A Computational Model Approach to Cerebral Palsy Gait: Mechanical Work with and without a Pediatric Posterior Walker

---

A Thesis

Presented to

the faculty of the School of Engineering and Applied Science

University of Virginia

---

in partial fulfillment
of the requirements for the degree

Master of Science

by

Marshall W Tumperi

May 2019

APPROVAL SHEET

This Thesis
is submitted in partial fulfillment of the requirements
for the degree of
Master of Science

Author Signature: Marshall Tumperi

This Thesis has been read and approved by the examining committee:

Advisor: Shawn D. Russell

Committee Member: Mark F. Abel

Committee Member: Silvia S. Blemker

Committee Member: Jason R. Kerrigan

Committee Member:

Committee Member:

Accepted for the School of Engineering and Applied Science:

Craig H. Benson, School of Engineering and Applied Science

May 2019

Abstract

Cerebral palsy is a neuromuscular disorder affecting motor function and development caused by a lesion to the brain at or around the time of birth. If not given adequate care and treatment prior to adolescence, children may lose the ability to ambulate on their own and require a wheelchair for mobility. Treatment for individuals with cerebral palsy targets increasing time walking throughout the day to avoid muscular atrophy while minimizing excess energy usage often seen in pathologic gait. Posterior walkers are commonly recommended by clinicians to increase stability and posture during gait. The purpose of this thesis was to further our understanding of the use of posterior walkers in both normal gait and pathologic, cerebral-palsied gait.

This work developed a dynamic model of the entire body which incorporated the posterior walker interaction forces during the gait cycle. The model was used to compare baseline, unassisted gait to gait with a posterior walker in typically developed (TD) and cerebral palsy (CP) child populations. Children with CP benefited from the walker through improved torso stability and reduced mechanical work of the whole body, particularly in the lower body and torso. Increased upper body moments highlighted the cost of pulling the walker and the applied vertical loading for support in the CP group. The TD group and the less-effected subjects in the CP group were inhibited by the walker as they walked slower and with shorter strides in walker-assisted gait. The current design of posterior walkers requires manual propulsion of the walker to take advantage of the benefits associated with its use. The dynamic model provides the ability to perform simulations with an automated walker which could decrease the energy expense of pulling the walker while still providing necessary stability. The development of an automated walker could further aid the user by neutralizing the required forces to pull the walker.

Acknowledgements

This work is the culmination of my research performed in the Motion Analysis and Motor Performance Lab, but it could not have been completed without the many people who helped me along the way. I would like to thank the members of my thesis defense committee for the commitment of their time and patience. Jason Kerrigan went above and beyond his call by giving guidance and support from the beginning to the end of the writing of this work and encouraged me greatly. Silvia Blemker taught me in my undergraduate years at the University of Virginia and quickly became a mentor that prepared me well for graduate school. Continued association with Silvia and her lab has helped me keep in mind the bigger picture as I transition out of graduate school. Mark Abel was an amazing resource for the clinical aspect of this work and his determination drove me to perform at my best.

I would like to thank Shawn Russell, my advisor, for his patience and guidance on the path of graduate school. I am tremendously grateful for the challenges and opportunities that were given me at the University of Virginia and wish Shawn the best as his lab continues to grow. I would like to thank the members of the MAMP lab for their support of my research and their friendship, particularly Emily Dooley who was a great friend to me.

Additionally, I would like to thank my family and friends who were a source of constant support through this process. Through the ups and downs of me working on my graduate work my parents and siblings were great sources of mental rest and perspective. My friends in Charlottesville and beyond were tremendously supportive of me and helped me to see the good of the work I was doing.

Finally, I would like to thank my best friend and fiancée, Olivia. Your patience and love for me and God were and continue to be essential to me. You are my glue when I feel like falling

apart, my hammer when I am being too stubborn, and you love to pour syrup into the waffle of my brain.

Table of Contents

Introduction

Typical human gait involves coordination of the segments of the body to move from one location to another while maintaining balance. In typically developed children these motions are learned and developed from a young age and result in an energetically efficient motion as they reach adolescence. In pathologic gait, such as in cerebral palsy (CP), children develop complex, abnormal patterns of gait to maintain neuromuscular control of their bodies. For more involved patients with a gross motor function classification system (GMFCS) of level III or more, motor function deteriorates with age (Hanna et al., 2009). Children with CP have greater energy expenditure during gait (Russell, Bennett, Sheth, & Abel, 2011). Those that are ambulatory may not be able to walk for the same length of time as typically developing peers. Due to the effects of CP, clinicians monitor children as they grow to develop treatment plans that can include orthotics and walkers as assistive devices or interventional surgeries to positively affect the biomechanics of gait. Posterior walkers have been adopted by clinicians for their positive effects on posture during gait (Mattsson & Andersson, 2008).

Current clinical practice utilizes gait analysis and lower body assessments to measure the effectiveness of interventions in CP. The resultant full body effects of interventions have largely been ignored in previous work and the load of pulling a posterior walker along for use in gait has not been quantified. This thesis focused on the effects of a posterior walker in pediatric gait in both typically developed (TD) and CP populations. This was accomplished by:

1. Developing a dynamic model for subject gait trials with outputs for full body joint kinematics and kinetics.
2. Calculating mechanical work of individual joints and summing into body segments over the gait cycle.

This research examined the use of a posterior walker in gait for CP and typically developed children and quantifies energy expenditure through total mechanical work. It is necessary to include all external forces and energies which influence the subject to obtain accurate values for mechanical work of the subject. Previous work has not examined the forces involved at the handles of the walker and thus ignored the effect of the walker on the energy cost of ambulation during walker-assisted gait. This is a problem because the results do not incorporate the effect of the walker on the upper body segments. Previous work by Konop (Konop et al., 2009) looked at upper extremity kinetics in children with CP. The study found significant correlations between upper body kinetics spatiotemporal patterns in CP gait. The interaction between walker and user needs to be quantified when the kinetics of the system are considered, otherwise the outcomes ignore the reliance upon and the use of the assistive device. By incorporating the kinetics of the walker, patterns in usage of the walker can be observed as they correlate to the gait.

The dynamic model was developed with the intent to further analyze the efficacy of an automated control on the walker to decrease the pulling action required by the user. This work identified potential for the control of the automated walker to be based on the interaction of the user with the handles of the walker. A motorized walker could reduce the negative cost of the intervention while maintaining the positive effects of the posterior walker on posture and stability. Further use of this model will examine the effects of a motorized walker on gait outcomes in CP and the potential for increasing distance and time of ambulation.

Reference List

Hanna, S. E., Rosenbaum, P. L., Bartlett, D. J., Palisano, R. J., Walter, S. D., Avery, L., & Russell, D. J. (2009). Stability and decline in gross motor function among children and youth with cerebral palsy aged 2 to 21 years. Developmental Medicine and Child Neurology. https://doi.org/10.1111/j.1469-8749.2008.03196.x

Konop, K. A., Strifling, K. M. B., Wang, M., Cao, K., Eastwood, D., Jackson, S., Harris, G. F. (2009). Upper extremity kinetics and energy expenditure during walker-assisted gait in children with cerebral palsy. Acta Orthopaedica et Traumatologica Turcica, 43(2), 156–164. https://doi.org/10.3944/AOTT.2009.156

Mattsson, E., & Andersson, C. (2008). Oxygen cost, walking speed, and perceived exertion in children with cerebral palsy when walking with anterior and posterior walkers. Developmental Medicine & Child Neurology, 39(10), 671–676. https://doi.org/10.1111/j.1469-8749.1997.tb07361.x

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011). The gait of children with and without cerebral palsy: Work, energy, and angular momentum. Journal of Applied Biomechanics, 27(2), 99–107.

Background & Literature Review

2.1 Cerebral Palsy

An international workshop defined CP in 2006 as “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 behavior, by epilepsy, and by secondary musculoskeletal problems” (Rosenbaum et al., 2007). Cerebral Palsy (CP) has an incidence of approximately 1.5 to 3.5 per 1000 live births (Cans, 2000; Winter, Autry, Boyle, & Yeargin-Allsopp, 2002; Yeargin-Allsopp et al., 2008) and lifetime costs are estimated near $1 million per person (Honeycutt et al., 2004). Listed below are known risk factors for CP in fetal and infant stages of life (Reddihough & Collins, 2003).

- Antenatal: malformations in cortical development and maternal infections in first and second trimester
- Perinatal: obstructed labor, multiple pregnancy, and birth weight
- Neonatal: neonatal seizures, infections, and injuries

These risk factors are associated with a singular central nervous system damage event occurring around birth which result in complications in motor function development. CP is non-progressive in that it does not further deteriorate physiological functions through the lifetime of the person. Clinically, the progression of motor function development is observed in adolescents with a CP diagnosis. While the damage in the body is not increasing, the growth of the child can further aggravate symptoms of CP. A tight gastrocnemius muscle can become tighter when the leg grows, further decreasing motor function at the ankle and knee joint. It is important to observe and analyze both the development of the child and the impact of interventions to have a positive or minimized negative effect on the motor function of the child.

The analysis and treatment of CP is approached through the classification of the disorder through musculoskeletal and motor function assessments. Three sub-types of CP are used to categorize presentations of the disorder (Cans, 2000):

- Spastic: increased tone and/or pathological reflexes
- Ataxic: loss of orderly muscle coordination
- Dyskinetic: involuntary, uncontrolled, and recurring movements

Spastic CP accounts for approximately 70 to 80% of total CP cases (Yeargin-Allsopp et al., 2008) where it may present unilaterally or bilaterally in the upper and/or lower limbs (Cans, 2000). The resulting abnormal motion in bilateral spastic CP of the lower limbs has been divided into four groups: true equinus, jump gait, apparent equinus, and crouch gait (Armand, Decoulon, & Bonnefoy-Mazure, 2016a). Of these, true equinus is characterized by the ankle in plantarflexion during stance phase and the hips and knees extended. Crouch gait is designated by excessive dorsiflexion at the ankle in concert with excessive flexion of the knee and hip joints (Figure 2.1). These gait patterns all result in the patient expending more energy to walk compared to a similar distance walked by a typically developed peer. Pathological gait patterns increase the load on the lower body joints which over time will cause pain and damage to the joints. Management of these gait abnormalities is important to alleviate pain in the short-term, increase gait efficiency, and prevent long-term damage.

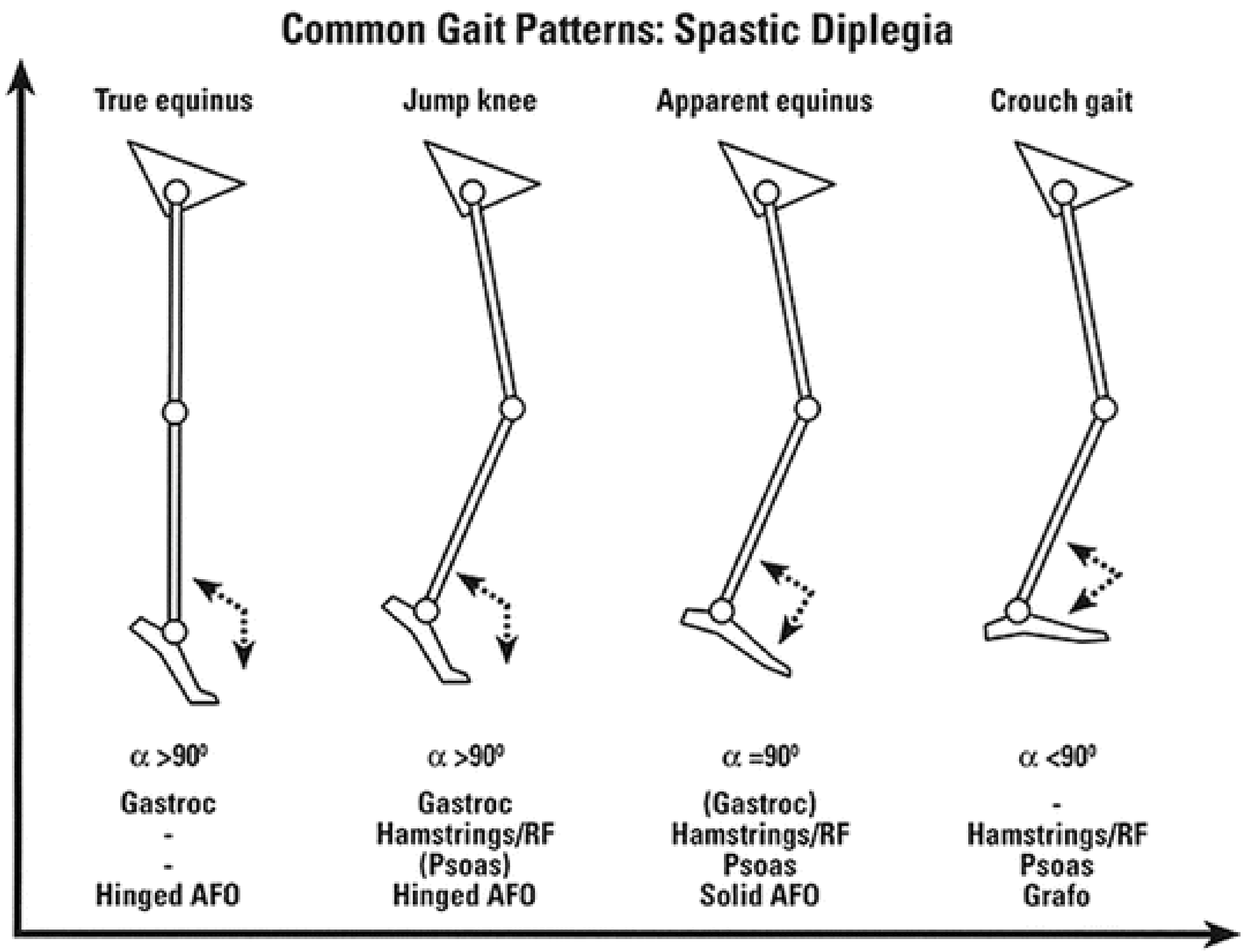

Figure 2.1: Gait patterns observed in cerebral palsy along with affected muscle groups and AFO recommendations. Reprinted with permissions from Wiley and Sons Inc. (Rodda & Graham, 2001).

