinematics, torso COM location, and total mass of the model (Delp et al., 2007). In addition, the trunk-pelvis angles were estimated to reduce residual forces and moments in this analysis. A computed muscle control (CMC) algorithm was used to solve for the muscle excitations that produced the experimentally-measured walking kinematics, thus generating muscle-driven simulations of walking (Thelen and

Anderson, 2006). Once the simulations were generated, an induced acceleration analysis (IAA) was performed to determine the function of individual muscles in accelerating the body COM and body segments. A hard kinematic constraint of rolling without sliding between the model foot and the ground was used to model the ground contact (Hamner et al., 2010). The contact point was modeled at the location of the center of pressure and the GRF was replaced with the kinematic constraint for the analysis at each time step.

3.1.6 Data Analysis

Mean dynamic muscle forces and muscle contributions to the three-dimensional COM accelerations were evaluated both before and after gait training. Individual muscles were categorized into groups for interpretation (Table 3.2) and were evaluated in four phases of the gait cycle (GC): Double Support 1 (~0-10% GC), Single Limb Stance (~10%-50% GC), Double Support 2 (~50%-60% GC), and Swing (~60%-100% GC). Accelerations were quantified in the three anatomical directions, (1) anterior/posterior, (2) superior/inferior, and (3) medial/lateral (Figure 3.2). Muscle contributions to COM accelerations in the three anatomical directions were characterized as contributions to (1) propulsion or braking, (2) positive or negative support, and (3) medial or lateral acceleration.

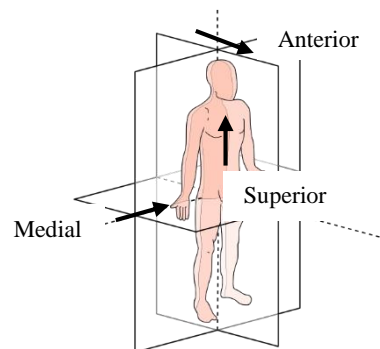


Figure 3.2 Anatomical acceleration directions, with positive directions indicated

A two-way, repeated-measures ANOVA was used to assess differences in muscle force and function (i.e., contributions to COM accelerations) between the left and right legs and as a result of training ($\alpha=0.05$). If significant leg or training main effects or a significant leg \times training interaction effect was found, a paired t-test was used to determine significant differences in specific muscle groups on each leg

as a result of training. Paired t-tests following a significant interaction effect tested each combination (e.g., (1) left leg: before/after training, (2) right leg: before/after training, etc.). The analysis of muscle contributions to the COM acceleration was limited to the muscle groups that provided substantial contributions (top 8 muscle contributions for each direction of acceleration) to the body COM accelerations. Significant main effects due to training indicated a change in muscle force production or function as a result of training, significant main effects due to leg differences indicated asymmetry in the specified muscle group, and interaction effects indicated that the child's legs were responding differently to training. Specifically, post-hoc results based on interaction effects observed in the ANOVA would indicate a change in symmetry by comparing t-test results for pre-training leg differences with post-training leg differences, or a training effect specific to a single leg by comparing t-test results for individual leg training differences.

Variability in dynamic muscle force production and muscle contributions to COM accelerations was also evaluated. Variation in stride-to-stride muscle force production or acceleration contribution was defined as the area between two curves indicating ± 1 standard deviation of the muscle force/acceleration contribution over the gait cycle for each child (Eq. 3.2). This integration technique allowed for the development of a single metric to characterize the overall variability of muscle force production or COM acceleration contribution throughout a full gait cycle for an individual.

$$I_{var} = \sqrt{\frac{1}{T} \int_{t_0}^{t_{end}} (F(t)_{upper\ lim} - F(t)_{lower\ lim})^2 dt} \quad (3.2)$$

Where: I_{var} is the variability index
 T is the total time for which the force is measured (0-100% GC)
 t_0 and t_{end} are the initial and final times of a gait cycle, respectively
 $F(t)_{upper\ lim}$ is the average muscle force plus 1 standard deviation for that normalized gait cycle time step for the three trials for each condition
 $F(t)_{lower\ lim}$ is the average muscle force minus 1 standard deviation for that normalized gait cycle time step for the three trials for each condition

A single representative variability index was defined as the sum of all individual muscle variability indices for each child before and after gait training. A paired student's t-test was used to test for statistically significant changes in the variability index after gait training ($\alpha=0.05$).

A Pearson correlation analysis was performed between the change in variability index for each child pre and post training and the change in total distance traveled on the 6-minute walk test pre and post training. Correlation strength was tested assuming a linear relationship between values.

Table 3.2 Muscle Group Characterization

Muscle Group	Muscle Group Abbr	Individual Muscles in Model
Ilio-psoas	IL	Iliacus Psoas
Adductors	AL	Adductor Longus Adductor Brevis Pectineus Quadratus Femoris
Adductor Magnus	AM	Adductor Magnus (Superior, Middle, and Inferior Comp)
Sartorius	SAR	Sartorius
Rectus Femoris	RF	Recuts Femoris
Vasti	VAS	Vastus Medialis Vastus Intermedius Vastus Lateralis
Anterior Gluteus Medius	GMEDA	Gluteus Medius Anterior and Middle Head Gluteus Minimus Anterior and Middle Head
Posterior Gluteus Medius	GMEDP	Gluteus Medius Posterior Head Gluteus Minimus Posterior Head Gemellus Piriformis
Tensor Fascia Lata	TFL	Tensor Fascia Lata
Gluteus Maximus	GMAX	Gluteus Maximus (Superior, Middle, and Inferior)
Hamstring	HAM	Semimembranosus Semitendinosus Gracilis Biceps Femoris Long Head
Biceps Femoris Short Head	BFSH	Biceps Femoris Short Head
Gastrocnemius	GAS	Gastrocnemius: Medial and Lateral Heads
Soleus	SOL	Soleus Tibialis Posterior Flexor Digitorum Longus
Tibialis Anterior	TA	Tibialis Anterior Extensor Digitorum Longus

3.2 Results

Musculoskeletal simulations developed for each walking trial were used to investigate changes in muscle force production and function for each child. Simulation quality was high for all simulations investigated ($<0.17\%$ BW average and $<15.6\%$ BW at the peak). Significant changes in muscle force production and function were found as a result of gait training (Table 3.3). A significant interaction effect in the GMEDA during swing phase was found, with an increase in force production on one leg and decrease in force production on the other leg, leading to significant asymmetry. Several other muscles were found to have overall significant asymmetry in dynamic muscle force production (Figure 3.3, Table 3.3). Overall changes found in dynamic force production were limited, and variability across children was high for all muscle groups.

Significant main and interaction effects were also found in the average muscle contributions to body COM accelerations (Table 3.3). Increased contribution to propulsion from SOL on only one leg during the second double support phase was found as a result of training. This increased contribution on a single leg led to increased asymmetry after training in SOL, however changes in muscle contribution after training and changes in symmetry did not reach statistical significance in a post-hoc analysis (Table 3.4). Increased contributions to support from VAS during the first double support phase were also found, with increases on one leg only. The increased contribution in the VAS resulted in increased symmetry after training. Decreased contributions to support from the RF during single limb stance were found only on one leg, resulting in decreased symmetry.

Asymmetries were also present in the braking contributions from the VAS and TA during the first double support. Several muscle contributions to the medial/lateral acceleration were also found to be asymmetric (Table 3.3, Figure 3.4).

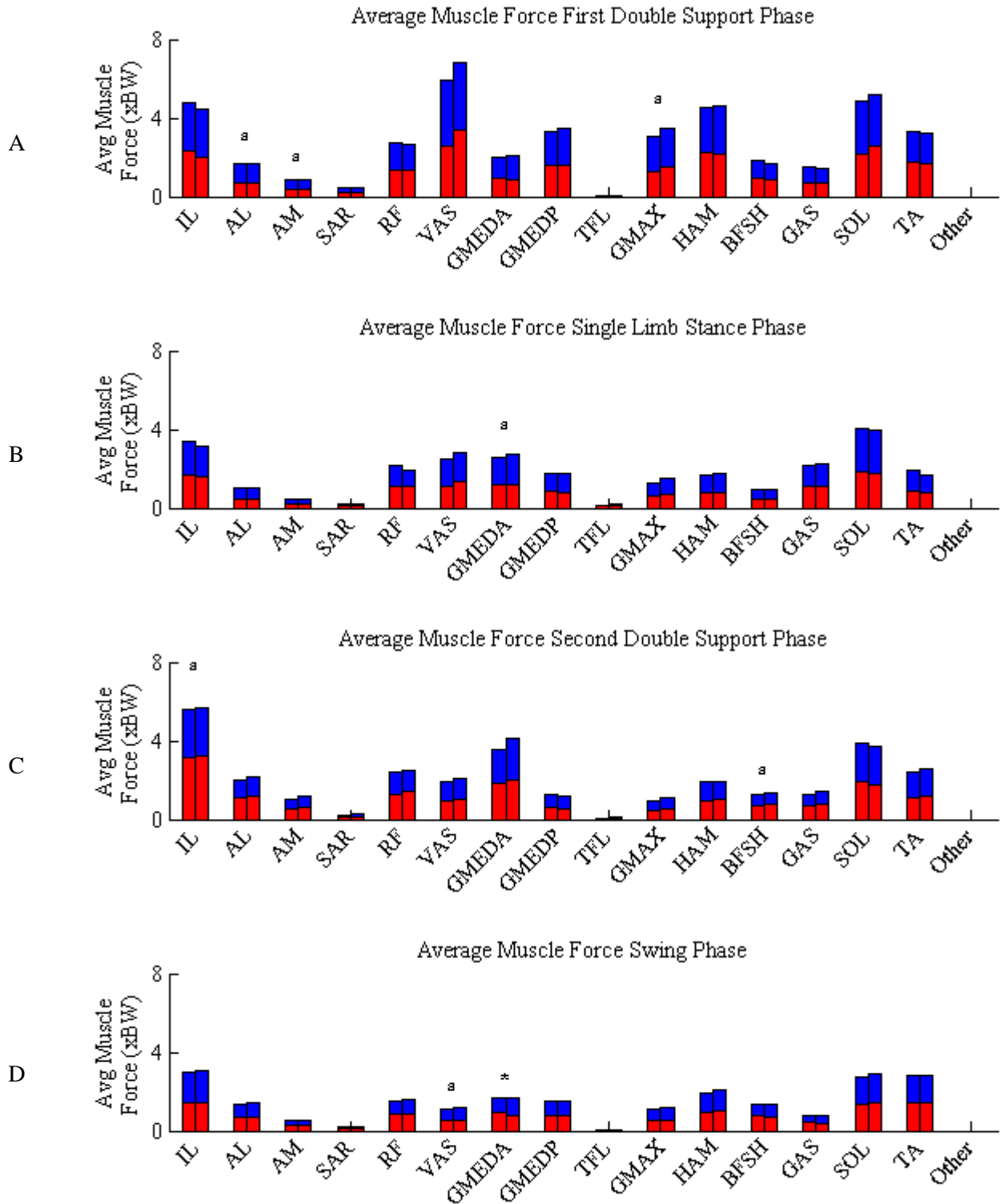


Figure 3.3 Average dynamic muscle force production during (a) first double support, (b) single limb stance, (c) second double support, and (d) swing phase of gait. Left leg (red) and right leg (blue) are shown before (left bar) and after (right bar) training for each muscle group. Significant training ($^{\circ}$), leg (a), and interaction (training \times leg) (*) effects from the ANOVA are shown.

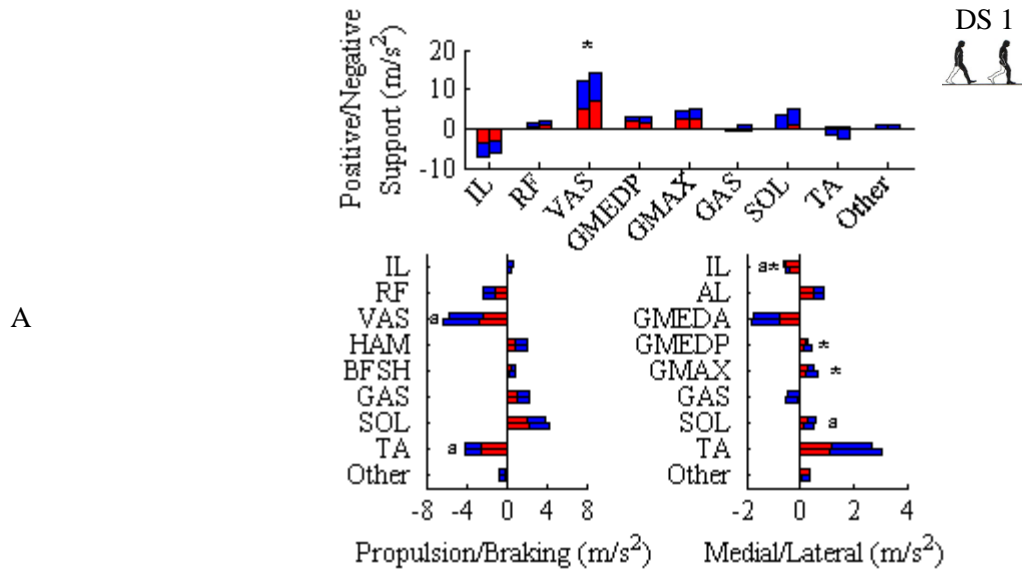


Figure 3.4a Average muscle contributions to COM acceleration during stance phases of gait including the (A) first double support phase, (B) single-limb stance, and (C) the second double support phase. Left leg (red) and right leg (blue) are compared before (left bar/top bar) and after (right bar/bottom bar) training for each muscle group. Only muscle groups contributing substantially to the overall acceleration were evaluated for significant changes (top 8 contributors were selected). Significant training ($^{\circ}$), leg ($^{\wedge}$), and interaction (training \times leg) ($*$) effects from the ANOVA are shown.

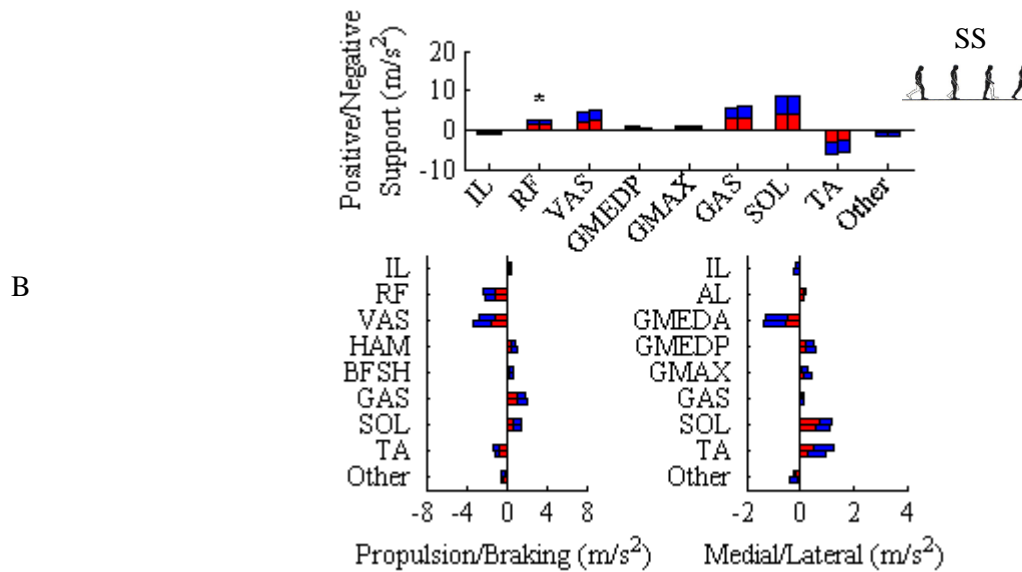


Figure 3.4b: continued

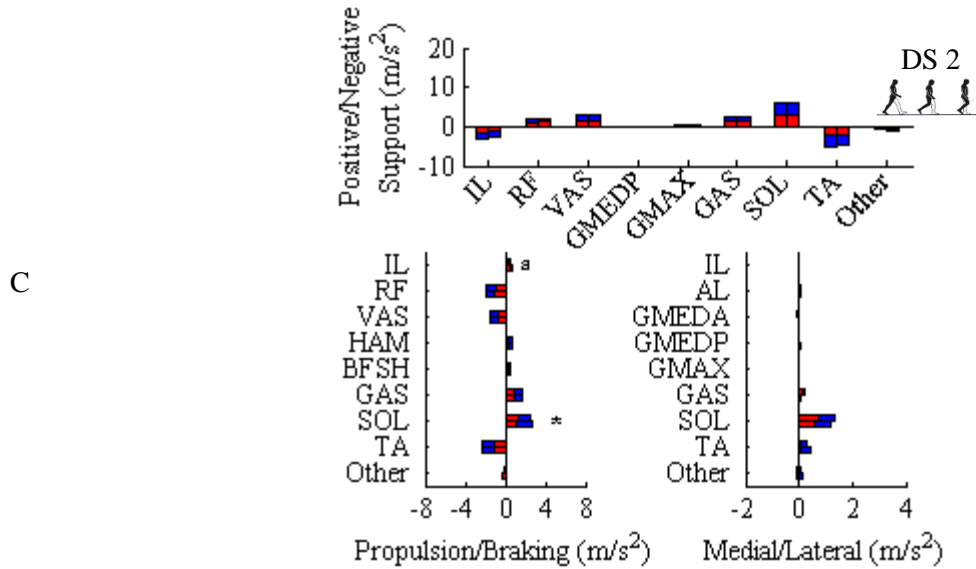


Figure 3.4c: continued

Table 3.3 Muscle force production and muscle contributions to COM acceleration ANOVA p-value results with significant main effects (leg, training) and interactions (training \times leg). All muscle groups for dynamic force production and the top 8 contributors to the COM acceleration were evaluated with the ANOVA. The phase where significant results were found is shown including the first double support phase (DS1), single-limb support (SS), second double support phase (DS2) and swing.

