

TRUNK AND HIP MUSCLE ACTIVITY PATTERNS
IN EARLY WALKERS WITH AND WITHOUT
CEREBRAL PALSY

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ABSTRACT

Trunk and Hip Muscle Activity Patterns in Early Walkers With and Without Cerebral Palsy

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Poor control of postural muscles is a primary impairment in cerebral palsy (CP), however little is known about the activity of postural muscles during walking in individuals with CP. The objective of this study was to investigate differences in trunk and hip muscle activation patterns during the early stages of walking in children with CP compared to children with typical development (TD). Thirty-one children (16 TD, 15 CP) with an average of 28.5 months of walking experience participated. Electromyographic (EMG) data were collected from 16 trunk and hip muscles as participants walked at a self-selected pace over an instrumented walkway. Custom-written computer programs were used to identify the onset of muscle activity, and to generate instantaneous mean frequency (IMNF) curves. A functional principal component analysis was performed to determine differences in IMNF curves between groups. Linear regression analyses were performed to investigate relationships between gait parameters, muscle activation, and musculoskeletal measures. Group means were significantly lower in the CP group than the TD group for all spatiotemporal gait parameters measured. The CP group had greater percent activation and coactivation for all muscles except the external oblique. Greater hip adductor spasticity was related to increased abdominal muscle activity in the CP group. The CP group also had higher mean frequency throughout the gait cycle for all muscles. Higher IMNF

can result from increased rates of motor unit activation, increased number of recruited motor units, or decreased synchrony of motor units, and may contribute to muscle fatigue in children with CP. Within the CP group, children classified as Gross Motor Function Classification System (GMFCS) level II demonstrated no differences in spatiotemporal parameters or percent muscle activation, but had greater gait symmetry and lower INMF for the trunk muscles, compared to the children classified as GMFCS level III. The potential influence of recording activity from adjacent trunk muscles is discussed, as well as the influence of the use of an assistive device by some children with CP. Postural muscle training during the early stages of walking in CP should be investigated to encourage the development of more functional and efficient movement strategies in these children.

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TABLE OF CONTENTS

ABSTRACT	ii
ACKNOWLEDGEMENTS	iv
LIST OF TABLES	x
LIST OF FIGURES	xi
CHAPTER	
1. INTRODUCTION	1
Cerebral Palsy	1
The EMG Signal	10
EMG Analysis	13
EMG and Children	16
EMG and CP	19
Analysis of Current Literature	24
Proposed Study	33
Specific Aims	36
2. SPATIOTEMPORAL GAIT CHARACTERISTICS IN EARLY WALKERS WITH CEREBRAL PALSY	39
Introduction	39
Methods	41
Participants	41
Procedures	43
Musculoskeletal Measurement	43

Walking Trials	44
Data Analysis	45
Results.....	46
Discussion.....	51
3. MUSCLE ACTIVATION PATTERNS FOR THE TRUNK AND HIP MUSCLES IN EARLY WALKERS WITH AND WITHOUT CEREBRAL PALSY.....	56
Introduction.....	56
Methods.....	57
Participants.....	57
Procedures.....	58
Musculoskeletal Measurement.....	58
EMG Instrumentation	59
Walking Trials	62
Data Analysis	63
Results.....	64
Discussion.....	72
4. TRUNK AND HIP MUSCLE ACTIVITY IN EARLY WALKERS WITH AND WITHOUT CEREBRAL PALSY – A FREQUENCY ANALYSIS.....	77
Introduction.....	77
Methods.....	79
Participants.....	79
Procedures.....	80
EMG.....	80
Walking Trials	82

Data Analysis	83
Results	85
Discussion	86
5. GAIT AND MUSCLE FUNCTION IMPAIRMENTS IN EARLY WALKERS WITH CEREBRAL PALSY – RELATIONSHIP TO GMFCS	92
Introduction	92
Methods	93
Participants	93
Procedures	94
Musculoskeletal Measurement	94
EMG	96
Walking Trials	97
Data Analysis	97
Spatiotemporal Measures	98
Muscle Activity	98
Results	100
Spatiotemporal Measures	100
Muscle Activity	103
Discussion	105
6. DISCUSSION	108
Review of Aims	108
Summary of Results	111
Aim 1	112
Aim 2	114

Aim 3	118
Limitations	119
Future Research	121
REFERENCES	123
BIBLIOGRAPHY.....	136
APPENDIX	
A. USE OF THE TEAGER-KAISER ENERGY OPERATOR FOR MUSCLE ACTIVITY	
DETECTION IN CHILDREN.....	15050
Introduction.....	15050
Methods.....	1522
Subjects.....	1522
Data Acquisition	1544
Onset/Offset Detection.....	1544
Analysis.....	1577
Results.....	1588
Discussion.....	16060
References.....	1666

LIST OF TABLES

Table	Page
1-1. EMG During Walking in Typically Developing Children	25
1-2. EMG During Walking in Children with Spastic CP	25
2-1. Participant Inclusion and Exclusion Criteria	42
2-2. Demographic and Anthropometric Data	47
2-3. Musculoskeletal Measures and Significant Relationships Between Musculoskeletal and Spatiotemporal Measures for Individuals With CP	48
2-4. Group Means for Spatiotemporal Parameters, Stride-to-Stride Variability, and Symmetry	50
3-1. Electromyogram Sensor Locations	61
3-2. Demographic and Anthropometric Data	66
3-3. Musculoskeletal Measures and Significant Relationships Between Musculoskeletal Measures and Activation for Individuals With CP	67
4-1. Electromyogram Sensor Locations	81
4-2. Demographic and Anthropometric Data	86
5-1. Participant Inclusion and Exclusion Criteria	95
5-2. Demographic and Anthropometric Data	101
5-3. Participant Musculoskeletal Characteristics	101
5-4. Group Means for Spatiotemporal Parameters, Stride-to-Stride Variability, and Symmetry	102
A-1. Demographic Data	153
A-2. Bland-Altman Agreement	159

LIST OF FIGURES

Figure	Page
1-1. Conceptual framework	7
1-2. Proposed study measures as related to constructs of the ICF	35
3-1. Mean total percent activation of trunk and hip muscles	68
3-2. Histograms for number of children with trunk muscle activity at each point in gait cycle	69
3-3. Histograms for number of children with hip muscle activity at each point in gait cycle	70
3-4. Muscle activity across the gait cycle	71
4-1. The principal component output for the gluteus maximus	87
4-2. Instantaneous mean frequency mean curves	88
4-3. Stride-to-stride variability in instantaneous mean frequency	89
5-1. Mean total percent activation of muscles for children who are level II and III on the GMFCS	102
5-2. Instantaneous mean frequency mean curves for children who were GMFCS level II and III	104
A-1. Representative data for the gluteus medius for one child with typical development and one with cerebral palsy	161
A-2. Representative data for the rectus abdominis for one child with typical development and one with cerebral palsy	162
A-3. Bland-Altman plots for the TKE and SD methods	163

CHAPTER 1

INTRODUCTION

Cerebral Palsy

Cerebral palsy (CP) is the most common neuromuscular disorder in children, with an increasing incidence throughout the world. In the early 1990s, incidence was approximately 2.5 cases per 1000 live births.¹ In 2002, the incidence in the United States increased to 3.6 per 1000.² In northwest Europe, the incidence has increased, from approximately 1.5 per 1000 live births in the 1960s to approximately 2.5 per 1000 in the 1990s.³ Increasing incidence is thought to be related to greater survival of infants born preterm and with a low birthweight.³ The estimated net lifetime cost of each case is over \$900,000 and the estimated total lifetime costs for all persons born with CP in 2000 in the United States is \$11.5 billion dollars.⁴

The impact of CP on quality of life is substantial. Children and adolescents with CP report lower quality of life and well-being than their peers^{5, 6} and adults have reported lower than average self-esteem/self-worth.⁷ Due to the incidence, cost, and impact on quality of life, improving existing methods of prevention, assessment, and treatment for CP are important clinical research objectives.

The definition of cerebral palsy has recently been revised to follow the concepts of the World Health Organization's International Classification of Functioning, Disability, and Health.⁸ Cerebral palsy is "... a group of permanent disorders of the development of movement and posture, causing activity limitation, that are attributed to non-progressive disturbances that occurred in the developing fetal or infant brain. The motor disorders of cerebral palsy are often accompanied by disturbances of sensation, perception, cognition, communication, and behaviour,

by epilepsy, and by secondary musculoskeletal problems.”⁹ The impairments of posture and movement control, caused by the primary neural insult in CP, include poor coordination,¹⁰ altered muscle activation^{11, 12} and spasticity,¹³ which lead to muscle weakness,^{12, 14} altered muscle physiology,¹⁵ and decreased range of motion,¹⁶ all of which contribute to compensatory movement patterns that limit gait and functional abilities. While the neurological lesion is non-progressive, the musculoskeletal sequelae often increase over time and with growth.¹⁶

Mobility limitations in CP range from mild difficulty with running to complete dependence on others for care. The severity and limb distribution of CP vary greatly and are dependent on the extent and location of the underlying central nervous system (CNS) injury. Clinical examinations, case histories, and magnetic resonance imaging (MRI) brain scan findings were analyzed in a large cross-sectional study of European children.¹⁷ Of the 351 children who had had MRI scans (mean age at scan was 38 months), 43% had damage to white matter tracts, including periventricular leukomalacia and/or periventricular hemorrhage. This pattern of injury accounted for over 80% of the children born prior to 34 weeks gestation.

Eighty-six percent of children with only posterior or posterior/middle periventricular white matter injury had a clinical diagnosis of spastic diplegia. Thirty-four percent of those with extensive damage that extended from posterior to anterior white matter tracts had a diagnosis of spastic quadriplegia. Thirteen percent of the children had basal ganglia damage, and 76% of this group was clinically classified as having dystonic CP. Seven percent had focal infarcts, and over 95% of these children had spastic hemiplegia, consistent with the typical damage and clinical presentation resulting from cerebrovascular accident (CVA). In addition, 9% had a watershed pattern of damage (ischemic damage in peri-ventricular arterial distribution), 9% had brain malformations, 7% had miscellaneous patterns of damage, and 12% had normal scans. These

latter groups were not associated with a specific clinical classification, although more than half of the ataxic group had normal scans. Incidence of specific damage to the cerebellum was not reported.¹⁷

Staudt, Krageloh-Mann and colleagues^{18, 19} similarly correlated the findings of MRI studies to diagnostic classification in spastic CP, and looked more specifically at the relationship of location of white matter damage to functional deficits. On semi-coronal plane reconstruction of the corticospinal projections to the more involved upper limb in children with hemiparesis, the degree of lateral extent of the white matter lesion strongly correlated with upper limb functional deficit ($r=0.816$).¹⁸ Semi-coronal plane reconstruction of the posterior corticospinal projections to the lower limbs of these children revealed that lateral white matter lesion extent again strongly correlated with functional deficit ($r=0.805$).¹⁸ Using the same reconstruction methods for MRI scans of children with bilateral periventricular leukomalacia and spastic diplegia, significant positive correlations were found between motor dysfunction of each of the four extremities and the severity of damage to the corresponding portion of the contralateral pyramidal tract (right upper limb $r=0.77$, left upper limb $r=0.65$, right lower limb $r=0.92$, left lower limb $r=0.80$).¹⁹

Krageloh-Mann and others²⁰ correlated the MRI findings at 5.5-7 years of age in 29 children born preterm to cerebral blood flow measurements from their first two days of life. Of the 19 children with MRI abnormalities, 63% had low oxygen delivery to the brain (67% of these children had CP), compared to only 12.5% of the 10 children with normal MRIs (none of whom had CP), indicating ischemia as a factor for white matter damage. The timing of brain injury during gestation is also indicative of the damage and subsequent clinical presentation.²¹ Damage during the first and second trimesters predominately causes cortical neurodysgenesis, which results in maldevelopments seen on MRI, such as lissencephaly and schizencephaly. Early third

trimester damage results in lesions to the white matter, particularly in the periventricular region, such as leukomalacia or hemorrhage. This is the most common form of damage associated with spastic CP classifications.²² Damage in the late third trimester generally results in gray matter injury, such as basal ganglia, thalamus, or cortico-subcortical lesions.²¹

Relationships between degree of white matter abnormality and general movement quality have been identified as early as one month of age in infants who were born prior to 30 weeks gestation.²³ The associations between diagnostic imaging findings and clinical manifestations in these studies are important not only for clinical prognosis and the development of valid animal models of CP, but also to give researchers and clinicians an understanding of the physical basis of observed movement difficulties so that the most efficacious neurorehabilitation strategies can be selected.

Interventions to both prevent and treat CP are not well developed nor well supported by research, in large part due to the complexity and heterogeneity of the condition. The majority of medical and surgical interventions to treat CP target secondary musculoskeletal deformities (e.g. spasticity or muscle contracture) rather than the primary issue of poor neuromuscular control. Medications such as baclofen are prescribed to decrease spasticity. Surgical approaches are typically reactive rather than preventative. Soft tissue releases lengthen contracted muscles to increase range of motion at a joint. Lower extremity osteotomies reduce bony abnormalities that result from years of abnormal muscle and positional forces during development, but do not correct the originating abnormal muscle forces. In fact, while kinematics may improve, patterns of muscular activation rarely change after orthopedic procedures.^{24, 25}

Therapeutic interventions, while often targeting secondary musculoskeletal deformities through flexibility,²⁶ strengthening,²⁷ and endurance training,^{28, 29} are the only avenue currently

available to clinically address the primary issue of poor neuromuscular control in CP. Strategies to address impairments in individuals with CP began to be widely practiced and taught in the 1960s.³⁰ Historically, these interventions were aimed at inhibiting abnormal muscle activity patterns. The hallmark of the neurodevelopmental treatment approach developed by Berta and Karl Bobath³⁰ was to inhibit abnormal movement synergies followed by hands-on facilitation of “normal” movement synergies. A criticism of this treatment is the lack of clarity in how facilitation of posture and movement by a therapist promotes learning and encourages progression to an independent, volitional motor ability by the individual.³¹ More work can now be done to design treatment strategies in accordance with recent findings of cortical motor plasticity^{32, 33} and how specific practice paradigms promote learning and lasting improvements in posture and movement control.

Collaborative neurorehabilitation research between neuroscientists and rehabilitation clinicians has helped translate work from basic neuroscience investigation into clinical practice. Much of this work was discussed and synthesized at the III Step summer institute in 2005.^{34, 35} Two basic themes have emerged across various fields of research from several client populations and in different countries – intervention after central nervous system injury is most effective when delivered both early and intensely after injury.

Specific, repeated, and challenging motor practice can induce training-dependent changes in the cortical structures of adults with other neurological conditions,^{32, 33, 36-38} which may address the primary issue of poor motor control. Physiological changes that occur with neural plasticity include increased synapses per neuron, increased synaptic density in cortical layers, and increased dendritic spine density.³³ These findings compliment rehabilitation research that supports targeted, intense practice, amounting to 1-14 hours per day for 2-12 weeks, to promote

learning of skilled movement patterns in adults with neurological injury.^{39, 40} Both final outcome and speed of recovery were enhanced with earlier versus later intervention in an animal model of motor cortex injury.⁴¹ Large-scale, multi-site observational cohort and clinical trial evidence further supports these themes. The most effective rehabilitation programs for patients with neurological injury include both intensive and early intervention.⁴²⁻⁴⁶

Children with CP use compensatory movement strategies as a result of decreased postural control, muscle weakness, poor coordination and spasticity. If motor patterns are reinforced with repetition, these abnormal patterns are reinforced daily with each functional movement, over time and throughout development. Changing the well-established abnormal motor patterns, after years of reinforcement, through conservative treatment with these children is difficult, if at all possible.

Abnormal movements are present from infancy^{23, 47} and postural control at 12 months predicts motor performance at 7 years.⁴⁸ An aggressive developmental pruning process occurs in infancy, eliminating pathways that are not used and reinforcing those that are used.⁴⁹ This supports the importance of targeted intervention early in CP to reinforce more functional motor pathways before they become increasingly difficult to access, and to alter abnormal posture and movement patterns before the neural pathways that support them become too rigid from years of reinforcement. Physical therapists are in an optimal position to intervene in this process. If effective strategies are identified, there may be greater potential to improve gait in younger children, who have a less reinforced abnormal gait pattern, and prevent the gait deviations typically seen in children in CP.⁵⁰ See Figure 1-1 for conceptual framework.

Applying the principle of intensive intervention to individuals with CP may not result in similar outcomes as in other client populations. First, the work to date in adults with spinal cord

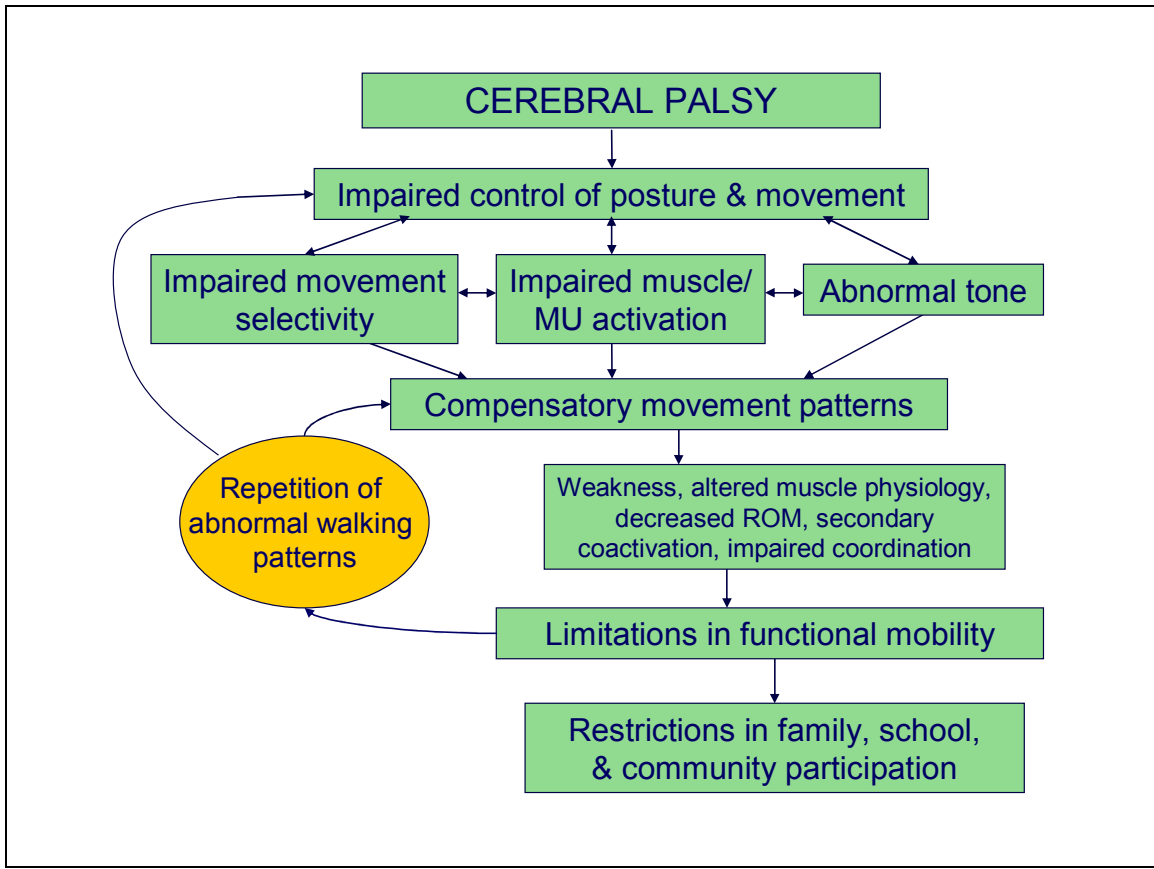


Figure 1-1. Conceptual framework. MU=motor unit, ROM=range of motion

injury and stroke involves participants who had a typically functioning nervous system until the time of injury in adulthood. Two factors are different when considering individuals with CP. To our knowledge, most have never functioned post-natally with a “typical” CNS.^{23, 47} We do not know if there is a difference in outcome between habilitation and rehabilitation in the application of neurorehabilitation treatment. We also do not understand the effects of a possible interaction between typical neuromaturational processes in children and the neuroplasticity that occurs after CNS damage. Second, if aggressive treatment is more beneficial the sooner it is provided after injury, interventions for CP would necessitate application during the first few years of life. Certainly, the age of the child affects the type of activities that are appropriate to address in therapy and also impacts the ability of the child to participate in aggressive rehabilitation programs. Early and aggressive treatment programs, however, can be feasible in infants and young children.⁵¹⁻⁵³

In order to evaluate the application of intense rehabilitation programs in these children effectively, we first need to understand more fully how they move at a young age during the development of the compensatory patterns of posture and movement. This study will be the first step in answering these questions by investigating performance during the early stages of walking. Only then will we be able to design appropriate interventions and measure their outcomes.

Advances in neuroimaging technology, such as functional MRI and transcranial magnetic stimulation, are allowing researchers to better understand how the brain and spinal cord work during cognitive and simple motor tasks. However, the neural templates used to analyze these complex data sets are based on typically functioning adults. We do not know if comparison with these templates is a valid procedure for analyzing locations of activity in children with brain

lesions. In addition, the scans must be collected while the individual maintains a primarily static posture (typically supine), restricting the amount of movement to a small excursion at one joint. With such small, isolated movement available during the procedure, there is no way to date to correlate the neural activity measured from these scans to the neural activity that occurs during everyday functional mobility, such as walking or transfers.

In fact, using evoked electromyographic responses in cats, Schneider et al⁵⁴ found that distal and proximal muscles were represented in several loci within the motor cortex, and the largest force responses of a muscle were often produced when stimulating an area of the cortex that was not believed to be the primary area for that muscle (i.e. some proximal muscles demonstrated strongest responses when stimulated in the lateral cortex, a distal forelimb region, and vice versa). Later work in primates⁵⁵ and humans^{56, 57} further supports the suggestion that muscles are mapped in numerous areas of a general cortical region, as part of functional synergies, rather than purely somatotopically.

With this in mind, we must exercise caution with the results of studies that report neural activation during motor tasks, because of a lack of evidence to suggest that neural activation of motor pathways is similar during recumbent, isolated, open-chain movements versus what occurs during upright, functional tasks. Thus, our best and most reliable information about neural activation during walking is collected through electromyography (EMG). While an indirect measure of neural activity, via the recording of the electrical signals of muscle contraction, EMG is portable and allows the acquisition of muscle activity information from several muscles at one time during numerous functional tasks. Furthermore, recent developments in the processing and analysis of EMG data have resulted in a greater understanding of the neural signals that produce a muscle contraction, such as the firing frequency of a particular motoneuron.^{58, 59}

As physical therapists, we have a responsibility to our patients and to our profession to make clinical decisions using the best scientific evidence available. We also have a responsibility to investigate the theoretical foundations of our practice in order to increase the volume and quality of evidence available in physical therapy science. Our understanding of the mechanisms of CNS dysfunction and plasticity after injury and intervention is in its infancy and this is an area in need of physical therapy investigators.

Studying EMG patterns during walking in young children with CP compared to experience-matched controls will give us information on how abnormal patterns of neuromuscular activation develop in CP. With this information, research programs that focus on understanding injury- and rehabilitation- induced plasticity in this population can be developed with greater measurement validity and subsequent confidence in the outcomes.

The EMG Signal

The electromyogram has been used for decades to detect the change in ionic potential that occurs during muscle contraction. Each motor unit action potential during a muscle contraction can be represented as a sinusoidal wave with distinct characteristics of amplitude, frequency, onset time, and offset time. The detected electrical signal is the temporal and spatial summation of all of the individual action potentials occurring in the range of the recording electrode during the recording session.⁶⁰

A number of intrinsic and extrinsic factors easily influence the EMG signal, some are the object of measurement, and others interfere with the signal and must be minimized. During a muscle contraction, the EMG signal will change with variations in number of active motor units in recording range, motor unit firing rate, muscle fiber size and type, and action potential

amplitude, duration, and shape. The signal is also joint position or muscle length dependent, movement velocity dependent, and is altered when a muscle is fatigued as a result of motor unit synchronization. Furthermore, the signal is affected by impedance (of skin, soft tissue, and the system), location and orientation of electrodes, contact pressure, and noise from nearby electronic equipment or motion artifact.⁶¹ This uncertainty of measurement led De Luca to call electromyography a “seductive muse.”⁶¹

The stochastic, oscillating trace of the raw EMG must be manipulated through several processing steps in order to yield useful information. Signal conditioning is performed by a preamplifier typically located near the recording surface of the EMG electrodes and an amplifier. The typical preamplifier is differentially configured in order to eliminate noise in the signal. The amplifier then increases the amplitude of the signal by a set gain in order to further reduce the influence of signal noise. Following signal conditioning, the EMG signal is further processed using software programs. The signal is initially filtered to remove high frequency noise (typically >400-500 Hz), then full wave rectified to convert oscillations in the negative direction to positive. The signal is smoothed with a low-pass filter (typically ≤ 5 Hz). The output at this stage is the linear envelope curve. Frequently, the data are processed a step further with either the root mean square or moving average method, which are mathematical steps believed to increase the reliability of the data.^{62, 63} Despite the chaotic nature of the EMG signal, with careful data collection practices and processing techniques, high repeatability in the timing of EMG signals over sessions 2-3 weeks apart has been established.^{64, 65}

Two different methods are used to collect the EMG signal, the placement of electrodes on the surface of the skin overlying the target muscle or the insertion of fine wires into the muscle belly. There are advantages and disadvantages of both methods. Surface electrodes are non-

invasive and have the ability to detect activation of the majority of the muscle, but are more likely to record activity from muscles adjacent to the target muscle. Their use is limited to muscles that have enough superficial surface area for the placement of the electrodes. Alternatively, fine wire electrodes are less likely to detect extraneous activity, but only make contact with muscle fibers from a single or a small number of motor units. As a result, the recorded signal may not be representative of the activity of the entire muscle. Additionally, the fine wires are able to access deep muscle tissue, but are considered invasive, and therefore their use in children for research purposes alone poses a ethical and practical dilemmas.⁶²

One important consideration is the relationship of EMG amplitude to force or torque generation. It is easy to reason that the measure of summation of motor unit action potentials would correlate linearly with the force output of the muscle, but this has been a topic of much debate. Tate and Damiano⁶⁶ tested this long-standing assumption in the quadriceps and hamstrings of children without neuromuscular pathology and children with CP during isometric (20%, 40%, 60%, and 80% maximum voluntary contraction, MVC) and isotonic (with resistance of 20%, 40%, and 60% of MVC) conditions, at several joint angles (20°, 45°, 65° knee flexion). Linearity, expressed as mean r^2 values, was strong (>0.90) for all isometric contractions in all conditions, with the CP group demonstrating slightly less linearity. Linearity was also >0.90 for all isotonic conditions in the control group, but decreased to mean r^2 values of less than 0.90 for the hamstrings and 0.80 for the quadriceps in the CP group. The authors concluded that the assumption of linearity may be valid for normal muscle and for isometric conditions in CP, but does not extend to dynamic contractions in individuals with impaired neurological function.

EMG Analysis

The majority of early analysis of EMG data was focused on measuring the on-off timing and amplitude of muscle activity,^{67, 68} often accomplished by visual examination of the raw trace. Mathematical approaches are now used to provide more standardization in measuring these characteristics. Onset and offset times are typically determined as a preset number of standard deviations above a quiet resting baseline.⁶⁹

Measuring amplitude is more difficult because anthropometrics, precise electrode placement, and body tissue impedance cannot be standardized from person to person, contributing to a great deal of variability in the EMG signal between individuals. For this reason, EMG amplitude, as measured in millivolts, cannot be compared across individuals. There are several methods of normalization, however that are used to compare data between persons.⁷⁰ One common method is to normalize each muscle recorded to the amplitude of the EMG during an isometric MVC for that muscle. Another approach is to use the maximum amplitude recorded during the functional activity that is being studied for normalization.^{62, 70} While the MVC method is often considered the “gold-standard”, this is not the best method in individuals with neurological involvement because amplitudes during dynamic movement have been shown to be higher than during MVC.⁷¹ Furthermore, eliciting a true MVC in children is often not feasible due to their inability to understand and follow the instructions.

Various efforts to develop methods that glean more meaningful information from the EMG signal have resulted in several more sophisticated approaches to analysis. Pattern recognition is a rapidly growing collaborative discipline that uses statistical analysis and decision rules to group complex data sets into similar patterns or measure how far a naïve pattern (e.g. new patient) is from known patterns (e.g. normative sample). Analyzing the frequency

component of the EMG signal using pattern recognition, in addition to the amplitude and timing components, has offered a great deal of useful information.

The frequency spectrum of an EMG signal contains information about the active motor units in the volume of muscle recorded by the electrode. Each motor unit in the recording area has its own unique action potential shape based on its spatial orientation with respect to the recording electrode. The action potential conduction velocity also influences the shape of the action potential detected by the electrode. Decomposing the EMG signal into its original sinusoidal components and generating a power density spectrum allows the determination of mean and median frequency measures, which are used to describe the detected motor unit activity. The Fast Fourier Transform method⁷² was first used to determine frequency characteristics of EMG signals, but is reliable only during static, isometric contractions.⁷² More recently, investigators have used continuous wavelet transform (CWT) because it is a reliable method for frequency analysis of dynamic EMG data.^{58, 59, 73}

Lauer and colleagues⁵⁸ retrospectively classified muscle activity patterns during walking in 21 children with CP (5 hemiplegia, 16 diplegia) and 16 typically developing (TD) children. Four muscles were analyzed bilaterally: vastus lateralis (VL), medial hamstrings (MH), medial gastrocnemius (MG), and tibialis anterior (TA). CWT produced a 3 dimensional scalogram (frequency x amplitude x time), which was then reduced to a time-frequency curve by instantaneous mean frequency calculations. “Muscle-muscle” plots were created to examine left and right symmetry and agonist-antagonist coactivation. These relationships were then quantified by principal component analysis and the subsequent harmonic output was used to calculate an EMG index for each muscle.