The gait patterns in spastic CP are best quantified through clinical gait analysis (Armand et al., 2016a). The pathologic gait of children with CP is analyzed as a critical indicator of their motor function which can help in advising patient-specific interventions. The Gross Motor Function Classification System (GMFCS) is widely accepted and utilized to provide quick classification criteria for clinicians and families to describe motor function and aid in clinical management decisions (Palisano, Rosenbaum, Bartlett, & Livingston, 2008). Gait analysis is a valuable tool for classification of CP both through the GMFCS and other assessments as it has a greater accuracy than visual analysis (DeLuca, 1991; Gage, DeLuca, & Renshaw, 1996; Kienast, Bachmann, Steinwender, Zwick, & Saraph, 1999).

CP directly affects the motor function of children with the disability, but downstream activities are also impeded. A study on the quality of life of children with CP reports that their disability inhibits their ability to create meaningful relationships with their peers (Colver et al., 2015). The capacity of a child to move and play with their peers is affected by how much pain they experience as a result of CP; thus, it is important to engage in treatment options both for basic motor function and for social quality of life.

2.2 Assistive Devices

Intervention options for CP are considered based on the classification and analysis of each individual case. Invasive options such as botulinum toxin injections and surgical intervention are usually considered in cases with decreasing motor function with growth (Gough, Eve, Robinson, & Shortland, 2004; Paul, Siegel, Malley, & Jaeger, 2007) and often used in combination with orthotics and assistive devices. CP cases classified on the GMFCS scale from level I through level IV are often prescribed orthotics and/or assistive devices, whether invasive interventions were considered or not (Palisano et al., 2008). Children on the GMFCS scale at level II or III use assistive devices for the majority of the day for mobility across various terrains. Assistive devices have been proven to increase stability and decrease lower extremity load during movement (Fast et al., 1995; Yepremian et al., 2009). For a pathological gait that is typically unbalanced an assistive device provides independence in their mobility.

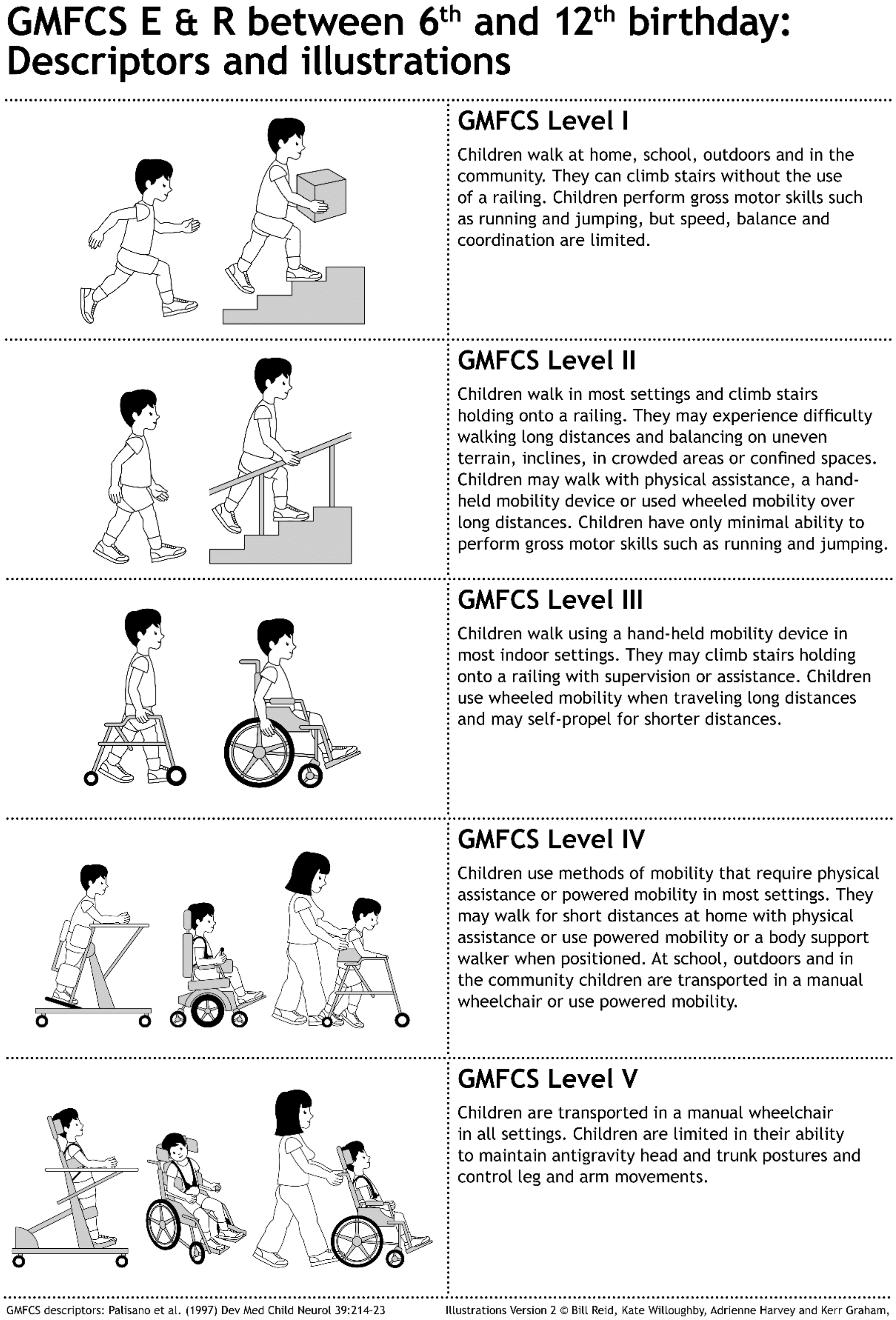

Figure 2.2: The Gross Motor Function Classification System for clinical evaluation of children aged six to twelve years old. (Burns et al., 2014)(Courtesy of Kerr Graham)

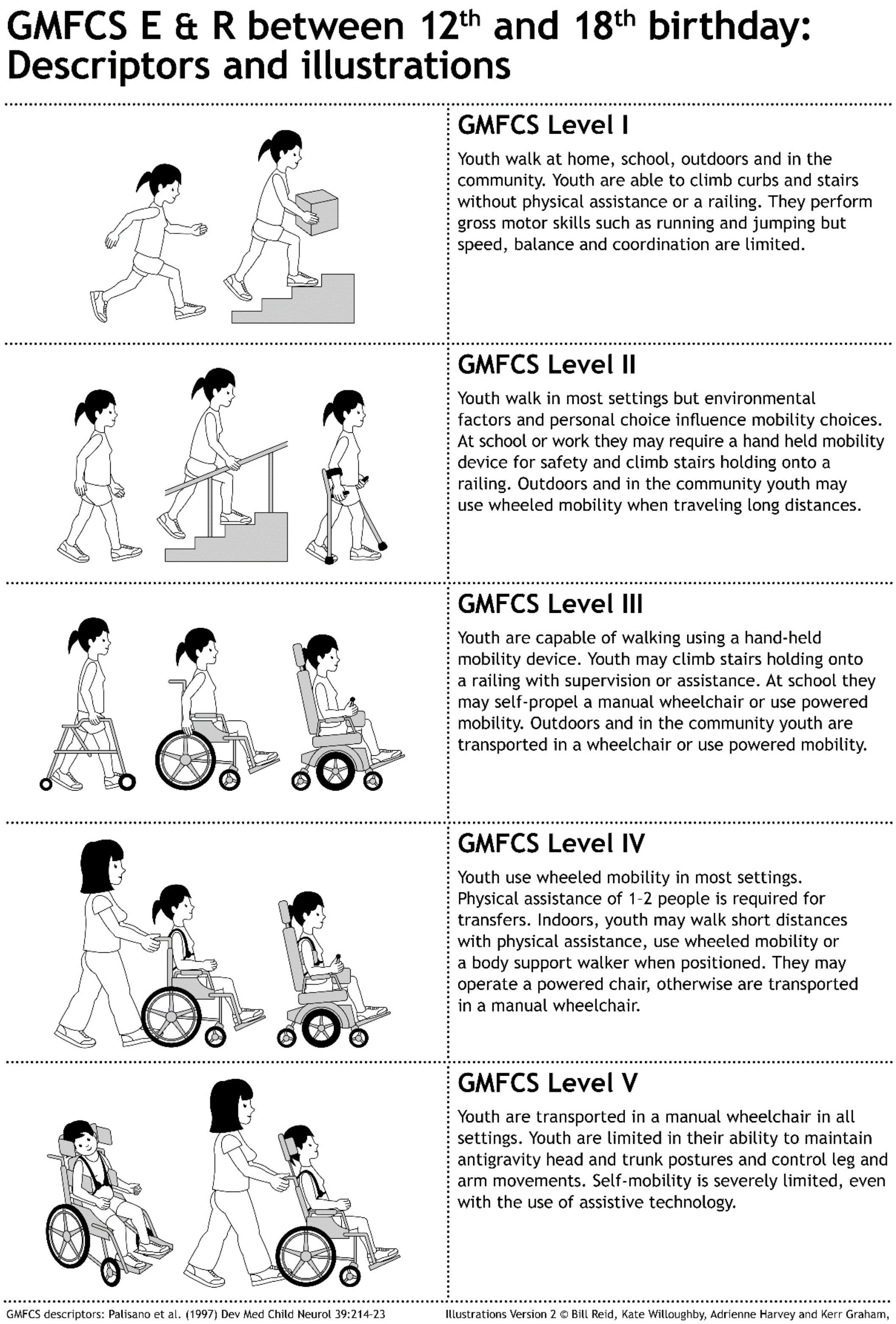

Figure 2.3: The Gross Motor Function Classification System for clinical evaluation of children aged twelve to eighteen years old. (Burns et al., 2014) (Courtesy of Kerr Graham)

Assistive devices, such as walkers and elbow crutches, have different preferential characteristics based on kinematic and energy expenditure effects of use as well as user preference (Bateni & Maki, 2005). Walkers assist in balance by providing steady stabilizing points of contact for the upper body. Elbow crutches are versatile and can be used on different terrains while providing dynamic support options. User preference for different devices influences the usage pattern for various aids (Lephart, Utsey, Wild, & Fisher, 2014) and in many instances preferences also align with the suggestion resulting from clinical gait and energy expenditure analysis.

The effect on gait kinematics and energy expenditure is important in analyzing the efficacy of an assistive device. The gait of crutch users can vary greatly depending on their adaptation to the device (Thys, Willems, & Saels, 1996). This research potentially introduces new devices to users because this was not part of the exclusion criteria for subject recruitment; thus, elbow crutches do not provide consistent data between subjects. Posterior walkers assist balance and position the user to be more upright, with their center of mass directly in-line with their legs, whereas anterior walker use results in an anterior pelvic tilt and poor posture (B M Greiner, Czerniecki, & Deitz, 1993; Logan, Byers-Hinkley, & Ciccone, 1990; Park, Park, & Kim, 2001). Posterior walkers also may increase walking speed and step length (Bachschmidt et al., 2000) although not significantly in all studies (Park et al., 2001). An increase in walking speed and step length indicates a better adaptation to the device and a preferable subject-device kinematic interaction.

Energy expenditure during motion, calculated from gait analysis, is significantly higher in CP cases when compared to their typically developed peers (Dziuba, Tylkowska, & Jaroszczuk, 2014; S. Russell, Bennett, Sheth, & Abel, 2011a; van den Hecke et al., 2007). It is necessary for assistive devices to not add excessive energy cost in order to increase stability. Generally, anterior and posterior walkers increase oxygen consumption during ambulation (Holder, Haskvitz, & Weltman, 1993; Protas, Raines, & Tissier, 2007). In CP, posterior walkers are more efficient than anterior walkers when comparing oxygen cost and consumption rates (Park et al., 2001).

2.3 Gait Efficiency Analysis

The analysis of gait efficiency through energy cost metrics is used to quantify the effect of pathology and different treatment options in CP. Many studies have shown that energy use is significantly greater in children with CP than typically developed children (Bolster, Balemans, Brehm, Buizer, & Dallmeijer, 2017a; Norman, Bossman, Gardner, & Moen, 2004; Raja, Joseph, Benjamin, Minocha, & Rana, 2007). Metabolic energy consumption considers the whole body's energy use and is measured through O_2 consumption or heart rate. The measure of metabolic energy cost through O_2 consumption is often used as the standard for comparison and viewed as the most accurate metric for gait efficiency. According to metabolic measures, children with CP use two to three times more energy than their typically developed peers (Bolster et al., 2017a; Ries & Schwartz, 2018). Norman (Norman et al., 2004) found energy expenditure index calculated from heart rate follows similar trends to O_2 measurements. Heart rate measurement is typically less expensive and more practical in clinical CP analysis. Metabolic energy metrics quantify the whole-body cost and lack the ability to look at specific body segments. Some literature has tried to define specifically what causes this extra energy usage in CP gait through alternate energy expenditure analyses. Oxygen consumption is a delayed metric for measuring metabolic energy expenditure and only gives the whole-body energy usage. This work required individual breakdown between segments of the body and between gait cycle events for in depth analysis of the kinetics.

Unnithan et al. found the cocontraction of muscle groups plays a role in the increased energy cost in CP gait using electromyographic (EMG) data (Unnithan, Dowling, Frost, & Bar-Or, 1996). EMG data yields individual data for muscle groups, but it can be difficult to quantify the whole-body muscle activation without significantly impeding the ambulation of the subject. Mechanical energy transfer has also been used to compare to metabolic energy cost, but a direct relationship has not been agreed upon in literature. Both mechanical work and metabolic cost increase with the degree of disability in CP (Dziuba et al., 2014; Johnston, Moore, Quinn, & Smith, 2004). Mechanical energy expenditure models can be further broken down into segments and joints of the body to more completely observe where energy cost may change due to pathology or assistive devices. Frost (Frost, Dowling, Bar-Or, & Dyson, 1997) had mixed results in comparing mechanical model outputs to metabolic energy cost. Their models did not incorporate kinetic data from the subjects and fell short of explaining a significant amount of change that was observed in metabolic energy cost. Van de Walle (Van de Walle et al., 2012) found the sum of the Vicon Plug-in-Gait model outputs for joint powers provided validity and sensitivity to pathologies like that of metabolic energy measures. By using mechanical energy calculations, this work was able to calculate the work of individual joints over the gait cycle.