Metric	Muscle Group	Phase	Training Main Effect	Leg Main Effect	Interaction Effect
Average Muscle Force	IL	DS2	-	0.0001	-
	AL	DS 1	-	0.0039	-
	AM	DS 1	-	0.0087	-
	VAS	Swing	-	0.0360	-
	GMEDA	SS	-	0.4470	-
		Swing	-	-	0.0246
	GMAX	DS 1	-	0.0030	-
	BFSH	DS 2	-	0.0105	-
Average Muscle Contribution to the Acceleration of the COM (Ant/Post)	IL	DS 2	-	0.0074	-
	VAS	DS 1	-	0.0061	-
	SOL	DS 2	-	-	0.0481
	TA	DS 1	-	0.0153	-
Average Muscle Contribution to the Acceleration of the COM (Sup/Inf)	RF	SS	-	-	0.0473
	VAS	DS 1	-	-	0.0225
Average Muscle Contribution to the Acceleration of the COM (Med/Lat)	IL	DS 1	-	0.0190	0.0330
	GMEDP	DS 1	-	-	0.2760
	GMAX	DS 1	-	-	0.0167
	SOL	DS 1	-	0.0029	-

Table 3.4 Group mean (SD) of muscle force production and muscle contributions to COM acceleration that had significant ANOVA main (leg, training) or interaction (training × leg) effects during phases of the gait cycle. The phase where significant results were found is shown including the first double support phase (DS1), single-limb support (SS), second double support phase (DS2) and swing. The significant results from post hoc pairwise t-tests are also shown in bold.

Metric	Muscle Group	Phase	Mean and Standard Deviation				POST HOC Analysis					
			Muscle Force (%BW) or Function (m/s ²)				Training	Leg	Training x Leg			
			Pre Training		Post Training				Right leg	Left leg	Pre Training	Post Training
Left	Right	Left	Right	(Pre/Post)	(Left/Right)	(Pre/Post)	(Pre/Post)	(Left/Right)	(Left/Right)			
Average Muscle Force	IL	DS2	2.480 (1.161)	3.107 (0.966)	2.508 (1.066)	3.186 (1.049)	-	0.0001	-	-	-	-
	AL	DS 1	0.963 (0.488)	0.734 (0.354)	0.974 (0.481)	0.697 (0.310)	-	0.0027	-	-	-	-
	AM	DS 1	0.526 (0.292)	0.337 (0.179)	0.498 (0.225)	0.335 (0.173)	-	0.0063	-	-	-	-
	VAS	Swing	0.608 (0.368)	0.504 (0.283)	0.650 (0.350)	0.504 (0.260)	-	0.0297	-	-	-	-
	GMEDA	SS	1.430 (0.546)	1.171 (0.434)	1.581 (0.380)	1.155 (0.456)	-	0.0384	-	-	-	-
		Swing	0.784 (0.323)	0.900 (0.363)	0.870 (0.255)	0.766 (0.274)	-	-	0.1068	0.1076	0.1804	0.0354
	GMAX	DS 1	2.300 (1.005)	2.223 (0.855)	2.447 (0.878)	2.193 (0.903)	-	0.0022	-	-	-	-
	BFSH	DS 2	0.570 (0.283)	0.690 (0.257)	0.597 (0.255)	0.762 (0.274)	-	0.0081	-	-	-	-
Average Muscle Contribution of COM Accel. (Ant/Post)	IL	DS 2	0.211 (0.218)	0.369 (0.275)	0.196 (0.121)	0.434 (0.224)	-	0.0003	-	-	-	-
	VAS	DS 1	-0.185 (0.298)	-0.004 (0.203)	-0.253 (0.249)	0.008 (0.100)	-	0.0002	-	-	-	-
	SOL	DS 2	1.251 (0.512)	1.243 (0.506)	1.487 (0.698)	1.154 (0.588)	-	-	0.0516	0.7128	0.7738	0.3157
	TA	DS 1	-1.514 (1.066)	-2.514 (1.332)	-1.616 (0.781)	-2.465 (1.016)	-	0.0004	-	-	-	-
Average Muscle Contribution of COM Accel. (Sup/Inf)	RF	SS	1.307 (0.295)	1.255 (0.473)	1.035 (0.233)	1.364 (0.255)	-	-	0.5777	0.1056	0.7323	0.0317
	VAS	DS 1	7.079 (3.125)	5.030 (3.174)	7.334 (2.278)	7.049 (3.428)	-	-	0.0565	0.7379	0.0482	0.8185
Average Muscle Contribution of COM Accel. (Med/Lat)	IL	DS 1	0.046 (0.262)	-0.537 (0.334)	0.162 (0.176)	-0.375 (0.290)	-	0.0003	0.0095	0.3596	0.0255	0.0053
	GMEDP	DS 1	-0.090 (0.247)	0.226 (0.269)	-0.245 (0.180)	0.156 (0.251)	-	-	0.0711	0.0650	0.0749	0.0292
	GMAX	DS 1	-0.232 (0.333)	0.261 (0.211)	-0.437 (0.233)	0.216 (0.200)	-	-	0.1354	0.0661	0.0216	0.0007
	SOL	DS 1	0.294 (0.665)	0.561 (0.511)	0.383 (0.663)	0.549 (0.510)	-	0.2953	-	-	-	-

High variability in the results from this study was found in both group level metrics (Table 3.4) and for individual children. Group level evaluations of variability index found no significant changes after training. However, changes in stride-to-stride intra-subject variability, captured with the overall variability index, were found at an individual child level (Figure 3.5). Both Child 1 and Child 6 had positive changes in index value, indicating less variable muscle control of COM acceleration in all three directions. Child 7 had consistently negative changes in variability index, indicating more variable index values after training, except in the support direction in the first double support phase.

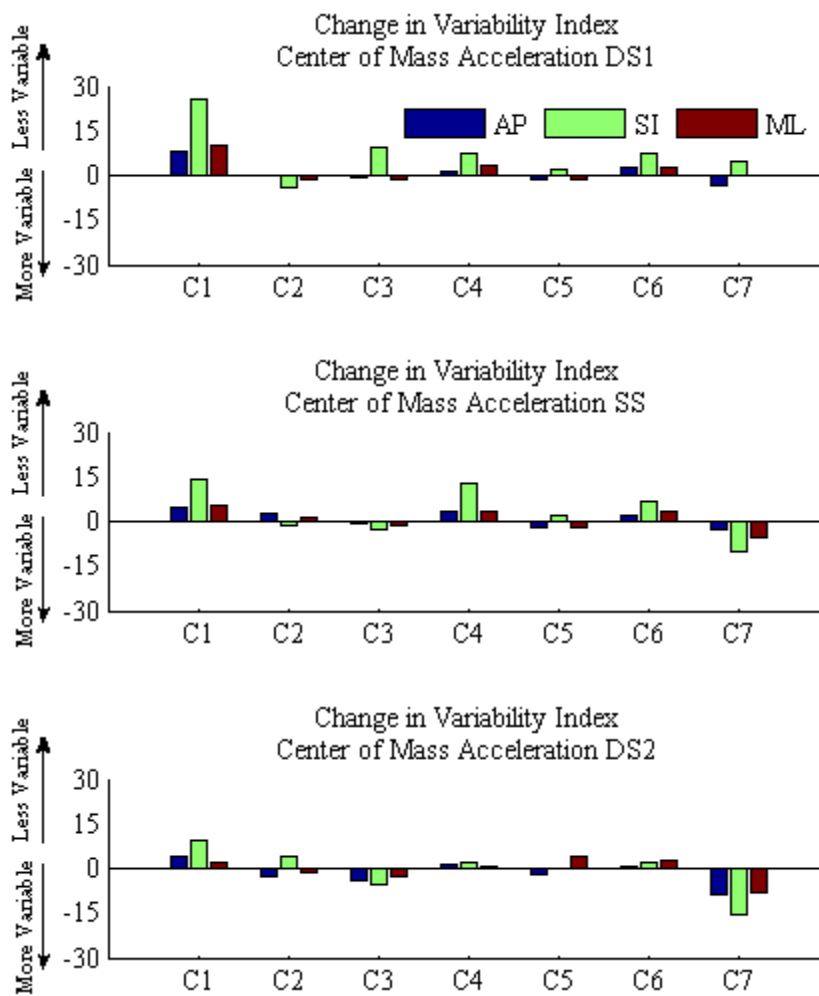


Figure 3.5 Total variability index of induced acceleration contributions including all muscle groups for each child during the three phases of stance including the first double support phase (DS1), single-limb stance (SS), and the second double support phase (DS2). The three anatomical acceleration directions anterior/posterior (AP, blue), superior/inferior (SI, green), and medial/lateral (ML, red) are indicated. Positive values indicate a reduction in variability, negative values indicate an increase in variability following gait training.

Changes in the total muscle intra-subject variability index were correlated to changes in the distance traveled in the 6-minute walk test (Table 3.5) before and after training. The variability index for muscle contributions to the COM acceleration in anterior/posterior direction during single limb stance correlated well with changes in the distance traveled after gait training (Figure 3.6). Correlations for the anterior/posterior acceleration contributions from the two double support phases also approached significance.

Table 3.5 Change in variability index for muscle contributions to COM acceleration correlated with changes in distance traveled for each child in the 6-minute walk test measured before and after gait training. Significant correlations are indicated with (*).

Phase of Gait	Acceleration Direction	R	p-value
Double Support 1	Anterior/Posterior	0.7197	0.0682
	Superior/Inferior	0.4388	0.3246
	Medial/Lateral	0.6708	0.0991
Single Limb Stance	Anterior/Posterior	0.8738	0.0101*
	Superior/Inferior	0.6191	0.1382
	Medial/Lateral	0.7525	0.0509
Double Support 2	Anterior/Posterior	0.6427	0.1195
	Superior/Inferior	0.7254	0.0650
	Medial/Lateral	0.2145	0.6442

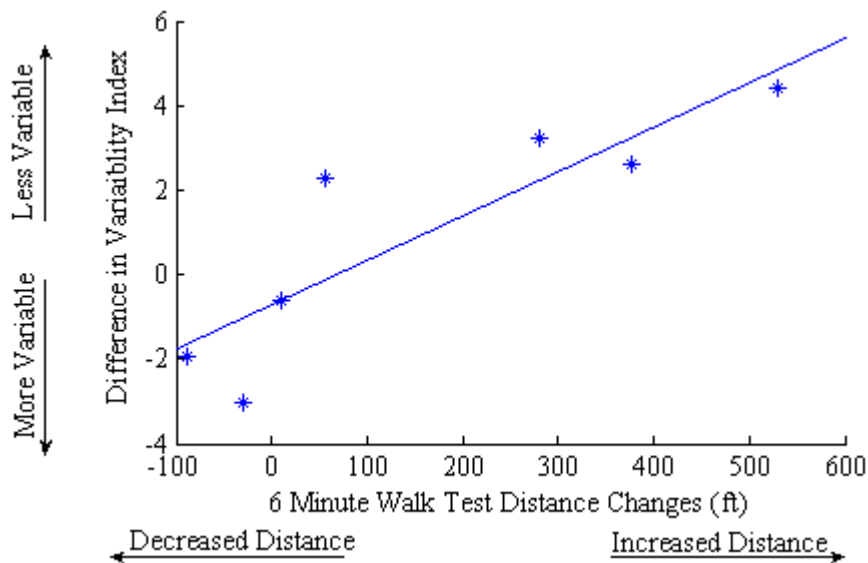


Figure 3.6 Total variability index changes after gait training for anterior/posterior acceleration contributions during single limb stance correlated to changes in distance traveled during the 6-minute walk test for all children. Positive variability index values indicated a reduction in variability, positive distance changes indicated increased distance traveled after gait training.

3.3 Discussion

The purpose of this study was to use modeling and simulation to characterize changes in individual muscle force production and function, quantified as contributions to COM acceleration, as a result of gait training. Through a modeling and simulation approach, the muscles responsible for specific changes to a child's gait pattern were identified.

Overall, musculoskeletal modeling identified important post-gait training changes in muscle function in the SOL, VAS, and RF muscles during stance. Changes in these muscle contributions were not consistent between legs, producing more or less symmetric accelerations after gait training. Different muscular responses between legs may suggest that one leg has a greater capacity to change as a result of gait training. In addition, we found that gait training produces a change in overall variability of muscle function; however, this change was not consistent for all children, and reduced variability in muscle contributions to A/P COM acceleration was correlated with increased performance in the 6-minute walk test.

Group level changes in the production of force potentially indicate specific muscle groups that are largely affected by gait training therapy. Changes in muscle force, however, were not always observed on both legs, increasing asymmetry in force production. Increased force production in the GMEDA during swing may be required to add power to the leg to assist the primary hip flexors (IL). Increased hip flexor power has also been attributed to increased walking speed in people with stroke (Nadeau et al., 1999) who have significant ankle plantarflexion weakness. The increased dynamic force production may be a mechanism to help increase walking speed in this population in the absence of typical ankle plantarflexor function. This ankle plantarflexion function may be altered as a result of muscle weakness, limitations to selective motor control, or AFO stops, reducing the ability for these children to produce plantarflexion torque when the ankle is in a neutral or plantarflexed position.

Asymmetric dynamic muscle force production was observed in several muscle groups, including GMAX, GMEDA, IL and BFSH. Asymmetry in the hip abductors and hip extensors during the first double support phase and single limb stance suggest that these muscles may be important targets for

increased therapy. As generalized gait training did not significantly affect these muscles across all children, strength training targeted specifically towards these muscle groups may be required to return symmetric muscle force production. The hip abductors play a significant role in supporting the body during the stance phase and may be a contributor to crouch gait when significant weakness exists (Steele et al., 2012b). Asymmetry in IL during the second double support may provide additional rationale for the asymmetric increases in force in GMEDA, which can act to flex the hip, during swing phase. Without adding power to the leg from IL during the second double support phase, force production in GMEDA may be required in swing phase for hip flexion. IL may also be a target for specific therapy to ensure power is being provided to the leg during the critical swing-initiation phase of gait in addition to GAS to increase walking speed (Goldberg et al., 2004b; Neptune et al., 2001). Targeted training such as functional electrical stimulation may be used during dynamic tasks to encourage specific timing of muscle activations. Observed asymmetries may also be a result of compensatory mechanisms related to reduced function in other muscle groups, and therefore, special care should be taken when evaluating the root cause of gait changes to establish if changes in individual muscle groups are primary or secondary contributors to altered gait. If a primary contributor can be established, targeted training may be employed to help improve walking ability. These targeted trainings may include targeted strength training to increase overall muscle strength, or functional electrical stimulation to develop a more normalized activation timing for affected muscles.

Group level changes in the muscle contributions to the COM acceleration indicate possible changes in muscular strategies that children are learning to provide propulsion, support, and mediolateral balance during walking. Increased contributions to propulsion in late stance from SOL suggest a walking pattern that is more similar to able-bodied walking (Neptune et al., 2001). Asymmetry in the soleus propulsion during the second double support phase suggests a target for additional therapy to maximize bilateral symmetry.