The indices were a measure of the deviation of muscle activation patterns relative to a comparison, typically developing group, calculated by squaring the four harmonic scores, summing them, and taking the square root of this value. This generated a “Euclidean” deviation from the control group, with larger values representing greater deviation. From these individual muscle EMG indices, a composite EMG index was calculated for each individual. The investigators were able to use the index to classify patterns with respect to the children’s functional status using the Gross Motor Function Classification System (GMFCS)⁷⁴ and correlate the patterns with respect to cadence ($r=-0.74$), velocity ($r=-0.62$), the Gillette Gait Index ($r=0.62$), and Gross Motor Function Measure (GMFM) Dimensions D ($r=-0.70$) and E ($r=-0.65$). This EMG index also has the potential to detect muscle activity changes following intervention. For example, muscle activation timing has not been observed to change following tendon lengthening, whereas significant shifts in signal frequency have been observed.⁷⁵

Another valuable measure, particularly for individuals with neurological involvement, is that of coactivation. A ratio of the muscle activity between an agonist and antagonist, measures of coactivation give information on the presence or lack of efficiency and coordination in movement patterns.⁷⁶ During functional movement, coactivation is often reported in terms of timing or amplitude. Timing coactivation ratios report the amount of time an antagonist is active during agonist contraction. Amplitude ratios are typically normalized either to the maximum amplitude recorded during a MVC or during the functional movement itself.

EMG analysis remains a critical element in the biomechanical analysis of gait in individuals with CP and its utility has improved in recent years. It is currently the most useful approach to indirectly study neuromuscular activation during functional movement. When collected carefully and processed and analyzed appropriately, researchers can make valuable

contributions to further characterize patterns of activation in different muscle groups, and to quantify the effect of various interventions. This type of instrumentation can be used to advance current practice by addressing the primary impairment in CP, i.e. deficits in control of posture and movement.

EMG and Children

Collecting EMG data from children increases the complexity of obtaining reliable information. Children characteristically have greater variability in spatiotemporal⁷⁷ and muscle activity patterns⁷⁸ than adults, and it is more difficult to maintain their attention and keep the electrodes and other instrumentation in place, all while minimizing influence on their gait pattern. Despite these challenges, the repeatability of sEMG in children was found to be comparable to adult values for between-day sessions (n=11, aged 6.5 ± 2.3 years)⁷⁸ when data from quadriceps, hamstrings, and ankle musculature were analyzed. The authors calculated variance ratios that represented the variance in waveform shape at each normalized time increment summed over the entire gait cycle divided by the total variance. Values closer to 0 indicate identically repeatable waveforms (no variability), and values closer to 1 indicate dissimilar waveforms (poor repeatability). The variance ratios ranged from 0.328 to 0.572 for within-day trials and from 0.343 to 0.657 for between-day trials (7 days apart). The rectus femoris (RF) was the most variable of the muscles tested and the MH demonstrated the greatest repeatability. There was no significant difference in within-day and between-day variance for the pediatric sample, whereas within-day variability was approximately twice that of adults, between-day values were similar to previously reported adult values.

Studies evaluating muscle patterns during in TD children are summarized in Table 1-1. These include studies accessible in a National Library of Medicine (Pubmed) online search reporting EMG data collected during independent, non-restricted, level walking in individuals under the age of 19 years with no known neuromuscular or orthopedic conditions. One study is a longitudinal case report that begins during the neonatal stepping phase and ends at the age of 7 years.⁷⁹ This study is included because it continued after independent walking was achieved.

Sutherland and colleagues performed a landmark study on the development of mature gait in a gait analysis laboratory.^{80, 81} They report the temporal-spatial gait characteristics, kinematics, kinetics, and muscle activation patterns of 186 children from ages 1 to 7 years (18 to 40 children at each age interval) in a cross-sectional design. When temporal-spatial gait characteristics were normalized for body height or leg length, referred to as dimensionless gait measurements, there was no difference between values at 4 years of age and adult values. Kinematic changes demonstrate a similar, perhaps even earlier maturation time, with the majority of joint motions resembling adult values by the age of 3. All data in the age 7 group were similar to adult values, consistent with later work describing gait maturation.⁸²

The lower extremity EMG profiles include data from the TA, vastus medialis (VM), gluteus medius (GMed), MH, lateral hamstrings (LH), gluteus maximus (GMax), and gastroc-soleus (TS) in 369 children. The primary differences in early walkers were the premature activity of the VM in early swing and TS in mid-swing, and the prolonged activity of the TA in late stance in children before the age of 2 years. After 2 years of age, muscle activation patterns in TD children were similar to those of adults.⁸⁰

Sutherland's work contradicts other work⁶⁸ reporting EMG activity from the VL, MH,, MG, and TA during gait in children from 1 to 7 years. Gait cycle events were determined from

four plantar footswitches which were covered with a sock. Some trends were identified, but no statistically significant differences in amplitude were observed between the 7 age groups. Lack of statistical significance may be a result of several factors. EMG amplitude was normalized to individual muscle activity, systematically reducing inter-subject variations in amplitude. Furthermore, footswitches are not the most optimal for detecting gait cycle events in this population because early walkers do not demonstrate the same pattern of foot contact as more mature walkers. An offset in the point of heelstrike identification caused by lack of precision of the footswitches could have shifted the entire muscle pattern curve, impacting the relative amplitude at each time point analyzed.⁸⁰ Low statistical power could also explain the statistical results, as trends were identified, but only 10 subjects were included in each age group.

A recent study in an older group of children and adolescents (3-18 years) reports no relationship between age and muscle activity onset time, cessation time, or duration for the majority of muscles.⁸³ VL median activation time and GMax and gastrocnemius cessation time were weakly but significantly correlated with age, height, velocity and stride length ($-0.24 < r < -0.40$), with excessive activity in the younger group in each case. Two possibilities explain these results. Age differences in EMG activity may only exist in very early walkers (age 1-2 years, as Sutherland found), and the youngest child in this study was 3 years old. Alternately, any differences that may exist in 3-4 year old children compared to older children were likely missed because several ages were grouped together for analysis, such that data from 3-6 year old children were pooled and compared to 7-11 year and 12-18 year old groups.

Age again had no relationship to muscle activity patterns when recorded in the ankles of children over the age of 4 years.⁸⁴ In determining the influence of walking speeds ranging between 1.5-5.5 km/h, the TS and peroneus longus (PL) were found to have earlier onset in

stance as speed increased. At speeds greater than 4.0 km/h, the TS and PL activity began in terminal swing. Onset and offset times of the TA were not dependent on speed. This study is consistent with the others^{80, 83} in finding no relationship between muscle timing and age. The results related to walking speed are important when collecting data in this age group, as young children tend to “run” down the recording walkway during gait analysis when, after waiting patiently and fully instrumented, they are given the “okay” to proceed to the opposite end.

The final study is a longitudinal case report of the progression of one child from neonatal stepping at 3 weeks of age to mature walking at 7 years.⁷⁹ Patterns of activation and no activation above baseline in relation to foot contact and stance and swing phases (determined by video analysis) are reported for the GMax, RF, LH, VM, lateral gastrocnemius (LG), and TA. Eight phases in the development of walking were identified and grouped into 3 stages: neonatal stepping, supported infant walking, and independent walking. The beginning of each stage was characterized by excessive activation and co-activation followed by a progression to more reciprocal and efficient (less activity) activation patterns. This pattern of early excessive muscle activity may coincide with, and contribute to, the kinematic reduction, or “freezing,” of degrees of freedom that is known to accompany early skill acquisition.⁸⁵ As a whole, this study and those previously discussed provide some normative values of duration of activity and onset and cessation times, for comparison to pathological gait patterns in children of similar age for gluteal, thigh, and leg muscles.

EMG and CP

EMG has been used extensively in children with CP to characterize abnormal timing of muscle activation and quantify abnormal coactivation of antagonist muscles. This section will

focus on the findings of investigations of muscle activity patterns during walking in individuals with spastic diplegic and quadriplegic CP. Details of the studies are provided in Table 1-2.

The first large-scale, controlled study described EMG data from 14 lower extremity muscles in 113 individuals (1.7-31.1 years) with diplegia and 45 able-bodied individuals (7.9-35.6 years).⁷¹ The average EMG amplitude of each muscle during MVC and during walking was used to calculate a cocontraction ratio. With the exception of ankle muscles, children with CP demonstrated smaller amplitudes and greater cocontraction during MVCs, with greater amplitudes and less cocontraction during walking. The authors determined that it was possible to classify the 2 groups based on EMG pattern alone. While this study offered extensive information at the time it was conducted, we now understand that differences in non-normalized amplitude can occur for many reasons, one being the type of muscle contraction. Amplitude data are now normalized before comparing values between isometric and dynamic contractions. For example, Damiano and colleagues¹⁰ scaled EMG amplitude to MVC maximum, and then weighted the values based on the amount of individual muscle weakness relative to age-matched normative values. Incidentally, they did not find a greater degree of cocontraction during isometric contractions than during gait in their study of 10 children with CP.

EMG findings are an important part of clinical decision making for children with CP, and are often considered when surgical or orthotic interventions are being considered. Ankle muscle activity in school-aged children with diplegia and equinovarus deformity was studied prior to surgical intervention.²⁴ All participants demonstrated premature TS and posterior tibialis (PT) onset prior to heelstrike and reduced TA duration in the swing phase. Three additional patterns of MH activity were identified (see Table 1-2). Some children with prolonged PT activity

underwent successful transfer of this tendon to the dorsal surface of the foot, where it assisted the under-active TA with dorsiflexion.

EMG has also been used in combination with kinematic data for surgical decision making. Sixteen children with diplegia and crouch gait provided data for 3-dimensional motion analysis and a computer model was used to calculate the dynamic length of the iliopsoas and MH from joint position and the known anatomic points of muscle attachment.⁸⁶ Eighteen of the 32 hamstrings studied were identified as functional hip extensors during stance because they demonstrated prolonged and concentric (shortening) activity during stance. The authors caution against the surgical lengthening of hamstrings that are active hip extensors, as lengthening may increase anterior pelvic tilt and hip flexion contracture, further contributing to crouch gait.

Differential diagnosis of mild spastic diplegia and idiopathic toe walkers has also been possible using EMG, but not consistently with kinematic or kinetic information.^{87, 88} The degree of LG coactivation during seated resisted knee extension, supine isometric quad sets, and walking was measured in children with both conditions.⁸⁷ There was no difference in LG activity between the two groups during gait; both demonstrated premature activity in swing compared to the control group. With quadriceps activation during resisted knee extension in sitting and quad sets in supine, control subjects demonstrated LG coactivation of 0.4% and 3% for the two tests, respectively. Idiopathic toe walkers demonstrated coactivations of 20% and 35%, respectively. Children with mild CP demonstrated obligatory plantarflexor coactivation during quadriceps activity during 86% of the test duration for both types of contractions. Even variables relating to birth history, such as weeks gestation at birth and length of hospital stay after birth were not different in these two groups, suggesting that this measure could be a clinically useful distinguishing factor. In other work, 7 children with CP who were 5-17 years of age were each

examined by 6 clinicians, with and without surface EMG. The use of surface EMG resulted in an increase in inter-rater agreement as well as an increase in the self-reported confidence of the clinicians in their use of the instrumentation.⁸⁸

In addition to its use in clinical decision making and diagnosis, EMG has been shown to have strong relationships with measures of function. The composite EMG index developed by Lauer and colleagues described in the EMG analysis section (B) above had moderate and strong correlations with spatio-temporal gait characteristics and common measures of gross motor function (see Table 1-2).⁵⁸ The relationship between energy expenditure and EMG was investigated by Unnithan and colleagues at two walking speeds in 9 children with CP who had a mean age of 12.7 years.⁷⁶ At a walking speed of 3 km/hr, the thigh cocontraction index (VL/hamstrings) explained 51.4% of the variance in measured oxygen uptake and the lower leg cocontraction index (TA/soleus) explained 42.8% of this variance. Cocontraction index was not related to energy expenditure in the control group. These findings are supported by Damiano et al¹⁰ who reported no correlation between cocontraction and muscle strength, but negative correlations between total cocontraction magnitude and energy expenditure during self-selected and fast gait speeds ($r = -0.71$ and -0.90 , respectively).

Finally, a methodological consideration was also addressed by Unnithan and others.⁸⁹ Similar to Detrembleur's study⁸⁴ of ankle muscles in TD children, the authors report that EMG duration of the VL, hamstrings, TA, and soleus is influenced by walking speed in children with CP. Cocontraction values increased significantly in CP and TD groups from a walking speed of 3 km/hr to a speed of 90% of the participants' fastest walking velocity.

This body of literature reveals that muscle activation patterns have been well studied in a select group of muscles in a relatively homogenous group of children with CP. Participants were

primarily those with spastic diplegia and between the ages of 5 and 15 years, who were ambulatory and classified at GMFCS levels from I to III. The overwhelming majority of studies have limited their investigation of muscle activation patterns during walking to the quadriceps, hamstrings, ankle plantarflexors and ankle dorsiflexors.

Information on the activation patterns of muscles from the time of onset of walking ability in children with CP may give physical therapy clinicians and researchers a greater understanding of how abnormal muscle activation develops to result in the gait deviations observed in children with CP. Additionally, information on how trunk and gluteal muscles function during walking will assist our understanding of their role in controlling the body's center of mass and maintaining upright posture during walking, and could lead to the development of interventions to address deficits in their function in early walkers with CP.

In general, there is remarkable consistency among the findings of the studies included in this review. Typically developing children under the age of 2 or 3 years demonstrate more coactivation of antagonistic muscle pairs than older children. The pattern of muscle activity approximates that of an adult rather quickly after the complex motor skill of walking is learned, with children demonstrating adult patterns by age 3 or 4 years. Children with CP demonstrate prolonged muscle activity during walking, typically with earlier onset and later cessation times than TD controls of the same age. One exception to this generalization is the TA, which demonstrates reduced activation time during the swing phase, consistent with a corresponding kinematic pattern of increased plantarflexion. Increased activation duration results in a greater degree of coactivation of antagonist pairs, which is inversely related to function and efficiency.

The consistency in findings is most likely a result of the homogeneity of the participant samples and muscles selected for examination. While the replication of findings by different

investigators with a similar participant sample increases the validity of the evidence, lack of variability in participant samples across studies leads to gaps in our current knowledge.

Information on the pattern of muscle activation in children with CP from the onset of walking is not available. Additionally, the overwhelming majority of studies have limited their investigation to the quadriceps, hamstrings, ankle plantarflexors and ankle dorsiflexors.

Analysis of Current Literature

While abundant information exists for select muscles in select populations, there are particular age groups, diagnostic classifications, and muscle groups that have been virtually ignored in the study of gait in children with CP. Information on the pattern of muscle activation in children with CP from the onset of walking is not available. This information could assist in the prediction of future gait patterns and warrant earlier, more intensive intervention, before the abnormal patterns are reinforced, to prevent the need for later invasive procedures.

Of the 207 participants with CP included in the studies in Table 1-2, only one child had a diagnostic classification of spastic quadriplegia,^{76, 89} while the others had diplegia or hemiplegia. Biomechanical analysis of children with severely impaired gait, as is more likely in children with quadriplegia, increases the difficulties related to data collection. Limited endurance, cumbersome assistive devices, and the tendency for the child to lean on the assistive device can result in unusable trials and require additional time to collect repeat trials. However, with careful advance preparation, study of children with spastic quadriplegic CP is feasible. For example, giving seated rest breaks between trials reduces the possibility of fatigue. Using an instrumented walkway to gather spatio-temporal gait characteristics instead of full 3-D motion analysis will reduce encumbrance from the instrumentation, and eliminate some of the potential problems.

Table 1-1. EMG During Walking in Typically Developing Children

Study	Age of participants	Sample size (n)	Muscles (electrode type)	Analysis Method	EMG findings
Granata et al 2005 ⁷⁸	\bar{x} = 6.5±2.3 years	11	Bilateral RF, MH, TA, MG (surface)	EMG amplitude normalized by scaling to unit of total energy (area under curve). Gait cycle timing recorded from instrumented walkway.	Measured repeatability in normalized amplitude and timing for trials within same session and between sessions 7 days apart. Mean variance ratios ranged from 0.33-0.57 for within-session and 0.34-0.66 for between-session. RF demonstrated greatest variability, MH the greatest repeatability. Compared to adult values, within-session variability is approximately twice that of adults, but comparable for between-sessions.
Sutherland et al 1988 ⁸⁰	1, 1.5, 2, 2.5, 3, 3.5, 4, 5, 6, and 7 years	369 (24-44 in each age group)	GMax, GMed, VM, MH, LH, TA, TS (surface)	On and off timing as a percent of gait cycle, determined by footswitches that indicated foot strike and toe off.	TA – delayed onset during swing phase at 1 and 1.5 years, mature pattern by age 2. VM – early and prolonged activity at 1 and 1.5 years, mature pattern by age 4 y. GMed – Trend toward earlier onset during swing in ages 1-3.5 y, mature pattern by age 4 y. MH – prolonged stance phase activity at 1 year, mature pattern by age 2 y. LH – Slightly early and prolonged activity at 1 and 1.5 years, mature pattern by age 2 y. GMax – early onset during swing phase in 1, 1.5, and 2.5 y age groups, mature pattern by age 3 y. GS – Two patterns identified: 2/3 of those aged 1 and 1.5 y and ~1/4 of other age groups demonstrated “immature pattern” with onset in mid-swing. “Mature” pattern was onset after foot strike, resembling adult pattern.
Tata and Peat 1987 ⁶⁸	1,2,3,4,5,6, 7 years	70 (10 in each age group)	Right VL, MH, MG, TA (surface)	EMG amplitude normalized to minimum and maximum levels recorded. Gait cycle events determined from 4 plantar footswitches.	No differences between age groups in amplitude or timing of peak amplitude. Trend for prolonged activity in VL, MH, and TA in ages 1 and 2. 1 y age group demonstrated early and rapid onset of MG in early stance. Patterns of 4-7 y age groups resemble those of adults. Method of normalizing amplitude to individual max/min likely washed out amplitude differences. Study also potentially underpowered.

Table 1-1. (continued)

Study	Age of participants	Sample size (n)	Muscles (electrode type)	Analysis Method	EMG findings
Chang et al 2007 ⁸³	3-18 years	87 (ages 3-6=32, ages 7-11=29, ages 12-18=26)	GMax, GMed, Hip adductors, RF, VM, VL, MH, LH, TA, Gastrocnemius, PB (surface)	Amplitude normalized to maximum activity. Custom computer algorithm used for pattern recognition of peak activity periods. 3-D motion analysis for temporal and kinematic data.	13% of all curves were not functionally interpretable due to having no identifiable pattern (error or impedance). MH, LH, GMax, GMin, and Gastroc have the most consistent and clinically interpretable curves, VL and PB the least. Small correlations between age or anthropometrics and some muscles, but not most. VL median activation time, GMax and Gastroc cessation time accounted for majority of correlations with age, height, weight, velocity and stride length.
Detrembleur et al 1997 ⁸⁴	4-11 years	15	Right TS, TA, PL (surface)	Timing determined by visual analysis of linear envelope curves. Gait cycle determined by force platform and conductive soles worn by participant.	Determined influence of walking at speeds between 1.5-5.5 km/h on EMG. TS and PL were active during the stance phase at all speeds, with earlier onset as speed increases. At speeds > 4.0 km/h, TS and PL activity began in terminal swing. TA was active during swing and early stance independent of speed. Age had no effect on duration of ankle muscle activity.
Okamoto et al 2003 ⁷⁹	Longitudinal from 3 weeks to 7 years	1	Right GMax, RF, LH, VM, LG, TA (surface)	Used video analysis to determine patterns of activation above baseline in relation to foot contact and stance and swing phases.	8 phases in the development of walking identified and grouped into 3 stages: neonatal stepping, supported infant walking, and independent walking. The beginning of each stage was characterized by excessive activation and co-activation followed by a progression to more reciprocal and efficient patterns with less activity.

\bar{x} = Mean, y=years, GMax=Gluteus Maximus, GMed=Gluteus Medius, VM=Vastus Medialis, VL=Vastus Lateralis, MH=Medial Hamstrings, LH=Lateral Hamstrings, TA=Tibialis Anterior, MG= Medial gastrocnemius, TS=Triceps Surae, PL=Peroneus Longus, PB=Peroneus Brevis

Table 1-2. EMG During Walking in Children with Spastic Diplegic or Quadriplegic CP

Study	Age of participants	Sample size (n)	Diagnostic and GMFCS classifications	Muscles (electrode type)	Analysis Method	EMG findings
Gueth et al 1984 ⁷¹	CP: 1.7-31.1 years (\bar{x} =11.0) TD: 7.9-35.6 years (\bar{x} =19.8)	113 CP 45 TD	Diplegia No description of walking ability (possibly I-III)	Bilateral GMed, hip adductors, RF, MH, MG, TA, PL during walking and MVC (surface)	Timing and amplitude determined by visual analysis. No information on gait data collection.	Participants with CP had smaller amplitudes and greater cocontraction during MVC, and greater amplitude during walking in most muscles except the MG and TA.
Brunt & Scarborough 1988 ²⁴	\bar{x} = 7.7 ± 3.88 years	13 CP	Diplegia or hemiplegia, all with equinovarus (not further defined) No description of walking ability (possibly I-III)	MH, TS, TA (surface) PT (fine wire)	Timing determined by visual analysis. Gait events determined by custom foot switches.	All demonstrated premature TS and PT onset before heel strike and reduced TA duration in swing. 3 MH patterns identified- A: n=5, MH prolonged in stance. B: n=3, MH prolonged in stance with intermediate burst at toe-off. C: n=5, MH activity normal but TA prolonged in stance.
Damiano et al 2000 ¹⁰	5-14 years (\bar{x} =9.2)	10 CP	Diplegia (n=9) Hemiplegia (n=1) I (n=3), II (n=2), III (n=5)	Bilateral RF, BF during MVC and walking (surface)	Cocontraction ratio (CCR)= min/max amplitude. Cocontraction magnitude (CCM)= mean amplitude, after normalization to isometric max and to weakness relative to normal values. 3-D motion analysis used to collect gait data at free and fast walking speeds.	CCR was not related to strength of either muscle. CCR during MVC was moderately correlated (0.35-0.60) to CCR during free and fast walking. CCM during free and fast gait negatively correlated with energy expenditure (-0.71, -0.90).

Table 1-2. (continued)

Study	Age of participants	Sample size (n)	Diagnostic and GMFCS classifications	Muscles (electrode type)	Analysis Method	EMG findings
Unnithan et al 1996b ⁸⁹	TD: \bar{x} =13.6 \pm 2.1 years	9 CP 8 TD	Diplegia (n=7), Hemiplegia (n=1), Quadriplegia (n=1) *I-II (n=7), III-IV (n=2)	Right VL, hamstring, TA, soleus (surface)	Data collected for 5 seconds while walking on a treadmill and normalized to max amplitude, not analyzed in context of gait cycle. Cocontraction index (CI)= area of overlap under EMG curve to total area.	Data analyzed for two walking speeds, 3 km/hr and 90% of the fastest walking speed. CI values and activation time were higher in CP than TD for both speeds and all muscles. CI increased in both groups with increased speed.
Unnithan et al 1996a ⁷⁶	CP: \bar{x} =12.7 \pm 2.8 years TD: \bar{x} =13.6 \pm 2.1 years	9 CP 8 TD	Diplegia (n=7), Hemiplegia (n=1), Quadriplegia (n=1) *I-II (n=7), III-IV (n=2)	Right VL, hamstring, TA, soleus (surface)	Data collected for 5 seconds while walking on a treadmill and normalized to max amplitude, not analyzed in context of gait cycle. Cocontraction index (CI)= area of overlap under EMG curve to total area.	Data analyzed for two walking speeds, 3 km/hr and 90% of predetermined fast walking speed. Contribution of cocontraction to oxygen cost of walking determined. Thigh CI explained 51.4% (r=0.717) and lower leg CI 42.8% (r=0.663) of the variance in VO ₂ in the CP group. No relationship in TD group.
Hoffinger et al 1993 ⁸⁶	4-25 years (\bar{x} =10.5)	16 CP	Diplegia, all with crouch gait (not defined) No description of walking ability (possibly I-III)	Bilateral RF, MH, hip adductors, VL (surface)	Computer model calculated dynamic iliopsoas and MH muscle lengths. Kinematic data collected using 3-D motion analysis.	Assessed role of the MH in crouch gait by relating muscle activity to kinematic shortening or lengthening. 18 of the 32 hamstrings identified as functional hip extensors because they had prolonged and concentric activity in stance.

Table 1-2. (continued)

Study	Age of participants	Sample size (n)	Diagnostic and GMFCS classifications	Muscles (electrode type)	Analysis Method	EMG findings
Policy et al 2001 ⁸⁷	ITW: 3-10 years (\bar{x} =6.8±2.0) CP: 5-9 years (\bar{x} =6.7±1.2) TD: 3.5-9 years (\bar{x} =6.1±2.0)	8 CP 8 ITW 8 TD	Diplegia, mild *I-II	VL, LG (surface)	On and off times analyzed as percent gait cycle. Foot switches used to determine stance and swing phases.	Used EMG to diagnosis mild spastic diplegia and ITW. Coactivation of LG during resisted knee extension and quad set in CP was 86% and 86%, in ITW 20% and 35%, and in TD 0.4% and 3%.
Lauer et al 2007 ⁵⁸	CP: 6-20 years (\bar{x} =10.4) TD: 7-14 years (\bar{x} =10.8)	21 CP 16 TD	Diplegia (n=16), Hemiplegia (n=5) I (n=10), II (n=6), III (n=5)	Bilateral VL, MH, MG, TA (surface)	Used continuous wavelet transform to generate time-frequency curves. Calculated a composite EMG index using principal component analysis (PCA) weights. Gait data collected with 3-D motion analysis.	The EMG index is a measure of the degree of total muscle activation relative to normal (0). The index was able to classify patterns according to level of impairment, and correlated with cadence (-0.74), velocity(-0.62), GMFCS (0.65), Gillette gait index (0.62), and GMFMD (-0.70) and E (-0.65).
Wakeling et al 2007 ⁵⁹	CP: 4-21 years (\bar{x} =11.3) TD: 3-21 years (\bar{x} =10.8)	17 CP 36 TD	Diplegia *I-II (n=14) III (n=3)	Bilateral RF, MH, MG, TA (surface)	Used continuous wavelet transform to generate time-frequency curves. Analyzed using PCA and an EMG normalcy score generated. Gait data collected with 3-D motion analysis.	Mean frequency was higher in CP for all muscles and both phases of gait. PCA was more sensitive in distinguishing CP from TD for the MG-TA pairs than RF-MH. EMG normal scores were higher (less normal) in the CP group for both muscle pairs and all phases of gait (total, stance, swing).

*GMFCS levels not provided, but could be determined based on description of walking ability

\bar{x} = Mean, y=years, CP=cerebral palsy, TD=typically developing, ITW=idiopathic toe walking, GMFCS = Gross Motor Function Classification System, GMax=Gluteus Maximus, GMed=Gluteus Medius, VM=Vastus Medialis, VL=Vastus Lateralis, MH=Medial Hamstrings, LH=Lateral Hamstrings, TA=Tibialis Anterior, MG= Medial gastrocnemius, TS=Triceps Surae, PL=Peroneus Longus, PB=Peroneus Brevis, PT=Posterior Tibialis, MVC=maximal voluntary contraction, GMFM=Gross Motor Function Measure

Fifty-four percent of individuals under the age of 14 years with spastic quadriplegia are ambulatory⁹⁰ and their inclusion in research studies that lead to treatment interventions is important to prevent age and growth-related functional decline in these children.

Children with CP typically reach a plateau in gross motor function at 5 to 7 years of age.⁹¹ It is critical to investigate walking patterns prior to this plateau in order to develop effective interventions that can reduce the reinforcement of abnormal postural and movement patterns before the neural pathways that support them become too inflexible, and to encourage the development of functional motor pathways before they become too difficult to access.

It is also important that early walking in CP is compared prospectively to a control group who has similar amounts of walking experience, which may be a stronger influence on the development of walking and balance skill than age.^{92, 93} In a study of 210 infants aged 9-17 months with less than one to 8 months of walking experience, experience was more closely related to spatial measures of gait than age. Measures included step length ($r=0.43$ for age, $r=0.59$ for experience), step width (-0.42, -0.55, respectively), dynamic base (a measure of base of support combined with path deviation, 0.47, 0.62, respectively), and foot rotation angle (-0.29, -0.50 respectively).⁹²

The study of axial and proximal limb musculature is a surprising omission in the current literature because of the belief that poor postural control is a primary impairment in CP.⁹ Poor control of the head and trunk is thought to cause compensation by other muscles (particularly those of the shoulder and hips) to assist in maintaining upright posture, and limiting these muscles' ability to function as primary movers of the extremities.⁹⁴ Significant impairment in postural control likely contributes to functional deficits because children tend to be high functioning (GMFCS I) when only distal limb musculature is involved.^{58, 87} Additionally,

impairments in trunk motion in children with CP have been reported, which may result from or contribute to altered trunk muscle activation patterns.⁹⁵

Proximal muscles are important for supporting posture and upright functional tasks. The gluteus maximus, for example, is critical in maintaining an upright posture for standing and walking, and contributes to power generation for forward movement.⁹⁶ It is also more difficult to successfully correct for proximal muscle deficits through bracing or taping, than for distal muscle deficits, because of the larger size of the trunk/hip area and the need to accommodate multiple trunk and hip positions during functional mobility. More efficient use of proximal muscles through improved timing, coordination, and force generation may result in greater walking ability or decreased energy cost. Muscles that move the hips and trunk should be considered for future study, including the gluteus maximus, gluteus medius, erector spinae, latissimus dorsi, trapezius, rectus abdominus, and external abdominal oblique.

It is clear from the evidence that collecting reliable and consistent EMG signals is not easy, but certainly feasible. Care must be taken to assure proper size and placement of electrodes, reduce impedance, and arrange instrumentation in the least restrictive fashion. Reliable data acquisition will be more difficult with young children who are small and may not be as tolerant of instrumentation or understand test procedures. Some researchers use foot switches to collect gait cycle information to eliminate the need for bulky and distracting kinematic markers. However, altered foot contact patterns reduce the ability to consistently identify the point of phase transition and identify gait cycle events (e.g. foot contact, foot off) to synchronize the EMG data,⁹⁷ making this option less than ideal in the target population. This issue could be addressed by using an instrumented walkway rather than 3-dimensional motion analysis. This would reduce the number of instrumentation components on the child's body as well as the setup

time. Each walking trial could also be video-taped to relate EMG and timing information to findings from observational gait analysis.