2.4 Modelling

It is difficult to directly measure characteristics of motion in people and animals, but it is important to be able to quantify the loads and movements of the individual parts associated with the whole body. The human body can be represented as a mechanical system during walking.

The study of a mechanical system requires the synthesis of a valid model in order to return parameter predictions from given input. Models give access to measures that cannot be recorded directly, for example joint moments and center of mass movement. For this work, a rigid body model was created to examine the effect of pathology and assistive technology on gait outcomes in walking. There are two general approaches to modelling human gait, inverse and forward dynamics. An inverse dynamics approach utilizes known kinematics to solve for musculoskeletal forces. In this work, inverse dynamic methods were used to predict joint movements and torques. Forward dynamics applies given joint torques and muscle forces to solve for gait kinematics. The model developed by this work sets up future work to run forward dynamics to tune and develop a controller for a powered walker in cerebral palsy gait. The controller for the powered walker will be tuned based on the kinematic model outcomes in order to increase the utility of the walker for the user.

Many software packages offer the ability to create inverse dynamic models. While no model perfectly predicts the outcome, it is important to manage the assumptions made in the model to yield the results that are desired. The Vicon Plug-In-Gait (PIG) dynamic model is offered in addition to the Vicon motion capture system. Contacts can only be applied to the feet in the PIG model which would not allow the walker reaction forces to be included at the handles and wrists. The PIG model uses surface marker locations to approximate skeletal positions and orientations. Model outputs are calculated directly from these markers which presents multiple potential sources of error. Skin motion during ambulation introduces a skin artifact of high frequency motion at impact instances during gait and skin movement across muscles and bones. Because the markers can move independent of the skeletal frame they are tracking, the segments defined by the markers change length throughout a trial. Without further plugins and model development, segments in the upper body of the Plug-In-Gait model are not as accurate and would require validation to look at kinetics (Stambolian, Asfour, & Eltoukhy, 2014). Vicon is inherently a closed code source and thus cannot be directly altered and tuned for specific models. The Vicon Plug-In-Gait dynamic model was not used in this work because it is most applicable for generic walking trials with ground reaction forces without additional model elements.

An OpenSim inverse dynamic model examines the musculoskeletal outputs during gait (Delp et al., 2007; Pontonnier & Dumont, 2009; Todorov, Erez, & Tassa, 2012). Many dynamic models have been made in OpenSim and are publicly available and verified through use in multiple protocols. They are useful for analyzing specific muscle activation and joint torques. It is difficult to include multiple contacts in OpenSim models as computation time increases. OpenSim also has a limited selection of contacts for modeling. For these reasons, an OpenSim model was not used in this work.

MSC Adams is a multibody dynamics solver software that is used in academic and industrial settings for research and product development. In MSC Adams a multibody dynamic model of the human gait can be created using the Generator of Body Data program (Baughman, 1983). MSC Adams has multiple, robust contact models built in which this work needs in order to facilitate the modelling of the interaction of the human body and the walker and the body and the ground. Resultant models in MSC Adams can be queried for individual joint kinetics, center of mass movement, and other user-defined parameters. This work used MSC Adams LifeMod plug-in to develop an inverse dynamic model to examine mechanical work of the body under different gait conditions.

In order to build a complete dynamic model of human gait the system movement and forces must be fully defined. Kinematics of the rigid bodies are needed to define movement in the multibody dynamic model. The reaction forces of anything in contact with the body must be taken into account. This work incorporated ground reaction forces and walker handle reaction forces into the model of human gait. With a converged model, all the data is available for the joints and segments to do secondary analysis.

Once the model of human gait including the walker has been verified, the impact of the walker on user motion can then be analyzed. Assistive devices increase stability in CP but also increase energetic cost on the user (Holder et al., 1993; Park et al., 2001; Protas et al., 2007). Applied torque at the wheels of a posterior walker could offset this negative effect of walker usage and a validated model would allow us to examine and predict the effect on movement.

Reference List

Armand, S., Decoulon, G., & Bonnefoy-Mazure, A. (2016). Gait analysis in children with cerebral palsy. EFORT Open Reviews, 1(12), 448–460. https://doi.org/10.1302/2058-5241.1.000052

Bachschmidt, R., Harris, G. F., Ackman, J., Hassani, S., Carter, M., Caudill, A., … Klein, J. (2000). Quantitative study of walker-assisted gait in children with cerebral palsy: anterior versus posterior walkers. In Pediatric Gait: A New Millennium in Clinical Care and Motion Analysis Technology (pp. 217–223). IEEE. https://doi.org/10.1109/PG.2000.858901

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: Benefits, demands, and adverse consequences. Archives of Physical Medicine and Rehabilitation, 86(1), 134–145. https://doi.org/10.1016/j.apmr.2004.04.023

Baughman, L. D. (1983). Development of an Interactive Computer Program to Produce Body Description Data. Dayton, OH. Retrieved from http://www.dtic.mil/docs/citations/ADA133720

Bolster, E. A. M., Balemans, A. C. J., Brehm, M.-A., Buizer, A. I., & Dallmeijer, A. J. (2017). Energy cost during walking in association with age and body height in children and young adults with cerebral palsy. Gait & Posture, 54, 119–126. https://doi.org/10.1016/j.gaitpost.2017.02.026

Burns, F., Stewart, R., Reddihough, D., Scheinberg, A., Ooi, K., & Graham, H. K. (2014). The cerebral palsy transition clinic: administrative chore, clinical responsibility, or opportunity for audit and clinical research? Journal of Children's Orthopaedics. https://doi.org/10.1007/s11832-014-0569-0

Cans, C. (2000). Surveillance of cerebral palsy in Europe: a collaboration of cerebral palsy surveys and registers. Developmental Medicine & Child Neurology, 42(12), 816–824. https://doi.org/10.1111/j.1469-8749.2000.tb00695.x

Colver, A., Rapp, M., Eisemann, N., Ehlinger, V., Thyen, U., Dickinson, H. O., … Arnaud, C. (2015). Self-reported quality of life of adolescents with cerebral palsy: a cross-sectional and longitudinal analysis. The Lancet, 385(9969), 705–716. https://doi.org/10.1016/S0140-6736(14)61229-0

Delp, S. L., Anderson, F. C., Arnold, A. S., Loan, P., Habib, A., John, C. T., … Thelen, D. G. (2007). OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement. IEEE TRANSACTIONS ON BIOMEDICAL ENGINEERING, 54(11). https://doi.org/10.1109/TBME.2007.901024

DeLuca, P. A. (1991). Gait analysis in the treatment of the ambulatory child with cerebral palsy. Clinical Orthopaedics and Related Research, (264), 65–75. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/1997253

Dziuba, A. K., Tylkowska, M., & Jaroszczuk, S. (2014). Index of mechanical work in gait of children with cerebral palsy. Acta of Bioengineering and Biomechanics / Wroclaw University of Technology, 16(3), 77–87. https://doi.org/10.5277/abb140309

Fast, A., Wang, F. S., Adrezin, R. S., Cordaro, M. A., Ramis, J., & Sosner, J. (1995). The instrumented walker: Usage patterns and forces. Archives of Physical Medicine and Rehabilitation, 76(5), 484–491. https://doi.org/10.1016/S0003-9993(95)80584-2

Frost, G., Dowling, J., Bar-Or, O., & Dyson, K. (1997). Ability of mechanical power estimations to explain differences in metabolic cost of walking and running among children. Gait & Posture, 5(2), 120–127. https://doi.org/10.1016/S0966-6362(96)01108-3

Gage, J. R., DeLuca, P. A., & Renshaw, T. S. (1996). Gait analysis: principle and applications with emphasis on its use in cerebral palsy. Instructional Course Lectures, 45(10), 491–507. Retrieved from http://jbjs.org/content/77/10/1607

Gough, M., Eve, L. C., Robinson, R. O., & Shortland, A. P. (2004). Short-term outcome of multilevel surgical intervention in spastic diplegic cerebral palsy compared with the natural history. Developmental Medicine & Child Neurology, 46(02), 91–97. https://doi.org/10.1017/S0012162204000192

Greiner, B. M., Czerniecki, J. M., & Deitz, J. C. (1993). Gait parameters of children with spastic diplegia: a comparison of effects of posterior and anterior walkers. Archives of Physical Medicine and Rehabilitation, 74(4), 381–385. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8466419

Holder, C. G., Haskvitz, E. M., & Weltman, A. (1993). The Effects of Assistive Devices on the Oxygen Cost, Cardiovascular Stress, and Perception of Nonweight-Bearing Ambulation. Journal of Orthopaedic & Sports Physical Therapy, 18(4), 537–542. https://doi.org/10.2519/jospt.1993.18.4.537

Honeycutt, A., Dunlap, L., Chen, H., al Homsi, G., Grosse, S., & Schendel, D. (2004). Economic Costs Associated with Mental Retardation, Cerebral Palsy, Hearing Loss, and Vision Impairment --- United States, 2003. Retrieved April 17, 2018, from https://www.cdc.gov/mmwr/preview/mmwrhtml/mm5303a4.htm

Johnston, T. E., Moore, S. E., Quinn, L. T., & Smith, B. T. (2004). Energy cost of walking in children with cerebral palsy-- relation to the Gross Motor Function Classification System. Developmental Medicine & Child Neurology, 46(1), 34–38. https://doi.org/10.1111/j.1469-8749.2004.tb00431.x

Kienast, G., Bachmann, D., Steinwender, G., Zwick, E. B., & Saraph, V. (1999). Determination of Gait Patterns in Children with Cerebral Palsy Using Cluster Analaysis. Gait and Posture, 10, 57. Retrieved from http://www.gaitposture.com/article/S0966-6362(99)90401-0/pdf

Lephart, K., Utsey, C., Wild, D. L., & Fisher, S. R. (2014). Estimating Energy Expenditure for Different Assistive Devices in the School Setting. Pediatric Physical Therapy, 26(3), 354–359. https://doi.org/10.1097/PEP.0000000000000047

Logan, L., Byers-Hinkley, K., & Ciccone, C. D. (1990). Anterior Versus Posterior Walkers: A Gait Analysis Study. Developmental Medicine & Child Neurology, 32(12), 1044–1048. Retrieved from http://doi.wiley.com/10.1111/j.1469-8749.1990.tb08521.x

Norman, J. F., Bossman, S., Gardner, P., & Moen, C. (2004). Comparison of the Energy Expenditure Index and Oxygen Consumption Index During Self-Paced Walking in Children with Spastic Diplegia Cerebral Palsy and Children Without Physical Disabilities. Pediatric Physical Therapy, 16(4), 206–211. https://doi.org/10.1097/01.PEP.0000145930.84009.23

Palisano, R. J., Rosenbaum, P., Bartlett, D., & Livingston, M. H. (2008). Content validity of the expanded and revised Gross Motor Function Classification System. Developmental Medicine & Child Neurology, 50(10), 744–750. https://doi.org/10.1111/j.1469-8749.2008.03089.x

Park, E. S., Park, C. I., & Kim, J. Y. (2001). Comparison of anterior and posterior walkers with respect to gait parameters and energy expenditure of children with spastic diplegic cerebral palsy. Yonsei Medical Journal, 42(2), 180–184. https://doi.org/10.3349/ymj.2001.42.2.180

Paul, S. M., Siegel, K. L., Malley, J., & Jaeger, R. J. (2007). Evaluating interventions to improve gait in cerebral palsy: a meta-analysis of spatiotemporal measures. Developmental Medicine & Child Neurology, 49(7), 542–549. https://doi.org/10.1111/j.1469-8749.2007.00542.x

Pontonnier, C., & Dumont, G. (2009). Inverse dynamics method using optimization techniques for the estimation of muscles forces involved in the elbow motion. International Journal on Interactive Design and Manufacturing (IJIDeM), 3(4), 227–236. https://doi.org/10.1007/s12008-009-0078-4

Protas, E. J., Raines, M. L., & Tissier, S. (2007). Comparison of Spatiotemporal and Energy Cost of the Use of 3 Different Walkers and Unassisted Walking in Older Adults. Archives of Physical Medicine and Rehabilitation, 88(6), 768–773. https://doi.org/10.1016/j.apmr.2007.03.001

Raja, K., Joseph, B., Benjamin, S., Minocha, V., & Rana, B. (2007). Physiological cost index in cerebral palsy: its role in evaluating the efficiency of ambulation. Journal of Pediatric Orthopedics, 27(2), 130–136. https://doi.org/10.1097/01.bpb.0000242440.96434.26

Reddihough, D. S., & Collins, K. J. (2003). The epidemiology and causes of cerebral palsy. Australian Journal of Physiotherapy, 49(1), 7–12. https://doi.org/10.1016/S0004-9514(14)60183-5

Ries, A. J., & Schwartz, M. H. (2018). Low gait efficiency is the primary reason for the increased metabolic demand during gait in children with cerebral palsy. Human Movement Science, 57, 426–433. https://doi.org/10.1016/j.humov.2017.10.004

Rodda, J., & Graham, H. K. (2001). Classification of gait patterns in spastic hemiplegia and spastic diplegia: a basis for a management algorithm. European Journal of Neurology, 8(s5), 98–108. https://doi.org/10.1046/j.1468-1331.2001.00042.x

Rosenbaum, P., Paneth, N., Leviton, A., Goldstein, M., Bax, M., Damiano, D., … Jacobsson, B. (2007). A report: the definition and classification of cerebral palsy April 2006. Developmental Medicine & Child Neurology, 49(SUPPL.109), 8–14. https://doi.org/10.1111/j.1469-8749.2007.tb12610.x

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011). The gait of children with and without cerebral palsy: work, energy, and angular momentum. J Appl Biomech, 27(2), 99–107. https://doi.org/10.1123/jab.27.2.99