Increased contributions to body support from the VAS during the first double support phase may indicate an increased ability for the child to control the loading response. Also, increased VAS support,

without increased contributions in single limb stance and the second double support phase may indicate more normal muscle mechanics. Previous work has found increased support contributions from the VAS and ankle plantarflexors throughout stance with little modulation of the acceleration in children with crouch gait (Steele et al., 2013). Alternatively, increased contributions to VAS support only during the first double support phase may match more closely with able-bodied gait (Liu et al., 2006; Neptune et al., 2004). Increased symmetry in the VAS contributions to braking in early stance were not observed. Asymmetry of the braking contributions in the VAS and TA muscle group may indicate that gait training is targeting the control of supporting the body during leg-to-leg transitions but not the anterior/posterior accelerations.

Although group level changes have provided insight into possible targets for additional therapy, high variability in the muscle force production and muscle induced accelerations revealed limited training effects. Given the variable nature of altered gait patterns in children with CP (Wren et al., 2005), it is not unexpected that muscular force and function would be so variable across children as well.

These results indicate that assessment of muscle functional roles at the individual child level is needed, which would provide important information for developing individualized therapy protocols. Individualized analysis of each child revealed unique training effects in dynamic muscle force production and function. For example, Child 1 had substantial reductions in variability of muscle-induced accelerations after training; however, this child had little change in the magnitude of force production and function after training. Child 6 increased propulsion contributions from the SOL and GAS during stance while reducing the required VAS and RF contributions to support during mid-stance (Figure 3.7). These results for Child 6 were coupled with increased knee extension angle during walking trials. Before training, the VAS muscle group functional role was consistent with reported altered muscle patterns for children with crouch gait (Steele et al., 2013, 2010). After training, Child 6 established VAS and ankle plantarflexor functions more consistent with able-bodied gait (Liu et al., 2006; Neptune et al., 2004, 2001). Child 7 also produced increased propulsion from the plantarflexors during the second double

support phase after training (Figure 3.8) which may indicate increased strength and ability to accelerate the body forward (Neptune et al., 2001; Steele et al., 2012b).

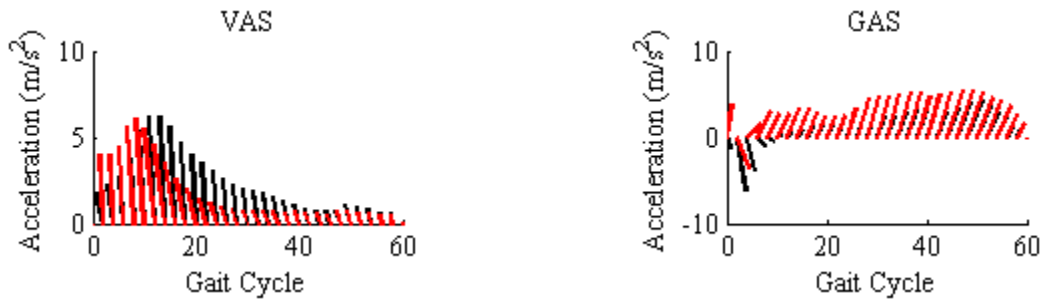


Figure 3.7 Child 6 GAS and VAS contributions to support and propulsion before (black) and after (red) gait training.

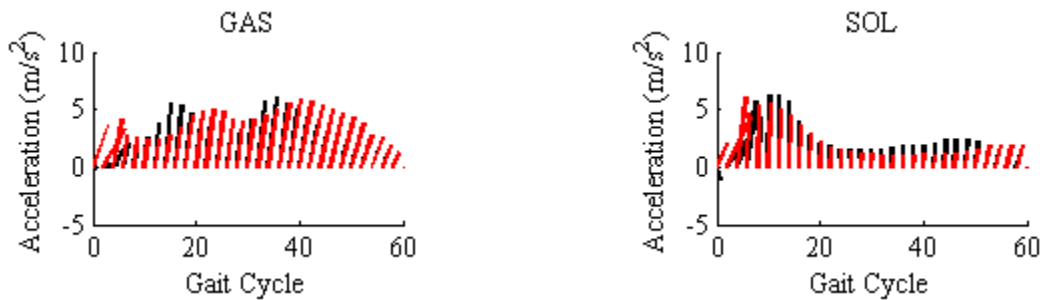


Figure 3.8 Child 7 plantarflexor contributions to support and propulsion before (black) and after (red) gait training.

Reductions in the variability of contributions to anterior/posterior acceleration correlated with increased walking endurance on the 6-minute walk test. Effectively modulating opposing propulsion/braking contributions from the VAS and ankle plantarflexors while maintaining support is important for effectively accelerating the body COM forward, and is often not achieved for children with crouch gait (Steele et al., 2013). Reductions in variability across gait cycles for each child specifically in the anterior/posterior transition from braking to propulsion in mid stance may be an important indicator for improved walking strategy and increasing endurance (Steele et al., 2013). Gait training also decreases stepping harmonics indicating a more consistent stepping pattern and increases the frequency of peak power also indicating more control of the stepping kinematics after training (Kurz et al., 2011b). Reductions in muscular contribution to COM acceleration variability may be associated with more consistent stepping patterns after training.

Musculoskeletal modeling and simulation approaches facilitate detailed analyses of dynamic muscle force production and function. This analysis provided possible targets for additional therapy including the hip extensors, hip abductors, and ankle plantarflexors, as well as possible factors for the effects of gait training on each child; however, there are limitations to the methods used in this study. A small sample size for the analysis with highly variable results may limit the ability to draw conclusions related to the effects of gait training for the group; however, as in many modeling studies, small sample size studies enable the development of detailed and patient specific results by evaluating simulation results at an individual level.

A generic musculoskeletal model, used for each child, was uniformly scaled in body size and mass based on a generic adult model. Additional factors including muscle weakness, bone deformities, spasticity, and previous surgical interventions were not accounted for in the model. Sensitivity studies evaluating the effects of using estimated body segment and muscle parameters found sensitivity in muscle force production magnitude and timing, most significantly, for changes in tendon slack length; however, the maximum effect on muscle force estimation was only approximately 100 N of force generation (Myers et al., 2014). An analysis of the significance of including tibial torsion was completed, and was in agreement with previous work (Hicks et al., 2007) finding that tibial torsion less than 30 degrees had small effects on model outputs, such as muscle force. All children in this analysis had limited tibial torsion (no more than 15 degrees from normal) and therefore, tibial torsion was not included in the model. Future work will further investigate incorporating muscle weakness and altered muscle mechanics on model outputs. The current musculoskeletal model has the ability to account for muscle weakness through the CMC optimization algorithm. Weak muscles for the patient will result in lower excitation values, however localized muscle weakness for certain muscle groups may require improved scaling techniques to fully capture the effects of weakness on the model.

Previous surgical interventions for the group of children in this study included muscle tendon lengthening surgeries for the soleus, hamstring, Achilles tendon, rectus femoris, and adductors; bony alignment surgeries for the tibia and femur, and rectus femoris transfer surgeries. Although this is a

limitation to the generic model used in this analysis, the aspect of comparing longitudinal results eliminates the effect of overall absolute error based on altered model properties. The effects of including children with rectus femoris transfers in the study had limited impact on the overall results of muscle force and function. Differences found in the RF contribution to support during single limb stance were still significant after eliminating the children with RF transfers from the statistical analysis; however, significant differences were only found between legs, instead of interaction effects. This may be a result of the smaller sample size or the altered function of the RF for the remaining children. Future work will include a model of the RF transfer surgery for those children included in this study who have undergone this procedure. Finally, a specific model for spasticity was not incorporated in the model, and thus spastic muscles in the children evaluated would result in increased estimations of excitations that produced muscle forces in the simulations.

The model results for children wearing AFOs were also evaluated carefully. A probabilistic analysis was completed to determine the effect of estimating AFO parameters on the resulting dynamic muscle force production and muscle function (CHAPTER 5). AFO parameters had little effect on muscles proximal to the ankle. The force estimates for muscles crossing the ankle were most sensitive to estimations of the neutral angle of the AFO and the dorsiflexion stiffness parameter. Although ankle muscle force production is sensitive to AFO parameter assignment, a rigorous literature review, in addition to implementing constraints based on the joint torque curve provides considerable reduction in the uncertainty level of the estimated parameters. As this was a longitudinal study looking only at changes between pre and post training, the absolute magnitude of the ankle muscles had little effect on the results of the study. By ensuring the AFO model was constant for both training condition simulations, we were able to eliminate the major effects of the AFO uncertainty.

Direct validation of estimated muscle excitation timing from the simulations was not possible in this analysis because electromyographic data collection was not included in the experimental protocol. In addition, torso kinematics were not collected during the experimental gait analysis. However, our simulations were of high quality in that residual forces and moments at the pelvis were small, indicating a

high level of dynamic consistency. In addition, joint reserve actuators were small and trunk-pelvis angles were estimated to further increase dynamic consistency in the simulations. Future work will incorporate electromyographic data collection and torso kinematics, to validate and improve our simulation generation methods. Even though muscle excitations could not be directly validated, our results indicated group level changes consistent with adaptations towards able-bodied gait in the VAS, RF and SOL despite high variability in the results.

The muscles that were affected the most by gait training varied across children, and at times the largest differences for an individual child were not apparent in the group statistical analysis. These results suggest that the neuromuscular impairments associated with individual muscles vary across children with CP, and gait training has the potential to affect many different muscle groups. The potential for each child to improve also varies greatly.

3.4 Conclusion

In this study, changes in muscle force and function were unique for each child. Overall changes as a result of gait training were found in the SOL propulsion and VAS support, indicating more normalized function towards able-bodied gait. Changes in the variability for each muscle's contribution to propulsion during single limb stance were correlated with increased distances on the 6-minute walk test. Specific muscles indicated for targeted training included the hip extensors, hip abductors, and ankle plantarflexors. Musculoskeletal modeling and simulation analysis can help greatly in identifying the muscular deficits for an individual child, and may thus be instrumental in developing targeted physical therapy protocols that are tailored to a specific patient.

CHAPTER 4

EFFECTS OF STRENGTH TRAINING ON MUSCLE FORCE AND FUNCTION

Strength training is a form of physical therapy commonly prescribed to children with cerebral palsy (CP). Muscle weakness is commonly observed in children with CP, and can be a contributing factor to altered gait patterns (Gage, 1990). Strength training is thought to reduce muscle weakness in these children, and therefore may provide improved walking ability. Improvements in walking speed (Dodd et al., 2002), strength (Steele et al., 2012a), and hip and knee gait kinematics (Damiano et al., 2010) have been seen after children with CP have participated in strength training. Individual improvements for each child, however, are highly variable with some children not responding or even reducing their ability to walk after training (Damiano et al., 2010; Steele et al., 2012a). Traditional strength programs have targeted knee extensor muscles; however, modeling studies have suggested that more appropriate targets may be the hip abductors and ankle plantarflexors (Steele et al., 2012a, 2012b). Little is known, however, about how isometric strength relates to dynamic muscle force production during gait.

CP is a neurological disorder that affects body movements and muscle coordination. Altered walking patterns are often attributed to changes in neuromuscular control and secondary factors, such as altered muscle mechanics and weakness. The connections between the neurological and muscular systems encompass complex interactions between muscle coordination, individual muscle mechanics, and skeletal dynamics. Due to the complexity of the musculoskeletal system, an evaluation of muscle mechanics at an individual muscle level is important. Effects of individual muscle force production on altered gait patterns before and after therapy are important in understanding the overall effects of therapy on children with CP. Musculoskeletal models and movement simulations have been previously used to assess altered muscle control and mechanics associated with CP (e.g., Hicks et al., 2008; Steele et al., 2013, 2010). In addition, musculoskeletal models have been used to investigate the potential effectiveness of different surgical interventions for children with CP (Fox et al., 2009) and a pilot study has evaluated the effects of strength training on a single child with CP (Damiano et al., 2010). However, modeling and simulation approaches have not been used to evaluate changes in the muscular force and function to provide body support,

propulsion and mediolateral balance associated with strength training for a group of children with CP. Nor has simulation muscle force results been compared with static strength measures after strength training.

Thus the purpose of the study was to characterize changes of individual muscle forces and function for multiple children with CP who have undergone strength training. Using musculoskeletal modeling and simulation, dynamic muscle force production and muscle functional roles can be investigated at the individual muscle level, and be used to help guide future strength training protocols that are focused on specific muscle groups.

4.1 Methods

Six children with spastic diplegic CP (13.5 ± 3.3 years, 152 ± 16.2 cm, 45.0 ± 13.7 kg) were selected from a group of children who underwent a six-week strength training protocol. All children were classified with Gross Motor Function Classification System (GMFCS) level I or II, indicating that the child could walk unassisted for at least short distances. Three of these children underwent a gait analysis and strength training protocol wearing their own prescribed ankle foot orthoses (AFO). Prior to and after the strength training protocol, a gait analysis session and clinical outcome assessments were completed.

4.1.1 Experimental Design

The strength training protocol included a training period of six weeks with training three times per week. Strength exercises were focused on improving strength of the hip, knee and ankle joints. Weighted cuffs were used to provide resistance, and four sets of five repetitions, at 65% of each child's maximum isometric strength, were completed in each training session. Each child's strength was measured every two weeks and the weight used in the training was altered accordingly. Isometric lower-limb strength was measured with a handheld dynamometer before and after strength training for each child. Strength measurements, collected with handheld dynamometry, have been shown to be reliable for measuring the strength of children with CP, and these measures were completed until a consistent result was achieved.

4.1.2 Gait Analysis and Model Development

Each gait analysis session consisted of collecting 3D kinematic and ground reaction force (GRF) data using the same protocol as described in the methods section of Chapter 3. Lower body kinematics were collected while the child walked back and forth along an instrumented walkway. The kinematics and GRFs were used to develop musculoskeletal simulations for each child before and after strength training in OpenSim 3.2. The model design including musculoskeletal geometry, passive joint torques and scaling methods were the same as described previously in Chapter 3. Three children in this study wore their own prescribed ankle foot orthoses during gait analysis sessions. The model described previously was used to apply a passive torque at the ankle and subtalar joints (Table 3.1). A probabilistic analysis, described in Chapter 3, was also used in this study to estimate the sensitivity of AFO parameters on model outputs.

4.1.3 Gait Simulations

Three walking trials for each child were selected for analysis from data collected before and after strength training, totaling 36 full gait cycle walking trials. For the walking trials, marker trajectories were filtered with a second order low-pass bidirectional Butterworth filter with a cutoff frequency of 6 Hz. The GRF data were filtered with the same filter at a low pass cut off of 20 Hz. A least-squares optimization algorithm (Lu and O'Connor, 1999) was used to determine an inverse kinematics solution that minimized the error between experimental marker positions and the rigid body model position, subject to kinematic constraints of the musculoskeletal model, in Visual3D (C-Motion Inc, Germantown, MD).

Simulations were generated in OpenSim 3.2 to reproduce measured kinematics for each child. First, a residual reduction algorithm (RRA) was used to improve the dynamic consistency between the GRFs and the inverse kinematics solution by making slight adjustments to the walking kinematics, torso center of mass (COM) location, and total mass of the model (Delp et al., 2007). In addition, the trunk-pelvis angles were estimated to reduce residuals in this analysis. A computed muscle control (CMC) algorithm was used to solve for the muscle excitations that produced the experimentally-measured walking kinematics, thus generating muscle-driven simulations of walking (Thelen and Anderson, 2006). Once the simulations

were generated, an induced acceleration analysis (IAA) was performed to determine the function of individual muscles in accelerating the body COM. A hard kinematic constraint of rolling without sliding between the model foot and the ground was used to model the ground contact (Hamner et al., 2010). The contact point was modeled at the location of the center of pressure and the GRF was replaced with the kinematic constraint for the analysis at each time step.

4.1.4 Data Analysis

Mean dynamic muscle forces and muscle contributions to the three-dimensional COM accelerations were evaluated both before and after strength training. Individual muscles were categorized into groups for interpretation (Table 3.2) and were evaluated in four phases of gait (Double Support 1 (~0-10% GC), Single Limb Stance (~10%-50% GC), Double Support 2 (~50%-60% GC), and Swing (~60%-100% GC)). Accelerations were quantified in the three anatomical directions, (1) anterior/posterior, (2) superior/inferior, and (3) medial/lateral (Figure 4.1). Change in muscle contributions to COM accelerations in the three anatomical directions were characterized as contributions to (1) propulsion or braking, (2) positive or negative support, and (3) medial or lateral acceleration.