After obtaining the raw data, careful decisions must be made on how to manage the signals. Simple on/off analyses are now a relatively basic method of analysis. We have much more sophisticated techniques^{58, 83} that give us more information about how the peripheral nerve is activating the muscle. Employing current mathematical and statistical methods of processing, normalizing, and analyzing the data, such as frequency analysis, will increase the validity of the findings and allow for comparison to the contemporary literature.

One final methodological consideration is the collection of the EMG signal of the gluteus medius in small children using surface electrodes. The three studies included in this review that report data from the gluteus medius in small children either provide no description of their placement or verification procedures,^{71, 80} or cite a reference for adult placement.⁸³ The authors of these studies are regarded as experts in biomechanical analysis of walking in children; however, this muscle has a small superficial area superior and lateral to the gluteus maximus. It is not clear that this area is large enough in small children to accommodate an EMG recording electrode without recording electrical activity from adjacent muscles, such as the gluteus maximus and tensor fascia lata.

A method to address this issue is to verify placement by an isolated isometric contraction of the gluteus medius after the electrode has been placed. There are two ways this can be done. First, in either supine or sidelying, with the hip in full extension (to place the gluteus maximus in a shortened position), the participant abducts the hip. Second, in standing with UE support (to aid balance and keep other muscles quiet), the opposite leg is lifted to activate the gluteus medius for pelvic stability in unilateral stance. Children with CP, however, may not be able to isolate the

gluteus medius in a volitional contraction. Manual muscle testing of hip abduction in this population often demonstrates this difficulty as hip flexion usually accompanies, and sometimes overpowers, the attempted hip abduction. Additionally, children aged 1-2 years may not be able to follow directions adequately to isolate this movement. For these reasons, it is suggested that both methods be used for young children, with and without CP, to provide verification of gluteus medius EMG sensor placement.

Proposed Study

CP is a common neuromuscular disorder that limits participation in family, school, work, and community environments, and negatively affects quality of life. With the primary impairment of poor control of posture and movement, physical therapists have the responsibility to explore new avenues in the measurement and treatment of movement dysfunction in CP.

Recent neuroplasticity studies in adults have demonstrated the importance of targeted and intensive intervention after injury to the central nervous system, *prior* to the establishment of maladaptive movement patterns.^{33, 42-45} However, systematic and thorough applications of aggressive intervention concepts have not been applied to individuals with CP. Motor patterns are reinforced with repetition,^{33, 36, 38} and young children with less walking experience may be more responsive to intervention than older children, adolescents, and adults with well-established compensatory motor patterns.

Poor control of postural muscles is considered a primary impairment in CP, which contributes to compensation by other muscles to assist in maintaining upright posture, and limiting those muscles from functioning effectively as primary movers of the extremities.⁹⁴ Surprisingly, axial and hip muscle activity during the emergence of walking skills has not been

systematically investigated in individuals with CP.⁹ Furthermore, the majority of research on gait in CP includes participants who are between the age of 5 years and adolescence.

The objective of this project is to investigate differences in trunk and hip muscle activation patterns during the early stages of walking in children with CP compared to children with TD. This study will serve as a foundation from which theoretically-grounded interventions will be developed, including programs that address deficits in trunk and proximal limb muscle function, adapting neurorehabilitation interventions to young children, and appropriately measuring the outcomes and mechanisms of these interventions. Figure 1-2 depicts the proposed measures of the study and how each corresponds to a construct in the World Health Organization's framework for the International Classification of Functioning, Disability, and Health.⁸

Information on the activation patterns of muscles from the onset of walking may give clinicians and researchers a greater understanding of how abnormal muscle activation develops to result in the gait deviations observed in children with CP. Additionally, information on how the trunk and gluteal muscles function during walking will assist our understanding of their role in maintaining upright posture, and could lead to the development of more effective interventions to address deficits in their function. This could allow the prediction of future gait patterns and warrant more intensive, targeted physical therapy interventions, before the compensatory postural patterns are reinforced and possibly preventing the need for corrective surgery.

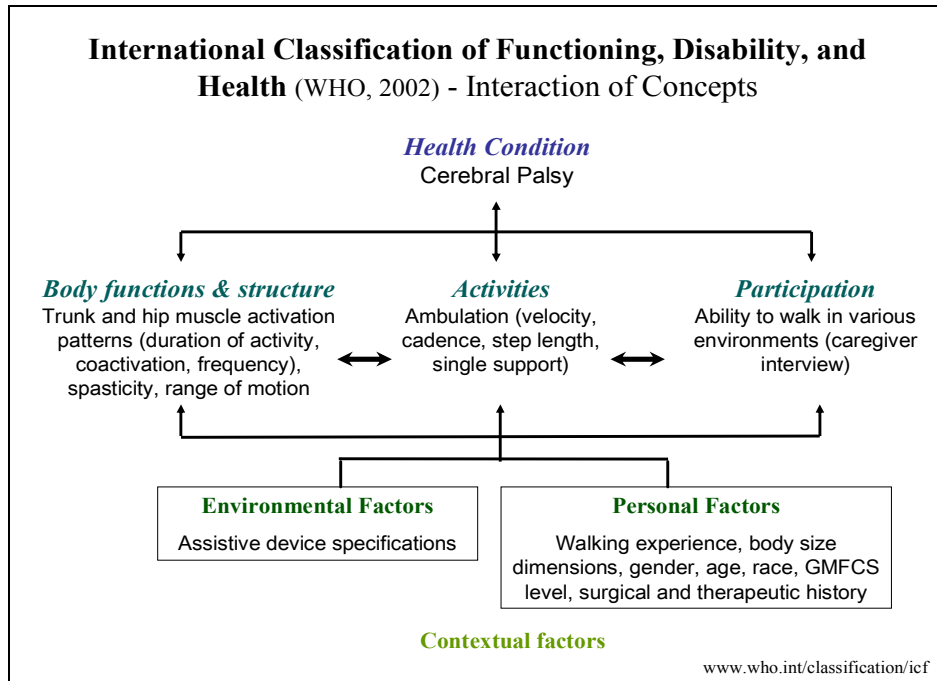


Figure 1-2. Proposed study measures as related to constructs of the World Health Organization's International Classification of Functioning, Disability, and Health (ICF).

Specific Aims

Aim 1. To determine if differences exist in spatiotemporal gait parameters between early walkers with spastic CP and children with TD with similar amounts of walking experience.

Hypothesis 1.1. Differences exist in spatiotemporal gait parameters (walking velocity, cadence, step length, base of support, and single support time) in children with spastic CP who are classified as Gross Motor Functional Classification System (GMFCS) level II-III and have 0.5-60 months of walking experience compared to children with TD and equivalent walking experience.

Hypothesis 1.2. Early walkers with spastic CP will demonstrate greater stride-to-stride variability in spatiotemporal gait parameters (walking velocity, cadence, step length, base of support, and single support time) than early walkers with TD.

Aim 2. To determine if differences exist in trunk and hip muscle activation patterns in reference to gait cycle events and phases (initial contact, foot off, stance phase, swing phase) between early walkers with spastic CP and children with TD with similar amounts of walking experience.

Hypothesis 2.1. Differences exist in trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosus) muscle activation timing patterns during walking in children with spastic CP who are classified as GMFCS level II-III and have 0.5-60 months of walking experience compared to children with TD with equivalent walking experience. Differences will also exist in the timing of muscle coactivation between antagonist pairs (erector spinae/rectus abdominus, rectus femoris/semitendinosus) between the two groups.

Hypothesis 2.2. Differences exist in the frequency characteristics of trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosis) muscle activation patterns during walking in children with spastic CP who are classified as GMFCS level II-III and have 0.5-60 months of walking experience compared to children with TD with equivalent walking experience.

Hypothesis 2.3. Early walkers with spastic CP will demonstrate greater stride-to-stride variability in trunk and hip muscle EMG frequency characteristics than early walkers with TD.

Aim 3. To determine if differences exist in spatiotemporal gait parameters and trunk and hip muscle activation patterns during walking in early walkers with spastic CP with similar amounts of walking experience but different levels of ambulatory ability.

Hypothesis 3.1. Children with spastic CP who are classified as GMFCS level III and have 0.5-60 months of walking experience will demonstrate differences in spatiotemporal gait parameters (walking velocity, cadence, step length, base of support, and single support time), compared to children with spastic CP who are classified as GMFCS level II with equivalent walking experience.

Hypothesis 3.2. Children with spastic CP classified as GMFCS level III will demonstrate greater stride-to-stride variability in spatiotemporal gait parameters compared to children with spastic CP who are classified as GMFCS level II.

Hypothesis 3.3. Children with spastic CP who are classified as GMFCS level III and have 0.5-60 months of walking experience will demonstrate differences in trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip

(bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosus) muscle activation and coactivation timing patterns, compared to children with spastic CP who are classified as GMFCS level II with equivalent walking experience.

Hypothesis 3.4. Children with spastic CP who are classified as GMFCS level III and have 0.5-60 months of walking experience will demonstrate differences in the frequency characteristics of trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosus) muscle activation patterns during walking, compared to children with spastic CP who are classified as GMFCS level II with equivalent walking experience.

CHAPTER 2

SPATIOTEMPORAL GAIT CHARACTERISTICS IN EARLY WALKERS WITH CEREBRAL PALSY

Introduction

Cerebral palsy (CP) is the most common neuromuscular disorder in children and has an increasing prevalence,³ a high economic cost⁴ and a negative impact on quality of life.⁵ CP is characterized by impairments in the development of movement and posture, that are attributed to disturbances that occurred in the developing fetal or infant brain.⁹ Children with CP use compensatory movement strategies as a result of decreased postural control, muscle weakness, poor coordination and spasticity. These abnormal motor strategies are reinforced with repetition,^{33, 36} over time and throughout development. Recent neuroplasticity studies in adults have demonstrated the importance of targeted and intensive intervention after injury to the central nervous system, *prior* to the establishment of maladaptive movement patterns.^{33, 42, 44} Similar increases in cortical activation were reported in a pilot study of children with CP undergoing an intensive 2 week locomotor training program.⁹⁸ However, systematic and thorough applications of intensive intervention concepts have not been applied to individuals with CP.

Young children with less walking experience may be more responsive to intervention than older children, adolescents, or adults with well-established compensatory gait patterns. To develop appropriate treatment programs for young children, it is critical to examine the gait patterns of individuals with CP during the development of compensatory movement and abnormal gait patterns. The majority of gait research in CP includes participants who are

between the age of 5 years and adolescence, which are many years after the causative brain injury. Young children are rarely included as research participants, in part because of limited attention, cognition, and tolerance to cumbersome instrumentation involved in gait analysis.

The immaturity of movement patterns in early walkers also complicates data interpretation. Mature gait is characterized by low variability⁹⁹ and a high degree of symmetry¹⁰⁰ from stride to stride. Children have greater stride to stride variability than adults.¹⁰¹ This variability should be reported in addition to parameter means in order to better characterize immature walking patterns. Investigating the more variable, less mature gait parameters of early walkers is necessary so that we can learn how to best intervene before the maladaptive compensations are less flexible and difficult to change.

To understand the process of the establishment of compensatory gait patterns in CP, walking must be studied from the onset of skill development. An initial step is to study spatiotemporal gait parameters at the time walking skills are being learned. It is also critical to establish methods to reduce the complexity of the study of gait to make participation feasible for younger children. This study lays the groundwork for earlier study of gait in CP when gait skill is emerging.

The objective of this study was to investigate differences in spatiotemporal characteristics of gait during the early stages of walking in children with CP compared to children with similar amounts of walking experience and typical development (TD). In addition, the variability and symmetry of these gait parameters, and their relationships with musculoskeletal measures, were also investigated.

Methods

Participants

Participants with CP were recruited through the CP clinic at Shriners Hospital for Children in Philadelphia, PA, Children's Specialized Hospital in Mountainside, NJ and through other area rehabilitation facilities. Participants with TD were recruited from siblings of the participants with CP, children of people known to the investigators, and a local day care. The institutional review board (IRB) of Temple University Hospital (for Shriners Hospital), and the IRBs of additional data collection sites as needed, approved all procedures. All data collection procedures were explained and parents gave informed consent to the research and to publication of the results. Assent of a minor was also obtained from participants 7 years of age or older.

Inclusion and exclusion criteria are listed in Table 2-1. Children with CP were able to walk with a hand held assistive device if it did not restrict movement of the trunk or pelvis (i.e. walker without pelvic guide or crutches). The selection of months of walking experience rather than age as a primary inclusion criterion was chosen based on reports that experience is a stronger predictor of walking and balance skill than age in early walkers.^{92, 93} The onset of walking was operationally defined as the age in months at which an infant or child was able to take at least 3 continuous independent steps on a consistent basis.¹⁰² Walking experience, in months, was calculated as the difference between the participant's age on the day of the study and the age of onset of walking. This is a novel approach for CP research because the majority of investigations compare children with CP to children with TD who are the same age, rather comparing groups who have similar amounts of walking experience, despite later onset of walking in CP.

Table 2-1. Participant Inclusion and Exclusion Criteria

Inclusion	Exclusion
<ul style="list-style-type: none"> • 0.5-60 months of walking experience • Able to ambulate barefoot at least 15 feet in a forward direction with supervision (children with CP could use assistive device) • Able to follow 1-step verbal instructions <p>For children with CP only:</p> <ul style="list-style-type: none"> • Spastic diplegia or quadriplegia • GMFCS II-III classification⁷⁴ 	<ul style="list-style-type: none"> • Lower extremity bony or soft tissue surgery or fracture in the past 12 months <p>For TD children only:</p> <ul style="list-style-type: none"> • History of any orthopedic, neuromuscular, or cardiovascular condition <p>For children with CP only:</p> <ul style="list-style-type: none"> • Spastic hemiplegic or non-spastic classification • History of dorsal rhizotomy • History of tendon transfer to a target muscle • Botulinum toxin injection to a lower extremity muscle in the past 6 months • Secondary orthopedic, neuromuscular or cardiovascular condition

Procedures

Musculoskeletal Measurement

All anthropometric and musculoskeletal measurements were taken by one pediatric physical therapist. The anthropometric measures included height, seated height, weight, and bilateral leg length (anterior superior iliac spine to the apex of the medial malleolus). All distances were measured with a Harpenden anthropometer (Holtain Limited, Pembrokeshire, UK) with the exception of those in two children who were fearful of the device. A standard cloth tape measure was used in those cases.

Range of motion (ROM) measures in children with CP included bilateral hip extension, hamstring length, and ankle dorsiflexion. Hip extension was measured using the Thomas test method.¹⁰³ Hamstring length was measured using the popliteal angle method. The hip was flexed 90° with the contralateral hip in extension and stabilized. The knee was then extended to end range. The angle between the vertical and the tibial shaft was recorded. Ankle dorsiflexion was measured in prone with the knee extended. Interrater reliability of goniometric measurements in children with spastic diplegic CP has been reported, with ICCs ranging from 0.92-0.95 for hip extension and 0.88 for ankle dorsiflexion.¹⁰⁴

Muscle stiffness was measured bilaterally in the hamstrings and hip adductors in children with CP using the modified Tardieu scale.¹⁰⁵ As described by Yam and Leung,¹⁰⁶ the hip adductors were tested with the hip in extension and knee in flexion (thigh is supported and leg hangs over the edge of the examination table) and the hamstrings were tested with the hip in flexion (contralateral hip in extension). Muscles were stretched at 2 velocities and 2 goniometric angles were recorded. R1 is the angle at which a change in muscle resistance, or “catch”, was detected when the limb segment was moved at a fast velocity. R2 is the end range of the muscle

length and was measured by moving the limb segment at a slow passive velocity. Additionally, the quality of the muscle reaction (X) was recorded.¹⁰⁶

The Tardieu scale has been found to discriminate between spasticity and muscle contracture in individuals who sustained a unilateral stroke.¹⁰⁷ Inter-rater reliability of the Tardieu scale for the hip adductors in children with CP was 0.71. Intra-rater reliability was not reported in the study by Yam and Lueng.¹⁰⁶ Test-retest intra-rater reliability for the hamstrings in adults with traumatic brain injury has been reported to be 0.81.¹⁰⁸

To increase tolerance and compliance with musculoskeletal measurements, children were distracted with watching videos or playing with a research aide during the examination.

Walking Trials

Children walked barefoot down an instrumented walkway (GAITRite®, CIR Systems, Havertown, PA) at a self-selected pace. Three to five trials, each consisting of one walk down the walkway with at least 4 consecutive footfalls, were collected depending on participant tolerance to testing procedures and fatigue. Footfall information was collected at 30 Hz.

Webster and colleagues¹⁰⁹ established high criterion validity (ICCs of 0.92-0.99) of the GAITRite® parameters with a 3-dimensional motion capture system. Reliability of the GAITRite® walkway was investigated in 19 children with motor disability by Wondra et al.¹¹⁰ Single trial reliability for stride length, base of support, velocity, and cadence ranged from 0.84 to 0.97.

Start and stop targets in child-friendly colors and patterns were placed on the floor approximately 5 feet beyond either end of the instrumented walkway to minimize acceleration or deceleration while walking on the walkway. A walking trial started by having the child stand on

the start target. Data collection was initiated through the GAITRite® software and the child was instructed to walk to the target beyond the opposite end of the walkway. If needed, children were motivated and rewarded with stickers, small snacks, or favorite toys. Children had the opportunity to sit on a chair in between walking trials to minimize fatigue. All walking trials were videotaped for later gait cycle selection and parents/caregivers signed a separate consent to allow medical videography.

Data Analysis

Video of each walking trial was reviewed to determine the most appropriate gait cycles to select for data analysis. Five gait cycles for each side (left and right) were selected based on the observation of typical walking (child was not distracted, did not stop walking, and was not moving arms toward an object). Individual footfall data generated by the GAITRITE® analysis software for these strides were used for analysis.

Five primary spatiotemporal parameters were analyzed: walking velocity, cadence, step length, base of support, and single support time (as a percent of the gait cycle). Step length and walking velocity measures were normalized to leg length and converted to dimensionless values.¹¹¹ Step length was divided by leg length. Walking velocity was scaled using the following equation: $v' = v(g\ell_0)^{-0.5}$ where v' represents dimensionless walking velocity, v represents measured walking velocity (m/s), g is the gravitational acceleration constant (9.8 m/s²), and ℓ_0 represents leg length (m).

Individual means, standard deviations (SD) and coefficients of variation (CV) were calculated for each parameter by averaging the 10 values (5 left, 5 right) from the individual gait cycles. The CV can be more meaningful than the SD as a measure of variability when there is the

potential for large differences in group means.¹¹² In addition, a symmetry ratio was calculated for each parameter by dividing the smaller left or right value by the larger value. This resulted in a value between 0.0 and 1.0, with values closer to 1.0 indicating greater symmetry.

Group means were then calculated for each parameter, and each variability and symmetry measure. Assuming unequal variance between groups, a general linear model (GLM) was used to determine differences in spatiotemporal gait measures. Group differences in walking experience and body size dimensions were examined using t-tests and considered covariates in the GLM if different ($p < 0.05$).

Linear regressions and the corresponding coefficients of determination (r^2) were used to account for the variance in spatiotemporal parameters within the CP group in relation to range of motion limitations and spasticity.

Results

Thirty-four children enrolled in this study. Data from three children were excluded due to one having a questionable diagnosis of CP and two who were unable to walk without additional assistance from an investigator during the testing session. Data for the remaining 31 children (17 male, 15 with CP) were used for analysis. In the group of children with CP, seven were classified as GMFCS level II and eight were level III. One was classified as spastic quadriplegic, and 14 as spastic diplegic. Three walked without assistive devices, nine used posterior rolling walkers, one used bilateral forearm crutches, and two used unilateral forearm crutches. ROM data were not obtained from one child due to time constraints. Demographic and musculoskeletal data are provided in Tables 2-2 and 2-3, respectively.

Table 2-2. Demographic and Anthropometric Data

		Onset of walking (mo)	WE (mo)	Sex	Age (mo)	Weight (kg)	Height (m)	BMI (kg/m ²)	Seated Height (m)
TD (n=16)	Mean	11.7	28.6	9F	39.7	15.1	0.97	15.9	0.55
	(SD)	(3.1)	(19.6)	7M	(19.5)	(3.9)	(0.13)	(1.8)	(0.06)
	Range	8.0-20.0	1.0-58.0		13.0-67.5	10.0-21.9	0.75-1.18	11.2-18.8	0.46-0.69
CP (n=15)	Mean	34.8	28.4	5F	63.1	19.6	1.06	17.2	0.56
	(SD)	(10.2)	(17.0)	10M	(23.2)	(5.9)	(0.14)	(2.4)	(0.06)
	Range	18.0-55.0	2.0-60.0		25.0-108.0	10.9-31.2	0.83-1.32	14.7-22.9	0.48-0.65
Total (n=31)	Mean	22.9	28.5	14F	51.0	17.3	1.10	16.5	0.56
	(SD)	(13.8)	(18.1)	17M	(24.1)	(5.4)	(0.14)	(2.2)	(0.06)
	Range	8.0-55.0	1.0-60.0		13.0-108.0	10.0-31.2	0.75-1.32	11.2-22.9	0.46-0.69

TD=typically developing, CP=cerebral palsy, WE=walking experience, mo=months, SD=standard deviation, M=male, F=female, kg=kilograms, m=meters, BMI=body mass index

Walking experience did not differ between groups ($p=0.969$). However, because the onset of walking was later in the group with CP, they were older than the group with TD, and were larger in some anthropometric measures. Height, BMI, and seated height did not differ between groups ($p=0.080$, 0.102 , 0.601 , respectively). The CP group was heavier ($p=0.017$) and had longer legs ($p=0.029$) than the TD group. Weight was, therefore, used as a covariate for the GLM for comparison of all spatiotemporal measures. Leg length was used as a covariate for cadence, base of support, and single support time measures (walking velocity and step length were normalized to leg length prior to statistical analysis).

Group means for the primary spatiotemporal parameters, the individual cycle to cycle variability measures (SD, CV), and the symmetry ratios are presented in Table 2-4. Group means

Table 2-3. Musculoskeletal Measures and Significant Relationships Between Musculoskeletal and Spatiotemporal Measures for Individuals With CP

Musculoskeletal measure		Mean (SD)*	Range*	Spatiotemporal measure	r	R ²	p value
Hip extension ROM		-2 (2)	-9-0	cadence symmetry	0.660	0.435	0.020
Dorsiflexion ROM		7 (9)	-5-27	cadence CV	0.602	0.363	0.023
				step length	-0.548	0.301	0.042
				step length symmetry	-0.582	0.339	0.047
				single support symmetry	-0.682	0.465	0.010
Hamstrings ROM		133 (9)	118-150	none	n/a	n/a	n/a
Hip adductor Spasticity	R1	17 (8)	2-33	none			
	R2	31 (12)	15-62	none	n/a	n/a	n/a
	<i>X</i>	1.7 (0.4)	0.5-2	none			
Hamstring spasticity	R1	94 (10)	68-109	none			
	R2	133 (9)	118-150	none	n/a	n/a	n/a
	<i>X</i>	1.9 (0.2)	1.5-2	none			

SD=standard deviation, ROM=range of motion, CV=coefficient of variation

*average of left and right sides, measured in degrees except for *X* value

were significantly lower in the CP group than the TD group for all main parameters: walking velocity, cadence, step length, base of support, and single support time.

Stride to stride coefficients of variation were significantly higher in the CP versus TD group for cadence, base of support, and single support. The stride to stride standard deviations were also higher for base of support and single support time. Symmetry ratios between left and right sides were significantly lower in the CP group for base of support.

For all participants, the covariate of weight was significantly related to cadence symmetry ($p=0.001$) and base of support symmetry ($p=0.013$) with lighter children demonstrating less symmetry. Weight was also significantly related to base of support ($p=0.045$), base of support SD ($p=0.007$), and base of support CV ($p=0.028$) with lighter children demonstrating greater base of support and more variability. The covariate of leg length was related to cadence symmetry ($p=0.008$) with children with shorter legs demonstrating less symmetry.

Few of the spatiotemporal characteristics were related to musculoskeletal measures in the CP group. Decreased average hip extension range of motion was related to decreased symmetry in cadence ($p=0.020$). Increased average ankle dorsiflexion range of motion was related to decreased step length ($p=0.042$), increased cadence variability ($p=0.023$), and decreased symmetry in step length ($p=0.047$) and single support time ($p=0.010$). Hamstring range of motion was not related to any of the spatiotemporal characteristics. Similarly, Tardieu test X values for both hamstrings and adductors were not related to spatiotemporal measures. However, the small range in X values (0-2) likely hinders the ability to identify a relationship if one exists.

Table 2-4. Group Means (and SD) for Spatiotemporal Parameters, Stride-to-Stride Variability (SD, CV), and Symmetry

	TD				CP				<i>p</i> values			
	Value	SD	CV	SR	Value	SD	CV	SR	Value	SD	CV	SR
Walking velocity [^]	4.15 (0.62)	0.54 (0.37)	0.13 (0.09)	0.93 (0.07)	2.19 (1.05)	0.59 (0.30)	0.32 (0.29)	0.78 (0.26)	0.000*	0.611	0.082	0.091
Cadence (steps/min)	155.2 (19.8)	14.3 (6.9)	0.09 (0.04)	0.95 (0.03)	103.3 (36.3)	21.7 (19.5)	0.21 (0.15)	0.87 (0.14)	0.001*	0.194	0.019*	0.153
Step length [^]	0.76 (0.07)	0.08 (0.05)	0.11 (0.07)	0.93 (0.06)	0.55 (0.16)	0.12 (0.06)	0.26 (0.32)	0.82 (0.29)	0.001*	0.074	0.222	0.266
Base of support (cm)	10.5 (2.4)	1.7 (0.6)	0.16 (0.06)	0.92 (0.06)	6.5 (7.2)	4.0 (2.3)	0.40 (0.20)	0.72 (0.27)	0.025*	0.000*	0.001*	0.001*
Single support (% gait cycle)	84.8 (2.8)	4.1 (2.5)	0.05 (0.03)	0.95 (0.04)	73.8 (17.7)	10.4 (8.6)	0.16 (0.14)	0.85 (0.13)	0.049*	0.021*	0.018*	0.077

TD= typically developing, CP= cerebral palsy, SD= standard deviation, CV= coefficient of variation, SR= symmetry ratio

[^]walking velocity and step length are dimensionless values, normalized to leg length

*indicates significant difference (p<0.05)

Therefore, the more specific R1 and R2 values were investigated, but again no relationship was present for those spasticity values. Table 2-3 presents correlation coefficients, coefficients of determination, and p values for all significant regressions ($p < 0.05$).

Discussion

This study investigated differences in spatiotemporal characteristics of gait, and the variability and symmetry of these measures, during the early stages of walking in children with CP and children with TD. Children with CP achieved independent walking at a later age than those with TD. The method of using walking experience, rather than age, for inclusion in the study controlled for the maturation of walking ability that occurs with practice after the onset of walking. Because of delayed motor development in individuals with CP, however, there were some body size differences between groups (children with CP were bigger) that needed to be accounted for during analysis of the data.

Spatiotemporal gait parameters for the TD group are consistent with normative values previously reported.^{81, 113, 114} Children with CP demonstrated slower walking velocity, decreased cadence, shorter step length, narrower base of support, and reduced single limb support compared to children with TD who had similar amounts of walking experience. Spatiotemporal parameters for the CP group were slightly lower than those reported by Wondra and colleagues for a group of children with mixed motor disabilities.¹¹⁰ However, the children in that study were an average of 1.5 years older and the sample was more heterogeneous (including children with diplegia, hemiplegia, ataxia, Angelman syndrome, and arthrogryposis) than the sample in the present study. Walking speed in the current study was also less, approximately half, than that reported by Sorsdahl and colleagues¹¹⁵ for a group of children with CP, but who were less

impaired than the current sample. None of the children in Sorsdahl's study walked with an assistive device and all were classified as GMFCS I or II.

Base of support (or step width) was highly variable in the CP group. The mean was lower than the TD group, which warrants discussion. Base of support values ranged from -6.3 to 17.5 cm. Two children had negative values, representing foot placement that crossed the midline of the body, or "scissoring". In the CP group, base of support was not related to hip adductor spasticity or the use of an assistive device, which would provide assistance with balance. To our knowledge, base of support in young children with CP has not been reported.

Little work has been done investigating stride to stride variability in spatiotemporal parameters in children. Hausdorff and colleagues found that stride time is more variable in 3-4 year old children compared to 6-7 year old children, who demonstrate more variability than 11-14 year old children.⁹⁹ However, these investigators did not report any of the parameters measured in our current study. Looper and colleagues reported CV for step length and step width in new walkers with TD and those with Down syndrome.¹¹⁶ For the 9 new walkers in their study, the CV for step length averaged 0.38. In our study, the mean step length CV was 0.11, but our participants had more walking experience. Step width CV, however, in their study averaged 0.10, and was similar to our measurement for the children with CP (0.16).

Mean stride to stride variability values were higher in the CP versus TD group for all parameters but only reached significance for cadence (CV), base of support (SD, CV), and percent single support (SD, CV). To our knowledge, stride to stride variability for spatiotemporal measures has not been reported for children with CP.

The symmetry ratios for the TD group ranged from 0.92 to 0.95. This is lower than the values reported by Wheelwright and colleagues for step length and swing time in children

(median 0.99 for both), and similar to the median symmetry ratio for double support time (0.91).¹¹⁷ The sample in Wheelwright's work was older, consisting of 134 children aged 3-18 years, and thus would be expected to have greater symmetry than the children in the present study based on previous work reporting greater symmetry in adolescents compared to children.¹¹⁸ In our study, symmetry ratios were lower for all parameters in the CP group compared to the TD group, but only significantly lower for base of support. Side to side symmetry has not been previously reported for spatiotemporal gait parameters in children with CP.