Stambolian, D., Asfour, S., & Eltoukhy, M. (2014). Using vicon bodybuilder and plug-in-gait to generate L5/S1 angles, forces and moments. In 2014 IEEE Aerospace Conference (pp. 1–7). IEEE. https://doi.org/10.1109/AERO.2014.6836487

Thys, H., Willems, P. A., & Saels, P. (1996). Energy Cost, Mechanical Work and Muscular Efficiency in Swing-Through Gait with Elbow Crutches. Journal of Biomechanics, 29(11), 1473–1482. Retrieved from https://ac.els-cdn.com/002192909684543X/1-s2.0-002192909684543X-main.pdf?_tid=c556cdc2-c711-4ddb-8055-89990e0db1fe&acdnat=1520428515_718582a0cd09299825d02b5e9e888582

Todorov, E., Erez, T., & Tassa, Y. (2012). MuJoCo: A physics engine for model-based control. In 2012 IEEE/RSJ International Conference on Intelligent Robots and Systems (pp. 5026–5033). IEEE. https://doi.org/10.1109/IROS.2012.6386109

Unnithan, V. B., Dowling, J. J., Frost, G., & Bar-Or, O. (1996). Role of cocontraction in the O2 cost of walking in children with cerebral palsy. Medicine and Science in Sports and Exercise, 28(12), 1498–1504. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8970144

Van de Walle, P., Hallemans, A., Schwartz, M., Truijen, S., Gosselink, R., & Desloovere, K. (2012). Mechanical energy estimation during walking: Validity and sensitivity in typical gait and in children with cerebral palsy. Gait & Posture, 35(2), 231–237. https://doi.org/10.1016/j.gaitpost.2011.09.012

van den Hecke, A., Malghem, C., Renders, A., Detrembleur, C., Palumbo, S., & Lejeune, T. M. (2007). Mechanical Work, Energetic Cost, and Gait Efficiency in Children With Cerebral Palsy. Journal of Pediatric Orthopaedics, 27(6), 643–647. https://doi.org/10.1097/BPO.0b013e318093f4c3

Winter, S., Autry, A., Boyle, C., & Yeargin-Allsopp, M. (2002). Trends in the Prevalence of Cerebral Palsy in a Population-Based Study. Pediatrics, 110(6), 1220–1225. https://doi.org/10.1542/peds.110.6.1220

Yeargin-Allsopp, M., Van Naarden Braun, K., Doernberg, N. S., Benedict, R. E., Kirby, R. S., & Durkin, M. S. (2008). Prevalence of cerebral palsy in 8-year-old children in three areas of the United States in 2002: a multisite collaboration. Pediatrics, 121(3), 547–554. https://doi.org/10.1542/peds.2007-1270

Yepremian, D., Desailly, E., Lejeune, L., Hareb, F., Jarrige, M., Bouchakour, D., & Khouri, N. (2009). The effects of walkers on lower limb kinematics of children with cerebral palsy. Gait & Posture, 30, S90. https://doi.org/10.1016/j.gaitpost.2009.08.134

Spatiotemporal Outcomes of Typical and Cerebral Palsied Gait: With and Without a Posterior Walker

3.1 Introduction

Cerebral Palsy (CP) is a group of disorders that affect motor and posture development caused by nonprogressive damage to the brain occurring near birth (Rosenbaum et al., 2007). Children with CP experience ranges of motor impairment due to muscle spasticity, weakness, and contraction control issues. During adolescence, CP can further affect children as they grow by tightening muscles and making ambulation difficult. Clinical intervention for CP is performed when it is seen to be most advantageous for the child's motor development. Gait analysis has become a crucial outcome measure to assess CP gait and influence intervention options for children with CP. Spatiotemporal characteristics of the gait cycle have been used to compare between CP populations and typically developed (TD) peers (Ju Kim & min Son, 2014) as well as to observe the change or deterioration of motor function over time for individuals (Johnson, Damiano, & Abel, 1997).

Clinical gait analysis is a powerful tool to aid in intervention and treatment of CP. It produces quantitative analysis of the gait cycle to provide specific data for clinicians on what muscle groups and joints are particularly affected by the individual case of CP. Johnson et al. observed that the velocity and stride length of children with CP decreased as the child grew (Johnson et al., 1997). This highlights the importance of interventions during adolescence to increase motor function. Prosser et al. showed that children with CP have lower walking velocity, cadence, step length, and percent of gait cycle in single support than their typically developed peers (Prosser, Lauer, VanSant, Barbe, & Lee, 2010). A smaller time in single support and smaller step length indicate a lower stability that children with CP have due to impaired muscular development. Walking velocity and stride measurements decrease over time for children with CP and thus the deficit observed by Prosser et al. would only grow through adolescence. Comparing gait parameters between children with CP and typically developed peers highlights the different characteristics of gait in the two populations. This analysis can inform clinical interventions in CP with quantifiable values that can be tracked over time to show progression of motor function and development.

Assistive walkers are typically recommended to increase stability and promote better posture of various users during gait (Bateni & Maki, 2005). Walkers provide stability to the torso through constant contact points to the ground and act to decrease the load on the lower body joints. Previous work has examined the effect of posterior and anterior walkers on CP gait. A posterior walker improves posture through lower flexion angles in the knees, trunk, and pelvis (B M Greiner et al., 1993; Logan, Byers-Hinkley, & Ciccone, 1990) and lowers energy expenditure (Park et al., 2001) compared to gait with an anterior walker. Assistive devices have been shown to lower the cadence of the user compared to their gait without an assistive device (Krautwurst, Dreher, & Wolf, 2016a). Many positive effects from a posterior walker have been seen through spatiotemporal and posture analysis but a child with CP still has to pull the walker along to become more stable in their gait. The load of the walker inherited by the user has not been examined through quantifying the forces where the user and walker interact at the handles of the walker.

Gait analysis in CP gait with a posterior walker has been limited and has not quantified the walker's positive and negative impact on CP gait. Previous work has examined CP gait without assistive devices to define the differences from TD gait (Ju Kim & min Son, 2014; Pauk, Ihnatouski, Daunoraviciene, Laskhousky, & Griskevicius, 2016; Prosser et al., 2010). Krautwurst et al. examined the effect of walkers on CP gait compared to baseline gait to inform the decisions of clinicians in surgical interventions for CP (Krautwurst et al., 2016a). Krautwurst et al. calculated velocity, cadence, step length, and step width as spatiotemporal metrics of gait. Previous work has focused on the effect of a few metrics on one feature of gait as opposed to a larger scope which more fully defines the effect of the walker on gait. No one has quantified the full effect of posterior walkers on spatiotemporal parameters and mechanical load on the body in CP gait compared to unassisted gait.

This work developed a method to quantify the positive and negative effects of a posterior walker used in CP gait. This was done through calculation of spatiotemporal characteristics for TD and CP children while walking with a posterior walker and without an assistive device. Over the gait cycle, calculated values include: stride length, stride time, cadence, velocity, step length, foot off and contact percentages of gait cycle, and single and double support percentages. Ground and walker handle reaction forces were analyzed to further define how the walker was used for support. The results are critical indicators of stability and balance in dynamic gait and were used to compare within populations to examine the effect of the walker on gait. Comparisons between populations then examine the incorporation of the walker into pathologic and TD gait. It is expected that the spatiotemporal parameters and ground reaction forces would align with what has been reported in previous literature under similar gait conditions. The handle reaction forces will show the reliance of the users on the walker and how the walkers were used by different groups.

3.2 Methods

3.2.1 Subjects and Procedure

The kinematic data for 17 children was collected in the Motion Analysis and Motor Performance Laboratory at the University of Virginia. There were two populations represented within the group. The first was composed of 8 children with a diagnosis of spastic, diplegic Cerebral Palsy (CP). The second group of 9 children consisted of controls with no known musculoskeletal pathologies. The group with CP had prescribed ankle foot orthotics (AFOs) and wore them during testing. Six of the children had a GMFCS of III, two typically used a walker to move. Four children used forearm crutches regularly for movement in the community. The other two children had a GMFCS II and did not require assistive devices except for long-distance settings. The children were community ambulators and had ranges of experience with assistive devices. The children were asked to walk over the short distance of the testing area without the aid of an assistive device. Four children with a GMFCS III used crutches to walk for the baseline trials and one child used a walker for baseline trials. The data for the subject who used a walker in baseline trials was not used for spatiotemporal analysis because it was assumed the results would be similar between baseline and the custom walker gait. The data for the instrumented walker was used later in the study for analysis. Subject assent and parental consent were approved by the University of Virginia’s Human Investigation Committee and were obtained for all subjects.

A full body marker set of 38 kinematic markers following the Plug-in-Gait marker set was attached to all subjects. Subjects walked along a straight path for 15 meters at their self-selected comfortable walking speed. Three-dimensional kinematic data were collected using a 12-camera Vicon Nexus Motion Capture System (Oxford Metrics, UK) sampling at 100Hz.

Ground reaction force (GRF) was collected with five Bertec force plates sampling at 1000Hz. The force plates were embedded in the floor in the center of the path for each trial. Each subject completed a minimum of three trials for each walking condition where two sequential steps were labelled for events of foot strike and foot off. The first condition for walking was a baseline, where the subject walked without any device aid other than their AFO's if prescribed or with crutches as necessary. For the second condition the children were instructed to walk with an instrumented, pediatric posterior walker. A typical posterior walker was instrumented at both handles with a 6 DoF load cell (ATI) to capture all forces (1000Hz) applied through the handles. The walker wheels were designed wide enough to avoid interference with force plate readings of GRFs. To correct for periodic interference when subjects walked off the center line down the walkway, MATLAB code was written to cancel out the interference of the walker. The code defined the location of the walker wheels and defined time periods when the walker was on a force plate. The vertical forces on the handles and the weight of the walker were combined to subtract from the GRF if interference was occurring. Subjects with no prescribed AFOs walked barefoot for all trials and those with prescribed AFOs wore their AFOs and shoes. Prescribed AFOs were worn for all baseline and walker-assisted trials. The data collected in Vicon was processed with the Nexus program where a total of 7 gait cycle events were labelled: foot strike, contralateral toe off, contralateral foot strike, toe off, foot strike, contralateral toe off, and contralateral foot strike. Events were labelled using recorded force plate data over the trial. Foot strike occurred when the vertical force on the respective plate was greater than a 15 N threshold. Toe off was marked when the vertical force fell below the 15 N threshold.

3.2.2 Spatiotemporal Analysis

Spatiotemporal analysis was conducted for trial outputs as presented in previous work from kinematic data (Johnson et al., 1997; Ju Kim & min Son, 2014; Pauk et al., 2016). Stride length for the left and right leg stride were calculated as the distance travelled by the respective foot between the first and second heel strike of the same foot. The two locations of the foot were calculated as the average of the ankle, heel, and toe markers at the heel strike time points. Stride length was normalized by the length of the respective leg so that the results could be compared across different subjects. Step length for the left and right step was calculated as the distance from the contralateral to the ipsilateral foot at heel strikes of the respective feet. The location of the feet was the average of the three foot markers on the foot. Step length was normalized by the length of the respective leg for which the step was taken.

Stride time was calculated as the period from initial heel strike to terminal heel strike and reported in seconds (Figure 3.1). The cadence was calculated as the inverse of the stride time and converted to steps per minute. Velocity was calculated by dividing the non-normalized stride length by the stride time.

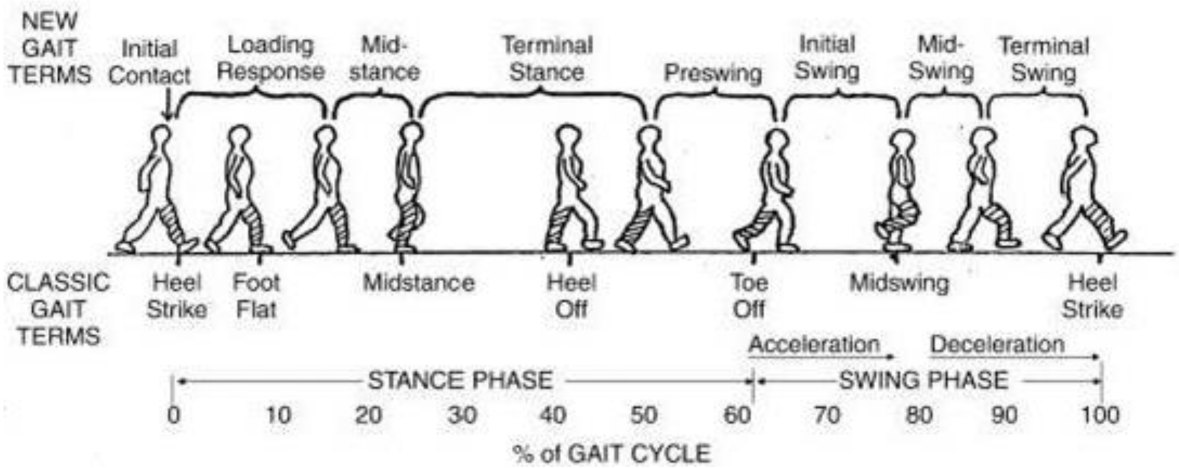

Figure 3.1: Gait cycle events and terminology.

Contralateral foot off and contact and ipsilateral toe off events were calculated as a percentage of the gait cycle from initial to terminal heel strike. Two values for each parameter were generated for each trial as left and right strides were used for this analysis. Single and double support time for both left and right strides were calculated using events labelled from force plate data. Single support time was calculated from adding the time period between contralateral toe off and contralateral foot strike and the time period between ipsilateral toe off and ipsilateral foot strike. Double support time was calculated by adding the time period between initial foot strike and contralateral toe off and the time period between contralateral heel strike and ipsilateral toe off. Single and double support were calculated as a percentage of the gait cycle by dividing the time of support by the respective stride time for the left and right stride.

Parameters were compared using statistical t-tests where the samples were assumed to have equal variance for comparing within groups and unequal variances for comparisons between groups.