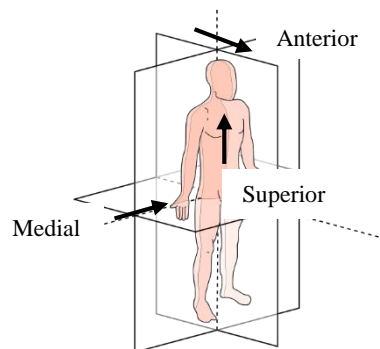


Figure 4.1 Anatomical acceleration directions, with positive directions indicated

A two-way, repeated-measures ANOVA was used to assess differences in muscle force and function (i.e., contributions to COM accelerations) between the left and right legs and as a result of training ($\alpha=0.05$). If significant leg or training main effects or a significant leg \times training interaction effect was found, a paired t-test was used to determine significant differences in specific muscle groups on each leg

as a result of training. Paired t-tests following a significant interaction effect tested each combination (e.g., (1) left leg, before/after training, (2) right leg, before/after training, etc.). The analysis of muscle contributions to acceleration of the COM was limited to the muscle groups that provided substantial contributions (top 8 muscle contributions for each direction of acceleration) to the body COM accelerations. Significant main effects due to training indicated a change in muscle force production or function as a result of training, significant main effects due to leg differences indicated asymmetry in the specified muscle group, and interaction effects indicated that the child's legs were responding differently to training. Specifically, post-hoc results based on interaction effects observed in the ANOVA would indicate a change in symmetry by comparing t-test results for pre-training leg differences with post-training leg differences, or a training effect specific to a single leg by comparing t-test results for individual leg training differences.

Changes in isometric strength measurements were also evaluated before and after gait training. Isometric strength measurements were collected with the child in a gravity neutral position, and an evaluator with a hand held force transducer applied resistance to the child's isometric force contraction about each lower limb joint. A two-way repeated measures ANOVA and associated post hoc analyses were similarly applied to the isometric strength measurements.

4.2 Results

Musculoskeletal simulations developed for each walking trial were used to investigate changes in muscle force production and function for each child. Simulation quality was high for all simulations investigated ($<0.2\%$ BW average and $<12.8\%$ BW at the peak). Few significant training or interaction effects on muscle force production and function were found, although there were several significant leg effects (Table 4.1). The only muscle group with dynamic muscle force production significantly affected by training was TFL unilaterally during the second double support phase. Asymmetry in muscle force production was found in several muscle groups including the IL, GMEDP, and GMAX during the first double support phase, the BFSH during the second double support phase, and the VAS during the swing

phase (Figure 4.2, Table 4.1). Asymmetry in the BFSH muscle force production was also reflected in this muscle's contribution to anterior acceleration in the second double support phase. Overall changes as a result of training found in dynamic force production were limited, and variability across children was high for all muscle groups.

Significant main and interaction effects were also found in the average muscle contributions to body COM accelerations (Table 4.1). Significant interactions after strength training were found in GAS contributions to support during single limb stance and the second double support phase. Decreased contributions to support from the GAS on one leg were found during single limb stance, producing increased symmetry, however this did not reach statistical significance. Increased contribution to support from the GAS on one leg was found during the second double support phase, trending towards increased symmetry. Increased negative contributions to support were also found in the IL during single limb stance resulting in increased symmetry, however these changes did not reach statistical significance. Asymmetry in muscle contributions to COM accelerations was also found, including the BFSH contribution to propulsion during single limb stance and the second double support (Figure 4.3, Table 4.1).

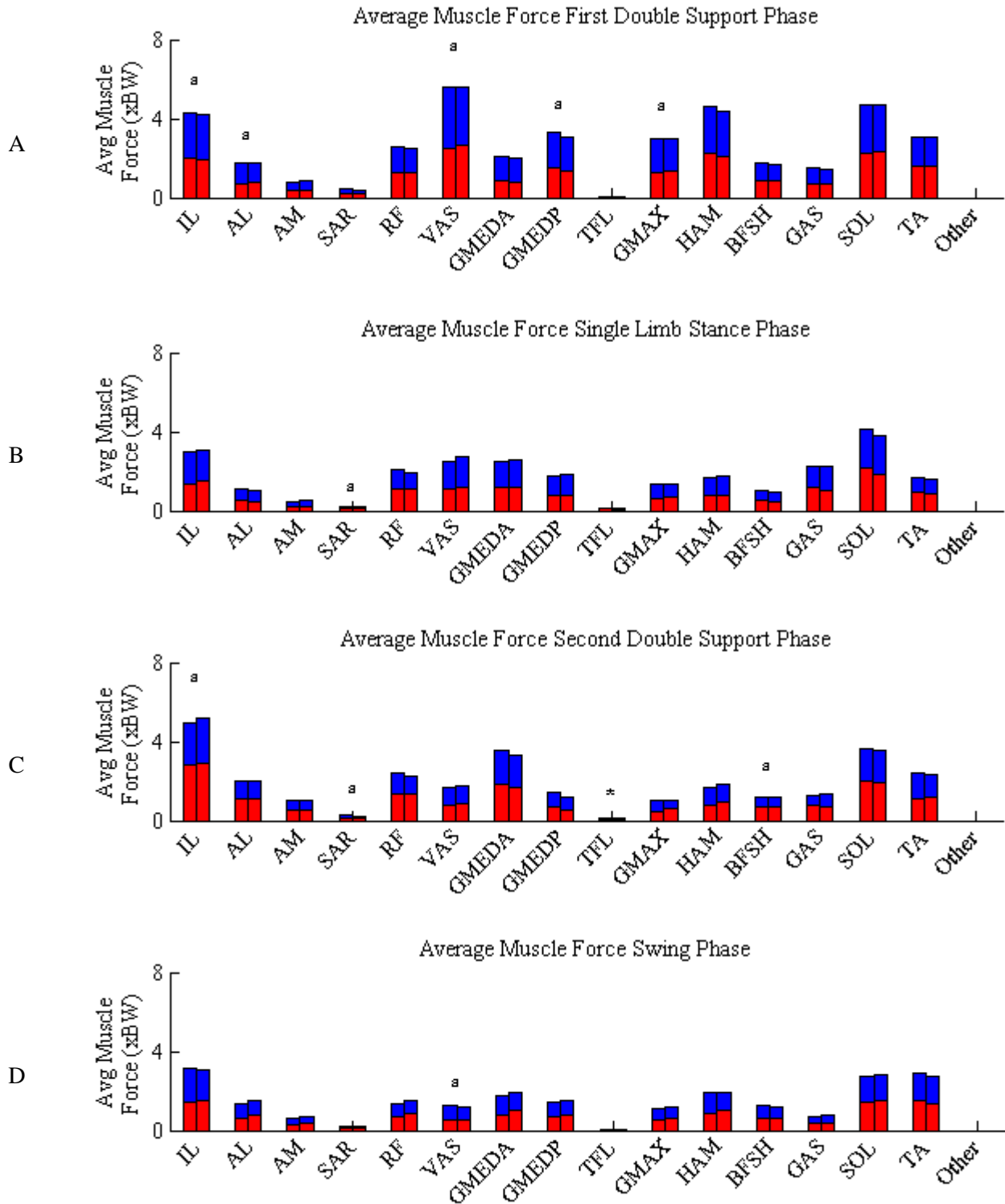


Figure 4.2 Average dynamic muscle force production during (a) first double support, (b) single limb stance, (c) second double support, and (d) swing phase of gait. Left leg (red) and right leg (blue) are shown before (left bar) and after (right bar) training or each muscle group. Significant training ($^{\circ}$), leg (a), and interaction (training \times leg) (*) effects from the ANOVA are shown.

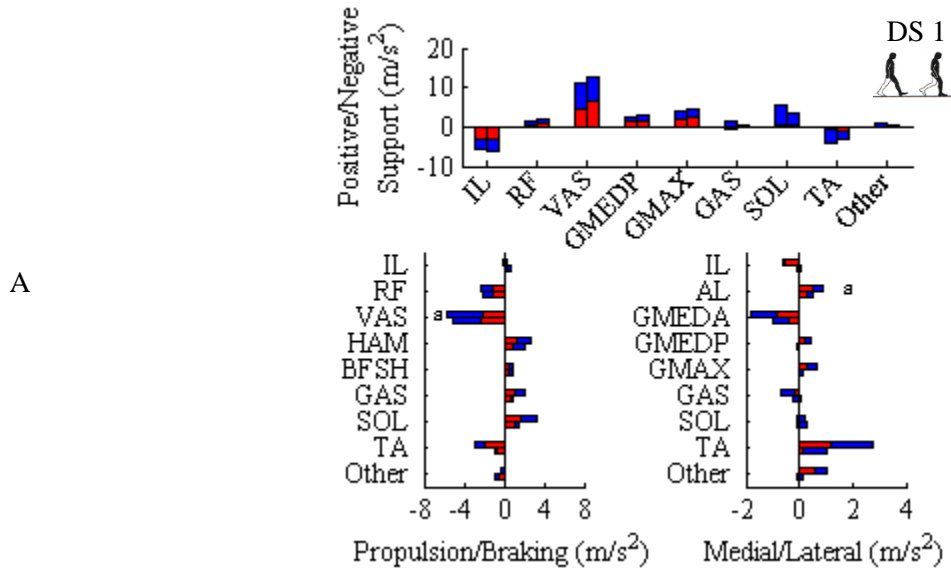


Figure 4.3 Average muscle contributions to COM acceleration during stance phases of gait including the (A) first double support phase, (B) single-limb stance, and (C) the second double support phase. Left leg (red) and right leg (blue) are compared before (left bar/top bar) and after (right bar/bottom bar) training for each muscle group. Only muscle groups contributing substantially to the overall acceleration were evaluated for significant changes (top 8 contributors were selected). Significant training (°), leg (^), and interaction (training × leg) (*) effects from the ANOVA are shown.

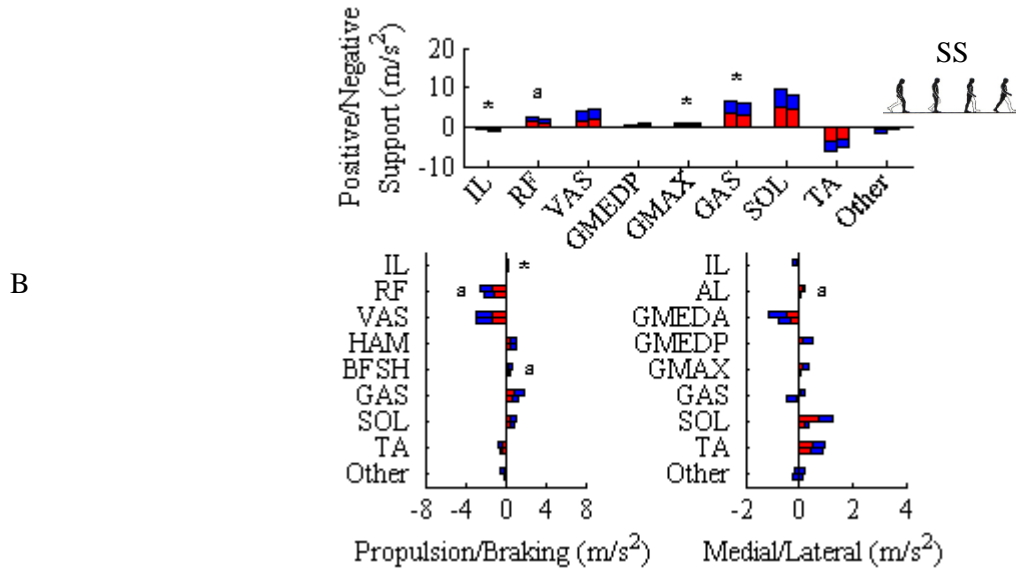


Figure 4.3b: continued.

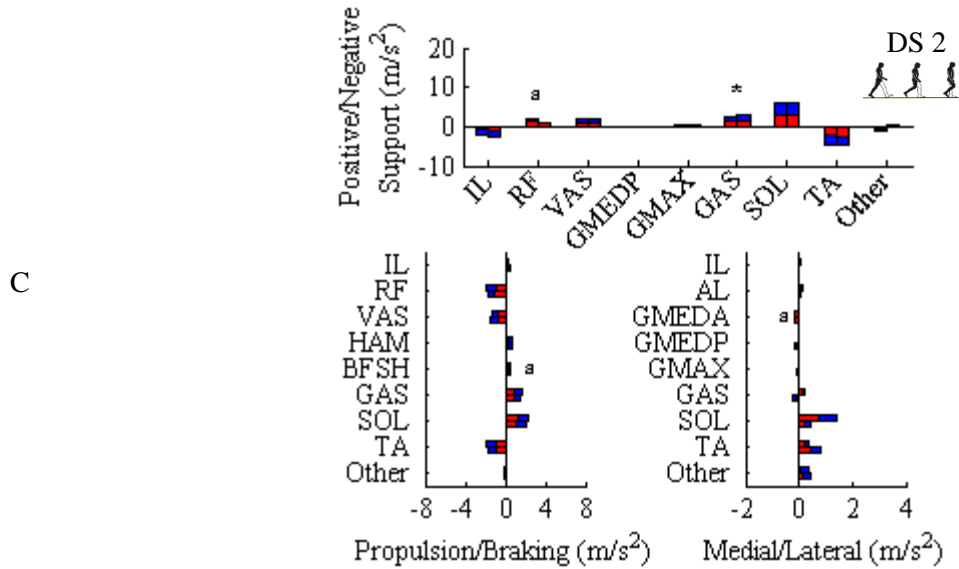


Figure 4.3c: continued.

Table 4.1 Muscle force production and muscle contributions to COM acceleration ANOVA p-value results with significant main (leg, training) and interaction (training \times leg) effects. All muscle groups for dynamic force production and only those muscles with substantial contributions to the COM acceleration were evaluated with the ANOVA. The phase where significant results were found is shown including the first double support phase (DS1), single-limb support (SS), second double support phase (DS2) and swing.

Metric	Muscle Group	Phase	Training Main Effect	Leg Main Effect	Interaction Effect
Average Muscle Force	IL	DS 1	-	0.0321	-
		DS 2	-	0.0314	-
	AL	DS 1	-	0.0018	-
	SAR	SS	-	0.0469	-
	VAS	DS 1	-	0.0017	-
		Swing	-	0.0351	-
	GMEDP	DS 1	-	0.0188	-
	TFL	DS 2	-	-	0.0212
	GMAX	DS 1	-	0.0114	-
	BFSH	DS 2	-	0.0090	-
Average Muscle Contribution to the COM (Ant/Post)	IL	SS	-	-	0.0263
	RF	SS	-	0.0462	-
	VAS	DS 1	-	0.0371	-
	BFSH	SS	-	0.0096	-
		DS 2	-	0.0057	-
Average Muscle Contribution to the COM (Sup/Inf)	IL	SS	-	-	0.0341
	RF	SS	-	0.0398	-
		DS 2	-	0.0321	-
	GMAX	SS	-	-	0.0222
	GAS	SS	-	-	0.0358
		DS 2	-	-	0.0221
Average Muscle Contribution to the COM (Med/Lat)	AL	DS 1	-	0.0151	-
		SS	-	0.0304	-
	GMEDA	DS 2	-	0.0250	-
Isometric Strength Measures	Knee Ext	-	0.0079	-	-
	Hip Flex	-	0.0012	-	-
	Ankle DF	-	-	-	0.0301

Table 4.2 Group mean (SD) of muscle force production and muscle contributions to COM acceleration that had significant ANOVA main (leg, training) or interaction (training × leg) effects during phases of the gait cycle. The phase where significant results were found is shown including the first double support phase (DS1), single-limb support (SS), second double support phase (DS2) and swing. Results from ANOVA post hoc pairwise comparisons are also shown.