Spasticity was not related to spatiotemporal gait parameters which is consistent with the findings of Ross and Engsborg.¹¹⁹ Some of the parameters were related to lower extremity range of motion. While greater hip extension range was associated positively with more cadence symmetry, greater dorsiflexion range was associated with greater variability (cadence), less symmetry (step length, single support), and shorter step length. While the maximal range of hip extension in the CP group approached normal values, maximal ankle dorsiflexion range exceeded normal range of motion. A crouch gait pattern is common in CP and is characterized by excessive dorsiflexion during the stance phase.⁵⁰ It has been suggested that a crouch gait pattern limits the ability to extend the hip and knee,¹²⁰ which would contribute to shorter step length.

Body size measures and walking experience were related to spatiotemporal parameters, variability, and symmetry in the TD group, but not the CP group. For the TD group, this is consistent with studies showing that older children have more mature gait patterns than younger children.^{81, 99} The larger children had more walking experience in both groups, but factors other than experience and size determine the extent of impairment in walking ability in CP. As shown

in Table 2-3, musculoskeletal factors explain some of this variance, but muscle activation patterns or severity of CP may explain additional variance in walking ability. The participants in this study were a fairly homogeneous group children with bilateral spastic CP, GMFCS II and III. These results may not be generalizable to children with greater walking experience or those with greater or lesser ambulatory ability.

Children with CP demonstrate reduced walking ability during the early stages of walking even when walking experience is equal between groups. These differences are consistent with those seen in older children with CP, which suggests that waiting several years after the onset of walking to intervene with gait impairments may not be necessary or optimal. The increased stride to stride variability in early walkers with CP could suggest that the neural circuitry supporting their gait patterns may not yet be well-established. Early in the development of walking skill may be a more optimal time to intervene, or may be a time when intensive treatments are most effective in changing compensatory gait patterns and lead to improved walking ability.

Differences are present in the walking performance of children of CP compared to children with TD during the early stages of walking ability, even when children are grouped by walking experience rather than age. Investigation of movement patterns early in their development may lead to more appropriate intervention in CP. Addressing abnormal patterns of movement before the neural pathways that support them become too rigid from years of reinforcement may produce more effective outcomes. Stride to stride variability and side to side asymmetry of spatiotemporal gait characteristics is increased in early walkers with CP. Treatment programs with aggressive practice paradigms may result in less variability, and improved efficiency, of movement in CP. Future investigation should examine treatment

programs delivered during the early stages of walking in CP as well as the variability and symmetry in other early walking characteristics, such as kinematics, kinetics, and muscle activation patterns.

CHAPTER 3

MUSCLE ACTIVATION PATTERNS FOR THE TRUNK AND HIP MUSCLES IN EARLY WALKERS WITH AND WITHOUT CEREBRAL PALSY

Introduction

Poor control of trunk postural muscles is a primary impairment in cerebral palsy (CP)^{9, 95} which causes compensation by other muscles to assist in maintaining upright posture.

Compensation by accessory muscles to aide in posture reduces their effectiveness in functioning as primary movers of the extremities.⁹⁴ Evidence to support this notion includes observations that children with CP have greater ambulatory ability when the distal limb musculature is primarily affected, and proximal limb musculature is less affected.^{58, 87} Additionally, proximal limb muscles, such as those of the hip, are also critical for maintaining upright mobility. Compared to knee and ankle muscles, the strength of the hip abductors explained the largest variance in gait and gross motor function in CP.¹¹⁹

Given their importance for ambulatory ability, it is surprising that direct study of activation patterns of trunk and hip muscles during walking have not been investigated in individuals with CP. To date, the majority of research investigates postural control on a macroscopic level by studying center of pressure (COP) trajectories during standing and walking, or by studying lower extremity muscle responses to balance perturbations.¹²¹ Direct study of trunk and hip muscle activity during the development of compensatory movement strategies is desirable because these movement patterns are reinforced with repetition,^{33, 36, 38} over time and throughout development. Young children with less walking experience may be more responsive to intervention than older children or adults who have well-established

compensatory patterns of muscle activation. Thus, developing successful treatment programs to improve postural control in individuals with CP requires a better understanding of trunk and hip muscle behavior during the early years of compensatory postural control development.

Electromyographic (EMG) analysis is a critical component in the examination of muscle function in individuals with CP. Several factors can be extracted from the EMG signal to provide insight into muscle activation patterns.^{58, 122, 123} The most commonly employed clinical use of muscle EMG is to determine the onset and offset timing of muscle activity during movement. This type of temporal information identifies periods of muscle activity and inactivity throughout the gait cycle and is used to determine coactivation of antagonistic muscle groups. The objective of this study was to investigate differences in the timing characteristics of trunk and hip muscle activity during the early stages of walking in children with CP compared to children with typical development (TD). Identifying early compensations in postural muscle function may lead to the development of more focused interventions that have the potential to improve movement and function before the usual postural deficits are reinforced.

Methods

Participants

Participants with CP were recruited through the CP clinic at Shriners Hospital for Children in Philadelphia, PA, Children's Specialized Hospital in Mountainside, NJ and through other local rehabilitation facilities. Participants with TD were recruited from siblings of the participants with CP, children of people known to the investigators, and from a local day care center. The institutional review board of Temple University Hospital (for Shriners Hospital), and the IRBs of the additional data collection sites approved all procedures. All data collection

procedures were explained and parents gave their informed consent to the research and to publication of the results. Assent of a minor was also obtained from participants 7 years of age or older.

The inclusion criteria for all children were: 1) 0.5-60 months of walking experience; 2) ability to ambulate barefoot at least 15 feet with supervision (children with CP could use their usual assistive device if it did not stabilize or restrict movement of the trunk or pelvis, i.e. walker without pelvic guide or crutches); and 3) ability to follow 1-step verbal directions. Children with CP additionally had 4) a diagnosis of spastic diplegia or quadriplegia; and 5) a Gross Motor Function Classification System (GMFCS) level of II or III.⁷⁴ Children were not considered for the study if they had a lower extremity surgery or fracture in the past 12 months, botulinum toxin injection in the past six months, or a history of dorsal rhizotomy or lower extremity tendon transfer.

The selection of months of walking experience rather than age as a primary inclusion criterion was chosen based on reports that experience is a stronger predictor of walking and balance skill than age in early walkers.^{92, 93} The onset of walking was operationally defined as the age when the child was able to take at least 3 continuous independent steps on a consistent basis.¹⁰² Walking experience, in months, was calculated as the difference between the participant's age on the day of the study and the age of onset of walking.

Procedures

Musculoskeletal Measurement

All anthropometric and musculoskeletal measurements were taken by one pediatric physical therapist. Anthropometric measurements included height, seated height, weight, and

bilateral leg length (anterior superior iliac spine to the apex of the medial malleolus). All anthropometric lengths were measured with a Harpenden anthropometer (Holtain Limited, Crosswell, Crymych, Pembs., SA41 3UF, UK) with the exception of those in two children who were fearful of the device. For these children, a standard tape measure was used in lieu of the anthropometer.

Range of motion (ROM) measures in children with CP included bilateral hip extension, hamstring length, and ankle dorsiflexion. Hip extension was measured using the Thomas test method and hamstring length was measured using the popliteal angle method.¹²⁴ Muscle stiffness was measured bilaterally in the hamstrings and hip adductors in children with CP using the modified Tardieu scale.¹⁰⁵ As described by Yam and Leung, the hip adductors were tested with the hip in extension and knee in flexion (thigh is supported and leg hangs over the edge of the examination table) and the hamstrings were tested with the hip in flexion (contralateral hip in extension).¹⁰⁶ Muscles were stretched at two velocities and two goniometric angles were recorded. R1 was the angle at which a change in muscle resistance, or “catch”, was detected when the limb segment was moved at a fast velocity. R2 was the end range of the muscle length and was measured by moving the limb segment at a slow passive velocity. Additionally, the quality of the muscle reaction (X) was recorded.¹⁰⁶

EMG Instrumentation

Surface EMG data from trunk, gluteal, and thigh muscles were acquired using a 16-channel recording system (Myomonitor III, Delsys Inc., P.O. Box 15734 Boston, MA, 02215) with preamplified silver-silver chloride parallel bar surface electrodes with a 10.0 mm

interelectrode distance. EMG data were collected at 1200 Hz, preamplified with a gain of 10, and band pass filtered between 20 and 450 Hz.

EMG data were collected from eight muscles bilaterally (Table 3-1). The rectus femoris (RF) and semitendinosus (ST) were chosen in addition to the trunk and gluteal muscles because these muscles anatomically cross the hip joint. Sensor placement for the abdominal muscles was determined using the methods described by Ng et al.¹²⁵ Sensor placement for all other muscles (back, gluteal, and thigh) was determined in accordance with SENIAM recommendations (Table 3-1).¹²⁶

The skin areas were cleaned with alcohol and the sensors were affixed to the skin with a double-sided adhesive interface. The electrodes were further secured using hypoallergenic tape or a flexible, latex-free, non-adhesive wrap encircling the waist and thighs (Coflex-NL®, Andover Healthcare, Inc., 9 Fanaras Drive, Salisbury, MA 01952). Self-adhesive reference electrodes (Axelgaard Manufacturing Co., Ltd., Lystrup 8520, Denmark) were placed on the skin over the patella bilaterally. A volitional contraction of each muscle, when possible, was elicited to verify placement and confirm the absence of electrical activity from an adjacent muscle. The children were asked to perform specific movements to elicit the corresponding muscle contractions, such as leaning backward in sitting to activate the rectus abdominis muscles, and standing on one leg (with hands held if needed) to activate the contralateral gluteus medius muscle.

Two static baseline EMG trials were collected prior to the walking trials to establish baseline muscle activity. For these trials, the child laid still in supine for 5 seconds. The trial with the least muscle activity was selected for analysis.

Table 3-1. Electromyogram (EMG) Sensor Locations

Muscle	Sensor Location
Trapezius (middle) ¹²⁶	50% of the distance from the medial border of the scapula to the T3 spinous process, parallel to the line from T5 to the acromion
Erector Spinae (longissimus) ¹²⁶	1-2 finger widths lateral from L1 spinous process, oriented vertically
Rectus Abdominus ¹²⁵	At the level of the ASIS, 1-2 cm lateral to the midline, oriented vertically
External Oblique ¹²⁵	Just below the rib cage at the inferior angle of the ribs, oriented obliquely
Gluteus Maximus ¹²⁶	50% of the distance from sacral vertebrae to greater trochanter, at greatest prominence of the middle buttocks, parallel to line from PSIS to middle posterior thigh
Gluteus Medius ¹²⁶	In sidelying, 50% of the distance from iliac crest to greater trochanter, parallel to this axis
Quadriceps Femoris (rectus femoris) ¹²⁶	50% of the distance from ASIS to superior patella, parallel to this axis
Semitendinosus ¹²⁶	50% of the distance from ischial tuberosity to medial tibial epicondyle, parallel to this axis

Walking Trials

Children walked barefoot down an instrumented walkway (GAITRite®, CIR Systems, 60 Garlor Drive, Havertown, PA 19083) at a self-selected pace. Webster and colleagues¹⁰⁹ established high criterion validity (ICCs of 0.92-0.99) of the GAITRite® parameters with a 3-dimensional motion capture system with adults who had recent knee arthroplasty. Reliability of the GAITRite® walkway was investigated in 19 children with motor disability by Wondra et al.¹¹⁰ Single trial reliability for stride length, base of support, velocity, and cadence ranged from 0.84 to 0.97.

Three to five trials, each consisting of one walk down the walkway with at least 4 consecutive footfalls, were collected depending on participant tolerance to testing procedures and fatigue. Start and stop targets in child-friendly colors and patterns were placed on the floor approximately 5 feet beyond either end of the instrumented walkway to minimize acceleration or deceleration while walking on the walkway. A walking trial started by having the child stand on the start target. Data collection was initiated through the GAITRite® software, which triggered EMG collection through a trigger module (Delsys Inc., Boston, MA) for synchronous recording. The child was then instructed to walk to the target beyond the opposite end of the walkway. Children had the opportunity to sit on a chair in between walking trials to minimize fatigue.

During the walking trials, the EMG preamplification unit that is typically worn on a backpack was carried behind all participants by an assistant so as not to add additional weight, which could affect muscle activity in the smaller children. If needed, children were motivated and rewarded with stickers, small snacks, or favorite toys. Walking trials were videotaped for later gait cycle selection and parents/caregivers signed a separate consent to allow medical videography.

Data Analysis

Video footage of each trial was reviewed to determine the most appropriate gait cycles to select for data analysis. Ten gait cycles (5 left, 5 right) were selected based on the observation of each individual's typical walking (child was not distracted, did not stop walking, and was not moving arms toward an object). The ten selected gait cycles were extracted from the EMG files using the time-synchronized marker data (initial foot contact for consecutive footfalls) collected from the instrumented walkway. EMG data were processed using custom-written programs in MATLAB software (The Mathworks Inc., 3 Apple Hill Drive, Natick, MA 01760). All signals were normalized to 1000 points, representing the gait cycle from 0 to 100% in 0.1% increments.

To identify muscle activity throughout the selected gait cycles, the Teager-Kaiser Energy (TKE) Operator was applied to the EMG data.¹²⁷ This method uses both the amplitude and frequency components of the signal and was found to better detect the onset of muscle activity than a standard amplitude threshold method.¹²⁸ A detailed description of the technique and its use in this study is included in Appendix A.

EMG data from the selected gait cycles for each participant were filtered with a second-order low-pass Butterworth filter with phase correction and a cut-off frequency of 10 Hz, and averaged across cycles. To determine the onset/offset threshold, the TKE operator was applied to the static EMG baseline signal. The resulting output was then rectified, and the mean and standard deviation were calculated. The mean, plus one standard deviation, was used as the threshold level to determine muscle activity during walking.

Total activation and coactivation were first analyzed as a percent of the gait cycle. A percent gait cycle of activation was calculated for each muscle by summing the duration (in percents) of all periods of muscle activity. Coactivation was determined by calculating the total

time (in percents) antagonistic muscles were simultaneously active. Coactivation was calculated for the ipsilateral rectus abdominis/erector spinae and rectus femoris/semitendinosus muscle pairs. Group means were calculated for percent activation and coactivation and 95% confidence intervals (CIs) were determined. These measures allowed for comparison of total relative time a particular muscle was active in both the CP and TD groups. Assuming unequal variance between groups, Mann-Whitney tests were used to determine differences in percent activation and coactivation ($p < 0.05$).

An additional analysis was then performed to determine when during the gait cycle muscle activity varied between groups. For this analysis, the gait cycle was reduced to 100 points (1% increments) and the number of children in each group who had activity in the muscle at each point in the gait cycle was determined. The chi square test, χ^2 , was performed at each point in the gait cycle to determine if significant differences ($p < 0.05$) existed between groups in the number of children who had activity in the particular muscle.¹¹²

Finally, linear regressions were performed to account for the variance in percent total activation and coactivation within the CP group associated with range of motion limitations and spasticity. All statistical analyses were performed using SPSS software (Version 11.0, SPSS Inc. Headquarters, 233 S. Wacker Drive, Chicago, Illinois 60606).

Results

A power analysis was performed after the initial 9 children (6 TD, 3 CP) completed the study. A sample size of 28 individuals (14 in each group) was predicted to sufficiently power each variable. An additional 20% (6 children) were enrolled to account for anticipated difficulties with participant tolerance and for cases of unusable data. Thus, a total of 34 children

were enrolled in the study. Data from three children were excluded due to one having a questionable diagnosis of CP and two who were unable to walk without additional assistance from an investigator during the testing session. Data for the remaining 31 children were used for analysis.

Walking experience did not differ between groups ($p=0.969$). The children with CP were heavier (0.017) and had longer legs (0.029) than the children with TD, due to a later onset of walking, and therefore older age at the time of testing, in the CP group. In the group of children with CP, seven were classified as GMFCS level II and eight were level III. One was classified as spastic quadriplegic, and 14 as spastic diplegic. Three walked without assistive devices, nine used posterior rolling walkers, one used bilateral forearm crutches, and two used unilateral forearm crutches. ROM data were not obtained from one child with CP due to time constraints. Demographic and musculoskeletal data are provided in Tables 3-2 and 3-3, respectively.

Activation from left and right sides did not differ within groups, so left and right side data were combined for percent total activation and coactivation. The CP group had significantly more total activation time for each muscle (ranging from $p<0.001$ to $p=0.024$) except for the external oblique (EO), which was not different from the TD group ($p=0.593$). Group means for activation, including 95% CIs are shown in Figure 3-1.

The CP group also had significantly more total coactivation time for both the rectus abdominis/erector spinae (RA/ES, $p=0.007$) and rectus femoris/semitendinosus (RF/ST, $p<0.001$) muscle pairs. Coactivation for the RA/ES averaged 20% (95% CI = 5-36%) for the CP group and 1% (95% CI = 0-3%) for the TD group. Coactivation for the RF/ST averaged 75% (95% CI = 61-89%) for the CP group and 20% (95% CI = 10-30%) for the TD group.

Table 3-2. Demographic and Anthropometric Data

		Onset of walking (mo)	WE (mo)	Sex	Age (mo)	Weight (kg)	Height (m)	BMI (kg/m ²)	Seated Height (m)
TD (n=16)	Mean (SD)	11.7 (3.1)	28.6 (19.6)	9F 7M	39.7 (19.5)	15.1 (3.9)	0.97 (0.13)	15.9 (1.8)	0.55 (0.06)
	Range	8.0-20.0	1.0-58.0		13.0-67.5	10.0-21.9	0.75-1.18	11.2-18.8	0.46-0.69
CP (n=15)	Mean (SD)	34.8 (10.2)	28.4 (17.0)	5F 10M	63.1 (23.2)	19.6 (5.9)	1.06 (0.14)	17.2 (2.4)	0.56 (0.06)
	Range	18.0-55.0	2.0-60.0		25.0-108.0	10.9-31.2	0.83-1.32	14.7-22.9	0.48-0.65
Total (n=31)	Mean (SD)	22.9 (13.8)	28.5 (18.1)	14F 17M	51.0 (24.1)	17.3 (5.4)	1.10 (0.14)	16.5 (2.2)	0.56 (0.06)
	Range	8.0-55.0	1.0-60.0		13.0-108.0	10.0-31.2	0.75-1.32	11.2-22.9	0.46-0.69

TD=typically developing, CP=cerebral palsy, WE=walking experience, mo=months, SD=standard deviation, M=male, F=female, kg=kilograms, m=meters, BMI=body mass index

To determine where in the gait cycle the children with CP had excessive muscle activity, histograms were generated to show the number of occurrences of muscle activity at each point in the gait cycle. Because of differences in left and right symmetry in individuals, and the inability to average nominal data (activated vs. not activated), each side was counted individually, resulting in a maximum count of 30 for the CP group, and 32 for the TD group. Figures 3-2 and 3-3 shows histograms for trunk and hip muscles, respectively, for both groups. The asterisks (*) identify the ranges in the gait cycle when the CP group had significantly more occurrences of muscle activity compared to the TD group, as determined by the chi square tests.

The trapezius (TZ) was more active in the CP group compared to the TD group throughout the majority of the gait cycle, except for the period from mid- to late stance. The ES was more active just prior to initial contact through mid-stance. The RA was also more active in

Table 3-3. Musculoskeletal Measures and Significant Relationships Between Musculoskeletal Measures and Muscle Activation for Individuals With CP

Musculoskeletal Measure		Mean (SD)*	Range*	Muscle Activation	r	R ²	p value
Hip extension ROM		-2 (2)	-9-0	None	n/a	n/a	n/a
Dorsiflexion ROM		7 (9)	-5-27	None	n/a	n/a	n/a
Hamstrings ROM		133 (9)	118-150	None	n/a	n/a	n/a
Hamstring spasticity	R1	94 (10)	68-109	None			
	R2	133 (9)	118-150	None	n/a	n/a	n/a
	<i>X</i>	1.9 (0.2)	1.5-2	None			
Hip adductor spasticity	R1	17 (8)	2-33	Rectus abdominis	-0.68	0.46	0.005
				External oblique	-0.55	0.31	0.033
				ES/RA	-0.59	0.34	0.021
	R2	31 (12)	15-62	Rectus abdominis	-0.54	0.29	0.038
				ES/RA	-0.53	0.28	0.043
	<i>X</i>	1.7 (0.4)	0.5-2	None	n/a	n/a	n/a

SD=standard deviation, ROM=range of motion, CV=coefficient of variation, ES/RA=erector spinae and rectus abdominis coactivation, *average of left and right sides, measured in degrees except for *X* value

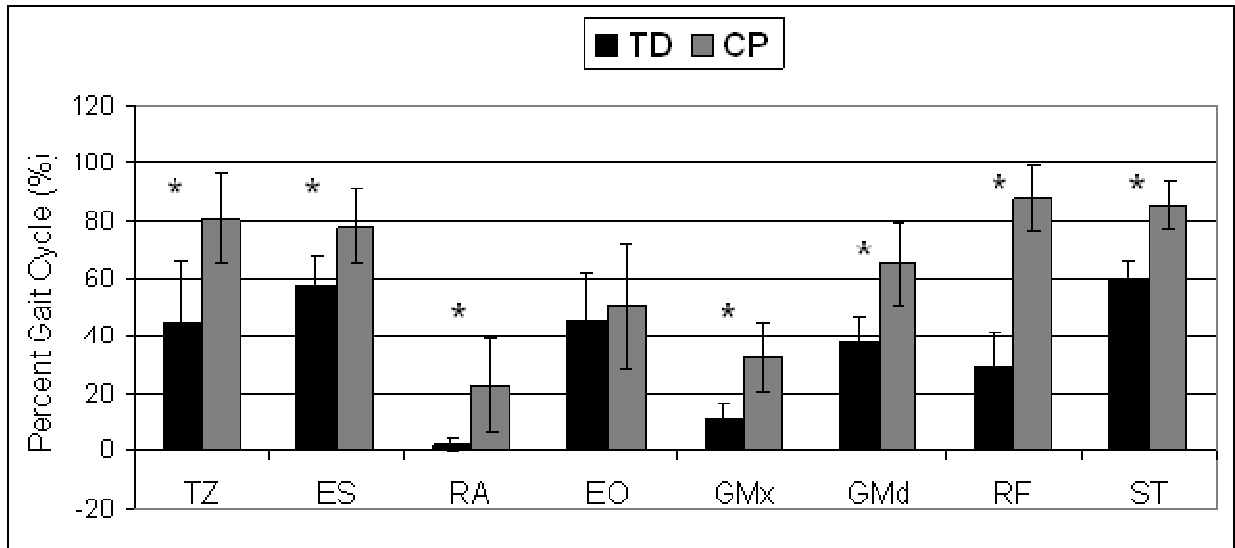


Figure 3-1. Mean total percent activation of trunk and hip muscles for children with cerebral palsy (CP) and typical development (TD). Bars represent upper and lower bounds of 95% confidence intervals. Asterisks (*) indicate muscles that were significantly different between groups. TZ=trapezius, ES=erector spinae, RA=rectus abdominis, EO=external oblique, GMx=gluteus maximus, GMd=gluteus medius, RF=rectus femoris, and ST=semitendinosus.

the CP group throughout most of the gait cycle, while there were no differences in EO activity between groups at any point in the stride. The gluteus maximus (GMx) was more active during both the stance and swing phases of gait, but not during the transitions between the phases. The gluteus medius (GMd) and ST were more active from mid-stance through early swing in the CP group. Except for a short period of time around initial contact, the RF was more active in the CP group throughout the gait cycle.

A summary figure was generated to generalize the periods of activity for each muscle in each group. Each group was considered to have activity at a particular point in the gait cycle if the number of occurrences was at least one-half of the number of possible occurrences at that point (15 for CP, 16 for TD). Figure 3-4 shows generalized periods of activity for each muscle in

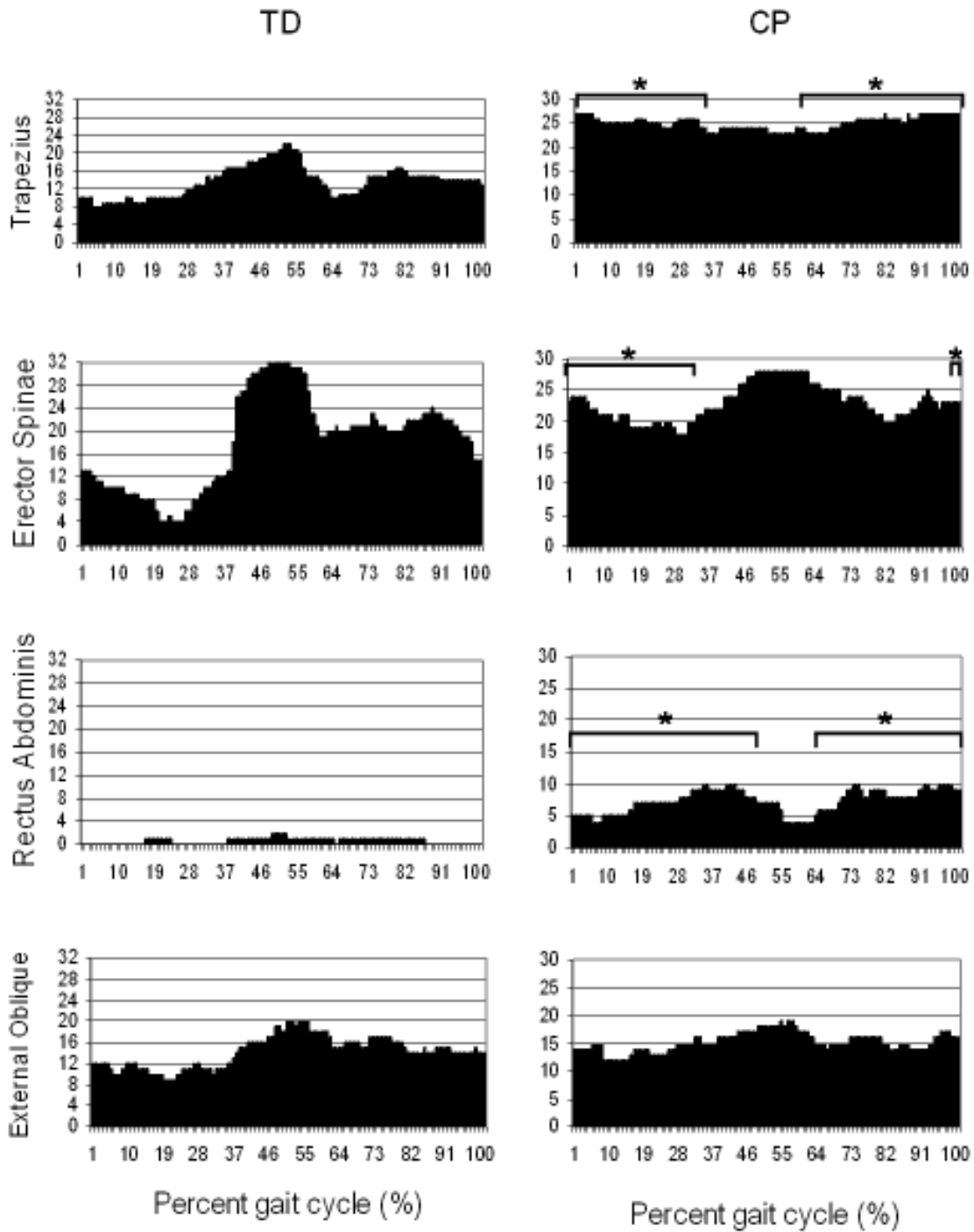


Figure 3-2. Histograms for number of occurrences of trunk muscle activity at each point in gait cycle (1% increments) in typical development (TD) and cerebral palsy (CP) groups. Left and right sides are counted individually, for a maximum of 32 in the TD group and 30 in the CP group. Asterisks (*) indicate periods of activity where the CP group had significantly more occurrences of muscle activity than the TD group.

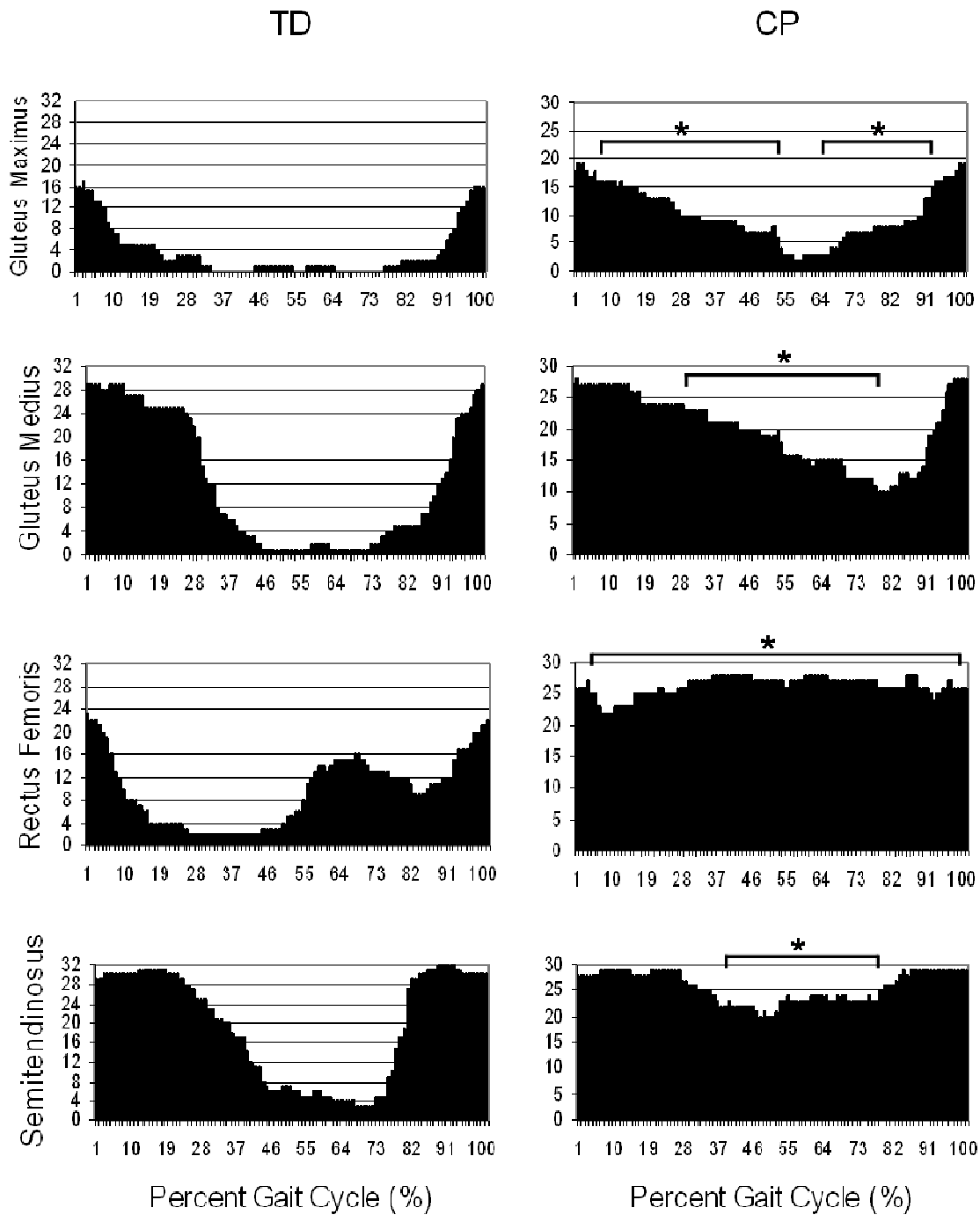


Figure 3-3. Histograms for number of occurrences of hip muscle activity at each point in gait cycle (1% increments) in typical development (TD) and cerebral palsy (CP) groups. Left and right sides are counted individually, for a maximum of 32 in the TD group and 30 in the CP group. Asterisks (*) indicate periods of activity where the CP group had significantly more occurrences of muscle activity than the TD group.