3.2.3 Reaction Forces

The ground and walker handle reaction forces were recorded for baseline and walker-assisted gait trials. Posterior walkers reduce the load on the lower body joints in populations which rely on the walker for ambulation (Bateni & Maki, 2005; Fast et al., 1995). The GRF was examined so that the reliance on the walker could be quantified at different times over the gait cycle. A greater reliance on the walker would present as a lower magnitude GRF over the gait cycle. Handle reaction forces were examined to define the dependence on the walker and the applied force to move the assistive device by each group. The CP group was expected to have a higher reliance on the walker and use more force to move the walker along through their gait cycle.

3.2.4 Statistical Parametric Mapping

Baseline and walker trial outputs for reaction forces were compared using one-dimensional statistical parametric mapping (SPM) originally designed for neurological testing (Friston, Ashburner, Kiebel, Nichols, & Penny, 2011). SPM has been shown to be a valuable statistical test for biomechanical comparisons by Pataky in a series of studies and SPM1D software interface in Python and Matlab (Pataky, 2010, 2012). SPM analysis performs better over time periods than individual tests of each point because it considers the surrounding data points in calculating the p-value in post hoc analysis (Pataky, Robinson, & Vanrenterghem, 2016). SPM allowed for the comparison of ground reaction forces and walker handle reaction forces over the gait cycle and identified time periods where walker-assisted gait deviated from baseline gait.

3.3 Results

3.3.1 Subjects

Subject anthropometrics were recorded during collection and are presented in Table 3.1. Body Mass Index was calculated based on subject mass, height, and age. Two groups were created to compare outcomes, one group of children with a diagnosis of cerebral palsy (CP) and another of typically developed (TD) peers as a control group. A range of children and young adults participated in the study for both groups. Through t-test comparison, weight ($p=0.019$) and height (p=0.049) were different between the groups. This is indicative of the differences between the populations, but for the purposes of this work the populations were considered similar as significance was not found in age or body mass index (BMI).

Table 3.1: Subject Anthropometrics. Significance between TD and CP groups is marked as * (p<.05), ** (p<0.01), *** (p<0.001).

Group	**Age (years)**	**Weight (kg)**	**Height (cm)**	**BMI**
TD	16±5.4	57.4±20.4	160.6±19.7	21.2±4.8
CP	13.5±2.4	36.2±5.6 *	144.0±6.6 *	17.2±2.1

3.3.2 Spatiotemporal Gait Parameters

The effect of a pediatric posterior walker on gait was quantified through spatiotemporal parameters (Table 3.2). Comparisons were also made between TD and CP groups to quantify the pathology of gait. One CP subject with a GMFCS of III was excluded from the analysis because they were unable to walk in their baseline trial without their own walker that they used on a daily basis. There was no significant difference in stride length between CP subject self-selected most and least affected limb so sides were combined for all data reported. The CP group was less stable and walked worse than the TD group in baseline trials based on the calculated spatiotemporal parameters. Normalized stride length was shorter in the CP group compared to the TD group when walking under normal conditions ($p<0.01$). The CP group walked at a lower velocity than the TD group for baseline gait ($p<0.01$). The CP group also spent more time in double support indicating a less stable gait.

The TD and CP groups also had significantly different parameters when walking with the walker as compared to their baseline walk. Velocity and stride length were worse for the TD group when walking with the walker. The CP group had no significant changes to their gait when walking with the walker but they trended towards walking slower. The TD group had shorter strides when walking with the walker than in their baseline trials. Percentage of time in single and double support were not different for either group.

Table 3.2: Spatiotemporal Parameters. All metrics are significant between TD and CP except opposite foot contact. Significance between baseline and walker trials is marked as * ($p<0.05$), ** ($p<0.01$), *** ($p<0.001$) and TD to CP marked as □ ($p<0.05$), □□ ($p<0.01$), □□□ ($p<0.001$).

Parameter Type	**TD (n=9)**		**CP (n=7)**	
Parameter	**Baseline**	**Walker**	**Baseline**	**Walker**
Spatial				
Stride Length (m)	1.29±0.15	1.17±0.17	0.92±0.17□□□	0.88±0.09
Normalized Stride Length	1.54±0.13	1.39±0.12*	1.20±0.18□□	1.15±0.14
Step Length (m)	0.63±0.07	0.58±0.08	0.45±0.08□□□	0.42±0.04
Normalized Step Length	0.75±0.06	0.69±0.06*	0.58±0.08□□□	0.56±0.06
Velocity (m/s)	1.23±0.11	1.07±0.14*	0.83±0.27□□	0.68±0.16
Temporal (gc = gait cycle)				
Single Support (% of gc)	77.9±3.40	77.3±3.28	71.4±5.36□	72.7±6.24
Double Support (% of gc)	22.1±3.40	22.7±3.28	28.6±5.36□	27.3±6.24
Stride Time (sec)	1.05±0.08	1.10±0.08	1.18±0.32	1.36±0.32
Cadence (step/min)	115.1±9.7	110.3±8.4	107.3±24.6	92.6±20.5
Single Support (sec)	0.82±0.05	0.85±0.05	0.84±0.19	0.98±0.21
Double Support (sec)	0.23±0.05	0.25±0.05	0.35±0.14	0.38±0.14
Opposite Foot Off (% of gc)	11.1±1.68	11.2±1.78	14.1±2.74□	13.8±3.26
Opposite Foot Contact (% of gc)	49.9±0.25	50.0±0.26	49.9±0.54	50.2±0.80
Toe Off (% of gc)	60.9±1.67	61.6±1.46	64.4±3.06□	63.7±3.37

The CP group was composed of a broad range of children with different GMFCS levels and different assistive device needs. For comparison between these subgroups, two groups were made based on their GMFCS level (Table 3.3). Two subjects had a GMFCS score of II and five subjects had a GMFCS score of III. Ranges for the GFMCS II group were reported as an average and standard deviation would not be representative. In baseline trials, the GMFCS II subgroup walked with greater velocity and lower stride time in their baseline gait compared to the whole CP group. Their stride and step lengths were also longer than the whole CP group. With the incorporation of the walker into their gait, the GMFCS II subgroup walked slower and with a shorter stride length. In walker-assisted gait, the percentage of time spent in single support decreased for the GMFCS II subgroup.

The GMFCS III subgroup walked slower and took shorter strides than the whole CP group. During walker-assisted gait the GMFCS III subgroup was more stable as they walked with an increased percentage of time in single support. The increase in single support resulted from earlier toe off times and no change in contralateral contact times over the gait cycle. Stride length trended towards being longer and velocity trended towards being slower in walker-assisted gait but was not significant.

The CP group with GMFCS III was further separated by the type of assistive device used for ambulation in their baseline trials in addition to their prescribed AFOs (Table 3.3). The subgroup that used crutches on a daily basis also used crutches in their baseline walk. No changes were significant in the “Crutches” subgroup (CS) when walking with the walker as compared to walking with crutches but the group size was small and trends were observed. The CS walked slower than the whole group and reciprocal effects on temporal parameters were calculated. Normalized stride length was also shorter in the baseline trials compared to the whole

CP group. When walking with a walker, the CS group increased the percentage of time they were in single support. The ipsilateral and contralateral toe off events occurred earlier in the gait cycle and shortened the time in double support.

Table 3.3: Spatiotemporal Parameters for subgroups of CP group based on GMFCS level. Significance between baseline and walker trials is marked as * ($p<0.05$), ** ($p<0.01$), *** ($p<0.001$) and baseline CP total to CP subgroup marked as □ ($p<0.05$), □□ ($p<0.01$), □□□ ($p<0.001$).

Parameter Type	**GMFCS II (n=2)**		**GMFCS III (n=5)**		**Crutches (n=3)**	
Parameter	**Baseline**	**Walker**	**Baseline**	**Walker**	**Baseline**	**Walker**
Spatial						
Stride Length (m)	0.96-1.27	0.89-0.99	0.84±0.07	0.86±0.10	0.87±0.02	0.90±0.08
Normalized Stride Length	1.34-1.50	1.17-1.23	1.11±0.10	1.14±0.17	1.12±0.10	1.16±0.20
Step Length (m)	0.47-0.60	0.42-0.48	0.41±0.03	0.41±0.04	0.42±0.01	0.43±0.04
Normalized Step Length	0.65-0.71	0.57-0.58	0.54±0.05	0.55±0.08	0.54±0.05	0.55±0.10
Velocity (m/s)	0.96-1.32	0.62-0.84	0.71±0.15	0.66±0.18	0.61±0.12	0.63±0.24
Temporal (gc = gait cycle)						
Single Support (% of gc)	67.6-78.4	61.0-71.6	70.8±5.20	75.2±3.99	68.9±6.04	75.3±5.62
Double Support (% of gc)	21.6-32.4	28.4-39.0	29.2±5.20	24.8±3.99	31.1±6.04	24.7±5.62
Stride Time (sec)	0.96-1.01	1.18-1.46	1.26±0.35	1.38±0.38	1.47±0.28	1.54±0.41
Cadence (step/min)	118.9-124.7	83.5-102.0	101.4±27.5	92.6±24.3	84.1±16.2	83.1±25.3
Single Support (sec)	0.68-0.76	0.84-0.89	0.88±0.21	1.03±0.24	1.00±0.17	1.14±0.24
Double Support (sec)	0.21-0.33	0.34-0.57	0.38±0.16	0.35±0.14	0.46±0.15	0.40±0.18
Opposite Foot Off (% of gc)	10.6-15.8	14.5-19.8	14.4±2.70	12.4±2.11	15.3±3.25	12.4±2.89
Opposite Foot Contact (% of gc)	49.7-49.9	51.0-51.1	49.9±0.66	49.8±0.62	50.0±0.66	49.9±0.71
Toe Off (% of gc)	60.7-66.5	65.1-70.2	64.7±3.08	62.2±1.74	65.8±3.43	62.2±2.44

3.3.3 Reaction Forces

Average GRFs over the gait cycle were compared between baseline and walker trials within groups. Average handle reaction forces for the walker assisted trials were splined to the length of the gait cycle for each group and plotted on a second axis alongside the GRFs. SPM analysis was performed to find differences between the average GRFs for both groups. The TD group walked with lower magnitude GRFs in all three directions when walking with the walker (Figure 3.2). Peak vertical GRF was lower during weight acceptance and in preparing for toe off for the TD group. Similar effects were observed over the same ranges for lateral and longitudinal forces in the TD group.

The CP group had a lower peak vertical force during loading in stance phase (Figure 3.3). Lateral GRF was lower over the terminal stance phase. Vertical loading on the walker was greater and more variable in CP gait compared to TD gait. Cyclical loading occurred in the vertical and longitudinal directions in the CP group with maxima during double support times following ipsilateral and contralateral foot strikes. The TD group applied comparably constant, low-magnitude forces to the walker.

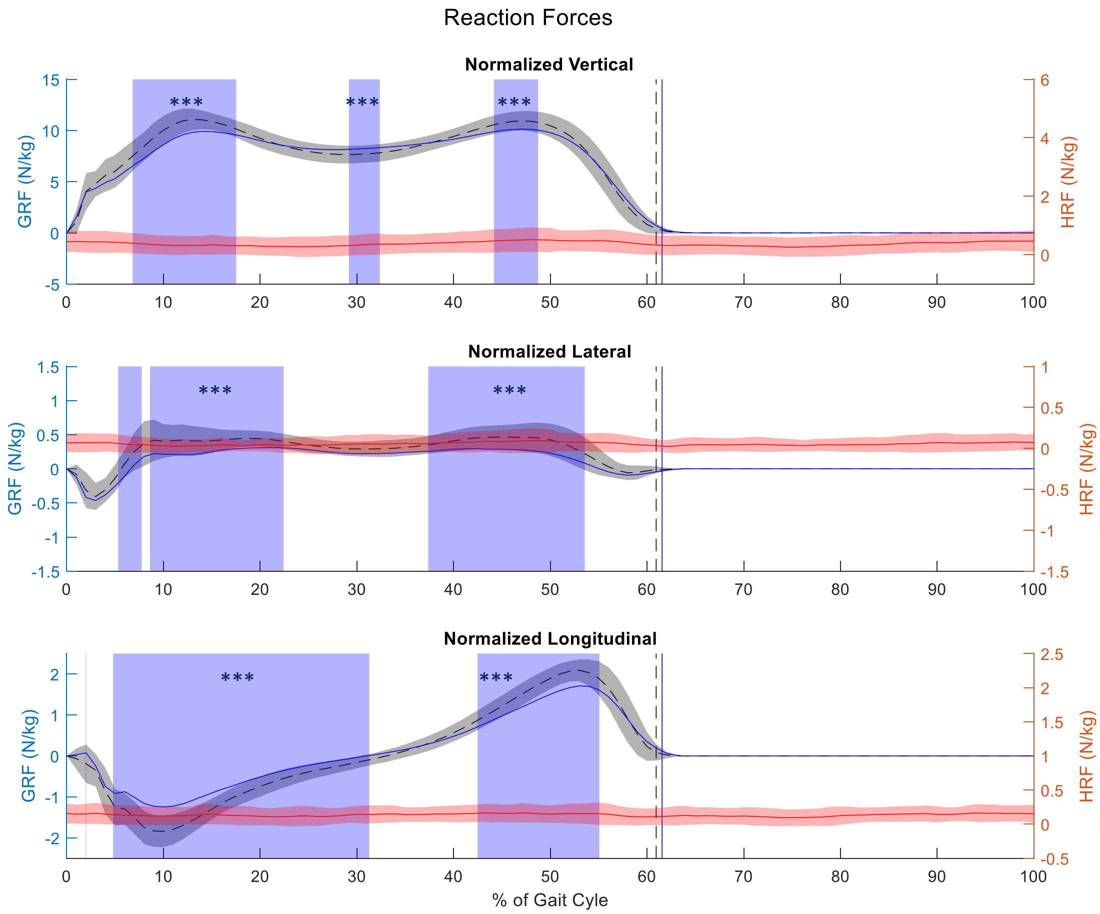

Figure 3.2: TD group 3-dimensional ground reaction forces over gait cycle. Significance between baseline (black and gray) and walker (blue) trials highlighted by blue shaded boxes. SPM significance marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Toe-off time marked by vertical lines for baseline (black) and walker assisted (blue) trials.