Metric	Muscle Group	Phase	Mean and Standard Deviation				POST HOC Analysis					
			Muscle Force (%BW) or Function (m/s ²)				Training (Pre/Post)	Leg (Left/Right)	Training x Leg			
			Pre Training		Post Training				Right leg (Pre/Post)	Left leg (Pre/Post)	Pre Training (Left/Right)	Post Training (Left/Right)
Left	Right	Left	Right									
Average Muscle Force	IL	DS 1	2.052 (1.178)	2.442 (1.402)	2.011 (0.880)	2.323 (1.030)	-	0.0247	-	-	-	-
		DS 2	2.875 (1.356)	2.266 (0.962)	3.047 (1.156)	2.340 (1.252)	-	0.0239	-	-	-	-
	AL	DS 1	0.363 (0.175)	0.481 (0.245)	0.356 (0.194)	0.498 (0.250)	-	0.0013	-	-	-	-
	SAR	SS	0.133 (0.090)	0.086 (0.033)	0.107 (0.023)	0.078 (0.036)	-	0.0389	-	-	-	-
	VAS	DS 1	2.551 (1.097)	3.264 (1.333)	2.727 (1.254)	3.080 (1.058)	-	0.0019	-	-	-	-
		Swing	0.539 (0.283)	0.742 (0.480)	0.526 (0.330)	0.687 (0.443)	-	0.0273	-	-	-	-
	GMEDP	DS 1	1.609 (0.696)	1.800 (0.780)	1.399 (0.789)	1.756 (0.705)	-	0.0162	-	-	-	-
	TFL	DS 2	0.058 (0.048)	0.039 (0.022)	0.037 (0.023)	0.055 (0.035)	-	-	0.1626	0.2040	0.1574	0.0569
	GMAX	DS 1	1.330 (0.748)	1.770 (0.601)	1.395 (0.859)	1.706 (0.591)	-	0.0085	-	-	-	-
	BFSH	DS 2	0.772 (0.346)	0.576 (0.351)	0.738 (0.354)	0.689 (0.355)	-	0.0061	-	-	-	-
Average Muscle Contribution of COM Accel. (Ant/Post)	IL	SS	0.225 (0.198)	0.158 (0.138)	0.132 (0.113)	0.181 (0.137)	-	-	0.8264	0.4817	0.2041	0.1146
	RF	SS	-1.066 (0.261)	-1.340 (0.561)	-0.921 (0.323)	-1.166 (0.369)	-	0.0183	-	-	-	-
	VAS	DS 1	-3.481 (1.475)	-2.158 (1.166)	-2.797 (1.709)	-2.328 (1.31)	-	0.0094	-	-	-	-
	BFSH	SS	0.276 (0.127)	0.387 (0.245)	0.275 (0.136)	0.307 (0.095)	-	0.0326	-	-	-	-
		DS 2	0.208 (0.164)	0.303 (0.1584)	0.208 (0.177)	0.333 (0.210)	-	0.0040	-	-	-	-

Table 4.2: continued.

Metric	Muscle Group	Phase	Mean and Standard Deviation Muscle Force (%BW) or Function (m/s ²)				POST HOC Analysis					
			Pre Training		Post Training		Training (Pre/Post)	Leg (Left/Right)	Training x Leg			
			Left	Right	Left	Right			Right leg (Pre/Post)	Left leg (Pre/Post)	Pre Training (Left/Right)	Post Training (Left/Right)
Average Muscle Contribution of COM Accel. (Sup/Inf)	IL	SS	-0.542 (0.331)	-0.271 (0.196)	-0.456 (0.234)	-0.608 (0.532)	-	-	0.2182	0.7010	0.0731	0.3729
	RF	SS	1.104 (0.599)	1.247 (0.678)	0.736 (0.881)	1.083 (1.082)	-	0.0356	-	-	-	-
		DS 2	0.757 (0.549)	1.328 (1.110)	0.219 (1.162)	0.845 (1.566)	-	0.0038	-	-	-	-
	GMAX	SS	0.380 (0.225)	0.300 (0.195)	0.399 (0.283)	0.522 (0.401)	-	-	0.2901	0.9096	0.3624	0.2327
	GAS	SS	3.060 (1.562)	3.530 (1.884)	3.034 (0.974)	2.869 (0.947)	-	-	0.1770	0.9357	0.1546	0.3216
		DS 2	1.068 (0.759)	1.487 (0.836)	1.310 (0.734)	1.484 (0.662)	-	-	0.9797	0.0040	0.1075	0.3615
Average Muscle Contribution of COM Accel. (Med/Lat)	AL	DS 1	-0.324 (0.247)	0.528 (0.211)	-0.155 (0.211)	0.325 (0.186)	-	0.0002	-	-	-	-
		SS	-0.076 (0.078)	0.148 (0.115)	0.037 (0.095)	0.055 (0.041)	-	0.0360	-	-	-	-
	GMEDA	DS 2	0.034 (0.152)	-0.156 (0.270)	0.058 (0.340)	-0.139 (0.372)	-	0.0071	-	-	-	-
Isometric Strength	Knee Ext	-	0.300 (0.178)	0.301 (0.161)	0.429 (0.164)	0.449 (0.137)	0.0001	-	-	-	-	-
	Hip Flex	-	0.0638 (0.0544)	0.0587 (0.0429)	0.111 (0.0589)	0.123 (0.0599)	<0.0001	-	-	-	-	-
	Ankle DF	-	0.0829 (0.0872)	0.129 (0.121)	0.132 (0.117)	0.214 (0.117)	-	-	0.0465	0.2242	0.2672	0.1437

Changes in isometric strength measurements were significantly affected by training in the knee extensors, hip flexors, and unilaterally in the ankle dorsiflexors (Table 4.1, Figure 4.4). Significantly increased isometric joint strength was found in the knee extensors and hip flexors. In addition, the ankle dorsiflexors isometric joint strength was significantly increased after strength training, however significantly increased strength was not consistent between limbs. In addition, no significant asymmetry in muscle strength was observed before or after gait training at any lower limb joint.

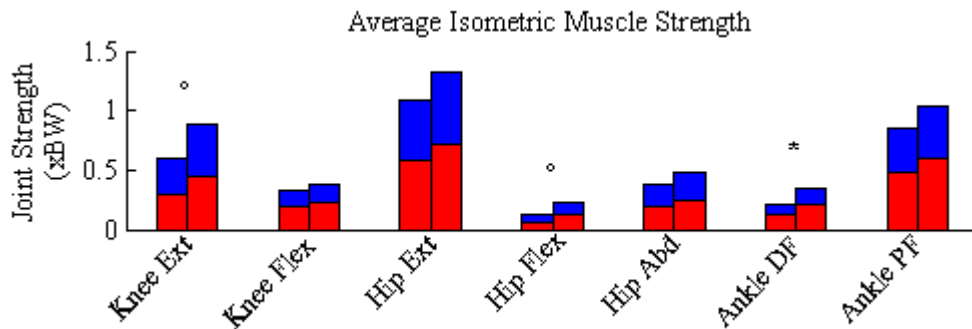


Figure 4.4 Average joint isometric strength for the left leg (red) and right leg (blue) are shown before (left bar) and after (right bar) training for each joint. Significant training ($^{\circ}$), leg (a), and interaction (training \times leg) ($*$) effects from the ANOVA are shown.

4.3 Discussion

The purpose of this study was to use modeling and simulation to characterize changes in individual muscle force production and function as a result of strength training. Through a modeling and simulation approach, the muscles responsible for specific changes to a child's gait pattern were identified.

Overall, musculoskeletal modeling of children with CP before and after strength training identified important changes in the function of the gastrocnemius to accelerate the body vertically during both single limb stance and the second double support. Greater negative support provided by the hip flexors during single limb stance was also found after training. Changes in muscle function were not consistent between legs for the children in this analysis; however, the individual leg changes did lead to an increase in symmetry after training. Several muscle groups were asymmetric and were not affected by training in both the contributions to body acceleration, and overall dynamic muscle force production. Changes in

isometric strength were also characterized with the knee extensors, hip flexors, and ankle dorsiflexors having significantly increased isometric strength after training. Changes in isometric strength measures were not consistent with changes in dynamic muscle force production or function after training.

Group level changes in dynamic muscle force production were primarily found only in leg differences, indicating asymmetry. Muscles with asymmetric force production in the first double support phase, including the IL, GMEDP, and GMAX all contribute substantially to positive (GMED, GMAX) or negative (IL) support of the body. Although no significant asymmetry in contributions to COM acceleration was evident for these muscles, they may be important targets for increasing strength, or selective motor control. Specifically, the GMED may benefit from increased targeted strength training or functional electrical stimulation as muscle weakness in this muscle may be a contributing factor in crouch gait for children with CP (Steele et al., 2012b). The GMAX may also substantially benefit from additional strength training as this muscle has been shown to accelerate the hip and knee into extension during stance, and the capacity of this muscle to produce joint torque is greatly reduced in crouched postures, leading to the potential for increased demand (Arnold et al., 2005; Hicks et al., 2008). Similar asymmetry was found as a result of gait training in the hip extensors and hip abductors. As neither gait nor generalized strength training significantly altered asymmetry in these muscles, a targeted therapy program for these muscles may help increase overall performance after therapy. Alternatively, other contributing factors may be limiting the ability of these children to normalize force production in these muscles, such as physiological properties including remodeling of motor units associated with spasticity (Marbini et al., 2002), and impaired rapid force generation in individual muscles (Moreau et al., 2012).

Significant changes in muscle function after strength training were found in a few muscle groups, including GAS. Decreased contributions from GAS to body support during single limb stance and increased contributions during the second double support phase suggest increased control of the timing of muscle force modulation during gait (Steele et al., 2010). Previous work has found that constant contributions to support and propulsion from the GAS were required in combination with constant support and braking contributions from the VAS. In able-bodied gait the contributions to support are

modulated with the VAS supporting the body in the first double support, the GMEDA, GMEDP, and skeleton supporting the body during single limb stance, and the GAS during second double support (Anderson and Pandy, 2003; Steele et al., 2013). Increased contributions to COM support during the second double support phase, in combination with decreased contributions to support in single limb stance indicate GAS function that is more closely associated with able-bodied function. Similar to gait training, strength training affects the main contributors to support during gait, and may produce more normalized function. While in gait training, significant differences were found in the SOL contributions to propulsion in second double support, and VAS contributions to support during first double support, both training types improved the function of the ankle plantarflexor – knee extensor muscle coordination. Both training types resulted in significant interaction effects in muscle function indicating that training results are not consistent bilaterally for a child with CP.

Asymmetry in dynamic muscle force production and muscle contributions to body COM acceleration were significant for several muscle groups throughout the gait cycle. This asymmetry was not, however, found in isometric strength measures for each child. Static strength measured before and after strength training did not indicate group level asymmetry. Although overall strength measured isometrically may have increased similarly for both limbs, the method the children are using to develop muscle force to drive motion during gait was asymmetric. These results may also be indicative of the ability of these children to selectively control a single joint motion, compared to controlling orchestration of several muscle groups during a dynamic task. During dynamic tasks, asymmetry may be a result of a combination of neuromuscular factors, such as altered coordination and muscle physiology (Damiano et al., 2002a; Marbini et al., 2002; Moreau et al., 2012). In addition, significantly increased strength measures after training were not consistent with changing dynamic muscle force production and function. For example, changes in the GAS contributions to support were not seen in isometric strength measurements at the joint level and changes in knee extension strength were not reflected in dynamic muscle force or function of the VAS as a result of training. The different results observed in dynamic muscle force production and isometric strength testing highlight the utility of modeling and simulation in assessing the underlying

biomechanics of children with CP. Similar results for dynamic joint range of motion and gait kinematics were found to be uncorrelated to corresponding static measures of joint range of motion (McMulkin et al., 2000) and static pose estimation of musculotendon lengths (Delp et al., 1996).

Although group level changes in muscle force and function have provided insight into possible targets for additional therapy, high variability in the muscle force production and muscle induced accelerations revealed few group level strength training effects. As with gait training, it is not unexpected that muscular force and function would be variable across children, given the variable nature of altered gait patterns in children with CP (Wren et al., 2005). Many studies have found variable results in functional measures including hip and knee extension during gait, walking speed, and stride length (Damiano et al., 2010; Steele et al., 2012a). These results indicate that assessment of muscle functional roles at the individual child level is needed, which would provide important information for developing individualized therapy protocols.

Individualized analysis of each child revealed unique training effects in dynamic muscle force production and function. There were two primary changes in function found after the training. Child 1 and Child 3 increased contributions to support from the GMEDA throughout stance. Child 1 also reduced support contributions from the VAS group and braking contributions from the VAS and RF during mid stance (Figure 4.5), suggesting reduced duration of knee extensor activity and increased reliance on the hip abductors during mid-stance, more consistent with able-bodied gait (Anderson and Pandey, 2003). Increased contributions from the GMEDA and decreased contributions from the VAS during mid-stance suggests more efficient muscle coordination and function after training (Steele et al., 2013).

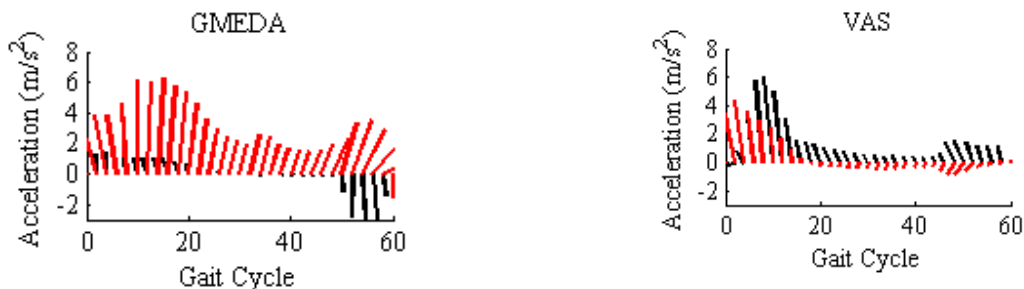


Figure 4.5 Child 1 GMEDA and VAS contributions to support and propulsion before (black) and after (red) strength training

Changes in the GMEDA for Child 1 after training may be related to increased strength measured after training, or better control of anti-gravity musculature, allowing modulation of lower limb muscle force. Alternatively, Child 5 increased ankle plantarflexion propulsion and support during the second double support phase (Figure 4.6). These increases were localized on only one side of the body, with the other leg providing little overall change. Both strategies (i.e., reducing VAS and RF braking contributions in mid stance versus increasing propulsion contributions from the ankle plantarflexors in the second double support phase) provided increased propulsion and are valid strategies for increased walking speed. Although, walking speed was not significantly different during the observed walking trials, maximum walking speed was increased after strength training, measured with the 10-meter walk test for both Child 1 and Child 5. The differing strategies may be why group level analyses showed little overall change after strength training, and why the results of individual muscles had high variability across children.

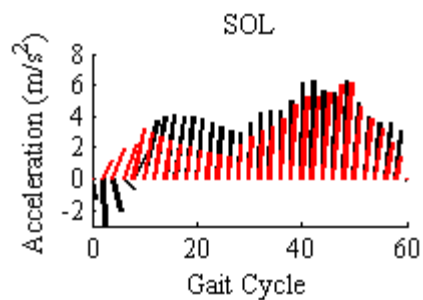


Figure 4.6 Child 5 SOL contributions to support and propulsion before (black) and after (red) strength training

Musculoskeletal modeling and simulation approaches facilitate detailed analyses of dynamic muscle force production and function, which can be used to help evaluate treatment protocols. Potential limitations noted previously for the musculoskeletal analysis of gait training (CHAPTER 3) are similarly applicable for this study of strength training. Small sample sizes for these analyses and variable results provided limited ability to draw conclusions at a group level; however many previous modeling studies performed useful analyses on a single simulation of group average data (e.g., Silverman and Neptune 2012). Our study builds on previous methods by simulating several walking trials for multiple individuals, and we look forward to expanding these results further in future work. Limitations to the

model, including the effects of bone deformities, muscle weakness, and AFO parameter estimations were also evaluated and the uncertainty associated with these limitations similarly apply to this strength training study, as in the gait training study (CHAPTER 3). As this was a longitudinal study, all modeling decisions were consistent across training conditions, and only changes in simulation outputs were evaluated in this study. The children in this study also had previous surgical interventions including musculotendon lengthening surgery, bony alignment surgery, and RF transfer surgeries. As the function of the RF is significantly affected by transfer surgery (Fox et al., 2009), the model used for this study has potential limitations for children who have undergone this procedure. The results for this study are relatively robust to limitations of this model, however. When children with RF transfer surgery were removed from the analysis, significant RF support contribution asymmetry remained significant during single limb stance, and shifted from significant to close to significant in the second double support phase. Future work will incorporate a model for the RF transfer into full body musculoskeletal simulations to account for the change in function of this muscle.