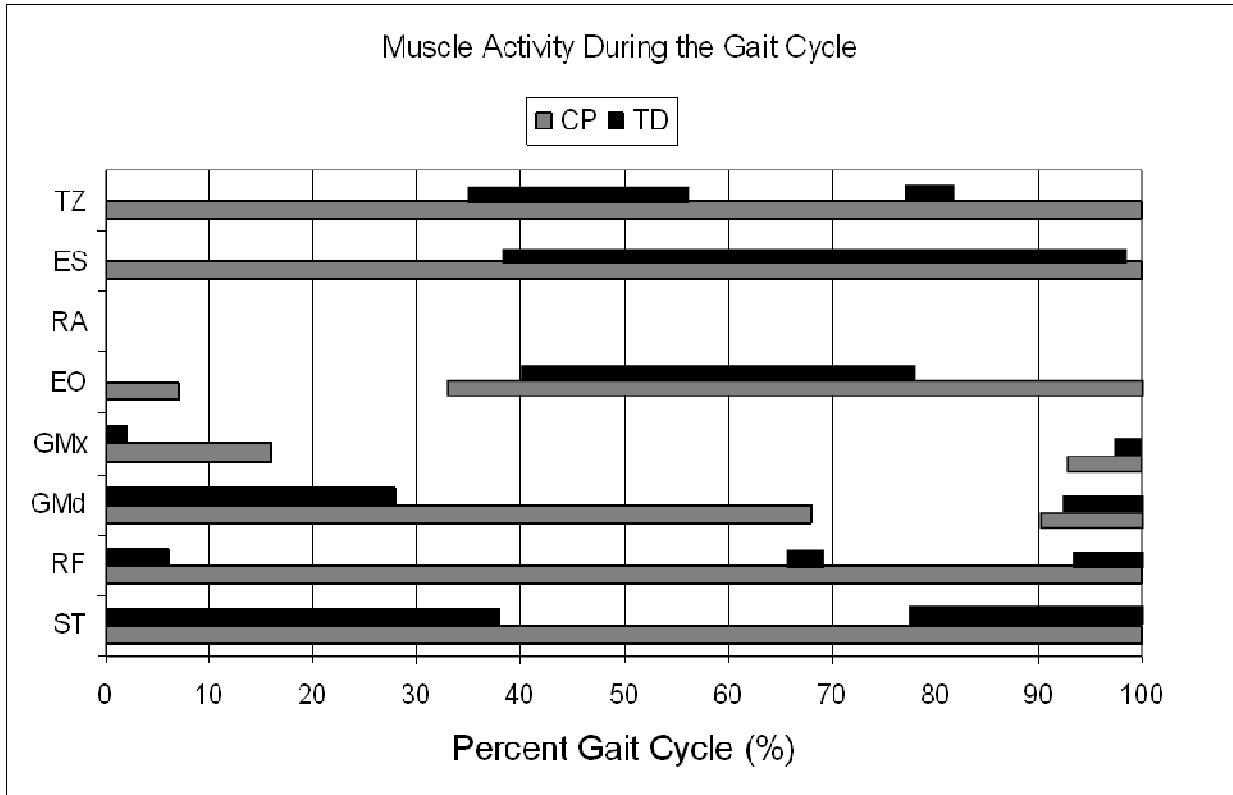


Figure 3-4. Muscle activity across the gait cycle for typical development (TD) and cerebral palsy (CP) groups. TZ=trapezius, ES=erector spinae, RA=rectus abdominis, EO=external oblique, GMx=gluteus maximus, GMd=gluteus medius, RF=rectus femoris, and ST=semitendinosus.

each group. The EO had several points in the gait cycle for each group when activity was present for one less than one-half of the maximum number of occurrences. Because these points were in the midst of large periods of activity, they were included within those periods of activation. The TZ, ES, RF, and ST demonstrated activity throughout the gait cycle in the majority of children with CP. The EO, GMx, and GMd demonstrated similar phases of activity in the CP group compared to the TD group, but they had longer periods of activation, including both earlier onset and delayed offset of activity. Although the RA demonstrated more total muscle activation in the CP group compared to the TD group, it was not active in most children with CP throughout the gait cycle.

The only musculoskeletal measure that explained any variance in muscle activation in the CP group was spasticity in the hip adductors. Both R1 and R2 values were related to total percent activation in the RA and total percent coactivation of the RA/ES. Additionally, the hip adductor R1 value was significantly related to total percent activation in the EO. All significant relationships were negative, indicating that increased spasticity (lower R1 and R2 values) was related to increased muscle activation. Table 3-3 presents correlation coefficients, coefficients of determination, and *p* values for all significant regressions ($p < 0.05$). Total percent activation and coactivation were not related to range of motion or hamstring spasticity.

Discussion

This study is the first to investigate muscle activation patterns of the trunk and hip muscles during the early years of walking in children with TD and with CP. It also demonstrates the use of objective methods of analyzing EMG signals to determine periods of activity and inactivity across the gait cycle. All muscles from early walkers with CP except for the EO

demonstrated more total activity than those from children in the TD group. Increased activity for the CP group was observed as prolonged durations of bursts of activity and muscles that were continuously activated throughout the gait cycle compared to the TD group. Even for the EO, which was not significantly different between groups, more than half of the CP group had activity during 80% of the gait cycle compared to 39% of the gait cycle in the TD group.

Excessive muscle activity has been reported in lower extremity muscles in children with CP,^{87, 89} but these studies compared children to TD children of the same age, who had more walking experience because walking onset is delayed in CP. TD children with less walking experience are known to have more muscle activity than TD children with more walking experience.⁸⁰ Using walking experience, not age, for inclusion in this study is a novel approach to control for the maturation of walking ability that occurs with practice after the onset of walking. The increased muscle activity in the CP group in this study, however, was present even as the groups were compared by walking experience rather than age.

The patterns of muscle activity in the GMd and ST in the TD group were consistent with those reported previously by Sutherland and colleagues in their study of over 300 children between the ages of 1 and 7 years.⁸⁰ The period of activity in the GMx was slightly shorter than that reported in the Sutherland study. This difference can likely be attributed to differences in the methods used to determine activation between the two studies. Sutherland and colleagues used visual inspection of the raw EMG signals, while we used advanced processing methods and objective rules to determine activation patterns. While these 3 muscles are the only ones in the current study that have been previously reported for young TD children, muscle activity patterns during walking have been shown to approximate adult patterns after the age of 3 years.⁸⁰ For this reason, comparison of the other muscles included in this study to data from adults is justified.

Timing patterns of activity for the RF in the TD group, as well as the gluteal muscles, are consistent with those reported in adults.¹²⁹ White and Nair investigated patterns of activity in the ES, RA, and EO in adults.¹³⁰ They did not use a threshold to determine the timing of muscle activity onset and offset, but identified different patterns by averaging normalized amplitude curves across participants. Therefore, exact comparison is not possible, but areas of increased normalized amplitude for the ES and EO in their study correspond to periods of activity for the ES and EO in the TD group in the current study. Also similar to the TD group in the current study, the majority of adult participants in the White and Nair study did not have periods of activity above baseline in the RA. The comparison of the TD muscle activity data from the current study to those mentioned above demonstrates that, similar to lower extremity muscles, activation patterns of the trunk muscles approximate those of adult patterns by a young age.

Greater spasticity of the hip adductors was related to greater total percent of activity in the abdominal muscles (RA, EO) and percent coactivation of ES/RA muscle groups in the CP group. A significant, but weaker ($r=0.28$), relationship was reported by Poon and Hui-Chan¹³¹ between the soleus stretch reflex and plantarflexor coactivation ratio during isometric contraction in children with CP. These data may indicate that increased muscle activity is an adaptive strategy to maintain postural control during walking in children with greater hip spasticity.

Despite our use of automated EMG processing techniques and objective rules to determine the duration of muscle activity in each group, there remains no flawless method to analyze EMG signals, particularly in children with neurological impairments. Unlike the larger and thicker muscles of the thigh and gluteal region, the superficial muscles of the trunk are thin and the recording sensors may have recorded some activity from the underlying muscles. The internal oblique, rhomboids, and transverse abdominis muscles are directly deep to the sensor

location for the EO, TZ, and RA muscles, respectively. The ES muscle was recorded from a location that is deep to the broad superficial fascia of the latissimus dorsi muscle. The use of fine wire needle EMG electrodes would avoid this potential issue, but application in young children has clear feasibility and ethical limitations. Additionally, needle electrodes only record from a single or small group of motor units, and as a result, the recorded signal may not be representative of the activity of the entire muscle.⁶²

An additional limitation exists in the comparison in the TZ muscle activity between the TD and CP groups because the majority (12/15) of the children in the CP group used an assistive device for walking. Of the 3 children who walked without a device, one had TZ percent activation greater than the CP group mean (88%), but the other 2 had less activation (22% and 11%). Use of an assistive device alone may have contributed to greater activation of the TZ in the CP group, because the shoulders were engaged during forward movement and bearing weight through the assistive device. This issue is difficult to avoid when studying early walkers with CP. According to the GMFCS classification,⁷⁴ only children classified as level I (the least impaired) begin to walk without the use of any assistive device. Therefore, to study any children with greater severity of CP during the early years of walking, the use of walking aids must be allowed.

Several other studies have examined postural control in CP during static standing, during reaching, and during external perturbation balance testing.¹³²⁻¹³⁶ The results of these studies demonstrated increased coactivation, prolonged latency of activation, altered muscle recruitment order following perturbations, and continuous activation of lower extremity and postural muscles. The results of the present study demonstrate similar findings for muscle activation during walking, including increased coactivation and continuous activation of postural muscles.

With the exception of the RA and GMx, all muscles in CP group were active over 75% of the gait cycle, which may restrict the child's ability to grade muscle activity and make fine adjustments to trunk position in relation to the lower extremities and the environment. Roerdink and colleagues report that after a stroke, individuals had less stability but also more regularity in frontal plane COP trajectories during standing than healthy peers.¹³⁷ With recovery and rehabilitation, COP trajectories became less regular. The authors suggest that after stroke, the participants attempt to limit variations in COP in order to decrease the degrees of freedom that they must control. A similar strategy may occur in early walkers with CP. Hsue et al report reduced anterior-posterior displacements of COP and center of mass (COM) during walking in CP.¹³⁸ Limiting excursion and variability in COP and COM, by excessively activating muscles of the trunk and hips, may be a strategy for children with CP to maintain upright posture against gravity and move the body forward despite the multitude of neurological impairments limiting typical movement patterns.

The results of this study suggest that postural muscle control training during the early stages of walking in CP should be investigated to encourage the development of more functional and efficient movement strategies in these children. Core stability is related to athletic performance and function in healthy adults,^{139, 140} and the effects of improving core muscle control should be examined in CP. Furthermore, the theoretical framework behind interventions designed to reduce postural sway in children with CP should be examined. Strategies to increase the child's ability to control greater variations in trunk movement through phasic trunk muscle coordination, rather than constant muscle activity, may encourage more effective and efficient patterns of postural muscle control.

CHAPTER 4

TRUNK AND HIP MUSCLE ACTIVITY IN EARLY WALKERS WITH AND WITHOUT CEREBRAL PALSY – A FREQUENCY ANALYSIS

Introduction

Cerebral palsy (CP) is the most common neuromuscular disorder in children with an increasing prevalence,¹⁻³ high economic cost⁴ and negative impact on quality of life.⁵⁻⁷ CP is characterized by impairment in the development of movement and posture attributed to disturbances that occurred in the developing fetal or infant brain.⁹

Poor control of trunk postural muscles is a primary impairment in cerebral palsy (CP)^{9, 95} which causes compensation by other muscles to assist in maintaining upright posture. Compensation by accessory muscles to aide in posture reduces their effectiveness in functioning as primary movers of the extremities.⁹⁴ Evidence to support this notion includes observations that children with CP have greater ambulatory ability when the distal limb musculature is primarily affected, and proximal limb musculature is less affected.^{58, 87} Additionally, proximal limb muscles, such as those of the hip, are also critical for maintaining upright mobility. Compared to knee and ankle muscles, the strength of the hip abductors explained the largest variance in gait and gross motor function in CP.¹¹⁹ Given their importance for ambulatory ability, it is surprising that direct study of activation patterns of trunk and hip muscles during walking have not been investigated in individuals with CP.

Electromyographic (EMG) analysis is a critical component in the examination of gait in individuals with CP. Frequency analysis of EMG signals using the continuous wavelet transform allows the dynamic EMG signal to be decomposed into its individual frequency components as a

function of time. The frequency of an EMG signal contains information about the pattern of muscle fiber activation.¹⁴¹ These time-frequency characteristics of the EMG signal have been shown to correlate with functional measures in CP⁵⁸ and can indicate muscle function changes following interventions even when EMG timing information did not change.^{58, 75} Lastly, an attractive feature of EMG frequency analysis is that it does not require the participant to generate a maximal force contraction, as when using EMG signal amplitude to determine relative levels of activation between muscles, which is difficult to obtain in young children because of limited cognitive capabilities.

Individuals with CP use compensatory movement strategies as a result of decreased postural control, poor coordination, muscle weakness and spasticity. These abnormal motor strategies are reinforced with repetition,^{33, 36, 38} over time and throughout development. Therefore, it is critical to first examine muscle behavior in individuals with CP during the development of these compensatory postural and movement patterns to better design treatment interventions.

Early walkers are rarely included as research participants, in part because of limited attention, cognition, and tolerance to the cumbersome instrumentation involved in gait analysis. The immaturity of movement patterns in early walkers also complicates data interpretation. Mature gait is characterized by low variability⁹⁹ and a high degree of symmetry¹⁰⁰ from stride to stride. Children have greater stride to stride variability than adults.¹⁰¹ This variability should be reported in addition to mean values in order to better characterize immature walking patterns prior to the development of fixed maladaptive patterns. The variability of muscle activity has not been investigated during the early years of walking ability in individuals with CP. The objective of this study was to investigate differences in the time-frequency characteristics of the trunk and

hip muscles during the early stages of walking in children with CP compared to children with similar amounts of walking experience and typical development (TD). The variability of muscle activity was also investigated.

Methods

Participants

Participants with CP were recruited through the CP clinic at Shriners Hospital for Children in Philadelphia, PA, and through other local rehabilitation facilities. Participants with TD were recruited from siblings of the participants with CP, children of people known to the investigators, and from a local day care center. All procedures were approved by the IRB of Temple University Hospital (for Shriners Hospital) and also the IRBs of additional data collection sites as needed. Parental consent was obtained prior to participation. Assent of a minor was obtained from any participant 7 years of age or older.

The inclusion criteria for all children were: 1) 0.5-60 months of walking experience; 2) ability to ambulate barefoot at least 15 feet with supervision (children with CP could use their usual assistive device if it did not stabilize or restrict movement of the trunk or pelvis); and 3) ability to follow 1-step verbal directions. Children with CP additionally had 4) a diagnosis of spastic diplegia or quadriplegia; and 5) a Gross Motor Function Classification System (GMFCS) level of II or III.⁷⁴ Children were not considered for the study if they had a lower extremity surgery or fracture in the past 12 months, botulinum toxin injection in the past six months, or a history of dorsal rhizotomy or lower extremity tendon transfer.

The selection of months of walking experience rather than age as a primary inclusion criterion was chosen based on reports that experience is a stronger predictor of walking and

balance skill than age in early walkers.^{92, 93} The onset of walking was operationally defined as the age in months at which an infant or child was able to take at least 3 continuous independent steps on a consistent basis.^{102, 142} Walking experience, in months, was calculated as the difference in the participant's current age and the age of onset of walking.

Procedures

EMG

Surface EMG data from trunk, gluteal, and thigh muscles were acquired using a 16-channel recording system (Myomonitor III, Delsys Inc., Boston, MA) with preamplified silver-silver chloride parallel bar surface electrodes with a 10.0 mm interelectrode distance. EMG data were collected at 1200 Hz, preamplified with a gain of 10, and filtered with a high pass filter of 20 Hz and low pass filter of 450 Hz.

EMG data were collected from eight muscles bilaterally as listed in Table 4-1. The rectus femoris and semitendinosus were chosen in addition to the trunk and hip muscles because these muscles anatomically cross the hip joint, and they have been extensively investigated in CP, allowing for comparison of data to existing literature. Sensor placement for the abdominal muscles was determined using the methods described by Ng et al.¹²⁵ Sensor placement for all other muscles (back, gluteal, and thigh) was determined in accordance with SENIAM recommendations (Table 4-1).¹²⁶

The skin areas were cleaned with alcohol and the sensors were affixed to the skin with a double-sided adhesive interface (Delsys Inc., Boston, MA). The electrodes were further secured using hypoallergenic tape or a flexible, latex-free, non-adhesive wrap encircling the waist and thighs (Coflex-NL®, Andover Healthcare, Inc.). Self-adhesive reference electrodes (Axelgaard

Table 4-1. Electromyogram (EMG) Sensor Locations

Muscle	Sensor Location
Trapezius (middle) ¹²⁶	50% of the distance from the medial border of the scapula to the T3 spinous process, parallel to the line from T5 to the acromion
Erector Spinae (longissimus) ¹²⁶	1-2 finger widths lateral from L1 spinous process, oriented vertically
Rectus Abdominus ¹²⁵	At the level of the ASIS, 1-2 cm lateral to the midline, oriented vertically
External Oblique ¹²⁵	Just below the rib cage at the inferior angle of the ribs, oriented obliquely
Gluteus Maximus ¹²⁶	50% of the distance from sacral vertebrae to greater trochanter, at greatest prominence of the middle buttocks, parallel to line from PSIS to middle posterior thigh
Gluteus Medius ¹²⁶	In sidelying, 50% of the distance from iliac crest to greater trochanter, parallel to this axis
Quadriceps Femoris (rectus femoris) ¹²⁶	50% of the distance from ASIS to superior patella, parallel to this axis
Semitendinosus ¹²⁶	50% of the distance from ischial tuberosity to medial tibial epicondyle, parallel to this axis

Manufacturing Co., Ltd., Lystrup, Denmark) were placed on the skin over the patella bilaterally. The children were able to watch a video or play with a research assistant during sensor placement to increase tolerance and compliance.

A volitional contraction of each muscle, when possible, was elicited to verify placement. The children were asked to perform specific movements to elicit the corresponding specific muscle contractions, such as leaning backward in sitting to activate the rectus abdominis muscles, and standing on one leg (with hands held if needed) to activate the gluteus medius muscles. For synchronous data collection during walking trials, the EMG system was triggered by the instrumented walkway.

Walking Trials

Children walked barefoot down an instrumented walkway (GAITRite®, CIR Systems, Havertown, PA) at a self-selected pace. Three to 5 trials, each consisting of one walk down the walkway with at least 4 consecutive footfalls, were collected depending on participant tolerance to testing procedures and fatigue. Footfall information was collected at 30 Hz. All walking trials were videotaped.

Start and stop targets were placed on the floor approximately 5 feet beyond either end of the instrumented walkway so that the children were not accelerating or decelerating in the data capture space. During the walking trials, the EMG preamplification unit that is typically worn on a backpack was carried behind all participants by an assistant so as not to add additional weight, which could affect muscle activity in the smaller children. If needed, children were motivated and rewarded with stickers, small snacks, or favorite toys. Children had the opportunity to sit on a chair in between walking trials to minimize fatigue.

Data Analysis

Video footage of each trial was reviewed to determine the most appropriate gait cycles to select for data analysis. Ten gait cycles (5 left, 5 right) were selected based on the observation of each individual's typical walking (child was not distracted, did not stop walking, and was not moving arms toward an object).

The 10 selected gait cycles were extracted from the EMG files using the time-synchronized marker data (initial foot contact for consecutive footfalls) collected from the instrumented walkway. EMG data were processed using custom-written programs in MATLAB software (The Mathworks Inc., Natick, MA). All signals were normalized to 1000 points, representing the gait cycle from 0 to 100% in 0.1% increments.

A time-frequency pattern for each muscle was generated using the continuous wavelet transform (CWT) as described by Lauer and colleagues.⁵⁸ The CWT uses the Morlet wavelet to generate a 3-dimensional scalogram (frequency x amplitude x time). The instantaneous mean frequency (IMNF) was calculated at each interval of the gait cycle.

A functional principal component analysis (PCA) was completed using the IMNF curves from all gait cycles collected for each muscle. The PCA is a mathematical least squares maximization procedure that transforms a large number of correlated variables (IMNF curves for all participants) into a smaller number of uncorrelated variables (regions of variability across the gait cycle) called principal components. This allows for variability across an entire curve to be captured in a small subset of principal components. The first PC accounts for as much of the variability in the entire data set as possible, and each succeeding component accounts for the maximum of the remaining variability. Each individual IMNF curve (for each participant) is then assigned a weight for each PC. The value of the weight describes the degree of agreement or

disagreement between the individual IMNF curve and the variance identified by that particular PC.

To assess if the muscle IMNF curves generated from the data of the children with CP differed from children with TD, and at what points in the gait cycle, the PC weights were averaged for each group and tested using a Welch statistic to determine if the means between the groups were equal. Because the variances may be unequal, the Welch statistic is preferred to the F-statistic and is considered a more robust test.¹⁴³

Individual stride-to-stride standard deviations (SD) in mean frequency were calculated for the gait cycle as a function of time for each muscle and for each participant based on the 10 analyzed gait cycles (5 left, 5 right). Stride-to-stride variability was then calculated for each muscle across the gait cycle for each group. Statistical analyses of the variability curves were performed using a functional analysis of variance (fANOVA) model.¹⁴⁴ A Fourier basis function of 20 expansion terms was fit to each curve, imposing a roughness penalty to ensure smoothness of the function up to the second derivative. A linear model was constructed using the following formula:

$$\text{Variability}_{mg}(\%) = \mu(\%) + \alpha_g(\%) + \varepsilon_{mg}(\%) \quad (1)$$

where g indicated the group, m is the number of functions representative of that group, μ is the grand mean function across all the functions, α are the specific effects on the function of being within a group g , and ε is the unexplained variation specific to the m^{th} curve. The terms for each of the general linear models was determined using a least squares fit. To be able to define a set of terms unique to each group, an additional constraint was placed that $\sum \alpha_g(\%) = 0$ for all gait cycle increments. Overall differences in variability were assessed using the F-ratio function, based on the F-statistic, calculated for each interval of the gait cycle.

Results

Thirty-four children enrolled in this study. Data from three children were excluded due to one having a questionable diagnosis of CP and two who were unable to walk without additional assistance from an investigator during the testing session. Data for the remaining 31 children were used for analysis. Walking experience did not differ between groups ($p=0.969$). Of the children with CP, seven were classified as GMFCS level II and eight were level III. One was classified as spastic quadriplegia, and 14 as spastic diplegic. Three walked without assistive devices, 9 used posterior rolling walkers, 1 used bilateral forearm crutches, and 2 used unilateral forearm crutches. Demographic and anthropometric data are provided in Table 4-2.

The principal component output for the gluteus maximus is shown as an example in Figure 4-1. Principal components 1, 2, and 4 show the direction of variance for the CP group and PC 3 shows the direction of variance for the TD group. The group not shown for each PC (TD for 1, 2, 4 and CP for 3) demonstrates equal variance in the opposite direction. When combined, the major regions of variance encompass the entire gait cycle, with the CP group demonstrating significantly higher mean frequency throughout.

Average IMNF curves for each muscle are shown in Figure 4-2. The first four PCs accounted for 97.4% of the variability in the erector spinae IMNF curves, 98.4% of the variability for the semitendinosus, 98.6% for the rectus abdominus, 98.8% for the trapezius and external abdominal oblique, 98.9% for the gluteus medius, 99.3% for the quadriceps femoris, and 99.6% for the gluteus maximus. All four PCs were significantly different between the TD and CP groups for each muscle ($p<0.001$). Examination of the specific regions of variability identified by each principal component revealed that the CP group demonstrated higher IMNF than the TD group throughout the gait cycle for all muscles.

Table 4-2. Demographic and Anthropometric Data

		Onset of walking (mo)	WE (mo)	Sex	Age (mo)	Weight (kg)	Height (m)	BMI (kg/m ²)	Seated Height (m)
TD (n=16)	Mean (SD)	11.7 (3.1)	28.6 (19.6)	9F 7M	39.7 (19.5)	15.1 (3.9)	0.97 (0.13)	15.9 (1.8)	0.55 (0.06)
	Range	8.0-20.0	1.0-58.0		13.0-67.5	10.0-21.9	0.75-1.18	11.2-18.8	0.46-0.69
CP (n=15)	Mean (SD)	34.8 (10.2)	28.4 (17.0)	5F 10M	63.1 (23.2)	19.6 (5.9)	1.06 (0.14)	17.2 (2.4)	0.56 (0.06)
	Range	18.0-55.0	2.0-60.0		25.0-108.0	10.9-31.2	0.83-1.32	14.7-22.9	0.48-0.65
Total (n=31)	Mean (SD)	22.9 (13.8)	28.5 (18.1)	14F 17M	51.0 (24.1)	17.3 (5.4)	1.10 (0.14)	16.5 (2.2)	0.56 (0.06)
	Range	8.0-55.0	1.0-60.0		13.0-108.0	10.0-31.2	0.75-1.32	11.2-22.9	0.46-0.69

TD=typically developing, CP=cerebral palsy, WE=walking experience, mo=months, SD=standard deviation, M=male, F=female, kg=kilograms, m=meters, BMI=body mass index

Stride-to-stride variability curves are presented in Figure 4-3. Variability tended to increase during the points in the gait cycle when the muscle was more active. The $F(2,57)$ critical value was 3.16 for an alpha level of 0.05. Individual variability was statistically higher in the CP group for all muscles across the gait cycle.

Discussion

This study reports trunk and hip muscle activity during the early stages of walking in children with CP compared to children with TD. Mean frequency of muscle activation during walking was higher and more variable from stride-to-stride throughout the gait cycle for the CP group than TD for all eight muscles investigated. The higher mean frequency in the CP group suggests altered strategies of muscle activation. Higher IMNF can result from increased rates of

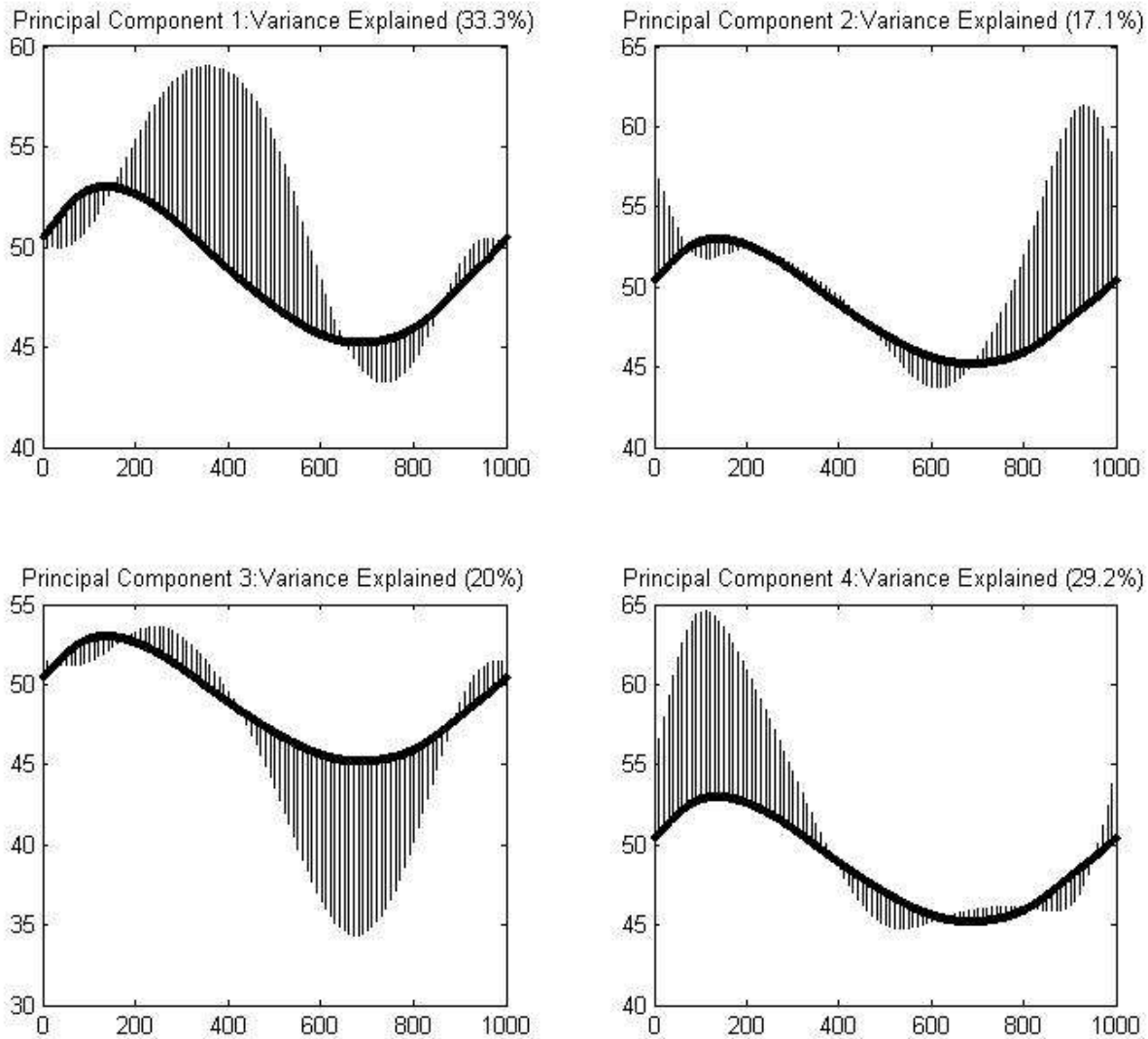


Figure 4-1. The principal component output for the gluteus maximus, with 99.6% of the variance explained. The dark curve in each graph is the mean instantaneous mean frequency for all participants, and is the same in each graph. The shaded areas indicate the regions of variability in the entire data set identified by each principal component. Principal components 1, 2, and 4 show the direction of variance for the cerebral palsy (CP) group and principal component 3 shows the direction of variance for the typical developing (TD) group. The group not shown for each principal component demonstrates equal variance in the opposite direction.