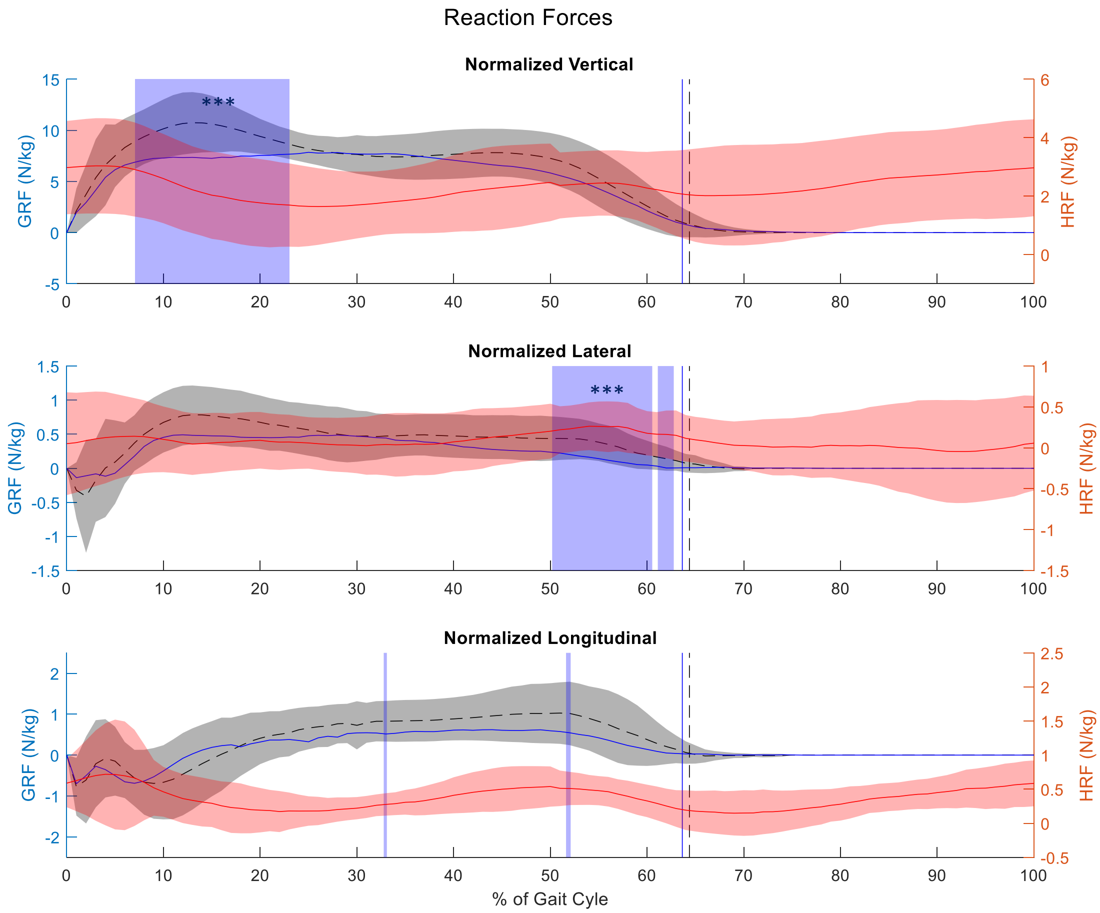

Figure 3.3: CP group 3-dimensional ground reaction forces over gait cycle. Significance between baseline (black and gray) and walker (blue) trials highlighted by blue shaded boxes. SPM significance marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Toe-off time marked by vertical lines for baseline (black) and walker assisted (blue) trials.

The CP group was separated into subgroups by GMFCS score and by assistive device used in baseline trials. The Crutches subgroup was created from members of the GMFCS III subgroup and provided further knowledge of the CP group as a whole. GRFs for baseline and walker-assisted gait and handle reaction forces for walker-assisted gait were examined within the subgroups (Figure 3.4). Vertical GRF was lower in all subgroups at the initial peak near the occurrence of contralateral foot off. The walker was loaded more vertically and laterally in the GMFCS III and Crutches subgroups compared to the GMFCS II subgroup. Further cyclical interaction with the walker was observed in the GMFCS III and Crutches subgroups. Vertical handle reaction force was greatest at ipsilateral and contralateral heel strikes. In the longitudinal direction peak force pulling the walker forward occurred during double support time following heel strikes during the gait cycle. The GMFCS II group applied forces to the walker that were greater than the TD group, but no significant pattern of use was evident.

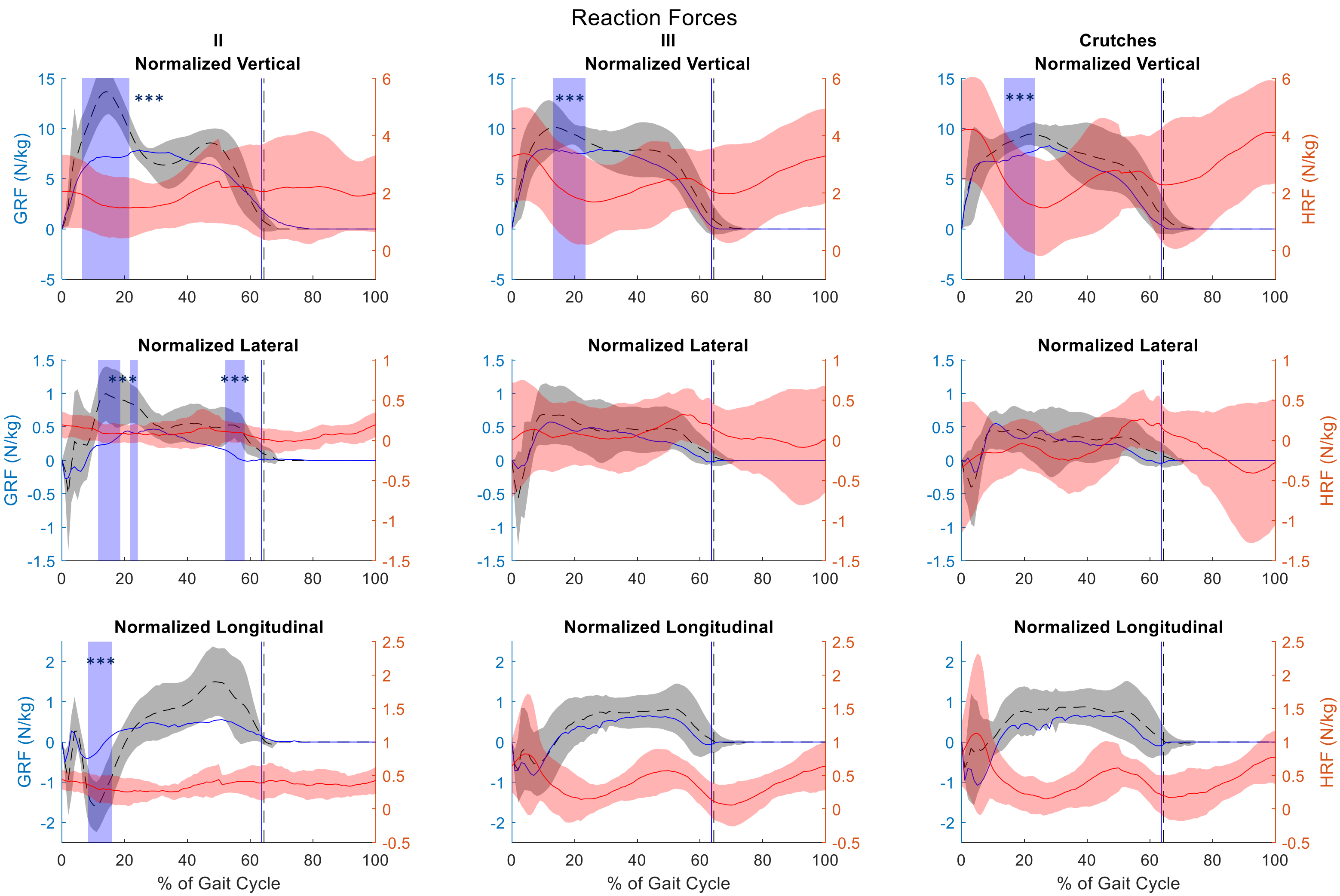

Figure 3.4: Average ground reaction forces for subgroups: GMFCS II (left), GMFCS III (middle), and Crutches (right) over a gait cycle for baseline (black and gray) and walker-assisted (blue) trials for the left oriented y-axis scale. The handle reaction forces (red) for walker-assisted gait are shown with the corresponding y-axis scale on the right. Toe-off time marked by vertical lines for baseline (black) and walker assisted (blue) trials. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

3.4 Discussion

Spatiotemporal gait parameters were used as outcome measures for analysis of the effects that posterior walkers have on TD and CP gait. The effect of pathological gait was seen by comparing baseline TD and CP gait. The incorporation of the walker into subject gait was examined by comparing the similarities and differences in spatiotemporal parameters between the two groups as well as reaction forces from the ground and handles of the walker. The TD group was inhibited by the walker as spatiotemporal parameters were worse in their gait with a walker. The whole CP group walked slower and had a shorter cadence in walker-assisted gait but was not as greatly affected by the walker on average as the TD group. Breaking the CP group up by GMFCS level revealed effects of the walker on the different groups within the CP group more decisively. The GMFCS II group experienced similar deleterious effects as the TD group which was expected as they would not typically be prescribed a walker for movement aid. The GMFCS III group increased their stability by walking with more time spent in single support with the walker compared to their baseline gait. They also applied more supporting force to the walker than the GMFCS II group. This analysis highlights the utility of the GMFCS for identifying populations of children with CP who can benefit from a posterior walker and separate those who would not be positively affected. The spatiotemporal parameters and reaction forces show a broad view of the walker's effect on gait and further analysis is performed in later chapters.

In baseline trials, most spatiotemporal parameters were significantly different ($p<0.05$) between the TD and CP groups. The results for stride length, velocity, and cadence agree with previous work (Armand, Decoulon, & Bonnefoy-Mazure, 2016b; Brégou Bourgeois, Mariani, Aminian, Zambelli, & Newman, 2014; Ju Kim & min Son, 2014), falling within the ranges of the presented values. A decreased velocity noted in the CP group was primarily a result of a decreased stride length. A lower cadence in CP gait also contributed to a lower velocity but the standard deviation of the CP group was very large and previous work has found mixed results on whether cadence is as much a contributor to overall velocity as stride length.

Temporal events also showed significance between the TD and CP groups. The control group performed similar gait patterns to that presented by Schwartz et al. (Schwartz, Rozumalski, & Trost, 2008). The CP group had a later occurrence of contralateral and ipsilateral foot off. The longer relative time until toe off for the CP gait cycle increases the time in double support indicating the need for more stability and balance with two points of contact with the ground. Children with CP are inherently less balanced and in control of their motor functions. The CP group showed an increase in double support time and decrease in velocity and stride length to enable a stable gait over the test area.

The CP group was split by GMFCS classification between II and III. The GMFCS II subgroup walked faster, at a faster cadence, and with longer strides than the whole CP group in baseline trials. The GMFCS III subgroup walked at a slightly slower speed and with a shorter stride length than the whole group. The GMFCS II subgroup was expected to walk with a greater velocity and longer stride length as this indicates a more stable gait in the population less affected by CP. While the subgroups differed in velocity, cadence, and stride length, they had similar gait event occurrence times and thus single and double support times.

The CP group was split into subgroups where the baseline walk was performed with assistive walking crutches and without additional support outside of prescribed AFOs. The CS group had a lower normalized stride length than the whole group. While using the crutches for assistance, the length of the CS group’s stride was limited by the cyclical placement of the crutches on the ground. Velocity and cadence were also negatively affected by using the crutches compared to the whole CP group. By breaking the CP group into subgroups by device used in baseline trials and GMFCS score, expected differences were observed but important similarities were also found. Particularly, the time spent in single and double support by all subgroups was similar to the whole group. This metric was used as a critical indicator of stability when the walker was introduced into the subject's gait. An increase in stability would be indicated by an increase in time spent in single support.

Posterior walkers and other assistive devices are used by many populations to increase stability by providing additional ground contact points throughout ambulation. Stride length, velocity, and cadence were all lower in walker-assisted gait for the TD group. The CP group showed similar trends in the stride parameters in walker-assisted gait except the stride length did not change. These values agree with that presented in previous work (Krautwurst et al., 2016a; Park et al., 2001). Along with the stride parameters, both group's total stride time was greater (p=0.005). The walker inhibited the gait of children with CP and TD children which is shown through the lower average velocity but it did not change the percentage of time spent in single support for either group. The time in single support for CP gait was greater in walker-assisted trials but the percentage of time spent in single and double support did not change, but by breaking up the CP group further analysis showed expected differences.

The GMFCS II group was negatively affected by the posterior walker. A shorter stride length, lower velocity, and shorter percentage of time in single support indicate the low utility the GMFCS II group received from the walker based on spatiotemporal parameters. The GMFCS III group increased time spent in single support but trended towards walking slower and with worse cadence. Walker-assisted gait was better than crutch-assisted gait for subjects who used crutches on a daily basis. Although the CS group typically did not use a walker, the walker enabled a more stable gait than walking with their crutches. Overall, a posterior walker had mixed effects on gait in the CP group, the GMFCS III users who would typically be prescribed a walker experienced both positive and negative effects on spatiotemporal metrics. To further define the walker interaction, ground and walker handle reaction force was compared between baseline and walker-assisted gait.

GRFs for TD and CP groups were different in vertical, lateral, and longitudinal directions. The TD group velocity with the walker was lower ($p<0.001$) compared to baseline and the changes in the TD group GRFs are characteristic of children walking with lower velocity (Schwartz et al., 2008). The CP group presented lower magnitude GRFs compared to baseline. The differences were similar to the TD group changes and the CP group also walked with a lower velocity ($p<0.001$) when using the walker than in baseline trials. While the changes seen in the GRFs could be attributed to the velocity decrease in walker trials, the handle reaction forces revealed a difference in how the two groups incorporated the walker in their gait. The TD group applied relatively constant force in all three directions to the walker over the gait cycle, which was expected as the walker does not increase the gait efficiency or stability through the TD group relying on it during their gait. The CP group applied greater and more variable downward force to the walker, indicating the reliance of the children with CP on the walker for cyclical use, thus decreasing the load on the lower body. The longitudinal force was relatively constant throughout the gait cycle except for time periods right after heel strike where the force peaked. The increased longitudinal force to pull the walker along coincided with the double support time for the CP group. The CP group used the walker as a relatively static support for their gait during single support and pulled the walker during times of greater stability in their gait.