Direct validation of estimated muscle excitation timing from the simulations was not possible in this analysis because electromyographic data collection was not included in the experimental protocol. In addition, torso kinematics were not collected during the experimental gait analysis. However, our simulations were of high quality in that residual forces and moments at the pelvis were small, indicating a high level of dynamic consistency. In addition, joint reserve actuators were also small and trunk-pelvis angles were estimated to further increase dynamic consistency in the simulations. Future work will incorporate electromyographic data collection and torso kinematics, to validate and improve our simulation generation methods. Even though muscle excitations could not be directly validated, our results indicated group level changes consistent with adaptations towards able-bodied gait in the GAS despite high variability in the results.

The muscles that were most greatly affected by strength training varied across children, and at times the largest differences for an individual child were not apparent in the group statistical analysis. However, changes in timing and magnitude of the support contributions from the GAS may indicate that strength

training most effectively targets this muscle to produce more normalized function. These results suggest that the neuromuscular impairments associated with individual muscles vary across children with CP; however, altered ankle plantarflexor function was found for multiple children. As with gait training, strength training has the potential to affect many different muscle groups, as was demonstrated by the different strategies adopted by Child 1 and Child 5 after training. The potential for each child to improve also varies greatly.

4.4 Conclusion

In this study, we found that strength training affects individual children differently, and significant changes related to training were only found in the actions of GAS during mid and late stance, producing accelerations more closely matched to able-bodied gait. Musculoskeletal modeling and simulation analysis used to determine dynamic muscle force production and muscle function may provide information to therapists that is not evident with static joint level isometric strength. Thus, musculoskeletal modeling and simulation analysis can help greatly in identifying the muscular deficits for an individual child, and may be instrumental in developing targeted physical therapy protocols that are tailored to a specific patient.

CHAPTER 5

SENSITIVITY ANALYSIS FOR ANKLE FOOT ORTHOSIS MODEL PARAMETERS

Cerebral palsy (CP) is a perinatal brain disorder that often results in mobility impairments.

Approximately 1 in every 323 children in the United States are affected by CP, with about 42% of those having limited to no walking ability (Christensen et al., 2013). The cause of common gait abnormalities in children with CP is often unclear, and results of treatment interventions are often variable. Recently, the use of gait analysis and three dimensional modeling has improved the probability of successful treatment interventions for these children (Chambers, 2001). Many children with CP require some kind of assistive device to walk (Knutson and Clark, 1991), which presents a challenge to develop musculoskeletal models that incorporate these assistive devices. Commonly, children with CP use ankle-foot orthoses (AFO) to assist in producing an appropriate range of motion at the ankle during gait and to prevent skeletal deformity (Knutson and Clark, 1991). The range of stiffness and the type of AFOs used in children with CP is diverse, and the purpose of this device can range from reducing fall risk in children who exhibit equinus gait, to supporting the body during stance in children with crouch gait, by preventing excessive forward translation of the tibia (Bregman et al., 2009; Radtka et al., 2005). Several experimental studies have been conducted to determine joint level stiffness and damping properties of these devices, reporting stiffness ranges from 0.2-2.667 Nm/deg with the evaluation of both polypropylene and carbon fiber orthoses with different design configurations (Cappa et al., 2005; Crabtree and Higginson, 2009; Sumiya et al., 1996). Without individual testing of these devices to determine each orthoses' mechanical properties, an accurate representation of an individual's AFO in a modeling environment can be difficult.

Although these AFOs affect walking speed, cadence, ankle kinematics, and ankle power, previous work also shows the timing of muscle activation in lower limb muscles is unchanged with the application of the AFO (Lam et al., 2005; Radtka et al., 2005, 1997). The use of modeling then becomes even more important to identifying the effects of AFOs on overall muscle function. A study conducted by Crabtree *et al.* developed a tunable AFO model to use in the investigation of neuromuscular effects of wearing AFOs during normal gait. The model used was a non-linear logarithmic model, tuned based on mechanical

testing of an individual AFO. Crabtree *et al.* established that the contribution of the soleus to ankle acceleration during gait was reduced, while the contribution from the gastrocnemius was increased (Crabtree and Higginson, 2009). Because many children with CP who have mobility impairments use assistive walking devices including AFOs, quantifying the sensitivity of AFO model parameters on gait simulation results is critical for effectively evaluating gait simulation results. Individual testing of a child's AFO may not be feasible in a clinical setting, thus the effect of estimating AFO model parameters is critical in musculoskeletal model development and simulation interpretation.

The purpose of this study, therefore, was to determine the effect of AFO mechanical property assumptions, including stiffness, damping, and equilibrium angle, on subject specific whole body musculoskeletal simulation of gait in children with CP.

5.1 Methods

Previously collected data from three dimensional gait analysis for one subject with spastic diplegic cerebral palsy who walked with AFOs was used in this analysis. Sixteen retro-reflective markers were placed on the lower limbs and pelvis of the subject. Four force plates (AMTI Inc., Watertown, MA) recorded six degree-of-freedom forces and torques in concert with lower limb kinematic data collected at 120 Hz (Vicon Motion Systems, Ltd., Oxford, UK). A single gait cycle was recorded for simulation. The subject wore his own posterior leaf spring AFOs during the data collection session. The subject walked with a mild crouch gait and was able to ambulate independently.

5.1.1 Musculoskeletal Model

A generic musculoskeletal model with 21 degrees of freedom and 92 musclotendon actuators with force-length and force-velocity properties (Delp et al., 1990) was scaled and adjusted to match the child's height and weight. In addition, passive torque actuators were added bilaterally to the ankle and subtalar joints to model the individual's AFOs. The passive actuator was modeled with a passive stiffness and damping factor. The stiffness was modeled as a torsional spring with a single stiffness factor (k_{θ}). The stiffness in each direction of motion at the joint was altered independently (for example, dorsiflexion

stiffness was independent of plantarflexion stiffness). First order torsional damping was also included in each joint, and was considered constant across the entire joint range. Finally, equilibrium points were included in the analysis to model the effect of changes in the “neutral” point of the AFO. The neutral point changed the angle where zero torque was applied by the AFO from the default of 0° (foot is flat on the ground and the shank is positioned vertically in space) to some altered angle. The equation for the passive AFO torque is shown in Equation 5.1.

$$\begin{aligned} \tau_{AFO} &= k_1 * |\theta| - d * |\dot{\theta}| \\ &\text{when } \theta > \Theta_{upper\ limit} \\ \tau_{AFO} &= -(k_2 * |\theta| - d * |\dot{\theta}|) \\ &\text{when } \theta < \Theta_{lower\ limit} \end{aligned} \quad (5.1)$$

Where: τ_{AFO} is the passive joint torque
 θ is the joint angle (ankle and subtalar)
 $\dot{\theta}$ is the joint velocity
 k_1 is a torsional stiffness factor applied above the upper limit joint angle threshold
 k_2 is a torsional stiffness factor applied below the lower limit joint angle threshold
 d is the damping coefficient
 $\Theta_{lower\ limit}$ and $\Theta_{upper\ limit}$ are bounds on the neutral region

5.1.2 Musculoskeletal Simulation

A muscle-driven simulation was developed to match the experimentally collected kinematics and kinetics. Using a 7 segment rigid body model with kinematic constraints at all lower limb joints, an inverse kinematics solution was produced for a single gait cycle based on the experimentally collected kinematic data (Lu and O'Connor, 1999). Musculoskeletal simulation methods previously described (CHAPTER 3) were used to improve dynamic consistency of experimentally collected data. Dynamic muscle force production was estimated using a computed muscle control algorithm (Thelen et al., 2003). The average muscle force was evaluated for each phase of the gait cycle, consistent with previously described methods (CHAPTER 3). Muscle contributions to body center of mass (COM) acceleration were calculated using an induced acceleration analysis, using a rolling without slipping hard constraint to model the foot-ground contact (Hamner et al., 2010). Average contributions to anterior/posterior, or propulsion/braking accelerations, and superior/inferior or positive/negative support accelerations were

determined for each phase of gait, as described previously (CHAPTER 3). Four muscle groups were evaluated in this analysis including the hip abductors (gluteus medius, gluteus minimus, and gluteus maximus); quadriceps (rectus femoris, vasti medialis, vasti intermedius, and vasti lateralis); hamstrings (semimembranosus, semitendinosus, biceps femoris long head, and biceps femoris short head); and ankle plantarflexors (medial gastrocnemius, lateral gastrocnemius, and soleus). As the child was diplegic and wore symmetric AFOs on each leg; only the right leg was analyzed to reduce the complexity of the analysis.

5.1.3 Sensitivity Analysis

A sensitivity analysis was completed to determine the effect of the AFO parameters on the resulting estimation of muscle forces and contributions to body COM accelerations during a gait cycle. A 200 trial Monte Carlo simulation was completed to develop a sample set of simulations that incorporated a range of stiffness, damping coefficients, and equilibrium angles for the AFO. In addition, an Advanced Mean Value (AMV) analysis was used and compared to the results developed from the Monte Carlo Simulation. The AMV method was only performed on the ankle plantarflexors because the results of the Monte Carlo simulation indicated that the other muscle groups were not substantially affected by the changes in AFO values. The AMV method was used to determine the sensitivity of the AFO parameters at the 5%, 10%, 25%, 50%, 75%, 90%, and 95% probability levels for the plantarflexors at all four phases of gait. A second 50 trial Monte Carlo simulation was performed to develop a probabilistic model for changes in muscle contributions to COM accelerations. The model parameters for all sensitivity analyses in this study were developed based on previous values from the literature (Bregman et al., 2009; Crabtree and Higginson, 2009; Sumiya et al., 1996; Yamamoto et al., 1993) and are reported in Table 5..

The sensitivity of the four major muscle groups influential during gait (hip abductors, quadriceps, hamstrings, and plantarflexors) were analyzed based on the results of the Monte Carlo simulation and the AMV analysis for each phase of gait.

Table 5.1 Ankle Foot Orthosis Model Parameters based on previously collected values for a range of AFOs. These parameters were used as the input for the probabilistic analysis.

Parameter	Average Value	Standard Deviation	Distribution Type
Ankle dorsiflex stiffness (ADSTIFF)	0.832 Nm/deg	0.822 Nm/deg	LogNormal
Ankle plantarflex stiffness (APSTIFF)	0.890 Nm/deg	0.802 Nm/deg	LogNormal
Ankle equilibrium point (AEQ)	1.023 deg	3.778 deg	Normal
Ankle damping (ADAMP)	0.01 Nm/deg/s	0.01 Nm/deg/s	LogNormal
Subtalar eversion stiffness (SESTIFF)	0.800 Nm/deg	0.349 Nm/deg	LogNormal
Subtalar inversion stiffness (SISTIFF)	0.812 Nm/deg	0.366 Nm/deg	LogNormal
Subtalar equilibrium point (SEQ)	0 deg	1 deg	Normal
Subtalar damping (SDAMP)	0.01 Nm/deg/s	0.01 Nm/deg/s	LogNormal

5.2 Results

Cumulative distribution plots were evaluated for the 200 trial Monte Carlo simulation for each muscle group and gait phase. The cumulative distribution plots indicate the change in muscle force production as a result of the variable AFO parameters (Table 5.1). The muscle group most significantly affected by the AFO was the ankle plantarflexors, specifically during stance. The second most important muscle group was the hip abductors; however, the range of muscle forces produced by this muscle group was greatly reduced, compared to the plantarflexors. The other muscle groups showed little change in the average muscle force produced over the phase of gait as a result of the perturbed AFO variables (Figure 5.1).

Based on the Monte Carlo simulation, the sensitivity of the output metrics was evaluated in relation to each input variable. In general the most influential input parameters for all four muscle groups were the ankle dorsiflexion stiffness and the subtalar damping parameters. The dorsiflexion stiffness had a large impact on the hamstrings, quadriceps and plantarflexors, with a positive correlation in the hamstrings and a negative correlation in the quadriceps and the plantarflexors, meaning that increased dorsiflexion stiffness increased hamstring muscle force production, and reduced quadriceps and plantarflexion muscle force. The subtalar damping was most influential to the hip abductors and hamstrings in the first double support and swing phases, and the quadriceps throughout the entire gait cycle. Additional important factors may include the ankle equilibrium point for the hamstrings, quadriceps, and plantarflexors (Figure 5.2).

The AMV method reasonably estimated the behavior of the model based on the Monte Carlo results for the plantarflexor muscle group throughout each phase of gait. Seven performance levels were evaluated using AMV including 5%, 10%, 25%, 50%, 75%, 90%, and 95%. The average difference between the two methods was determined by taking the difference at each probability level and then averaging the difference for each phase of gait. The differences for the four phases of gait for the plantarflexor muscle force were -8.25%, -8.53%, 4.88%, and -2.47%, respectively. Negative differences indicated a shift of the CDF to the left, or that the AMV method was underestimating the values seen in the Monte Carlo simulation.

Regardless of the small discrepancies in the two probabilistic results, the reported importance for each input parameter was evaluated. The importance factors for the input parameters were consistent with the Monte Carlo simulation sensitivity levels. The ankle dorsiflexion stiffness was the most significant factor; however, the equilibrium angles of both the ankle and subtalar joint indicated a much stronger relationship to plantarflexion muscle force in AMV compared to Monte Carlo.

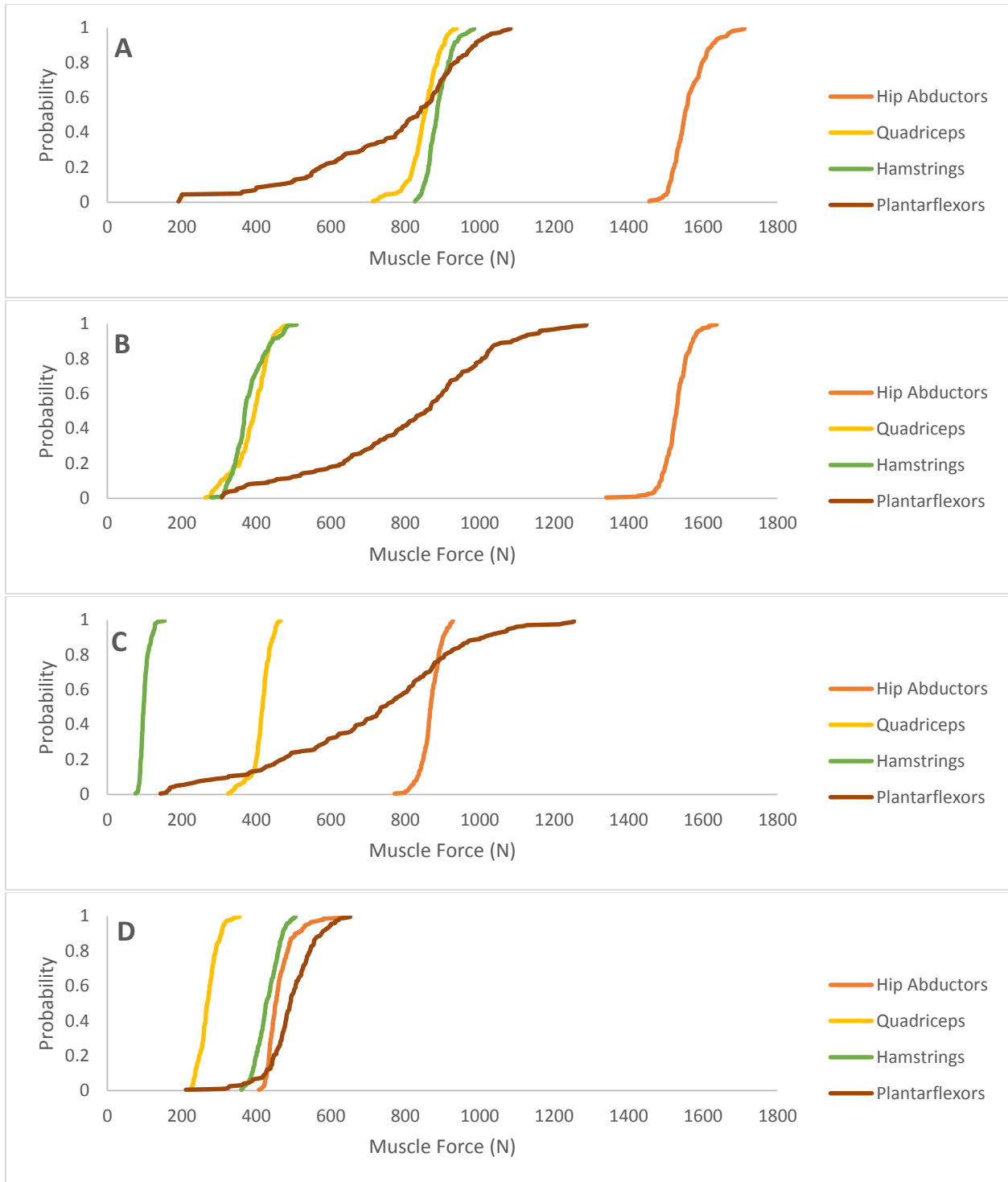


Figure 5.1 Cumulative Distribution Functions (CDF) for average muscle force production in (A) double support 1, (B) single limb stance, (C) double support 2, and (D) swing phases of gait based on a 200 trial Monte Carlo Simulation.