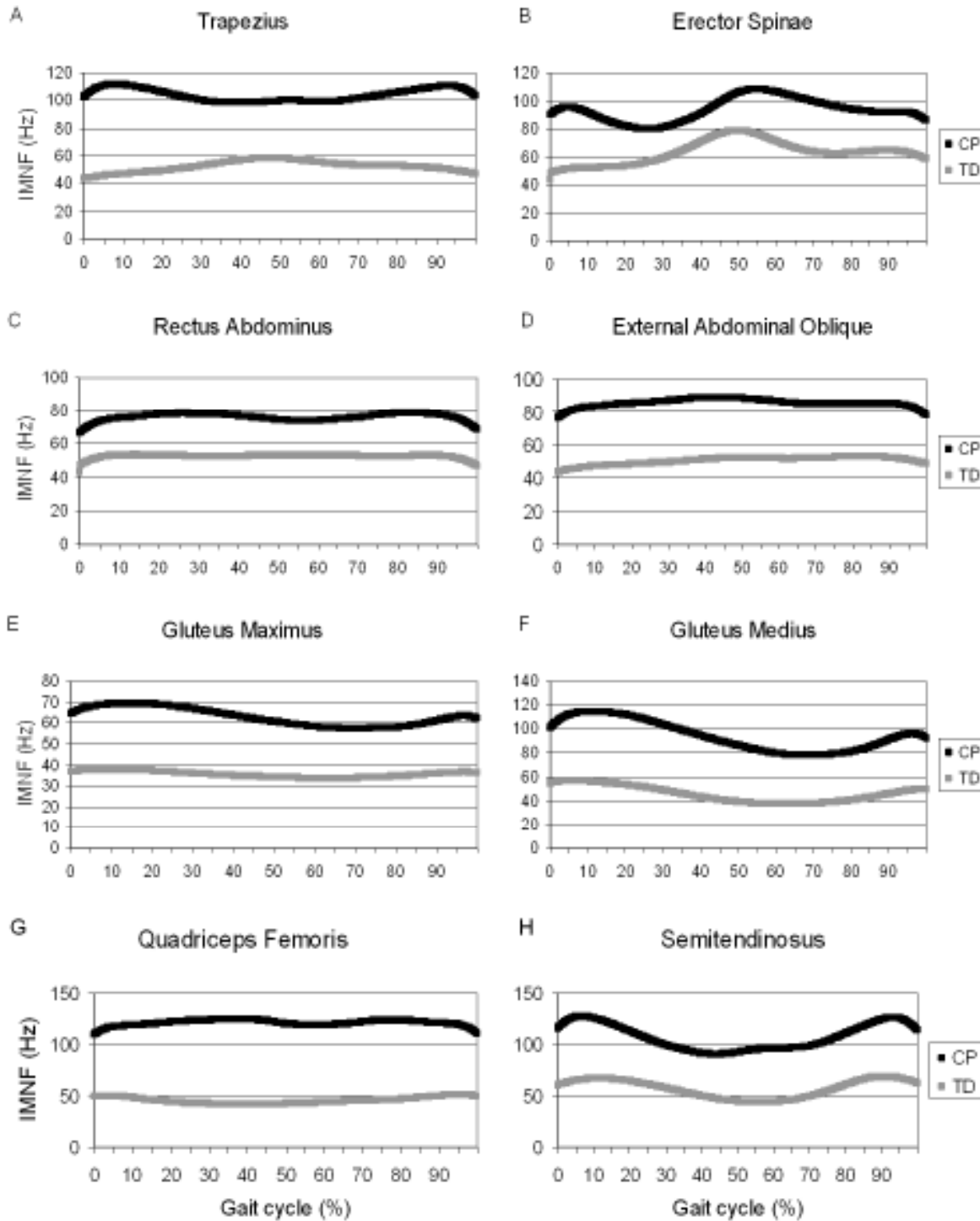


Figure 4-2. Instantaneous mean frequency (IMNF) mean curves across the gait cycle for children with typical development (TD) and cerebral palsy (CP) for the trapezius (A), erector spinae (B), rectus abdominus (C), external oblique (D), gluteus maximus (E), gluteus medius (F), quadriceps femoris (G), and semitendinosus (H).

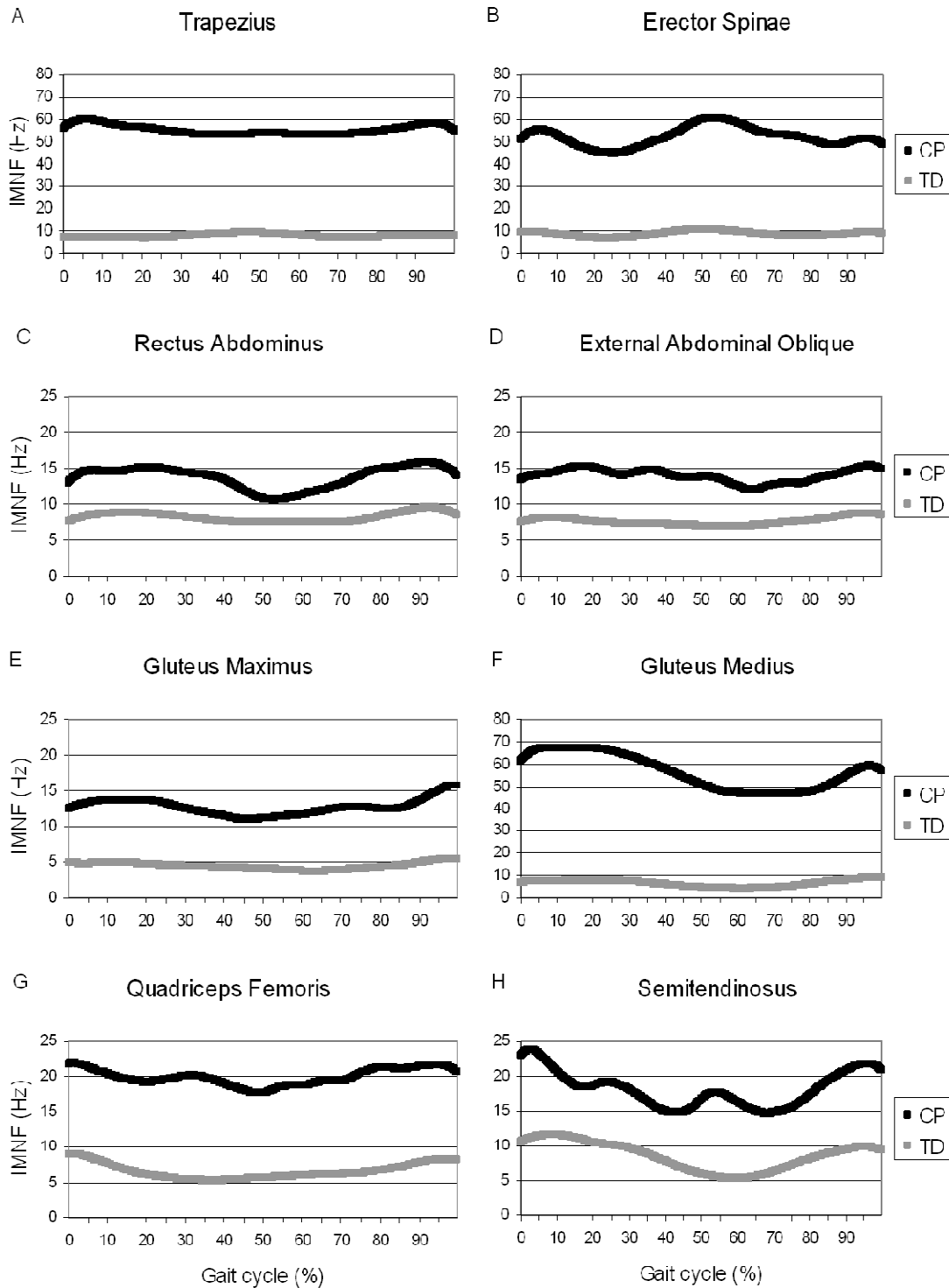


Figure 4-3. Stride-to-stride variability in instantaneous mean frequency (IMNF) across the gait cycle for children with typical development (TD) and cerebral palsy (CP) for the trapezius (A), erector spinae (B), rectus abdominus (C), external oblique (D), gluteus maximus (E), gluteus medius (F), quadriceps femoris (G), and semitendinosus (H).

motor unit firing, increased number of recruited motor units, or decreased synchrony of motor units.¹⁴¹ This is consistent with literature suggesting excessive and dyscoordinated muscle activity in CP.^{89, 145} Lam et al. has suggested that higher EMG median frequency contributes to muscle fatigue in children with CP.¹⁴⁶ Additionally, excessive muscle activity is related to decreased biomechanical efficiency.⁷⁶ The markedly increased stride-to-stride variability in muscle activation patterns in the CP group further implies dyscoordination and immaturity of muscle behavior.⁹⁹

Wakeling et al⁵⁹ reported mean frequency of lower extremity muscles for 36 children and adolescents with TD and 17 with spastic diplegic CP. The mean frequency for the rectus femoris in the CP group is consistent with that obtained in this study. The CP mean frequency reported for the semimembranosus is slightly higher than that obtained in the current study for the semitendinosus muscle. The TD data for both of these muscles in the current study is lower than that reported for the control group in Wakeling's study, which may be a result of the older age of the control group in that study (mean age 10.8 years).

Similarly, the IMNF values are consistent with those reported by Lauer and colleagues for the medial hamstrings and quadriceps in the CP group, and at the low end of the range for the TD group.⁵⁸ The control group in Lauer's study had the same mean age as the control group in Wakeling's study (10.8 years). Higher mean frequency in children with CP has also been reported during recumbent cycling.¹⁴⁷ To our knowledge, no previous studies have investigated mean frequency characteristics of the trunk or gluteal muscles in children with TD or with CP.

White and colleagues investigated trunk muscle activity during walking in 38 able-bodied adults.¹³⁰ Whereas they considered the amplitude of the EMG signal, not the frequency, the patterns of activity in the rectus abdominus, external oblique and erector spinae were similar to

patterns of increased frequency in both groups for this study. This indicates that the timing of muscle activation bursts alone may not be sufficient to examine muscle behavior in individuals with neurological impairments.

Based on this evidence of excessive and asynchronous trunk and hip muscle activity in CP, we suggest that postural muscle training should be a component of rehabilitation programs. Postural muscle training during the development of upright postural control may be the most effective time to aggressively intervene and change these compensatory movements, prior to years of reinforcing poor trunk movement patterns. Core muscle control training has been effectively applied to several other clinical populations,^{140, 148, 149} but has not been systematically investigated in children with neurological disorders.

The results of this study may not be generalizable to children with greater or lesser severity of CP or to other children with neurological disorders. Additionally, the data from the trapezius muscle should be interpreted with caution due to the use of an assistive device in the majority (12/15) of the children with CP. Activity of the trapezius muscle would be expected to be altered with the use of a hand-held assistive device.

This study demonstrates increased mean frequency in the trunk and hip muscles in early walkers with CP. Stride-to-stride variability in mean frequency was also increased in CP. The increased muscle activation and decreased synchrony in the trunk muscles are consistent with muscle behavior of the lower extremities in CP. The clinical implications of this work are that postural muscle training during the early stages of walking in CP should be investigated to encourage the development of more effective and stable movement strategies in these children.

CHAPTER 5

GAIT AND MUSCLE FUNCTION IMPAIRMENTS IN EARLY WALKERS WITH CEREBRAL PALSY – RELATIONSHIP TO GMFCS

Introduction

Cerebral palsy (CP) is the most common neuromuscular disorder in children with an increasing prevalence,³ a high economic cost,⁴ and a negative impact on quality of life.⁵ CP is characterized by impairment in the development of movement and posture attributed to disturbances that occurred in the developing fetal or infant brain.⁹

General mobility and ambulatory ability in children with CP are commonly classified by the Gross Motor Function Classification System (GMFCS).⁷⁴ GMFCS levels range from I to V with higher numerical classification indicating increasing degree of mobility limitations. Several authors have reported an increasing degree of severity in impairment-based measures consistent with increasing levels of the GMFCS, including energy cost of walking,¹⁵⁰ spasticity,¹⁵¹ and muscle activity.⁵⁸

These studies included participants who were between the ages of 6 and 20 years. There is a need to determine if differences in spatiotemporal gait measures and muscle activity patterns are present during the early years of walking, or only after years of repetition and reinforcement of compensatory movement patterns. Specifically, poor control of postural muscles is considered a primary impairment in CP, contributing to compensation by other muscles to assist in maintaining upright posture.⁹ However, it is unknown if impairments in core trunk and hip muscle activation differ with the GMFCS classification levels. Recent advances in

electromyographic (EMG) signal analysis to analyze the timing and frequency components of EMG signals may assist in identifying differences if they exist.⁵⁸

Musculoskeletal impairments and walking limitations and are known to increase over time in children with CP.¹⁵² It may be important to group children by walking experience rather than chronological age to control for changes in walking function that occur with walking experience. Previous work has reported that experience is a stronger predictor of walking ability and balance skill than age in typically developing infants.^{92, 93} This is a novel approach for CP research because the majority of studies compare groups of children who are the same age, rather than comparing groups who have similar amounts of walking experience, despite potential differences in the age when walking onset occurs.

The data for this study were collected during a larger study comparing early walkers with CP to those with typical development (TD). This analysis compares 2 groups, children classified as GMFCS level II and children classified as GMFCS level III. The objective of this study is to investigate differences in spatiotemporal gait parameters and muscle activity in early walkers with CP and different levels of classification on the GMFCS.

Methods

Participants

Participants were recruited through the CP clinic at Shriners Hospital for Children in Philadelphia, PA, Children's Specialized Hospital in Mountainside, NJ and through other local rehabilitation facilities. The institutional review board (IRB) of Temple University Hospital (for Shriners Hospital), and the IRBs of additional data collection sites as needed, approved all procedures. All data collection procedures were explained and parents gave informed consent to

the research and to publication of the results. Assent of a minor was also obtained from participants 7 years of age or older.

Inclusion and exclusion criteria are listed in Table 5-1. Children were permitted to walk with a hand held assistive device if it did not restrict movement of the trunk or pelvis (i.e. walker without pelvic guide or crutches). The selection of months of walking experience rather than age as a primary inclusion criterion was chosen based on reports that experience is a stronger predictor of walking and balance skill than age in early walkers.^{92, 93} The onset of walking was operationally defined as the age in months at which an infant or child was able to take at least 3 continuous independent steps on a consistent basis.¹⁰² Walking experience, in months, was calculated as the difference between the participant's age on the day of the study and the age of onset of walking.

Procedures

Musculoskeletal Measurement

All anthropometric and musculoskeletal measurements were taken by one pediatric physical therapist. The anthropometric measures included height, seated height, weight, and bilateral leg length (anterior superior iliac spine to the apex of the medial malleolus). All anthropometric lengths were measured with a Harpenden anthropometer (Holtain Limited, Pembrokeshire, UK). Range of motion (ROM) measures included bilateral hip extension, hamstring length, and ankle dorsiflexion. Muscle stiffness was measured bilaterally in the hamstrings and hip adductors in children with CP using the modified Tardieu scale.¹⁰⁵ A detailed description of these procedures is provided in Chapter 2. To increase tolerance and compliance

Table 5-1. Participant Inclusion and Exclusion Criteria

Inclusion	Exclusion
<ul style="list-style-type: none"> • 0.5-60 months of walking experience • Able to ambulate barefoot at least 15 feet in a forward direction with supervision (children with CP could use assistive device) • Able to follow 1-step verbal instructions • Spastic diplegia or quadriplegia • GMFCS II-III classification⁷⁴ 	<ul style="list-style-type: none"> • Lower extremity bony or soft tissue surgery or fracture in the past 12 months • Spastic hemiplegic or non-spastic classification • History of dorsal rhizotomy • History of tendon transfer to a target muscle • Botulinum toxin injection to a lower extremity muscle in the past 6 months • Secondary orthopedic, neuromuscular or cardiovascular condition

with musculoskeletal measurements, children were distracted with videos or played with a research aide during the examination.

EMG

Surface electromyographic (EMG) data from 8 trunk and lower extremity muscles were acquired bilaterally using a 16-channel recording system (Myomonitor III, Delsys Inc., Boston, MA) with preamplified silver-silver chloride parallel bar surface electrodes with a 10.0 mm interelectrode distance. Data were collected from the trapezius, erector spinae, rectus abdominis, external oblique, gluteus maximus, gluteus medius, rectus femoris, and semitendinosis. Sensor placement for the abdominal muscles was determined using the methods described by Ng et al.¹²⁵ Sensor placement for all other muscles (back, gluteal, and thigh) was determined in accordance with the SENIAM recommendations.¹²⁶ EMG data were collected at 1200 Hz, preamplified with a gain of 10, and band pass filtered from 20-450 Hz.

The skin areas were cleansed with alcohol and the sensors were affixed to the skin with a double-sided adhesive interface (Delsys Inc., Boston, MA). The electrodes were further secured using hypoallergenic tape or a flexible, latex-free, non-adhesive wrap encircling the waist and thighs (Coflex-NL®, Andover Healthcare, Inc.). Self-adhesive reference electrodes (Axelgaard Manufacturing Co., Ltd., Lystrup, Denmark) were placed on the skin over the patella bilaterally. A volitional contraction of each muscle, when possible, was elicited to verify placement. A 5-second resting EMG trial was collected in supine prior to the walking trials to establish baseline muscle activity. For synchronous data collection during walking trials, the EMG system was triggered by the instrumented walkway.

Walking Trials

Children walked barefoot down an instrumented walkway (GAITRite®, CIR Systems, Havertown, PA) at a self-selected pace. Three to 5 trials, each consisting of 1 walk down the walkway with at least 4 consecutive footfalls, were collected depending on participant tolerance to testing procedures and fatigue. Footfall information was collected at 30 Hz.

Start and stop targets in child-friendly colors and patterns were placed on the floor approximately 5 feet beyond either end of the instrumented walkway to minimize acceleration or deceleration while walking on the walkway. A walking trial started by having the child stand on the start target. Data collection was initiated through the GAITRite® software, which triggered EMG collection through a trigger module (Delsys Inc., Boston, MA) for synchronous recording. The child was then instructed to walk to the target beyond the opposite end of the walkway. If needed, children were motivated and rewarded with stickers, small snacks, or favorite toys. Children had the opportunity to sit on a chair in between walking trials to minimize fatigue.

During the walking trials, the EMG preamplification unit that is typically worn on a backpack was carried behind all participants by an assistant so as not to add additional weight, which could affect muscle activity in the smaller children. Walking trials were videotaped for later gait cycle selection and parents/caregivers signed a separate consent to allow videography.

Data Analysis

Video of each walking trial was reviewed to determine the most appropriate gait cycles to select for data analysis. Five gait cycles for each side (left and right) were selected based on the observation of typical walking (child was not distracted, did not stop walking, and was not moving arms toward an object).

Spatiotemporal Measures

The primary spatiotemporal parameters included: walking velocity, cadence, step length, base of support, and single support time (as a percent of the gait cycle). Step length and walking velocity measures were normalized to leg length and converted to dimensionless values.¹¹¹ Step length was divided by leg length. Walking velocity was scaled using the following equation: $v' = v(g\ell_0)^{-0.5}$ where v' represents dimensionless walking velocity, v represents measured walking velocity (m/s), g is the gravitational acceleration constant (9.8 m/s^2), and ℓ_0 represents leg length (m).

Individual means and stride-to-stride standard deviations (SD) were calculated for each parameter by averaging the 10 values (5 left, 5 right) from the selected gait cycles for each participant. In addition, a symmetry ratio was calculated for each parameter by dividing the smaller left or right value by the larger value. This resulted in a value between 0.0 and 1.0, with values closer to 1.0 indicating greater symmetry. Group means were then calculated for each parameter, and each variability and symmetry measure. Assuming unequal variance between groups, Mann-Whitney tests were performed to determine group differences between children classified as GMFCS level II and III ($p < 0.05$).

Muscle Activity

The ten selected gait cycles were extracted from the EMG files using the time-synchronized marker data (initial foot contact for consecutive footfalls) collected from the instrumented walkway. EMG data were processed using custom-written programs in MATLAB software (The Mathworks Inc., Natick, MA). All signals were normalized to 1000 points, representing the gait cycle from 0 to 100% in 0.1% increments.

To identify muscle activity throughout the selected gait cycles, the Teager-Kaiser Energy (TKE) Operator was applied to the EMG data.¹²⁷ A detailed description of the technique and its use in this study is provided in Appendix A. Total activation and coactivation were analyzed as a percent of the gait cycle. A percent of gait cycle activation was calculated for each muscle by summing the duration of all periods of muscle activity. Coactivation was determined by calculating the total time antagonistic muscles were simultaneously active. Coactivation was calculated for the rectus abdominis/erector spinae muscle pairs and the rectus femoris/ semitendinosus muscle pairs. Group means were calculated for percent activation and coactivation. These measures allowed for comparison of total relative time a particular muscle was active in each group. Assuming unequal variance between groups, Mann-Whitney tests were used to determine differences in percent activation and coactivation ($p < 0.05$).

A time-frequency pattern for each muscle was also generated using the continuous wavelet transform described by Lauer and colleagues⁵⁸ to examine the frequency component of the EMG signals. Examination of EMG mean frequency can provide insight into the neurological input that causes muscle activation. A functional principal component analysis (PCA) was performed using the instantaneous mean frequency (IMNF) curves from all gait cycles collected for each muscle. The first principal component accounts for as much of the variability in the data set as possible, and each succeeding component accounts for the maximum of the remaining variability. To assess if the muscle IMNF curves generated from the data of the children who were GMFCS level II differed from children who were GMFCS level III, and at what points in the gait cycle, the 4 PC weights for each muscle were tested using a Welch statistic ($p < 0.05$).

Results

Eighteen children participated in this study. Data from 3 children were excluded: one had a questionable diagnosis of CP, and two were unable to walk without additional assistance from an investigator during the testing session. Data for the remaining 15 children were used for analysis. Seven were classified as GMFCS level II and 8 were level III. One was classified as spastic quadriplegic, and 14 as spastic diplegic. Three walked without assistive devices, 9 used posterior rolling walkers, 1 used bilateral forearm crutches, and 2 used unilateral forearm crutches. ROM data were not obtained from 1 child due to time constraints. Demographic and musculoskeletal data are provided in Tables 5-2 and 5-3, respectively. Walking experience did not differ between children who were GMFCS level II and those who were level III ($p=0.168$). Anthropometric and musculoskeletal measures also did not differ between groups (p values ranging from 0.211-0.551 and 0.077-0.926, respectively).

Spatiotemporal Measures

Group means for the primary spatiotemporal parameters, the individual stride-to-stride variability measures (SD), and the symmetry ratios are presented in Table 5-4. There were no differences in the primary spatiotemporal parameters between children who were GMFCS level II and those who were level III. Variability was different between groups for cadence ($p=0.021$), with the GMFCS level III group demonstrating greater stride-to-stride variability. Left to right symmetry was significantly lower in the GMFCS level III group for walking velocity ($p=0.025$), cadence ($p=0.037$), and single support time ($p=0.035$).

Table 5-2. Demographic and Anthropometric Data

		Onset of walking (mo)	WE (mo)	Sex	Age (mo)	Weight (kg)	Height (m)	BMI (kg/m ²)	Seated Height (m)
GMFCS II (n=7)	Mean (SD)	30.9 (5.8)	21.8 (17.7)	4F 3M	52.6 (19.7)	17.6 (5.1)	1.02 (0.12)	16.7 (1.7)	0.55 (0.05)
	Range	23.0-36.0	2.0-43.0		25.0-79.0	12.5-27.0	0.83-1.17	14.9-19.7	0.48-0.64
GMFCS III (n=8)	Mean (SD)	38.2 (12.3)	34.1 (15.0)	1F 7M	72.3 (23.2)	21.4 (6.2)	1.10 (0.14)	17.6 (3.0)	0.57 (0.06)
	Range	18.0-55.0	8.0-60.0		37.0-108.0	10.9-31.2	0.83-1.32	14.7-22.9	0.49-0.65
Total (n=15)	Mean (SD)	34.8 (10.2)	28.4 (17.0)	5F 10M	63.1 (23.2)	19.6 (5.9)	1.06 (0.14)	17.2 (2.4)	0.56 (0.06)
	Range	18.0-55.0	2.0-60.0		25.0-108.0	10.9-31.2	0.83-1.32	14.7-22.9	0.48-0.65

GMFCS=Gross Motor Classification System, WE=walking experience, mo=months, SD=standard deviation, F=female, M=male, kg=kilograms, m=meters, BMI=body mass index

Table 5-3. Participant Musculoskeletal Characteristics

		ROM*		Spasticity*						
		HExt (deg)	Hams (deg)	Dorsi (deg)	Hip adductors			Hamstrings		
					R1 (deg)	R2 (deg)	X	R1 (deg)	R2 (deg)	X
GMFCS II	Mean (SD)	-1 (1)	132 (10)	5 (8)	19 (10)	34 (13)	3.1 (1.1)	93 (12)	132 (10)	3.9 (0.4)
	Range	-3-0	118-148	-4-16	6-33	22-62	1-4	68-105	118-148	3-4
GMFCS III	Mean (SD)	-3 (3)	134 (10)	10 (10)	15 (8)	29 (12)	3.8 (0.5)	95 (10)	134 (10)	3.9 (0.4)
	Range	-9-0	123-150	-5-27	2-27	15-54	3-4	81-109	123-150	3-4
Total	Mean (SD)	-2 (2)	133 (9)	7 (9)	17 (8)	31 (12)	1.7 (0.4)	94 (10)	133 (9)	1.9 (0.2)
	Range	-9-0	118-150	-5-27	2-33	15-62	0.5-2	68-109	118-150	1.5-2

GMFCS=Gross Motor Classification System, SD=standard deviation, ROM=range of motion, HExt= hip extension, Hams=hamstrings, Dorsi=dorsiflexion, *average of left and right sides, except X value is the sum of left and right sides

Table 5-4. Group Means (and SD) for Spatiotemporal Parameters, Stride-to-Stride Variability (SD), and Symmetry

	GMFCS II			GMFCS III			<i>p</i> values		
	Value	SD	SR	Value	SD	SR	Value	SD	SR
Walking velocity [^]	2.60 (1.15)	0.47 (0.27)	0.91 (0.07)	1.82 (0.87)	0.72 (0.30)	0.66 (0.32)	0.132	0.149	0.025*
Cadence (steps/min)	110.8 (39.2)	12.1 (8.4)	0.95 (0.04)	96.8 (34.8)	30.1 (23.0)	0.80 (0.16)	0.487	0.021*	0.037*
Step length [^]	0.63 (0.14)	0.10 (0.02)	0.92 (0.04)	0.48 (0.15)	0.14 (0.09)	0.75 (0.35)	0.064	0.873	0.688
Base of support (cm)	4.4 (8.6)	4.4 (3.3)	0.70 (0.33)	8.6 (5.4)	3.6 (1.5)	0.76 (0.21)	0.522	0.827	0.724
Single support (% gait cycle)	75.3 (13.8)	9.6 (10.4)	0.93 (0.04)	80.0 (9.0)	11.5 (6.1)	0.76 (0.15)	0.668	0.291	0.035*

GMFCS=Gross Motor Function Classification System, SD=standard deviation, SR=symmetry ratio, min=minute, cm=centimeters, %=percent, [^]dimensionless values normalized to leg length, *indicates significant difference ($p < 0.05$)

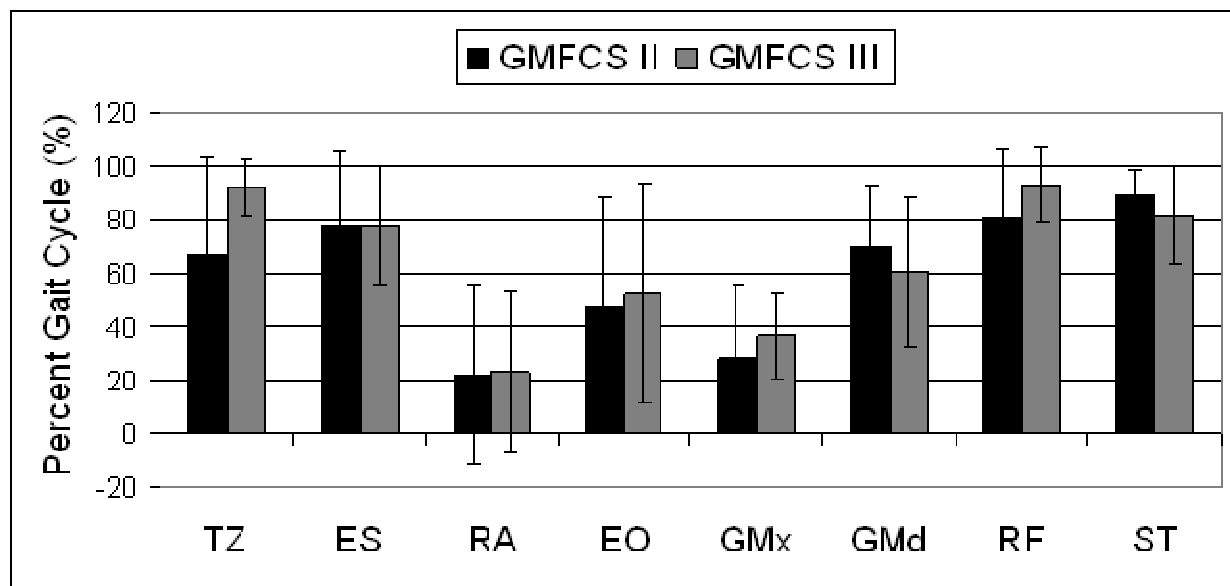


Figure 5-1. Mean total percent activation of muscles for children who are level II and III on the Gross Motor Function Classification System (GMFCS). Bars represent one standard deviation. There were no significant differences between groups. Trap=trapezius, ES=erector spinae, RA=rectus abdominis, EO=external oblique, GMax=gluteus maximus, GMed=gluteus medius, RF=rectus femoris, ST=semitendinosus.