The GMFCS III and CS groups had lower initial peak vertical GRF in walker assisted gait and cyclically loaded the walker vertically and longitudinally. Decreased vertical ground reaction forces were expected because the groups walked slower in walker-assisted gait than in baseline (Schwartz et al., 2008). The CP subgroups had greater magnitude decreases in peak vertical ground reaction forces than what is explained by their decrease in velocity and this can be attributed to their reliance on the walker. The subgroups relied on the walker most at heel strike and trended lower during double support followed by increased vertical loading during single support. GMFCS III and CS groups pulled the walker most during double support times. For the subgroups that relied on the walker, cyclic patterns of use were observed and loading on the walker coincided with gait event times associated with single and double support. This usage pattern is ideal for designing an intelligent motorized controller based on these forces to decrease the need for longitudinal pulling of the walker. The effect of the walker on the GRF of the GMFCS II group was like that seen in TD gait as the GRF was decreased in all directions. The GMFCS II group relied on the walker for load bearing by vertical loading but no cyclic pattern was observed. The walker was not positively incorporated into the gait of the GMFCS II group as they did not interact with it in a cyclic way. From this analysis, the GMFCS II subgroup did not need the walker, but relied on it for support throughout the gait cycle to the detriment of their velocity, stride length, and single support time. The subgroups have a lower sample size so it is likely that with a larger group there would be a more observable effect of the walker on reaction forces of the subgroups.

Overall, walker-assisted gait in the TD group was worse than baseline and the results show that the walker potentially makes the user less stable and unbalanced. The control group was not expected to use the walker for support, the spatiotemporal parameters were expected to change in walker-assisted gait and possibly contribute to an abnormal gait pattern. The CP group showed similar trends in spatiotemporal parameters between baseline and walker trials as the TD group. The CP group used the walker for support through downward force on the walker throughout the gait cycle. Longitudinal force applied by the CP group was low except during double support time where children with CP were most stable and pulled the walker along for the next static period where the longitudinal force is lower. The CP group (n=9) could lack power in some parameters as higher standard deviations occurred compared to the TD group. This is likely due to the broad group of children with spastic, diplegic CP that volunteered for the study and the nature of the pathologic gait presenting more variably than TD gait. Time in single support increased for both groups which indicates a more stable gait but the percentage of the gait cycle spent in single support was not significantly greater. There were mixed results as to the positive effects of the walker on gait in both groups.

The walker had similar effects on spatiotemporal parameters of both groups to different degrees and thus it was difficult to fully define the utility of walker-assisted gait in pathologic gait with spatiotemporal parameters alone. Subgroups within the whole CP group were positively and negatively affected by the walker in how much time they spent in single and double support. Negative effects of the walker on gait were evident from spatiotemporal parameters in all subjects. The force applied to the walker offered a glimpse into how the TD and CP groups differ in walker-assisted gait. There was no conclusive evidence that the walker had any positive effect on gait for either whole group, although the GMFCS III and CS subgroups had better stability and balance in walker assisted gait while they also cyclically used the walker for support and to pull it along during their gait.

3.5 Conclusion

The spatiotemporal analysis in TD and CP groups quantified the difference in various gait characteristics between and within the populations. Results confirmed previous work that CP gait is generally less stable without assistive devices compared to TD peers and that velocity during gait is typically lower in CP. Differences within populations were also observed when walking with and without a pediatric posterior walker. As they are currently designed, walkers benefit a small subgroup of children with CP while for other groups the walker offers some benefit but also some detriment. This work helps to identify these groups and offers information that could influence the future design of a better walker. The results presented in this work are only a piece of how the walker's effect can be quantified. More on the walker's effect on gait will be discussed in Chapter 4 using model outputs for joint kinematics and kinetics.

Reference List

Armand, S., Decoulon, G., & Bonnefoy-Mazure, A. (2016). Gait analysis in children with cerebral palsy. EFORT Open Reviews, 1(12), 448–460. https://doi.org/10.1302/2058-5241.1.000052

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: Benefits, demands, and adverse consequences. Archives of Physical Medicine and Rehabilitation, 86(1), 134–145. https://doi.org/10.1016/j.apmr.2004.04.023

Brégou Bourgeois, A., Mariani, B., Aminian, K., Zambelli, P. Y., & Newman, C. J. (2014). Spatio-temporal gait analysis in children with cerebral palsy using, foot-worn inertial sensors. Gait and Posture. https://doi.org/10.1016/j.gaitpost.2013.08.029

Fast, A., Wang, F. S., Adrezin, R. S., Cordaro, M. A., Ramis, J., & Sosner, J. (1995). The instrumented walker: Usage patterns and forces. Archives of Physical Medicine and Rehabilitation, 76(5), 484–491. https://doi.org/10.1016/S0003-9993(95)80584-2

Friston, K. J., Ashburner, J. T., Kiebel, S. J., Nichols, T. E., & Penny, W. D. (2011). Statistical parametric mapping: the analysis of functional brain images. Amsterdam: Elsevier.

Greiner, B. M., Czerniecki, J. M., & Deitz, J. C. (1993). Gait parameters of children with spastic diplegia: a comparison of effects of posterior and anterior walkers. Archives of Physical Medicine and Rehabilitation, 74(4), 381–385. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8466419

Johnson, D. C., Damiano, D. L., & Abel, M. F. (1997). The Evolution of Gait in Childhood and Adolescent Cerebral Palsy. Journal of Pediatric Orthopaedics, 17(3), 392–396. Retrieved from https://ovidsp.tx.ovid.com/sp-3.30.0b/ovidweb.cgi

Ju Kim, C., & min Son, S. (2014). Comparison of Spatiotemporal Gait Parameters between Children with Normal Development and Children with Diplegic Cerebral Palsy. Journal of Physical Therapy Science, 26(9), 1317–1319. Retrieved from https://www.jstage.jst.go.jp/article/jpts/26/9/26_jpts-2014-008/_pdf/-char/ja

Krautwurst, B. K., Dreher, T., & Wolf, S. I. (2016). The impact of walking devices on kinematics in patients with spastic bilateral cerebral palsy. Gait & Posture, 46, 184–187. https://doi.org/10.1016/j.gaitpost.2016.03.014

Logan, L., Byers-Hinkley, K., & Ciccone, C. D. (1990). Anterior Versus Posterior Walkers: A Gait Analysis Study. Developmental Medicine & Child Neurology, 32(12), 1044–1048. https://doi.org/10.1111/j.1469-8749.1990.tb08521.x

Park, E. S., Park, C. I., & Kim, J. Y. (2001). Comparison of anterior and posterior walkers with respect to gait parameters and energy expenditure of children with spastic diplegic cerebral palsy. Yonsei Medical Journal, 42(2), 180–184. https://doi.org/10.3349/ymj.2001.42.2.180

Pataky, T. C. (2010). Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. Journal of Biomechanics. https://doi.org/10.1016/j.jbiomech.2010.03.008

Pataky, T. C. (2012). One-dimensional statistical parametric mapping in Python. Computer Methods in Biomechanics and Biomedical Engineering, 15(3), 295–301. https://doi.org/10.1080/10255842.2010.527837

Pataky, T. C., Robinson, M. A., & Vanrenterghem, J. (2016). Region-of-interest analyses of one-dimensional biomechanical trajectories: bridging 0D and 1D theory, augmenting statistical power. PeerJ, 4, e2652. https://doi.org/10.7717/peerj.2652

Pauk, J., Ihnatouski, M., Daunoraviciene, K., Laskhousky, U., & Griskevicius, J. (2016). Research of the spatial-temporal gait parameters and pressure characteristic in spastic diplegia children. Acta of Bioengineering and Biomechanics, 18(2), 121–129. https://doi.org/10.5277/ABB-00393-2015-02

Prosser, L. A., Lauer, R. T., VanSant, A. F., Barbe, M. F., & Lee, S. C. K. (2010). Variability and symmetry of gait in early walkers with and without bilateral cerebral palsy. Gait & Posture, 31(4), 522–526. https://doi.org/10.1016/j.gaitpost.2010.03.001

Rosenbaum, P., Paneth, N., Leviton, A., Goldstein, M., Bax, M., Damiano, D., … Jacobsson, B. (2007). A report: the definition and classification of cerebral palsy April 2006. Developmental Medicine & Child Neurology, 49(SUPPL.109), 8–14. https://doi.org/10.1111/j.1469-8749.2007.tb12610.x

Schwartz, M. H., Rozumalski, A., & Trost, J. P. (2008). The effect of walking speed on the gait of typically developing children. Journal of Biomechanics, 41(8), 1639–1650. https://doi.org/10.1016/J.JBIOMECH.2008.03.015

Posterior Walker-Assisted Gait of Children With and Without Cerebral Palsy: Kinematics and Kinetics

4.1 Introduction

Children with CP typically are often prescribed posterior walkers to enable more stable gait and better posture. The positive effects of a posterior walker must outweigh the negative effects for the user to willingly incorporate it into their gait, but previous work has not quantified the full effect of a posterior walker gait. In Chapter 3, spatiotemporal characteristics revealed a generally negative effect on gait in both CP and TD groups. The GMFCS III subgroup walked with more time in single support with the walker than without and applied cyclic forces to the walker. Previous work has looked at the joint kinematics of the lower body and torso joints during gait to quantify posture (Brooke M. Greiner, Czerniecki, & Deitz, 1993; Krautwurst et al., 2016a; Logan, Byers-Hinkley, et al., 1990). In the clinical setting, some measurements are harder to accomplish directly. A model can be utilized to calculate joint kinematics and kinetics and examine the walker's effect on posture and overall gait with the data that was collected. No previous work has been able to perform full body kinetics on gait with a walker, this work developed a full kinetic model for gait with a posterior walker. Model utility for clinical gait analysis is dependent on the simplifying assumptions used to perform the dynamic analysis in the model.

Many different models have been created for pathologic gait to measure body segment dynamics. Vicon Nexus incorporates the conventional gait model (CGM) that has been used for many applications in gait analysis (Baker, Leboeuf, Reay, & Sangeux, 2017). The CGM relies on kinematic markers to model full body movement. The segments of the body change characteristic length and orientation based on marker movement during gait. The upper body model has not been validated using the CGM and this work requires an upper body model to complete the analysis. For these reasons the CGM formulation in Vicon Nexus's Plug-in-Gait model was not used. Baker et al. propose that inverse kinematic models are more suitable to clinical gait analysis (Baker et al., 2017). A rigid body model performs better in dynamic simulations by applying forces and torques to segments which don't change defining characteristics from frame to frame.

Inverse kinematic models use rigid bodies to represent body segments and drive the motion of the body through an optimization technique to minimize errors between the body's position and the measured kinematic markers. The kinematic markers do not dictate the exact movement of the body as they do in CGM but instead contribute to the model's calculations for minimizing the error between measured and expected kinematic marker positions to drive motion. Seth et al. describe an inverse kinematic, musculoskeletal model in OpenSim to observe muscle group activation in normal and pathological gait (Seth, Sherman, Reinbolt, & Delp, 2011). Musculoskeletal models in OpenSim examine lower body, upper body, or specific muscle groups but typically do not look at the whole body with all the incorporated musculoskeletal elements as it would be computationally inefficient. Additionally, OpenSim does not have a robust computational system to manage contact forces aside from ground contact forces. In this work, along with the analysis of the body's kinematics, the interaction of the body with the ground and with a posterior walker was modelled. In OpenSim, the necessary contact forces could not be easily developed to model the necessary interactions. Alternate approaches were considered which could quantify whole-body kinetics and perform individual segment analysis. Observation of muscle activation and force was outside of the scope of this work.

MSC Adams is a multibody dynamics solver software capable of inverse kinematic and kinetic modelling of systems that is used in academic and industrial settings for research and product development. In MSC Adams a multibody dynamic model of the human gait can be created using the Generator of Body Data program (Baughman, 1983). MSC Adams has multiple, robust contact models built in which this work required in order to facilitate the modelling of the ground contact force and walker handle contact force on the body. Inverse kinematic simulations in MSC Adams act to minimize the energy in spring-damper systems between measured and modelled kinematic markers to accomplish the motions with the given constraints of the system. Resultant models in MSC Adams can be queried for individual joint kinetics, center of mass movement, and other user-defined parameters. Additionally, Adams and Matlab Simulink have a co-simulation environment where dynamics and controllers can be developed and implemented. This will be used in future work to develop a controller for automating the walker to further increase the benefit to the user. This work used MSC Adams with the LifeMod plug-in to develop a dynamic model (S. Russell, Bennett, Sheth, & Abel, 2011b).

In this chapter, the methods used to create a dynamic walking model of human gait with the addition of an instrumented posterior walker are detailed. Subjects walked under two conditions: their typical walk (baseline) and gait with a posterior walker. The posterior walker was instrumented in a similar manner to the method of Konop et al. (Konop et al., 2009). Full body kinematics and kinetics were examined over a gait cycle for both TD and CP groups. Posture analysis was performed by examining joint flexion-extension angles at specific time points during gait and comparing to baseline TD gait which was assumed to have an optimal posture for gait. Through validating the model, we can quantify kinetics of the body which has not been done in previous work.

4.2 Methods

4.2.1 Subjects and Procedure

The kinematic data for 17 children was collected in the Motion Analysis and Motor Performance Laboratory at the University of Virginia. There were two populations represented within the group. The first was composed of 8 children with a diagnosis of spastic, diplegic Cerebral Palsy (CP). The second group of 9 children consisted of controls with no known musculoskeletal pathologies. The group with CP had prescribed ankle foot orthotics (AFOs) and wore them during testing. Six of the children had a GMFCS of III, two typically used a walker to move. Four children used forearm crutches regularly for movement in the community. The other two children had a GMFCS II and did not require assistive devices except for long-distance settings. The children were community ambulators and had ranges of experience with assistive devices. The children were asked to walk over the short distance of the testing area without the aid of an assistive device. Four children with a GMFCS III used crutches to walk for the baseline trials and one child used a walker for baseline trials. The data for the subject who used a walker in baseline trials was used for kinematic and kinetic analysis for walker-assisted trials only. Statistical analyses were performed assuming unequal variances with this additional subject. Subject assent and parental consent were approved by the University of Virginia's Human Investigation Committee and were obtained for all subjects.