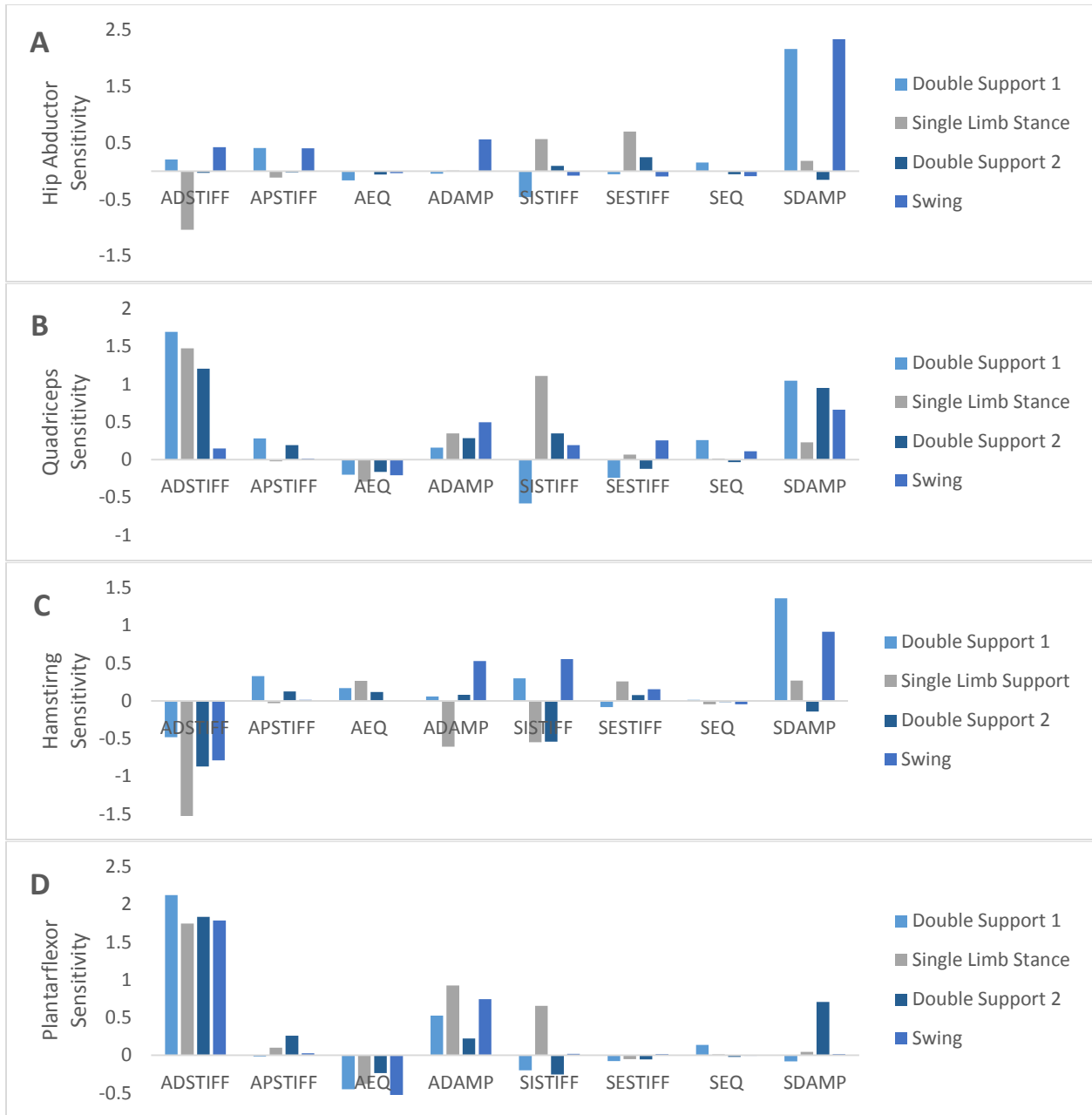


Figure 5.2 Sensitivity levels of muscle group forces to each input parameter (defined in Table 5.1). Positive sensitivity values indicate negative correlations, negative sensitivity levels indicate positive correlations. The sensitivity of the (A) hip abductors (B) quadriceps, (C) hamstrings, and (D) plantarflexors for all phases of gait for each input variable are shown.

The average importance levels were determined using AMV method for the seven probability levels (Figure 5.). The importance levels indicated in the AMV analysis do not specify if the input parameters have a positive or negative correlation with the output muscle force; rather, they indicate the relative contribution of each parameter on total variability in the results. To evaluate the direction of each

correlation, the sensitivity factors were evaluated. Strong negative correlations with plantarflexion force production existed for the ankle dorsiflexion stiffness, and strong positive correlations existed for the ankle and subtalar equilibrium angle.

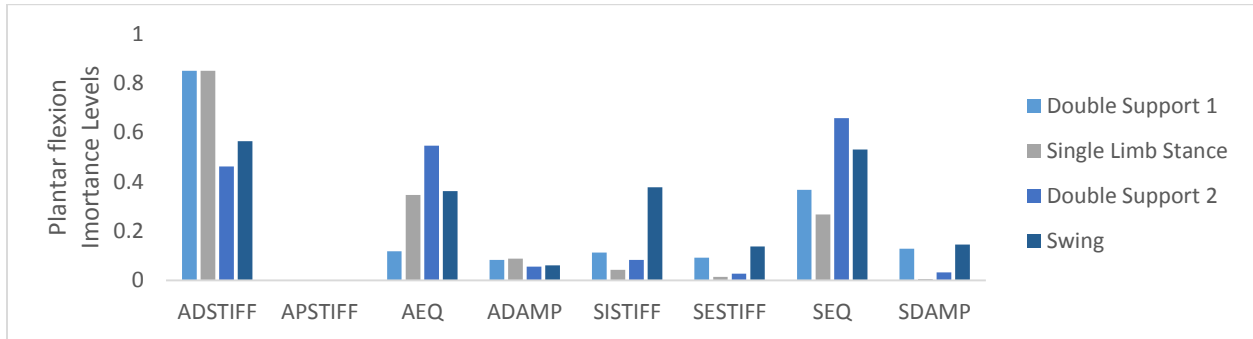


Figure 5.3 Advance Mean Value importance levels in plantarflexion muscle force metric for all phases of gait.

A statistical analysis was also completed for muscle contributions to COM accelerations. A more limited data set for the Monte Carlo was used, and was evaluated against the 200 trial simulation to ensure the more limited data set still captured the behavior of the system (Figure 5.4). The results from the limited data set closely match the larger data set, however the smaller sample set does not capture the farthest most parts of the tails on the distribution function.

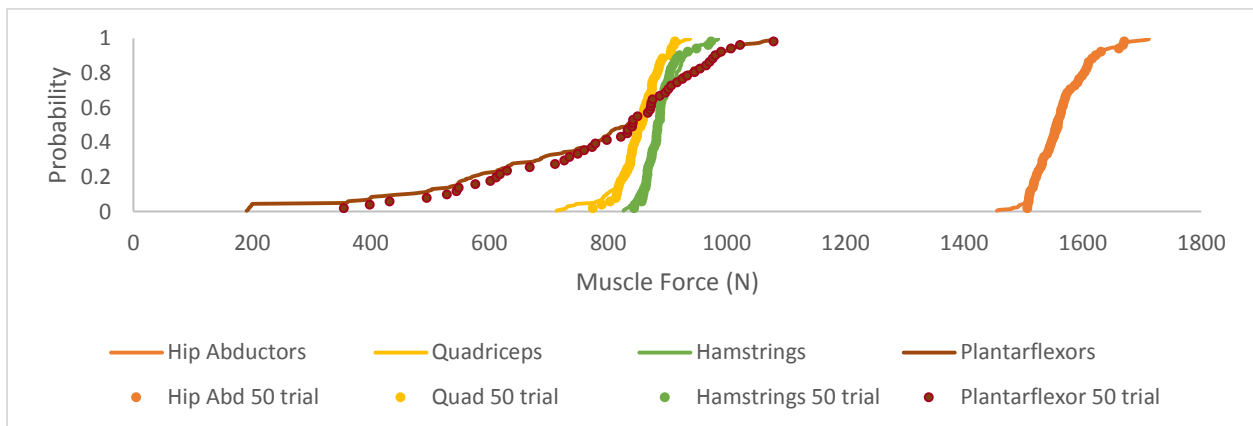


Figure 5.4 Cumulative distribution function for average muscle force production during the first double support phase. The 200 trial Monte Carlo simulation is used as the standard to compare the limited data set of only 50 trials.

Muscle contributions to propulsion (Figure 5.5) and support (Figure 5.6) responded similarly compared to changes in dynamic muscle force production. The ankle plantarflexor group was most affected, with the AFO limiting propulsion contributions completely during the second double support

phase at the tails of the probability analysis. Changes in AFO parameters have an influence on the overall function of the hip abductors, shifting their average function from braking in all three phases of gait to propulsion.

Ankle plantarflexor contributions to support were largely affected by changes in AFO parameter estimations. For muscular contributions to support, a shift in the overall function (positive/negative support) was not found for any muscle group based on AFO estimations.

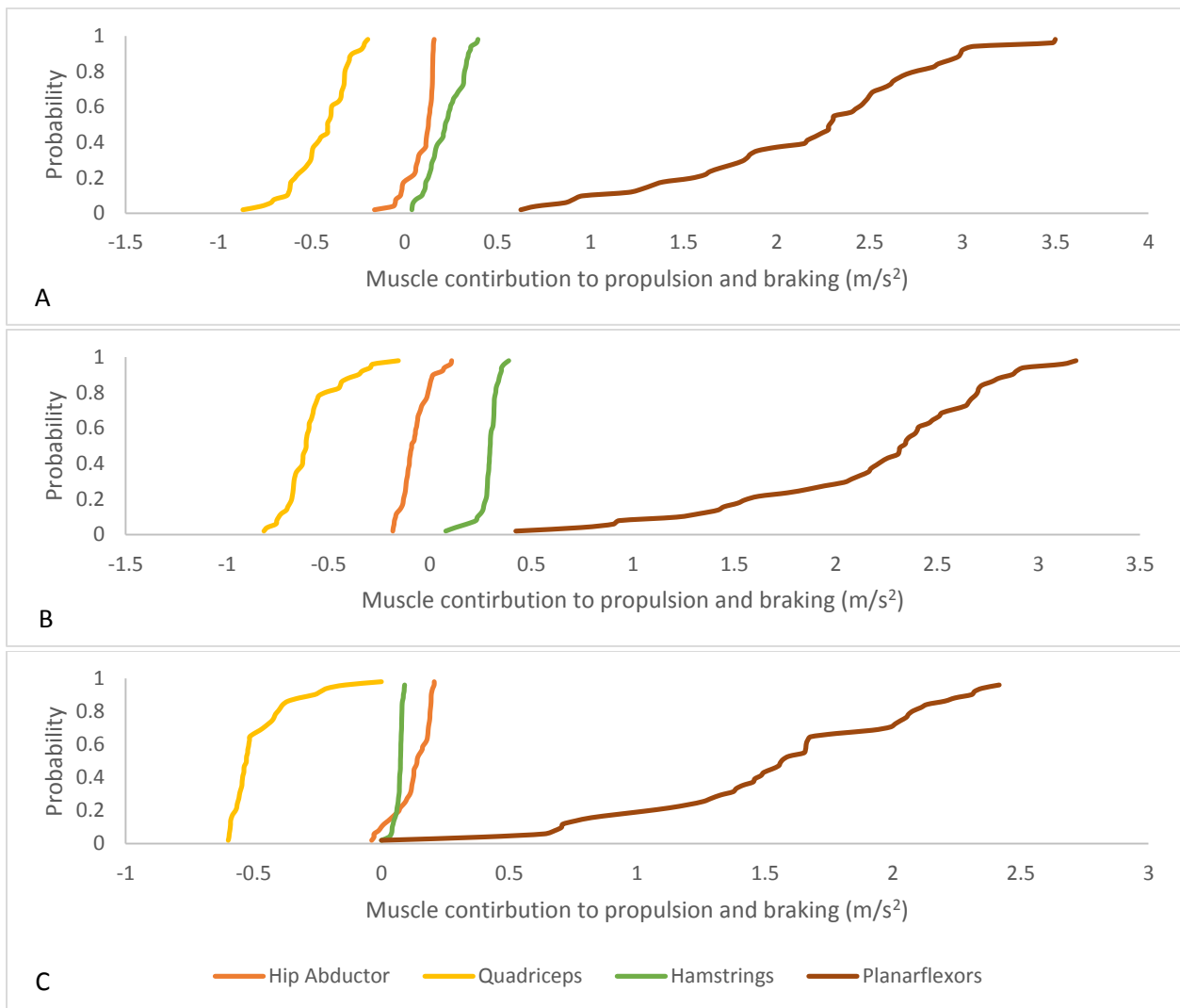


Figure 5.5 Cumulative distribution function for muscle contributions to center of mass accelerations in the anterior/posterior direction (propulsion/braking) for (A) first double support, (B) single limb stance, and (C) second double support for the four muscle groups of interest, and the ankle AFO contribution.

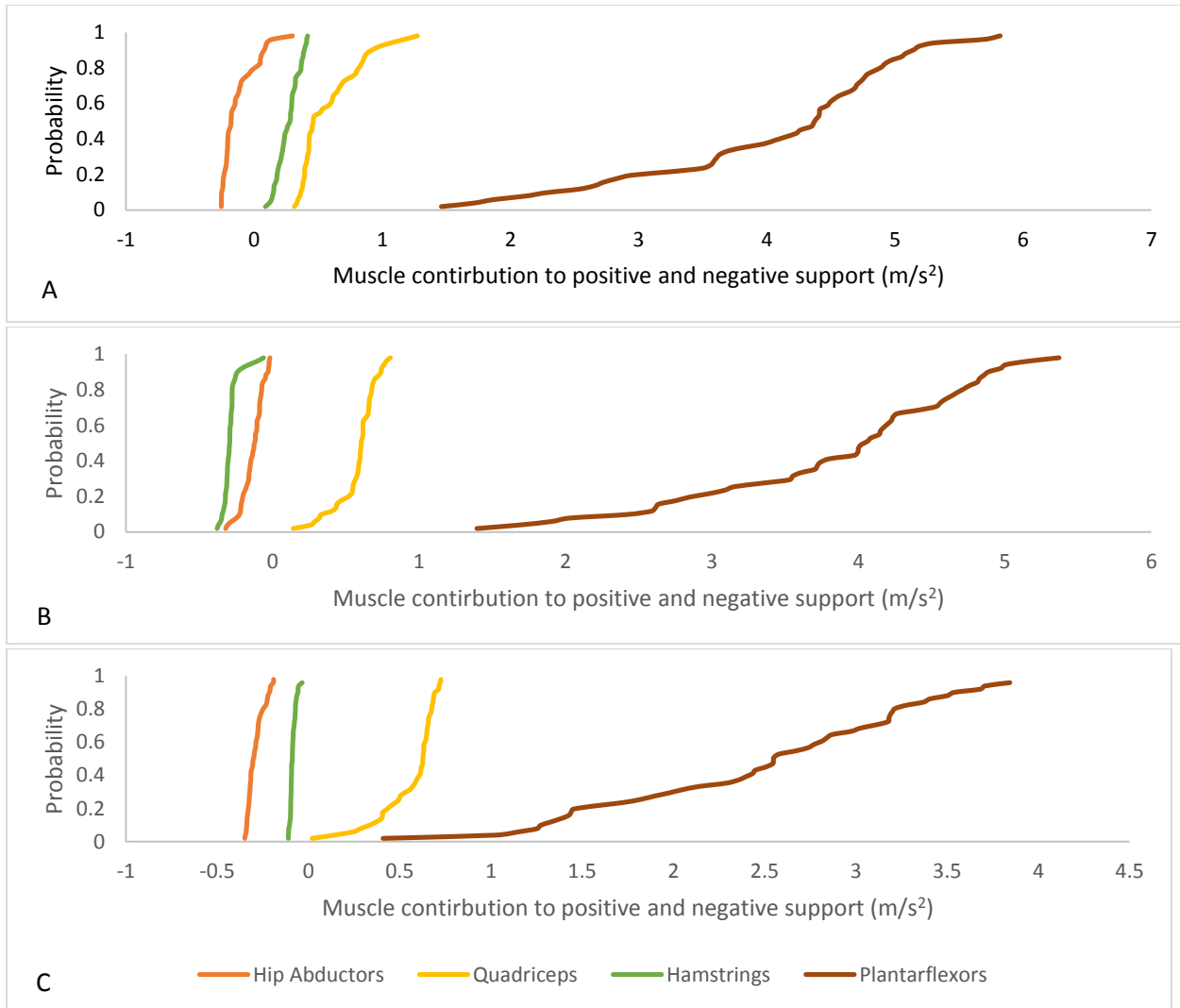


Figure 5.6 Cumulative distribution function for muscle contributions to center of mass accelerations in the superior/inferior direction (positive/negative support) for (A) first double support, (B) single limb stance, and (C) second double support for the four muscle groups of interest, and the ankle AFO contribution.