Muscle Activity

Activation from left and right sides did not differ within groups, thus left and right side data were combined for percent activation and coactivation measures. Group means for activation are shown in Figure 5-1. There were no differences between the GMFCS level II and level III groups for activation (p values ranging from 0.195-0.862). The groups were also not different in percent coactivation over the gait cycle. Coactivation was 19.1% and 20.2% in the GMFCS level

II and III groups, respectively, for the rectus abdominis/erector spinae (RA/ES, $p=0.637$).

Coactivation was 71.9% and 77.2% in the GMFCS level II and III groups, respectively, for the rectus femoris/semitendinosus (RF/ST, $p=0.772$).

In the EMG frequency analysis, the first 4 PCs accounted for 97.1-99.5% of the variability in the curves for each of the muscles. Average IMNF curves for each muscle are shown in Figure 5-2. Asterisks (*) indicate regions of significant difference between groups. There were no differences across the gait cycle between groups for the gluteus maximus, gluteus medius, and semitendinosus. The first PC for the RA was different between groups, indicating lower mean frequency in the GMFCS III group compared to GMFCS II from late stance through early swing. For the RF, the first and fourth PCs were different, accounting for higher mean frequency in the GMFCS III group during stance phase. For the ES and external oblique (EO), both the second and fourth PCs were different between groups, indicating higher mean frequency in the GMFCS III group from late stance through swing for the ES and during stance for the EO. All four PCs were significantly different between the two groups for the trapezius, accounting for higher GMFCS III mean frequency across the gait cycle compared to GMFCS II.

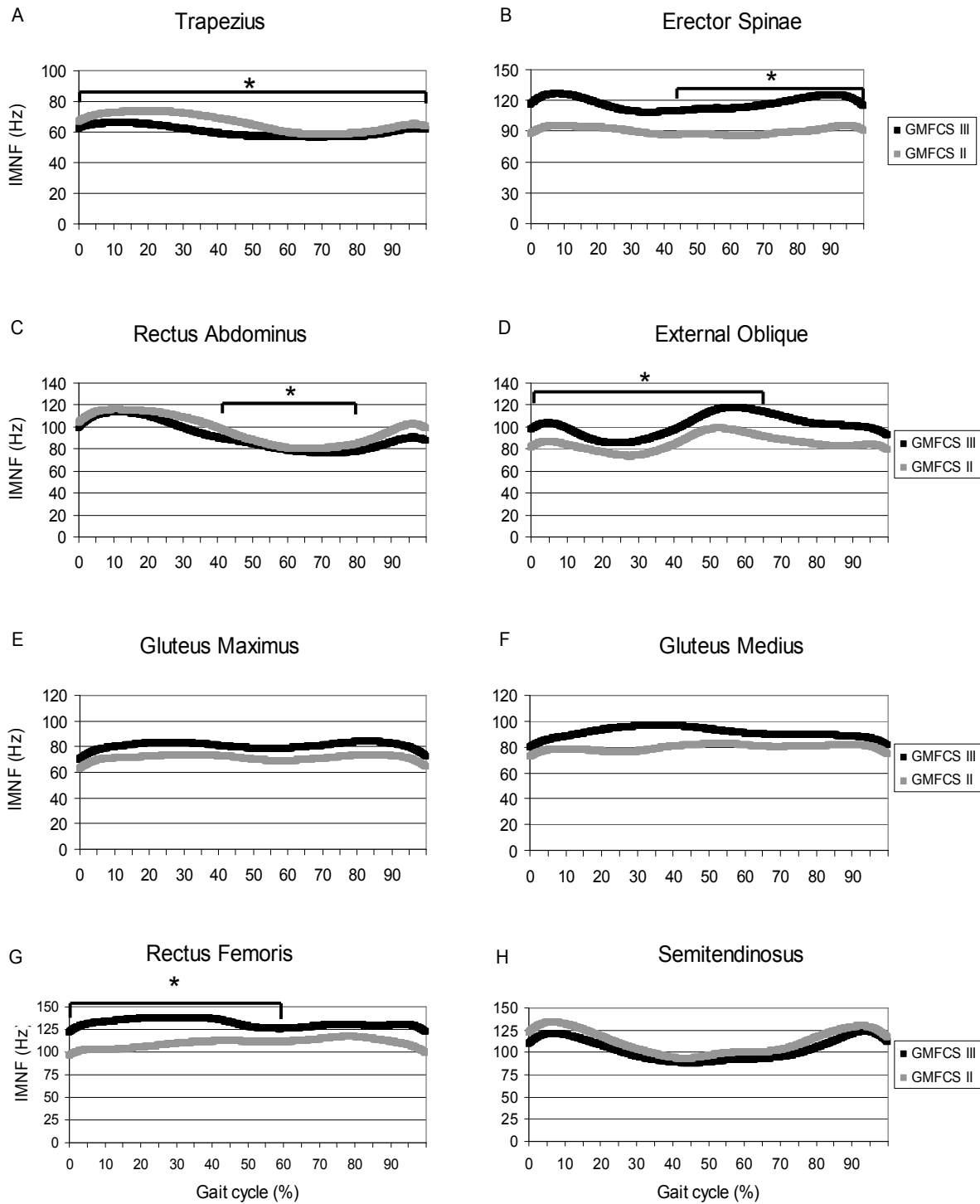


Figure 5-2. Instantaneous mean frequency (IMNF) mean curves across the gait cycle for children who were GMFCS level II and level III for the trapezius (A), erector spinae (B), rectus abdominus (C), external oblique (D), gluteus maximus (E), gluteus medius (F), quadriceps femoris (G), and semitendinosus (H). Asterisks (*) indicate areas of the gait cycle where IMNF differed between groups.

Discussion

This study investigated differences in impairment measures during the early years of walking in children with CP who had different levels of ambulatory ability. Other than variability in cadence, there were no differences in spatiotemporal parameters or spatiotemporal stride-to-stride variability between the children classified as GMFCS level II and level III. This is inconsistent with McDowell and colleagues' work that found walking speed differences by GMFCS level in older children with CP.¹⁵³ However, left-right symmetry was different between groups in the present study for walking velocity, cadence, and single support time. The GMFCS III group demonstrated less symmetry in each case. Symmetry is a characteristic of mature gait¹⁰⁰ and less symmetry indicates greater impairment in walking ability. These data suggest that symmetry, therefore, may be an important measure to identify impairments during the early years of walking, before differences in spatiotemporal parameters may be apparent.

For muscle activity, there were no differences in the general measures of total percent activation and coactivation over the gait cycle for any of the muscles studied. There were, however, differences in mean frequency of EMG signals for several of the muscles, including all 4 trunk muscles. In general, the group classified as GMFCS III had higher mean frequency than the GMFCS II group. This is consistent with previous work in older children indicating greater mean frequency with greater severity of functional ability.⁵⁸ Differences in mean frequency suggest altered strategies of muscle activation. Higher IMNF can result from increased rates of motor unit firing, increased number of recruited motor units, or decreased synchrony of motor units.¹⁴¹ These muscle activation results suggest that some measures of muscle activity may be more sensitive to degree of impairment than others, and that some muscles may demonstrate differences earlier than others. When available, it may be important to use advanced EMG

analysis techniques to detect early differences in muscle activation patterns. In fact, frequency characteristics of the EMG signal have indicated muscle function changes following interventions even when EMG timing information did not change.⁷⁵

Early detection of impairment in CP is important to appropriately tailor interventions during the early phases of walking skill development. Focused and aggressive intervention, before compensatory movement patterns are repeated and reinforced for years, is important to maximize neuroplasticity and functional gains after injury to the central nervous system.^{33, 42, 98} To identify which impairments to target, it is necessary to both use measures that are able to distinguish degree of impairment early in the skill development, and to measure the impairments that begin to differentiate by severity earlier than others.

This study is a beginning to investigate impairment differences by GMFCS level during the early years of walking in CP. However, there are several limitations that should be addressed in future work. Although the difference was not statistically significant, the GMFCS level III group had 12 months more of walking experience than the GMFCS II group. The differences between groups may have been greater if walking experience was equal between groups. Additionally, this study included a small sample with a limited range of GMFCS levels, which again may have limited the detection of group differences. Future work should include more participants and all 4 of the ambulatory levels of the GMFCS. Different impairment measures than the ones included in this study should also be investigated, including joint kinematics and kinetics during walking, muscle strength and spasticity, and energy expenditure.

Differences in spatiotemporal gait symmetry are apparent in the early years of walking between children with CP classified as GMFCS level II and level III. Differences in the instantaneous mean frequency of EMG signals are also apparent, but not in total muscle

activation time. This study determines that some, but not all, impairment-based measures are able to distinguish between degree of impairment early in the development of walking. Additionally, some impairments may differentiate by GMFCS level earlier than others. Clinicians and researchers should carefully consider the type and sensitivity of impairment measures when planning interventions for early walkers with CP.

CHAPTER 6

DISCUSSION

Review of Aims

Cerebral palsy (CP) is a common neuromuscular disorder that limits participation in family, school, work, and community environments, and negatively affects quality of life. CP is “... a group of permanent disorders of the development of movement and posture, causing activity limitation, that are attributed to non-progressive disturbances that occurred in the developing fetal or infant brain.”⁹ Poor control of postural muscles is a primary impairment in CP, which contributes to compensation by other muscles to assist in maintaining upright posture. To understand how compensatory postural patterns are established in CP, and later develop interventions that effectively diminish these impairments, postural muscle activity should be studied during the early years of walking. Core trunk and hip muscle activity during the emergence of walking skills has not been previously investigated in individuals with CP.

The objective of this project was to investigate differences in trunk and hip muscle activation patterns during the early stages of walking in children with CP compared to children with typical development (TD). Information on the activation patterns of muscles from the onset of walking ability may give physical therapy clinicians and researchers a greater understanding of how abnormal muscle activation develops to result in the gait deviations typically observed in children with CP. Additionally, information on how trunk and gluteal muscles function during walking will assist our understanding of their role in maintaining upright posture during walking, and could lead to the development of more effective interventions to diminish deficits in their function. Lastly, early assessment of trunk and hip muscle activity may assist in the prediction of

future gait patterns and warrant more aggressive, targeted physical therapy interventions, before the abnormal patterns are reinforced over time and possibly preventing the need for corrective surgery in the muscle or limb that compensates for poor postural control.

The specific aims and hypotheses for this study were:

Aim 1. To determine if differences exist in spatiotemporal gait parameters between early walkers with spastic CP and children with TD with similar amounts of walking experience.

Hypothesis 1.1. Differences exist in spatiotemporal gait parameters (walking velocity, cadence, step length, base of support, and single support time) in children with spastic CP who are classified as Gross Motor Functional Classification System (GMFCS) level II-III and have 0.5-60 months of walking experience compared to children with TD and equivalent walking experience.

Hypothesis 1.2. Early walkers with spastic CP will demonstrate greater stride-to-stride variability in spatiotemporal gait parameters (walking velocity, cadence, step length, base of support, and single support time) than early walkers with TD.

Aim 2. To determine if differences exist in trunk and hip muscle activation patterns in reference to gait cycle events and phases (initial contact, foot off, stance phase, swing phase) between early walkers with spastic CP and children with TD with similar amounts of walking experience.

Hypothesis 2.1. Differences exist in trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosus) muscle activation timing patterns during walking in children with spastic CP who are classified as GMFCS level II-III and have 0.5-60 months of walking experience compared to children with TD with equivalent

walking experience. Differences will also exist in the timing of muscle coactivation between antagonist pairs (erector spinae/rectus abdominus, rectus femoris/semitendinosus) between the two groups.

Hypothesis 2.2. Differences exist in the frequency characteristics of trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosus) muscle activation patterns during walking in children with spastic CP who are classified as GMFCS level II-III and have 0.5-60 months of walking experience compared to children with TD with equivalent walking experience.

Hypothesis 2.3. Early walkers with spastic CP will demonstrate greater stride-to-stride variability in trunk and hip muscle EMG frequency characteristics than early walkers with TD.

Aim 3. To determine if differences exist in spatiotemporal gait parameters and trunk and hip muscle activation patterns during walking in early walkers with spastic CP with similar amounts of walking experience but different levels of ambulatory ability.

Hypothesis 3.1. Children with spastic CP who are classified as GMFCS level III and have 0.5-60 months of walking experience will demonstrate differences in spatiotemporal gait parameters (walking velocity, cadence, step length, base of support, and single support time), compared to children with spastic CP who are classified as GMFCS level II with equivalent walking experience.

Hypothesis 3.2. Children with spastic CP classified as GMFCS level III will demonstrate greater stride-to-stride variability in spatiotemporal gait parameters compared to children with spastic CP who are classified as GMFCS level II.

Hypothesis 3.3. Children with spastic CP who are classified as GMFCS level III and have 0.5-60 months of walking experience will demonstrate differences in trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosis) muscle activation and coactivation timing patterns, compared to children with spastic CP who are classified as GMFCS level II with equivalent walking experience.

Hypothesis 3.4. Children with spastic CP who are classified as GMFCS level III and have 0.5-60 months of walking experience will demonstrate differences in the frequency characteristics of trunk (bilateral erector spinae, middle trapezius, rectus abdominus, external abdominal oblique) and hip (bilateral gluteus maximus, gluteus medius, rectus femoris, and semitendinosis) muscle activation patterns during walking, compared to children with spastic CP who are classified as GMFCS level II with equivalent walking experience.

Summary of Results

Thirty-four children enrolled in this study. Data from three children were excluded, resulting in 15 children with CP and 16 children with TD who were included for analysis. In the group of children with CP, seven were classified as GMFCS level II and eight were level III. These children had an average of 28.5 (18.1 SD) months of walking experience, which did not differ between groups. Children with CP achieved independent walking at a later age than those with TD. The method of using walking experience, rather than age, for inclusion in the study controlled for the maturation of walking ability that occurs with practice after the onset of walking. However, it must be considered that equal months of walking experience may not

necessarily reflect equal amounts of walking practice. Children with TD may have had more opportunities for walking throughout each day, and may in fact still have more hours of walking practice compared to peers with CP, despite equal time since onset of walking.

Aim 1

The children with CP demonstrated significantly lower values than the TD group for all primary spatiotemporal gait parameters: walking velocity, cadence, step length, base of support, and single support time. The CP group also demonstrated greater stride-to-stride variability in cadence, base of support, and single support time. Only a few of the spatiotemporal characteristics were related to musculoskeletal measures in the CP group. Increased average ankle dorsiflexion range of motion (ROM) was related to decreased step length and increased cadence variability. Hip extension ROM, hamstring ROM, and Tardieu spasticity measures were not related to the mean value or variability for any spatiotemporal measure. In the TD group, but not the CP group, larger anthropometric measures (height, weight, seated height and leg length) were related to more mature walking ability and decreased stride-to-stride variability.

Spatiotemporal gait parameters for the TD group are consistent with previously reported normative values.^{81, 113, 114} Spatiotemporal parameters for the CP group were slightly lower than those reported by Wondra and colleagues for a group of children with mixed motor disabilities¹¹⁰ and by Sorsdahl and colleagues¹¹⁵ for a group of children with less severe CP. Differences in age, diagnostic classification, and level of ambulatory ability, between the children in the present study and the participant samples from those studies, likely explain the variance.

Mean base of support was lower in the CP group than the TD group and was highly variable. Two children had negative values, representing foot placement that crossed the midline

of the body, likely caused by excessive hip adduction during swing phase. However, base of support in the CP group was not related to hip adductor spasticity, or to the use of an assistive device, which could compensate for a narrow base of support by providing balance stability. Future investigation should be done to examine the implications of base of support for walking ability and postural control, and if interventions can change base of support.

Little work has previously been published reporting stride-to-stride variability in spatiotemporal parameters in children. Hausdorff and colleagues found that stride time is more variable in younger children.⁹⁹ However, these investigators did not report any of the parameters measured in the current study. Looper and colleagues reported variability for step length and step width in new walkers with Down syndrome (DS) and TD.¹¹⁶ Step width variability in their children with DS was similar to our measurement for the children with CP, but step length variability was lower in our children. This could be explained by the greater months of walking experience that our participants had compared to their DS group.

Spasticity was not related to spatiotemporal gait parameters, which is consistent with the findings of Ross and Engsborg.¹¹⁹ Greater dorsiflexion ROM was associated with greater variability in cadence and shorter step length. Maximal ankle dorsiflexion ROM in the CP group exceeded normal range of motion. A crouch gait pattern is common in CP and is characterized by excessive dorsiflexion during the stance phase.⁵⁰ A crouch gait pattern limits the ability to extend the hip and knee,¹²⁰ which would contribute to shorter step length.

Anthropometric measures and walking experience were related to spatiotemporal parameters and variability in the TD group, but not the CP group. For the TD group, this is consistent with studies showing that older children have more mature gait patterns than younger children.^{81, 99} The larger children had more walking experience in both groups, but factors other

than walking experience, body size, or musculoskeletal impairments may explain the majority of the variance in walking ability in CP.

Children with CP demonstrate reduced walking ability during the early stages of walking even when walking experience is equal between groups. These differences are consistent with those seen in older children with CP, which suggests that waiting several years after the onset of walking to intervene with gait impairments may be detrimental. The increased stride-to-stride variability in early walkers with CP could suggest that the neural circuitry supporting their gait patterns may not yet be well-established and perhaps can be affected favorably with interaction. Addressing abnormal patterns of movement before the neural pathways that support them become too rigid from years of reinforcement may produce more effective outcomes. Early in the development of walking skill may be a more optimal time to intervene, or may be a time when intensive treatments are most effective in changing compensatory gait patterns and lead to improved walking ability. Treatment programs with intensive practice paradigms may result in less stride-to-stride variability, and improved efficiency, of movement in CP.

Aim 2

This study is the first to investigate muscle activation patterns of the trunk and hip muscles during the early years of walking in children with TD and with CP. It also demonstrates the use of objective methods of analyzing EMG signals to determine periods of activity and inactivity across the gait cycle.

The CP group had significantly more total muscle activation time for each muscle investigated except for the external oblique (EO), which was not different from the TD group. The CP group also had significantly more total coactivation time for both the rectus

abdominis/erector spinae (RA/ES) and rectus femoris/semitendinosus (RF/ST) muscle pairs. The trapezius (TZ), ES, RF, and ST demonstrated activity throughout the gait cycle in the majority of children with CP. The EO, gluteus maximus (GMx), and gluteus medius (GMd) demonstrated similar phases of activity in the CP group compared to the TD group, but they had longer periods of activation, including both earlier onset and delayed offset of activity. Although the RA demonstrated more total muscle activation in the CP group compared to the TD group, it was not active in most children with CP.

The patterns of muscle activity in the GMd and ST in the TD group were consistent with those reported previously by Sutherland and colleagues in their study of over 300 children between the ages of 1 and 7 years.⁸⁰ The period of activity in the GMx was slightly shorter than that reported in the Sutherland study. This difference can likely be attributed to differences in the methods used to determine activation between the two studies, visual inspection in their study compared to automated onset detection in our study. While these 3 muscles are the only ones in the current study that have been previously reported for young TD children, muscle activity patterns during walking have been shown to approximate adult patterns after the age of 3 years.⁸⁰ For this reason, comparison of the other muscles included in this study to data from adults is justified. Timing patterns of activity for the RF in the TD group, as well as the gluteal muscles, are consistent with those reported in adults.¹²⁹ Although differences in data analysis prevent exact comparison with our data, White and Nair reported areas of increased normalized amplitude for the ES and EO in their study of adults that correspond to periods of activity for the ES and EO in the TD group in the current study. Also similar to the TD group in the current study, the majority of adult participants in the White and Nair study did not have periods of activity above baseline in the RA. The comparison of the TD muscle activity data from the

current study to the studies mentioned above demonstrates that, similar to lower extremity muscles, activation patterns of the trunk muscles approximate those of adult patterns by a young age.

All muscles from children in the CP group except for the EO demonstrated significantly more total activity than those from children in the TD group. Even for the EO, more than half of the CP group had activity during 80% of the gait cycle compared to 39% of the gait cycle in the TD group. Excessive muscle activity has been reported in lower extremity muscles in children with CP.^{87, 89}

With the exception of the RA and GMx, all muscles in CP group were active over 75% of the gait cycle, which may restrict the child's ability to grade muscle activity and make fine adjustments to trunk position in relation to the lower extremities and the environment. Roerdink and colleagues report that after a stroke, individuals had less stability but also more regularity in frontal plane COP trajectories during standing than healthy peers.¹³⁷ With recovery and rehabilitation, COP trajectories became less regular. The authors suggest that after stroke, the participants attempt to limit variations in COP in order to decrease the degrees of freedom that they must control. A similar strategy may occur in early walkers with CP. Hsue and colleagues report reduced anterior-posterior displacements of COP and center of mass (COM) during walking in CP.¹³⁸ Limiting excursion and variability in COP and COM, by excessively activating muscles of the trunk and hips, may be a strategy for children with CP to maintain upright posture against gravity and move the body forward despite the multitude of neurological impairments limiting typical movement patterns.

The CP group also had higher instantaneous mean frequency (IMNF) throughout the gait cycle for all muscles compared to the TD group. Stride-to-stride variability in mean frequency of

muscle activation was also statistically higher in the CP group for all muscles across the gait cycle. The IMNF values for the RF and ST in the CP group are generally consistent with those reported by Wakeling et al⁵⁹ and Lauer et al.⁵⁸ However, the values for these muscles in the TD group of the current study are lower than those reported for the control groups in the studies by Wakeling et al⁵⁹ and Lauer et al⁵⁸ which may be a result of the older age of the control groups in those studies.

The higher mean frequency in the CP group suggests altered strategies of muscle activation. Higher IMNF can result from increased rates of motor unit firing, increased number of recruited motor units, or decreased synchrony of motor units.¹⁴¹ This is consistent with literature suggesting excessive and dyscoordinated muscle activity in CP.^{89, 145} The markedly increased stride-to-stride variability in muscle activation patterns in the CP group further implies dyscoordination and immaturity of muscle behavior.⁹⁹ Lam et al have also suggested that higher EMG median frequency contributes to muscle fatigue in children with CP.¹⁴⁶

The only musculoskeletal measure that explained any variance in muscle activation in the CP group was spasticity in the hip adductors. Tardieu R1 and R2 values for the hip adductors were related to activation of the RA and EO, and total percent coactivation of the RA/ES, with increased spasticity predicting increased abdominal muscle activity. These data may indicate that increased muscle activity is an adaptive strategy to maintain postural control during walking in children with greater hip spasticity. Total percent activation and coactivation were not related to range of motion or hamstring spasticity.

The results of this study suggest that postural muscle control training during the early stages of walking in CP should be investigated to encourage the development of more functional and efficient movement strategies in these children. Based on this evidence of excessive,

dyscoordinated, and compensatory trunk and hip muscle activity in CP, postural muscle training may be an effective component of rehabilitation programs.

Aim 3

Walking experience, anthropometric, and musculoskeletal measures did not differ between children with CP who were classified as GMFCS level II and those who were level III. There were no differences in the primary spatiotemporal parameters between children who were GMFCS level II and those who were level III. Gait variability was only different between groups for cadence, with the GMFCS level III group demonstrating greater stride-to-stride variability. However, left to right symmetry was significantly lower in the GMFCS level III group for walking velocity, cadence, and single support time. Despite no difference in the current study, previous work has reported differences in walking speed by GMFCS level in older children with CP.¹⁵³ Left-right symmetry measures were the only spatiotemporal variables that demonstrated consistent differences between groups. Symmetry is a characteristic of mature gait¹⁰⁰ and less symmetry indicates greater impairment in walking ability. These data suggest that symmetry, therefore, may be an important measure to identify impairments during the early years of walking, before differences in spatiotemporal parameters may be apparent.

There were also no differences between the GMFCS level II and level III groups for percent of muscle activation or coactivation over the gait cycle. In the IMNF analysis, there were no differences between groups for the GMx, GMd, or ST. The GMFCS III group had lower mean frequency for the RA from late stance through early swing. The GMFCS III group had higher mean frequency for the RF and EO during stance phase, for the ES from late stance through swing, and for the TZ throughout the gait cycle. This is consistent with previous work in

older children indicating greater mean frequency with greater severity of functional ability.⁵⁸ These muscle activation results suggest that some measures of muscle activity may be more sensitive to degree of impairment than others, and that some muscles may demonstrate differences earlier than others. When available, it may be important to use advanced EMG analysis techniques to detect early differences in muscle activation patterns.

This study determined that some, but not all, impairment-based measures are able to distinguish between degree of impairment early in the development of walking. Additionally, some impairments may differentiate by GMFCS level earlier than others. Early detection of impairment in CP is important to appropriately tailor interventions during the early phases of walking skill development. Focused and intensive intervention, before compensatory movement patterns are repeated and reinforced for years, is important to maximize neuroplasticity and functional gains after injury to the central nervous system.^{33, 42, 98} To identify which impairments to target, it is necessary to both use measures that are able to distinguish degree of impairment early in the skill development, and to measure the impairments that begin to differentiate by severity earlier than others.

Limitations

There are several limitations in this study that should be addressed in future work. The participants in this study were a fairly homogeneous group children with bilateral spastic CP, GMFCS II and III. The results may not be generalizable to children with CP who have greater walking experience or those with greater or lesser ambulatory ability.

Despite our use of automated EMG processing techniques and objective rules to determine the duration of muscle activity in each group, there remains no flawless method to

analyze EMG signals, particularly in children with neurological impairments. Unlike the larger and thicker muscles of the thigh and gluteal region, the superficial muscles of the trunk are thin and the recording sensors may have recorded some activity from the underlying muscles. The internal oblique, rhomboids, and transverse abdominis muscles are directly deep to the sensor location for the EO, TZ, and RA muscles, respectively. The ES muscle was recorded from a location that is deep to the broad superficial fascia of the latissimus dorsi muscle. The use of fine wire needle EMG electrodes would avoid this potential issue, but application in young children has clear feasibility and ethical limitations. Additionally, needle electrodes only record from a single or small group of motor units, and as a result, the recorded signal may not be representative of the activity of the entire muscle.⁶²

An additional limitation exists in comparing data between the CP and TD groups because the majority (12/15) of the children in the CP group used an assistive device for walking. This might be particularly important in the comparison in the TZ muscle activity. Of the 3 children who walked without a device, one had TZ percent activation greater than the CP group mean (88%), but the other 2 had less activation (22% and 11%). Use of an assistive device alone may have contributed to greater activation of the TZ in the CP group, because the shoulders were engaged during forward movement and bearing weight through the assistive device. This issue is difficult to avoid when studying early walkers with CP. According to the GMFCS classification,⁷⁴ only children classified as level I (the least impaired) begin to walk without the use of any assistive device. Therefore, to study any children with greater severity of CP during the early years of walking, the use of walking aids must be considered.

In the comparisons between the children who were classified as GMFCS level II and level III, there are additional considerations that may have limited the detection of group

differences. Although not statistically significant, the GMFCS level III group had 12 months more walking experience than the GMFCS II group. The differences between groups may have been greater if walking experience was equal between groups. Additionally, this study included a small sample with a limited range of GMFCS levels, which again may have limited the detection of group differences. Future work should include more participants and all 4 of the ambulatory levels of the GMFCS.

Future Research

The evidence from this study suggests that postural muscle control training during the early stages of walking in CP should be investigated to encourage the development of more functional and efficient movement strategies in these children. Children with CP demonstrate excessive, dyscoordinated, and compensatory trunk and hip muscle activity during the early years of walking. Postural muscle training may be an effective component of rehabilitation programs.

Core stability is related to athletic performance and function in healthy adults^{139, 140} and core muscle control training has been effectively applied to several other clinical populations,^{140, 148, 149} but has not been systematically investigated in children with neurological disorders. Furthermore, the theoretical framework behind interventions designed to reduce postural sway in children with CP should be examined. Strategies to increase the child's ability to control greater variations in trunk movement through phasic trunk muscle coordination, rather than constant muscle activity, may encourage more effective and efficient patterns of postural muscle control.

Future work should also include all four of the ambulatory levels of the GMFCS, and more participants and at each of the levels, which would strengthen the validity of the data in a

heterogeneous population such as CP. Future investigation should also examine the variability and symmetry in other walking measures, such as gait kinematics and kinetics. Variability and symmetry reflect neurological maturation and biomechanical efficiency, and give valuable insight into the development of movement patterns.

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APPENDIX

APPENDIX A

USE OF THE TEAGER-KAISER ENERGY OPERATOR FOR MUSCLE ACTIVITY

DETECTION IN CHILDREN

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Introduction

Poor control of the postural muscles is considered a primary impairment in cerebral palsy (CP).¹ Poor postural control contributes to compensation by other muscles to assist in maintaining an upright posture, and limiting those muscles from functioning effectively as primary movers of the extremities.²⁻⁴ Core trunk and hip muscle activation patterns have not been systematically investigated in individuals with CP. One method by which hip and trunk muscle function can be assessed is through the use of surface electromyography (sEMG).