The parameters that introduced the most variability into ankle plantarflexion propulsion and support estimates included the ankle dorsiflexion stiffness, ankle equilibrium point, and ankle damping. Increased estimations in ankle dorsiflexion stiffness led to decreased plantarflexor propulsion contributions. Increased estimations in ankle neutral angle, and ankle damping led to increased plantarflexor propulsion contributions. Increases in dorsiflexion stiffness lead to decreased contributions from the plantarflexors in stance. A similar trend was seen for plantarflexor contributions to support for dorsiflexion stiffness, ankle neutral angle, and ankle damping.

5.3 Discussion

The purpose of this analysis was to define uncertainty intervals and overall effects of estimating AFO parameters on musculoskeletal modeling results. Many populations with neuromuscular disorders, including children with CP, commonly wear AFOs to increase walking ability. Modeling these devices properly is important to analyze, interpret and treat gait pathologies. Unlike a prosthetic device that replaces a portion of the limb, an AFO is intended to augment the function of the ankle musculature. Modeling the interaction of an AFO with the biological joint and musculature can be difficult. In addition, previous research has shown that the use of an AFO significantly affects the contributions to body propulsion from the ankle plantarflexors (Crabtree and Higginson, 2009).

In this analysis, the sensitivity of the model to AFO mechanical parameters was evaluated. Most notably, the ankle muscles including the gastrocnemius and soleus had the greatest sensitivity to changes in the AFO model for both dynamic muscle force production values, and muscle contributions to COM acceleration. The most significant input parameter for this model was the ankle dorsiflexion stiffness. As the motion modeled in this analysis was a child with CP walking with crouch gait, it is reasonable that the dorsiflexion stiffness would be critical for the analysis. A crouch gait walking pattern is often associated with increased dorsiflexion during stance, and weak ankle plantarflexors has been cited as a potential cause of crouch gait (Steele et al., 2012b). In the AFO model, increasing the ankle dorsiflexion stiffness provided a larger passive torque to prevent dorsiflexion (plantarflexion torque), leading to a reduced muscle force required from the plantarflexors (gastrocnemius and soleus) to reproduce the same motion. In the probabilistic analysis, a negative correlation existed between the ankle dorsiflexion stiffness and plantarflexion muscle force, indicating increased stiffness would reduce the required muscle force, which is consistent with the purpose of a rigid AFO used to augment the function of the plantarflexors during stance. The muscle force results were similar to the results for muscle contributions to propulsion and support. It is important to note that although in the probabilistic model changes in AFO stiffness had large impacts on muscle force production, this does not necessarily indicate that children would have the same response if their personal AFOs were exchanged for different stiffness values. This analysis used the same

experimentally collected gait motion for all perturbations. By altering AFO parameters in the clinic; a child would change their gait pattern.

In addition, the analysis of the model using a Monte Carlo simulation indicated that subtalar damping was important during the first double support and swing phases of gait. This result may be due to these phases of gait surrounding the dynamic heel strike event. Interestingly, the muscles most affected by subtalar damping are the proximal muscles that control the action of the hip. This result suggests a relationship between the action of the subtalar and the hip to control internal/external rotation and foot placement, with potential implications for stability.

Finally, the equilibrium angles of the AFO model were highlighted in the Monte Carlo simulation as well as the AMV analysis for impacts on muscle force production. In theory, this variable shifts the passive torque curve horizontally so that only more extreme dorsiflexion angles would initiate the resistive torque. The ankle equilibrium angle had a positive correlation with the plantarflexors and quadriceps, meaning that as the equilibrium angle moves into dorsiflexion, the force must increase in the plantarflexor muscles.

Based on the results from this analysis, an estimation of at least the ankle dorsiflexion stiffness for a child walking with a crouch gait is important to build an accurate muscle-driven simulation. In addition, an analysis of the subtalar damping and stiffness equilibrium values could also improve the accuracy of a simulation. If these metrics must be estimated because the mechanical properties cannot be measured in clinic, the conclusions drawn from the results need to be critically evaluated for the potential effects of misrepresenting the AFO mechanical properties. Methods for measuring AFO parameters have been previously cited; for example, the BRUCE device designed to measure AFO stiffness values for individual child AFOs (Bregman et al., 2009).

The results of this analysis suggest important parameters for testing when designing AFOs for an individual child. Although this analysis was specific for a posterior leaf spring AFO, a similar analysis could be used to evaluate parameter estimation for other types of AFO's and different abnormal gait patterns. Because of the significant ankle plantarflexion force production and function sensitivity to the

dorsiflexion stiffness, subtalar damping, and neutral torque angle of the model, design considerations for these specific parameters may help customize AFOs for each child and could be customized to maximize the performance of the device.

Some of the limitations of this study included the simplified model used to approximate an AFO. The model of the AFO used in this analysis did not account for hysteresis and viscoelastic effects that have been seen in the behavior of AFOs (Crabtree and Higginson, 2009). In addition, assumptions regarding the distribution of the input parameters will have an effect on the results of the sensitivity analysis. With few previous studies evaluating the stiffness and damping effects of AFOs (Bregman et al., 2009; Crabtree and Higginson, 2009; Sumiya et al., 1996; Yamamoto et al., 1993) the approximations of the input distributions were limited. A single gait cycle for a single child with CP was simulated. Modeling additional children with different gait patterns commonly associated with CP, would assist in developing a broad understanding of the overall effect of AFO modeling for the larger CP population.

Although this analysis was specified for a child with posterior leaf AFOs, the range of tested AFO parameter indicates a more broad use of these results for a range of AFO stiffness values. Future work will include evaluating these limitations to develop a more diverse group of simulations that can be tested to ensure the analysis of the AFO model is robust for a large set of children with CP. In addition, a more detailed AFO model could be designed to account for non-linear stiffness and damping characteristics, observed most significantly at the end ranges of motion. Finally, and most importantly, this model should be implemented into a modeling and simulation study to investigate what possible insights regarding gait in children with CP can be gained by increasing the patient base from strictly unassisted ambulation to children who can ambulate with AFOs.

5.4 Conclusion

In conclusion, the modeling of AFOs is important for accurate analyses and interpretation of musculoskeletal simulations in children with CP; however, special care is required when estimating the parameters of the AFO, especially in ankle dorsiflexion stiffness. Caution is required when making

conclusions based on the muscle forces produced at the ankle in these models if the mechanical properties are estimated without experimental validation.

CHAPTER 6

SUMMARY AND CONCLUSIONS

The purpose of this work was to characterize the effects of gait and strength training on dynamic muscle force production and muscle contributions to body center of mass acceleration in children with cerebral palsy using a musculoskeletal modeling approach. Studies on gait and strength training were conducted to evaluate these effects on seven and six children respectively. Gait training may effectively alter the soleus contributions to propulsion, and the vasti contributions to support, developing gait patterns more closely matched with able-bodied function. Reduced stride-to-stride variability in muscle contributions to body center of mass propulsion and braking during single limb stance correlated closely to increased distance traveled during the 6-minute walk test. Results of both training types were quite variable with a range of effects, different for each child in the study, which may have limited the number of group level significant changes after training. An individualized analysis for each child revealed improved function of many muscle groups that were not consistent across all children. Despite these variable results, specific muscles indicated for targeted strength training included the hip extensors, hip abductors, and ankle plantarflexors due to pronounced asymmetry, or asymmetric changes in force production or function as a result of gait training, significant for all children in the study.

Similarly, changes in muscle force production and contributions to body COM accelerations were characterized after strength training. Changes in the gastrocnemius contributions to support produced accelerations more closely matched with able-bodied gait. No other overall changes were evident after strength training; however, individualized patient analysis similar to that performed for gait training revealed different strategies for body support and propulsion after training. Changes in muscle force and function after training were not consistent across all children in this analysis. Inconsistent results between dynamic muscle force production, estimated from the gait simulations, and isometric strength measurements suggest that musculoskeletal modeling and simulation provides important information about muscle mechanics that is not evident with only joint level isometric strength testing.

Physical therapy results in changes in muscular force and function, however these results are revealed more clearly by evaluating each child independently. Musculoskeletal modeling and simulation analysis can help greatly in identifying the muscular deficits for individual children, and thus, may be instrumental in developing targeted physical therapy protocols that are tailored to a specific patient.

LIST OF SYMBOLS

Cerebral Palsy	CP
Center for Disease Control	CDC
Gross Motor Function Classification System	GMFCS
Gross Motor Function Measure	GMFM
Electromyography	EMG
Single event multilevel surgeries	SEMLS
Partial body weight support treadmill training	PBWSTT
Proportional-derivative control	PD
Ankle foot orthoses	AFO
Center of mass	COM
Ground reaction force	GRF
Gait cycle	GC
First double support phase of gait	DS1
Single limb stance phase of gait	SS
Second double support phase of gait	DS2
Swing phase of gait	Swing
Anterior or Posterior anatomical direction	AP
Superior or inferior anatomical direction	SI
Medial or lateral anatomical direction	ML
Body weight	BW
Advanced Mean Value method	AMV

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APPENDIX A

CLINICAL ASSESSMENT RESULTS FOR PHYSICAL THERAPY

Table A-1 Average (SD) walking speed determined by gait analysis for each child before and after training measured in m/s. The model statistic was used to determine if walking speed significantly changed after training. Model statistic values are shown and significance is indicated (*).

	Gait Training			Strength Training		
	Pre Training	Post Training	Model statistic	Pre Training	Post Training	Model statistic
Child 1	0.902 (0.105)	0.987 (0.088)	0.085 < 0.160	1.106 (0.023)	1.071 (0.023)	0.035 < 0.037
Child 2	1.071 (0.084)	1.144 (0.073)	0.073 < 0.129	1.428 (0.020)	1.448 (0.010)	0.019 < 0.026
Child 3	1.311 (0.013)	1.360 (0.024)	0.049 > 0.036 *	0.959 (0.002)	1.003 (0.040)	0.045 < 0.047
Child 4	1.012 (0.053)	1.101 (0.031)	0.090 > 0.071 *	1.087 (0.017)	1.206 (0.051)	0.119 > 0.063 *
Child 5	1.319 (0.037)	1.263 (0.062)	0.056 < 0.084	1.253 (0.032)	1.241 (0.029)	0.012 < 0.051
Child 6	1.209 (0.044)	1.365 (0.060)	0.156 > 0.087 *	0.986 (0.025)	1.082 (0.031)	0.097 > 0.046 *
Child 7	1.077 (0.044)	1.070 (0.039)	0.007 < 0.068	-	-	-

Table A-2 Time (s) to complete the 10 meter walk test for both before and after training.

	Gait Training		Strength Training	
	Pre Training	Post Training	Pre Training	Post Training
Child 1	5.40	5.41	6.45	6.29
Child 2	6.76	6.86	5.18	5.18
Child 3	5.82	5.43	8.92	8.58
Child 4	8.23	6.13	6.42	6.49
Child 5	6.41	6.53	7.42	5.92
Child 6	6.13	5.95	9.14	8.32
Child 7	9.37	8.48	-	-

Table A-3 Distances traveled (m) during the 6-minute walk test for before and after training.

	Gait Training		Strength Training	
	Pre Training	Post Training	Pre Training	Post Training
Child 1	363.3	524.6	453.1	477.3
Child 2	397.2	512.1	542.9	573.7
Child 3	491.3	494.5	222.0	246.1
Child 4	318.7	404.4	437.7	476.0
Child 5	477.0	449.9	406.5	423.4
Child 6	466.5	483.9	315.0	324.8
Child 7	310.2	301.1	-	-

Table A-4 Isometric joint strength (lbs) measured using a handheld dynamometer for each child before and after gait training.

			Pre Gait Training		Post Gait Training	
			Left	Right	Left	Right
Child 1	knee	extension	35	47	42	54
		flexion	20	13	19	19
	hip	extension	42	44	55	54
		flexion	4	6	9	11
		abduction	25	26	22	19
	ankle	dorsiflexion	21	23	19	19
plantarflexion		52	28	65	49	
Child 2	knee	extension	29	28	35	34
		flexion	22	19	24	23
	hip	extension	41	43	61	63
		flexion	12	12	15	12
		abduction	31	25	24	23
	ankle	dorsiflexion	15	17	14	15
plantarflexion		47	65	54	60	
Child 3	knee	extension	62	69	52	90
		flexion	11	37	10	33
	hip	extension	63	96	78	109
		flexion	10	9	11	11
		abduction	26	29	22	39
	ankle	dorsiflexion	5	28	7	30
plantarflexion		38	96	59	102	
Child 4	knee	extension	45	37	49	33
		flexion	38	32	18	16
	hip	extension	60	64	48	44
		flexion	13	14	13	16
		abduction	34	32	15	32
	ankle	dorsiflexion	19	18	12	11
plantarflexion		94	89	56	33	
Child 5	knee	extension	33	34	29	28
		flexion	7	8	6	16
	hip	extension	49	47	50	49
		flexion	3	2	2	0
		abduction	13	19	20	18
	ankle	dorsiflexion	0	5	0	6
plantarflexion		22	23	22	30	
Child 6	knee	extension	44	44	49	47
		flexion	10.5	8	7	6
	hip	extension	65	62	68	78
		flexion	3	3.5	5	4
		abduction	15.5	16	13	14
	ankle	dorsiflexion	6	5	13	10
plantarflexion		28	27.5	28	29	
Child 7	knee	extension	7	10	7	11
		flexion	3	6	4	9
	hip	extension	17	17	28	27
		flexion	2	2	4	5
		abduction	7	7	7	7
	ankle	dorsiflexion	0	8	0	8
plantarflexion		11	16	13	24	

Table A-5 Isometric joint strength (lbs) measured using a handheld dynamometer for each child before and after strength training.

			Pre Strength Training		Post Strength Training	
			Left	Right	Left	Right
Child 1	knee	extension	37	35	43	42
		flexion	19	21	26	28
	hip	extension	55	58	63	71
		flexion	10	8	15	16
		abduction	20	20	31	30
	ankle	dorsiflexion	18	22	18	23
plantarflexion		52	58	55	78	
Child 2	knee	extension	53	56	71	73
		flexion	8	23	8	31
	hip	extension	62	101	82	110
		flexion	11	7	13	11
		abduction	22	34	29	33
	ankle	dorsiflexion	6	28	7	37
plantarflexion		64	107	62	102	
Child 3	knee	extension	18	16	48	46
		flexion	61	61	43	43
	hip	extension	50	51	62	65
		flexion	13	13	20	21
		abduction	23	29	38	37
	ankle	dorsiflexion	11	9	41	42
plantarflexion		52	64	96	93	
Child 4	knee	extension	24	24	51	48
		flexion	4	3	11	11
	hip	extension	44	40	68	75
		flexion	0	2	6	7
		abduction	16	14	26	19
	ankle	dorsiflexion	5	0	14	9
plantarflexion		20	21	30	28	
Child 5	knee	extension	40	33	35	41
		flexion	6	7	5	8
	hip	extension	49	51	31	46
		flexion	2	0	7	8
		abduction	14	17	11	17
	ankle	dorsiflexion	7	6	7	9
plantarflexion		29	27	17	23	
Child 6	knee	extension	6	11	13	19
		flexion	2	5	3	12
	hip	extension	26	36	40	45
		flexion	2	4	5	8
		abduction	7	8	8	9
	ankle	dorsiflexion	0	6	0	10
plantarflexion		7	12	13	26	