There are several variables that can be extracted from the sEMG signal that can provide insight into muscle activity in children with and without CP, including the use of time-frequency analyses,⁵⁻⁸ examination of the amplitude of the sEMG signal in response to force generation and task demands,⁹ and the occurrence of muscle onsets and offsets.¹⁰ While the combined examination of all components of the sEMG would provide the greatest understanding of muscle function,¹¹ the method most commonly employed clinically is the determination of the onset and offset of muscle activity during the movement. Determining the duration of muscle activity is

clinically useful in order to investigate differences between individuals with and without neuromuscular impairments, and to assess co-activation of the agonist and antagonist muscles.

The determination of the onset and offset of muscle activity is fairly straightforward when clear bursts of phasic activity are recorded. It is complicated when the amplitude of the activation bursts are low, the baseline activity is high, the muscles are tonically active or quiet throughout the movement, and when the quality of the signal is compromised by internal or external noise. These scenarios are likely to occur in children with CP because their movement is characterized by muscle weakness, prolonged activation times, spasticity, and poor neuromuscular activation.¹ In addition, cardiac electrical activity is commonly recorded with the sEMG signals of the trunk musculature, making the analysis of the signal difficult.¹²

Several automated methods have been proposed in the literature to facilitate the detection of muscle activity, such as the use of a standard deviation (SD) threshold with respect to a resting baseline,^{13,14} pattern recognition and template matching,^{15,16} or probabilistic criterion matching.^{11,17-19} These methods, however, rely primarily on changes in the amplitude component of the sEMG signal to determine the onset and offset of activity, and may fail to detect distinct bursts of activity when there is a poor signal to noise ratio.²⁰ Thus, visual inspection of the raw sEMG trace remains a popular method even though errors with this method and disagreement between experienced raters are known limitations.¹³

Recently the Teager-Kaiser Energy (TKE) operator was proposed as another method to detect muscle activity using both amplitude and frequency information from the sEMG. This method improves the signal to noise ratio²¹ by taking into consideration the modulations that occur both in amplitude and frequency within the sEMG signal during movement.^{5,6} When applied to sEMG signals from able-bodied adults during isometric contractions, the TKE was

more accurate in detecting the onset of EMG activity than a standard amplitude method.²² The method has also been successfully applied to sEMG signals during walking in older adults.²³ Use of the TKE method has not been examined with sEMG signals from children or from individuals with spasticity. Thus, the purpose of this study was to apply the TKE operator to the detection of sEMG activity from the hip and trunk muscles recorded during gait in children with and without CP to determine the validity of this method.

Methods

Subjects

Data were analyzed from ten children with typical development (TD), and five children with CP (Table 1). All of the children were part of a larger study examining hip and trunk muscle activity patterns in early walkers. The parents of all the children signed a university institutional review board approved consent form, and the children gave verbal assent. Inclusion criteria for the pilot study were: 1) 0.5-60 months of walking experience, 2) the ability to ambulate barefoot at least 15 feet with supervision, children with CP could use any hand held assistive device (i.e. canes, crutches, and walkers) that did not stabilize or restrict movement of the trunk or pelvis; and 3) the ability to follow 1step verbal directions. Children were excluded if they: 1) had surgery to or fracture of the lower extremities in the past 12 months; 2) had botulinum toxin injection in the lower extremities in the past six months; 3) had a history of dorsal rhizotomy; 4) had a history of tendon transfer to a target muscle; or 5) had a known orthopedic, neuromuscular or cardiovascular condition (or secondary condition for the children with motor impairment).

Table A-1. Demographic Data

ID	Age (months)	Gender	BMI (kg/m ²)	Walking experience (months)	Diagnosis	Assistive device
1	64	F	11.2	52	TD	none
2	41	F	16.6	21	TD	none
3	47.5	F	17.0	38	TD	none
4	16	M	13.3	5	TD	none
5	22	M	17.9	12	TD	none
6	67.5	M	15.8	58	TD	none
7	59	M	17.6	29	CP	PRW
8	53	M	22.9	35	CP	PRW
9	38	M	16.4	10	CP	PRW
10	28	F	16.5	12.5	TD	none
11	13	F	18.8	1	TD	none
12	51	F	17.5	42.5	TD	none
13	44	F	15.7	31	TD	none
14	75	F	16.0	42	CP	none
15	47.5	M	16.3	11.5	CP	FC

BMI: Body Mass Index, CP: cerebral palsy, TD: typical development
 PRW: posterior rolling walker, FC: forearm crutches

Data Acquisition

Spatiotemporal information, including heel strike events, were recorded as the children walked down a 15-foot long by 3-foot wide padded walkway (GAITRite®, CIR Systems, Havertown, PA USA) at a self-selected pace. Concurrent surface EMG (sEMG) activity was acquired from the following muscles bilaterally: middle trapezius, erector spinae, rectus abdominus, external abdominal oblique, gluteus maximus, gluteus medius, rectus femoris, and semitendinosus. The Myomonitor III recording system (Delsys Inc., Boston, MA) was employed with pre-amplified silver-silver chloride parallel bar surface electrodes with a 10.0 mm inter electrode distance. The sEMG data was collected at a sampling rate of 1200 Hz, pre-amplified with a gain of 10, and filtered with a high pass filter of 20 Hz to remove motion artifact, and low pass filter of 450 Hz. Placement for the abdominal electrodes was performed using the methods described by Ng et al,²⁴ while placement for all other muscles (back, gluteal, and thigh) was performed in accordance with the SENIAM recommendations.²⁵ Synchronization between the walkway and the EMG data collection system was achieved through a trigger signal from the walkway to initiate EMG data collection. A resting static EMG trial was collected in supine prior to walking trials in order to establish baseline muscle activity. The same baseline recording was used as a reference for all three methods of onset/offset detection.

Onset/Offset Detection

The discrete form of the TKE operator, Ψ , was used in this study. The discrete TKE operator is defined in time domain as:

$$\Psi_d[x(n)] = x^2(n) - x(n+1)x(n-1) \quad (1)$$

For a given signal:

$$x(n) = A \cos[\omega_0(n) + \Theta] \quad (2)$$

$$x(n+1) = A \cos[\omega_0(n+1) + \Theta] \quad (3)$$

$$x(n-1) = A \cos[\omega_0(n-1) + \Theta] \quad (4)$$

where n is the sequence index of a sampled signal, A is amplitude, ω_0 is the angular frequency, and Θ is the initial phase.

Since

$$\cos(\alpha + \beta) \cos(\alpha - \beta) = \frac{1}{2} [\cos(2\alpha) + \cos(2\beta)] \quad (5)$$

$$\cos 2\alpha = 1 - 2 \sin^2 \alpha = 2 \cos^2 \alpha - 1 \quad (6)$$

The cross term of $x(n-1)x(n+1)$ can be rewritten as being equal to:

$$A^2 \cos[\omega_0(n+1) + \Theta] \cos[\omega_0(n-1) + \Theta] \quad (7)$$

$$\frac{1}{2} A^2 [\cos(2\omega_0 n + 2\Theta) + \cos(2\omega_0 n)] \quad (8)$$

$$\frac{1}{2} A^2 [2\cos^2(\omega_0 n + \Theta) - 1 + 1 - 2\sin^2(\omega_0 n)] \quad (9)$$

$$A^2 \cos^2(\omega_0 n + \Theta) - A^2 \sin^2(\omega_0 n) \quad (10)$$

$$x^2(n) - A^2 \sin^2(\omega_0 n) \quad (11)$$

and substituting back into Equation 1 yields:

$$\Psi_d[x(n)] \approx A^2 \sin^2(\omega_0 n) \quad (12)$$

Thus, the output of the discrete TKE is proportional to the instantaneous amplitude and frequency of the input signal. While the dynamic sEMG signal cannot be assumed to be a Fourier function over its entire length, at the beginning and the end of the signal where few motor units are active the assumption of a Fourier function can be made.²¹

Five representative gait cycles (ipsilateral heel strike to ipsilateral heel strike) were extracted from the sEMG recordings and processed using MATLAB (The MathWorks Inc., Natick MA, USA). The TKE operator was applied to the sEMG for each gait cycle, filtered with a second-order low-pass Butterworth filter with phase correction and a cutoff frequency of 10 Hz, and averaged. To determine the onset/offset threshold, the TKE operator was applied to the static EMG baseline signal (5 second recording) that was recorded while all subjects were reclined on a mat in the supine position. The resulting output was then rectified, and the mean and standard deviation were calculated. The mean, plus one standard deviation, was used as the threshold level.

The assessment of the TKE operator for onset/offset detection was performed by comparing the duration of muscle activity obtained to both the standard deviation (SD) threshold method, and visual assessment by independent raters. Comparison of the TKE to these methods was done as they represented the most commonly used techniques for clinical analysis and reporting of muscle activity, even though both are known to contain errors.^{13,20} However, without a true gold standard for muscle activity assessment, these methods were selected because of their extensive use. For the SD method, the data were rectified and then smoothed using a second-order low-pass Butterworth filter with phase correction ($f_c=10$ Hz). A 20 ms long, moving average, square window was then applied to the output. If the mean voltage within the window was greater than the mean of the baseline sEMG signal plus three standard deviations, the muscle was identified as being active.

Visual assessment of muscle activity was performed by two examiners, both having previous experience with visual analysis of sEMG. The signals for a given muscle were rectified and averaged across all five trials and displayed on a computer screen with the baseline mean

and 2 standard deviations displayed concurrently. Using the graphical user interface commands with MATLAB, the examiners were able to indicate onset and offset times of the EMG signal with respect to the percent of the gait cycle to the nearest 0.1%. Each examiner determined muscle activity on the same set of traces on two separate days, three days apart to evaluate repeatability.

Analysis

The study was designed to examine the validity of the TKE method to assess hip and trunk muscle activity. However, the difficulty in establishing the validity of the TKE method in relation to the SD method and visual observation lies in the fact that the degree to which these methods also accurately reflect muscle activity in the hip and trunk musculature are also unknown. Therefore, to adequately compare all the methods, the relative and absolute reliability was calculated by using intraclass correlation coefficients (ICCs) and Bland-Altman plots using the visual observation as the basis for comparison.

Inter-and intra-rater reliability of the visual assessment method was established for the entire data set, as well as for the individual muscles, to establish the relative error of the visual raters and thus aid in comparison of the other techniques. All analyses were performed using SPSS, version 16.0 (SPSS Inc, Chicago, IL USA). An ICC_{2,k} was selected since the raters were considered as representative of a larger population of similar raters, and assessment was based on the mean of several measures.²⁶ ICC values above 0.8 were interpreted as good reliability, 0.6 to 0.8 as moderate, and below 0.6 as poor.²⁶ The same model (ICC_{2,k}) was then used to compare the SD and TKE methods to the raters assessment of activity. Agreement of muscle activity

across the assessment methods was determined through the use of Bland-Altman limits of agreement.²⁷

Results

Initial analysis of muscle activity was performed with the data separated between the group of children with TD and the group of children with CP. As no trend was observed between the groups for the muscles, the data was combined for full analysis. The intra-rater reliability across all muscles for the two raters (R1,R2) between the first and second assessment days was 0.93 for R1 and 0.95 for R2, indicating reasonable consistency and a low error rate for each rater in determining muscle activity. Inter-rater reliability across all muscles on the first day assessment was good (ICC = 0.89), with individual muscle assessments presented in Table 2. Good reliability was exhibited for all muscles except the rectus abdominus (ICC = 0.69) and semitendinosus (0.57).

Examples of the TKE and SD curves for select muscles, and compared to the results determined by the raters, are presented Figures 1 and 2. The data in Figure 1 represent an example of an EMG recording with a high signal to noise ratio with clear bursts of activity (gluteus medius) for one child with TD and one with CP. There was high agreement between R1, R2, and the TKE method, while the SD method underestimated activation time. The data in Figure 2 represent an example of a muscle with a low signal to noise ratio (rectus abdominus) characterized by low muscle activity and a variable firing pattern, for one child with TD and one with CP. R1, R2 and the TKE agreed with activation time for the child with TD (0%), while the SD method resulted in 100% activation. For the child with CP, R1 determined the muscle to be

Table A-2. Bland-Altman Agreement

Muscle	R1/R2 Day 1	TKE/R1	TKE/R2	SD/R1	SD/R2
Trapezius	0.92	0.89	0.89	0.30	0.57
Erector spinae	0.77	0.83	0.86	0.33	0.36
Rectus abdominus	0.69	0.51	0.53	0.17	0.20
External abdominal oblique	0.91	0.45	0.50	0.15	0.11
Gluteus maximus	0.83	0.51	0.72	0.56	0.76
Gluteus medius	0.84	0.76	0.82	0.41	0.60
Rectus femoris	0.89	0.76	0.89	0.11	0.12
Semitendinosus	0.57	0.70	0.65	0.11	0.40

R1: Rater 1, R2: Rater 2, TKE: Teager-Kaiser Energy Operator
SD: Standard Deviation Threshold Method

active for 0% of the gait cycle, R2 59%, the TKE 39%, and the SD method 100%. This example highlights limitations in the visual assessment method when activity bursts are not clear.

Inter-rater reliability (Table 2) ranged from 0.45 and 0.89 between the TKE and R1, and from 0.50 to 0.89 between TKE and R2, indicating poor to good agreement. The SD method, in comparison, indicated poor to moderate agreement across the data set with values ranging between 0.11 and 0.56 in relation to R1, and from 0.11 to 0.76 in relation to R2. Overall, the TKE method performed better in comparison to the raters than the SD method, with the poorest correlations for TKE method for the rectus abdominus, external abdominal oblique, and semitendinous.

Bland-Altman plots of the TKE and SD methods in comparison to the raters for all the muscles are shown in Figure 3. Bland-Altman plots indicated that most points (95.0% for R1 and R2 respectively) lie within the 95% limits of agreement for the TKE method, and for the SD method (97.5% and 98.1% for R1 and R2 respectively). However, the 95% limits of agreement were wide in relation to the mean difference of the two methods (TKE to R1: 113%; TKE to R2: 95%; SD to R1: 170%; SD to R2: 151%), but were lower for the TKE method. The biases for the TKE method (TKE to R1: -5%, TKE to R2: 4%) were also lower than for the SD method (SD to R1: -24%, SD to R2: -15%). The negative bias for the SD method to both raters indicated a consistent underestimation of muscle activity.

Discussion

Accurate detection of muscle activity from the hip and trunk musculature during walking in the pediatric population, both with and without disabilities, is of great importance to understand how these muscles work to maintain upright posture during the maturation of gait in the pathological condition. With this knowledge, researchers and clinicians may be able to develop interventions that address deficits in core muscle activation. However, the extraction of muscle activity patterns from these muscles is complicated by the fact that in children and, in particular, those with a pathological condition, the ability to recruit the muscle could be compromised,¹ and the quality of the signal could be compromised by internal or external noise sources.¹²

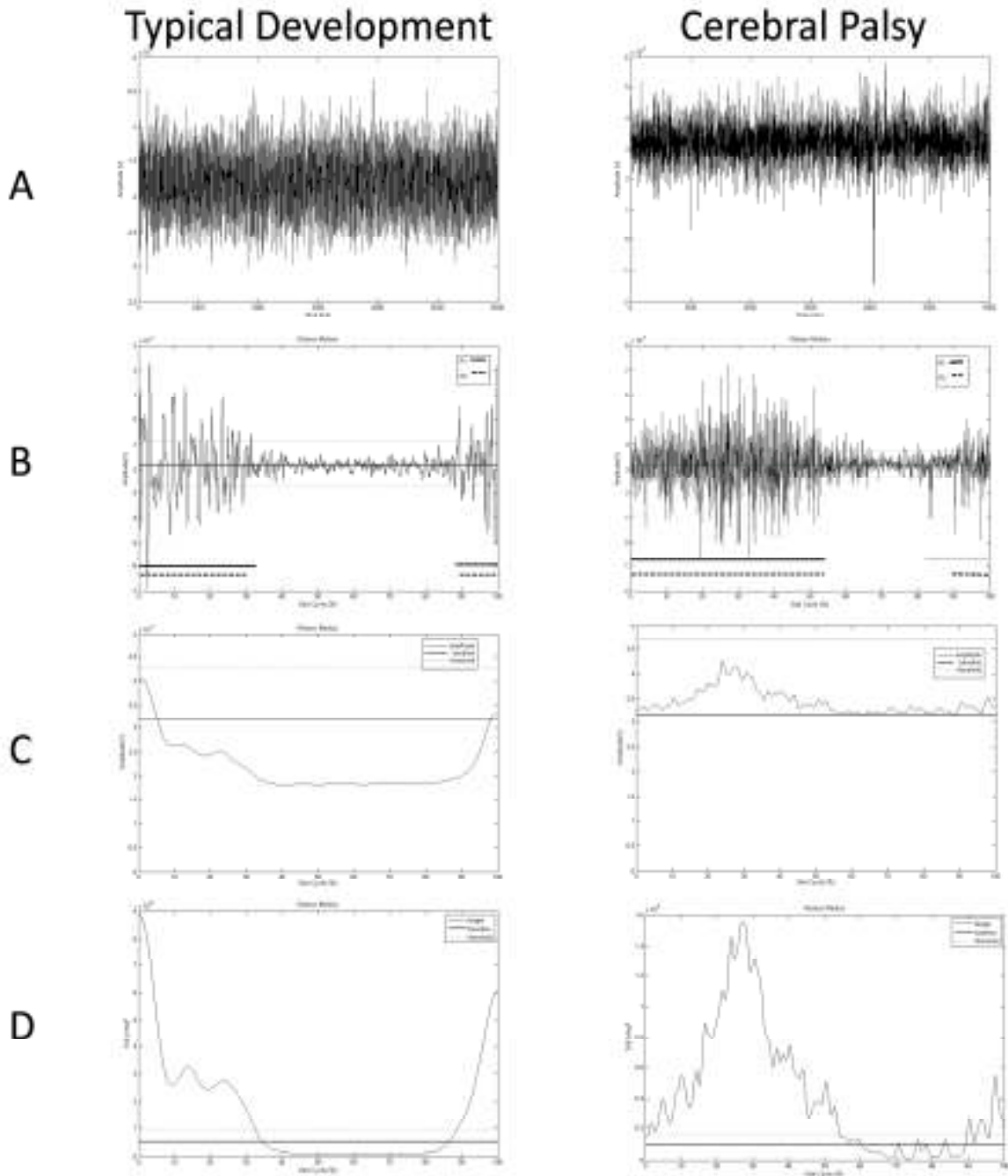


Figure A-1: Representative data for the gluteus medius for one child with typical development (TD) and one with cerebral palsy (CP). A. Raw EMG traces of the static baseline trial. B. Raw signals during walking. Thick solid and dashed lines parallel to x-axis represent muscle activation as determined by R1 and R2, respectively. C. SD amplitude curve. The dashed lines indicate the threshold level determined from comparison to the baseline. All activity above the threshold was considered “on”. D. TKE curves with dashed line again indicating threshold of “on” activity.

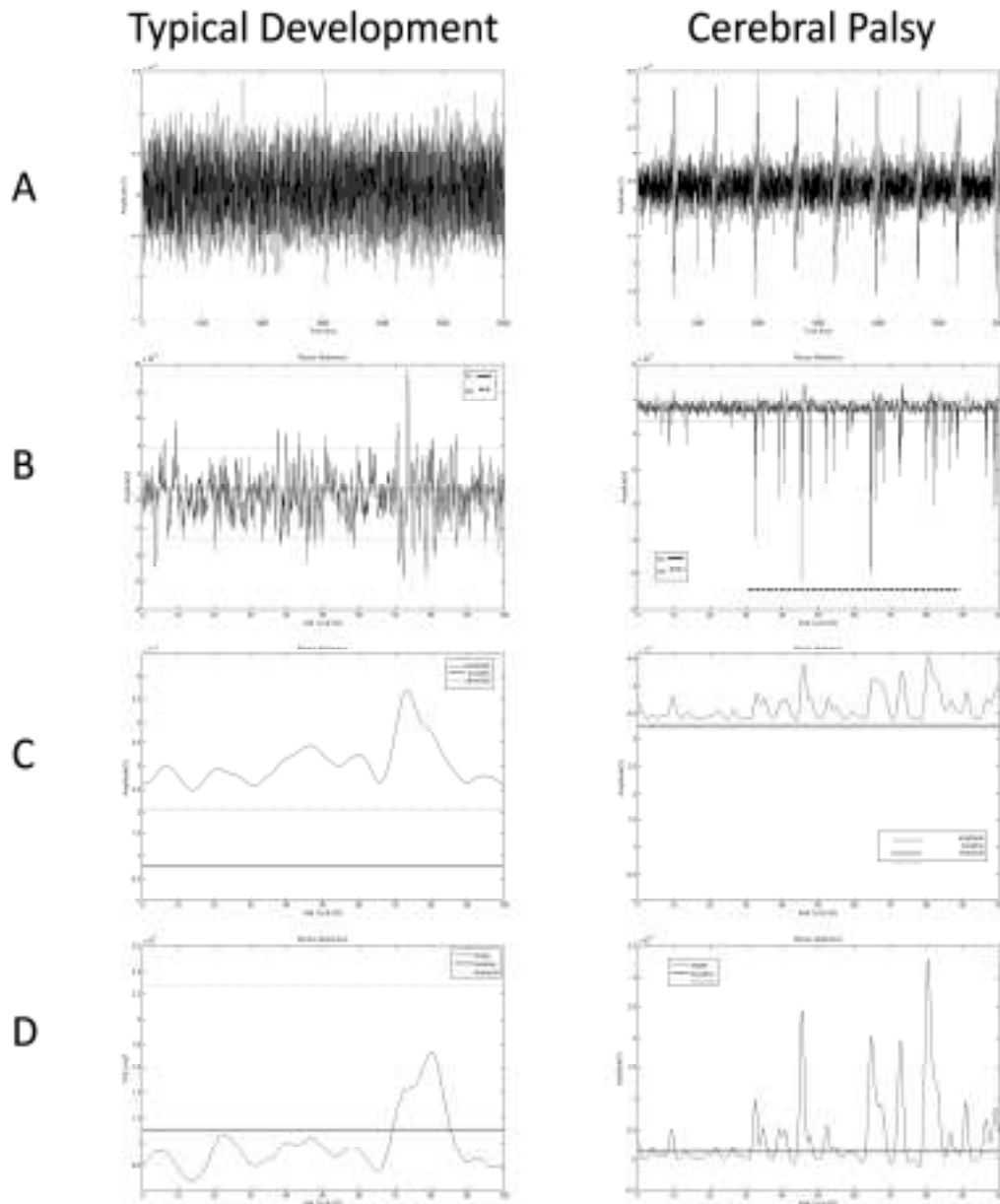


Figure A-2: Representative data for the rectus abdominis for one child with typical development (TD) and one with cerebral palsy (CP). A. Raw EMG traces of the static baseline trial. B. Raw signals during walking used by the raters for visual assessment. Thick solid and dashed lines parallel to x-axis represent muscle activation as determined by R1 and R2, respectively. The thick dashed line parallel to x-axis represents muscle activation as determined by R2. R1 did not determine that the muscle was activated during any period of the gait cycle for either child. C. SD amplitude curve. The dashed lines indicate the threshold level determined from comparison to the baseline EMG. All activity above the threshold was considered “on”. D. TKE curves with dashed line again indicating threshold of “on” activity.

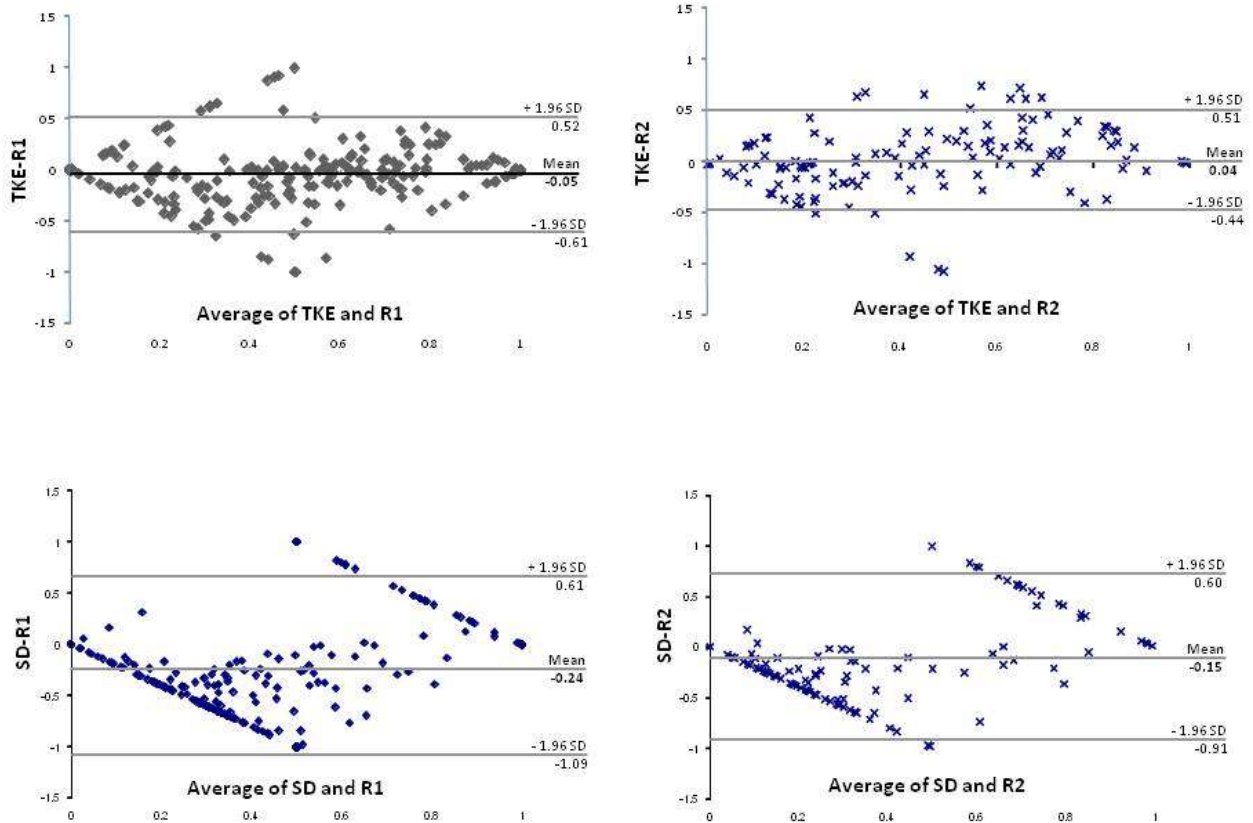


Figure A-3: Bland-Altman plots for the TKE and SD methods, which examine the agreement between the methods and the two raters (R1 and R2) with respect to overall muscle activity. The y axis represents the subtraction of method from the rater, whereas the x axis represents the average of the two. The 95% limits of agreement are represented as grey lines, and mean bias is represented as a solid black line.

Because of these complications, a majority of the methods proposed in the literature that rely upon amplitude measurements, such as the use of a standard deviation (SD) threshold with respect to a resting baseline,^{13,14} pattern recognition and template matching,^{15,16} probabilistic criterion matching,^{11,17-19} and visual assessment, may under or over report the activity level in the hip and trunk muscles. Likewise, the generalized likelihood ratio and the approximated generalized likelihood ratio methods,^{11,18,19} which work well in cases of poor signal to noise ratio, or low levels of muscle activity, depend upon a probabilistic change in the amplitude level of the signal to detect bursts of activity, and may fail if the muscle is tonically active with respect to amplitude.

Therefore, the examination of amplitude changes, as well as alterations within the frequency components of the sEMG signal, becomes necessary to accurately determine of the level of muscle activity. The TKE operator,²¹⁻²³ considers both amplitude and frequency changes in the EMG signal. By taking frequency into consideration for the detection of muscle activity for the hip and trunk muscles, the ability to correctly detect intervals of muscle activity may be enhanced, particularly in light of recent studies on the sensitivity of frequency information in the sEMG signal⁵⁻⁸ to detect differences across groups as well as the effects of a clinical intervention.

This study demonstrated that the TKE method of muscle activity detection more closely resembled the results obtained from visual analysis than the SD amplitude threshold method, and thus may be a more desirable, automated method of calculating muscle activity in the hip and trunk muscles of children (Figures A-1-A-2: Table A-2). Automated processing of muscle activity substantially reduces the time needed from clinicians and researchers to analyze sEMG data, and can provide a reliable, objective analysis. A perceived additional advantage of the TKE

method is the relatively minimal degree of computational complexity, which may make it more clinically feasible to implement and more accepted for use. The ease of use may be why the SD and visual methods are still reported for muscle activity assessment, in spite of the more robust methods that have been proposed.^{11,15-19}

Recording and interpreting EMG signals from small children is a challenge. Placement must be accurate to avoid crosstalk from adjacent muscles. In addition, abnormal motor unit firing patterns in individuals with neurological impairment can increase the variability and decrease the clarity of the signal. Our results are based on the assumption that visual analysis is an accurate method of determining the onset and offset of muscle activity. It is apparent from the representative data for the rectus abdominus (Figure A-2) that this is not always the case (R1 vs. R2 disagreement). While visual analysis is a clinically accepted standard, it is not a flawless method. Visual analysis, like most automated methods, primarily considers only the amplitude component of the EMG signal which is prone to error.²⁰ In instances where the TKE method is not consistent with the visual analysis of one or both raters, it remains unclear which method is more accurate and further investigation is needed.

Clinicians and researchers in numerous fields of work investigate muscle activity. A reliable, objective, and accurate method of EMG analysis is necessary to make sound clinical decisions and appropriately maintain the rigor of scientific research. The TKE method of onset and offset detection appears a valid tool to meet these goals, particularly in instances of low signal to noise ratio. Likewise, investigations into the affects of signal filtering and sample rate on the accuracy of TKE method, and comparison to a larger group of raters, needs to be explored. The application of this method to other clinical populations and other movement

patterns will further establish its feasibility and allow for more detailed recommendations on its use in those areas.

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