A full body marker set of 38 kinematic markers following the Plug-in-Gait marker set was attached to all subjects. Subjects walked along a straight path for 15 meters at their self-selected comfortable walking speed. Three-dimensional kinematic data were collected using a 12-camera Vicon Nexus Motion Capture System (Oxford Metrics, UK) sampling at 100Hz. Ground reaction forces were collected with five Bertec force plates sampling at 1000Hz. The force plates were embedded in the floor in the center of the path for each trial. For trials with the instrumented posterior walker, three-dimensional handle reaction forces were collected with 6-DOF ATI load cells sampling at 1000Hz. The load cells were connected in line between the walker and handle grips (Figure 4.1). Each subject completed a minimum of three trials for each walking condition where two sequential steps were labelled for events of foot strike and foot off. The first condition for walking was a baseline, where the subject walked without any device aid other than their AFO's if prescribed or with crutches as necessary. For the second condition the children were instructed to walk with a pediatric posterior walker. The data collected in Vicon was processed with the Nexus program where a total of 7 gait cycle events were labelled: foot strike, contralateral toe off, contralateral foot strike, toe off, foot strike, contralateral toe off, and contralateral foot strike. Events were labelled using recorded force plate data over the trial. Foot strike occurred when the vertical force on the respective plate was greater than a 15 N threshold. Toe off was marked when the vertical force fell below the 15 N threshold.

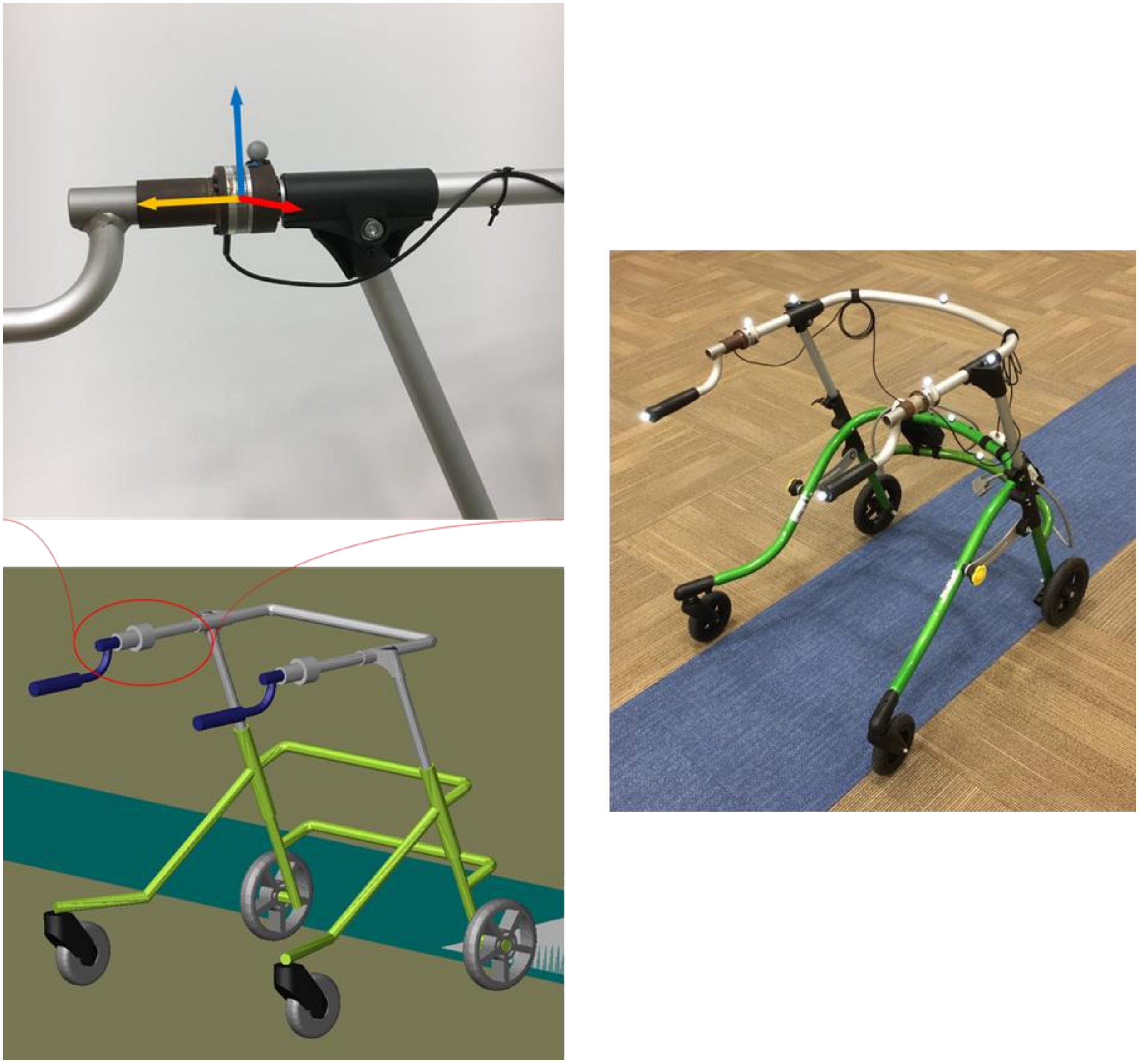

Figure 4.1: Instrumented posterior walker with 6 DoF load cells attached between handles and frame of walker.

4.2.2 Preparation for Modelling

Kinematic and kinetic data were exported from Nexus for processing in Matlab before the trials were modelled. Along with stored values for the subjects' anthropometric data the kinematic data for the body was written to a file for MSC.ADAMS to read. Force and moment data from the force plates and load cells was filtered using a lowpass, Butterworth filter with a cutoff frequency at 100Hz. Kinematic marker positions were filtered using a lowpass, Butterworth filter with a cutoff frequency of 15Hz. The global coordinate system (ŭ) in Vicon had the X axis aligned with the direction of travel, the Y axis normal to the sagittal plane, and the Z axis positive in the vertical direction. The model was created in a global coordinate system (ũ) where the X axis was aligned with the direction of travel, the Y axis was positive in the vertical direction, and the Z axis was normal to the sagittal plane. The coordinate system for the model is defined from the Vicon global reference frame by a 90-degree rotation about the positive X axis.

$$R = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 0 & 1 \\ 0 & -1 & 0 \end{bmatrix} \qquad \tilde{u} = R\breve{u}$$

4.2.3 Physical Model

A scaled model was created for each subject in MSC.Adams with the LifeMod plugin using subject-specific anthropometric data (age, height, mass, and gender). The model (Figure 4.2) consisted of 19 rigid body segments: head, neck, upper torso, central torso, lower torso, clavicle (2), upper arms (2), lower arms (2), hands (2), upper legs (2), lower legs (2), and feet (2). The physical properties of the body's segments were defined using the Generator of Body Data (GeBOD) database (Cheng, Obergefell, & Rizer, 1994). Center of mass was calculated from the segment's positions. 18 joints were defined between segments and allowed triaxial rotation except for the wrist and elbow which allowed two axes of rotation and the knee which allowed one. An additional joint in the upper neck existed in the model but was locked and so was not considered in the analysis.

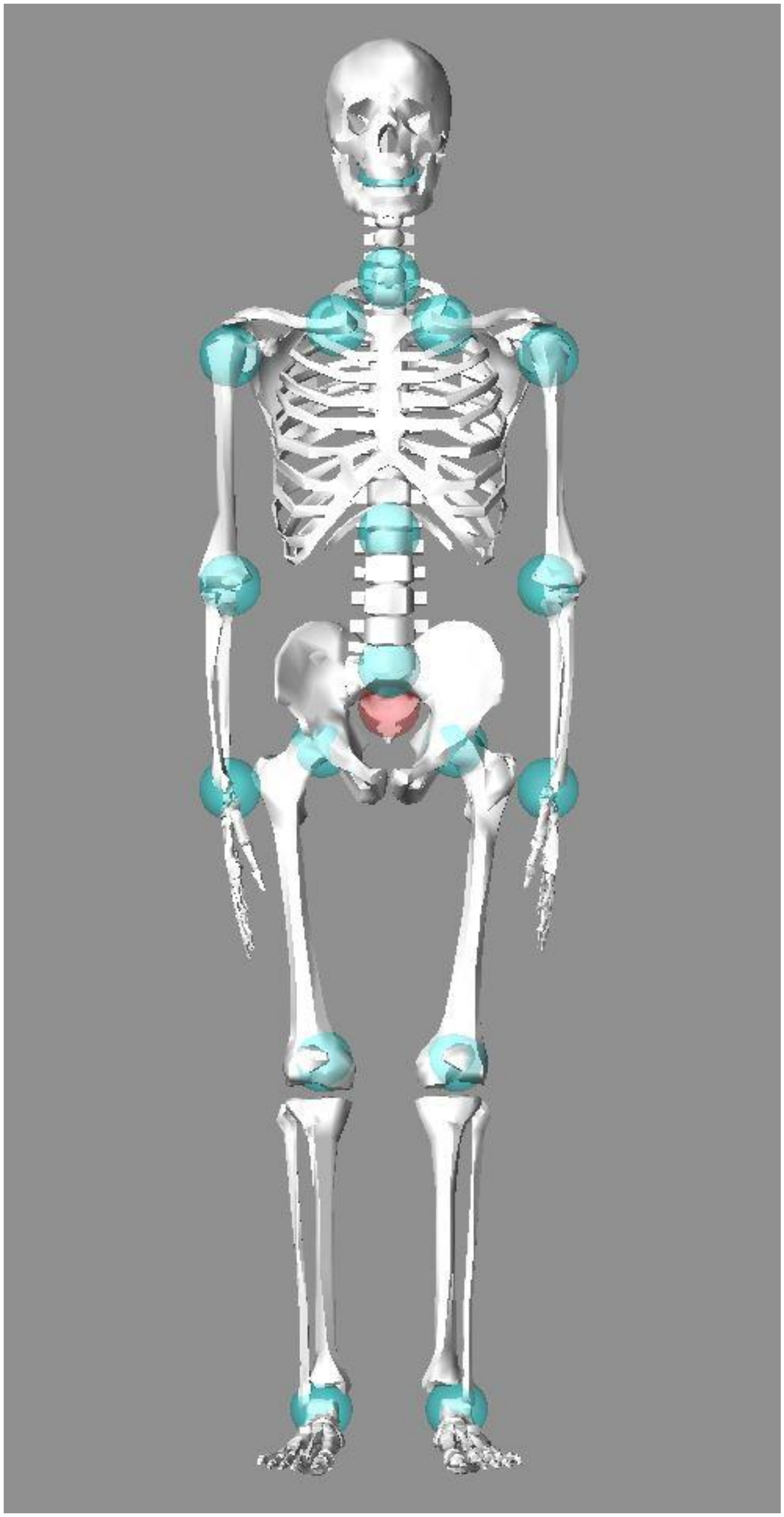

Figure 4.2: Rigid body model segments imported and scaled from GeBOD database with joints (blue spheres) and center of mass (red sphere).

For each subject a static trial was used to scale the body from the GeBOD database using age, weight, height, and gender as defining parameters. Three trials for each walking condition were modelled for kinematic and kinetic analysis of the body over the gait cycle. The motion of the body was driven by the kinematic markers attached to the model by spring-damper systems. Dynamic optimization of the springs to maintain minimum residual spring energy were used in inverse kinematic analysis to drive the motion of the body and joint kinematics were recorded. Recorded joint motion was used as input for PD controllers on the joints of the body during dynamic simulations. Integral components were not included in the controllers because there is no steady-state period for which they are most useful. An integral component would act to destabilize the model because kinetics in gait are periodic and dynamic. For the dynamic simulations, other necessary forces were added to the model along with fixed-position blocks signifying the floor and embedded force plates. Gravity was generally applied to the whole body. Ground reaction forces from force plates embedded in the floor had varying centers of pressure and multiple methods were considered to increase accuracy and simplicity for the model. Ground reaction forces were applied as contact forces from the center of the respective force plates to the center of mass of the respective foot. Moments were also applied which acted to correct the moment arm in the force application to the ankle. This method was found to be accurate for force application as three-dimensional ankle, knee, and hip moments were correctly applied in the PD controller. Ground reaction force was trimmed to include force plate data from the identified gait cycles within the trial. One gait cycle was considered a left foot heel strike to left foot heel strike and vice versa in the right foot. Twenty frames were included on either side of the gait cycles to avoid modelling error during the time period of interest. This placed time periods of instantaneous force application from the force plates outside of the gait cycle for the later analysis to not include impulse controlling moments in the joints.

The model analyzed joint kinematics and kinetics at discrete time points where kinematic motion drove the motion and the force application to the body was also discrete. The reaction forces could thus be applied at times off by one hundredth of a second and this would cause discrete error between the body and force application time. To account for this discrete error a spring-damper control on rotational and translational motion of the body center of mass was created to lock the center of mass motion to follow the recorded motion from the inverse kinematic simulation. The center of mass control applied the minimum force needed to correct the body for discrete error inherent to dynamic simulations. For all trials, joint kinematics and kinetics and center of mass movement were exported from dynamic simulations with the body, ground reaction forces, and gravity as inputs.

For trials with the instrumented walker a model was created to include the handle reaction forces. The creation of the walker model is detailed in Appendix 1 (Figure 4.3). The walker motion was driven by spring-dampers applied from the locations of the kinematic markers defined on the walker and each trial's kinematic data. A contact force between the walker handles and the center of mass of the hand was created for the application of the force from the load cells. The reference marker used for this force was aligned with the origin of the load cell. Joint kinematics and kinetics and center of mass motion were exported as model outputs from the dynamic simulations for walker trials.

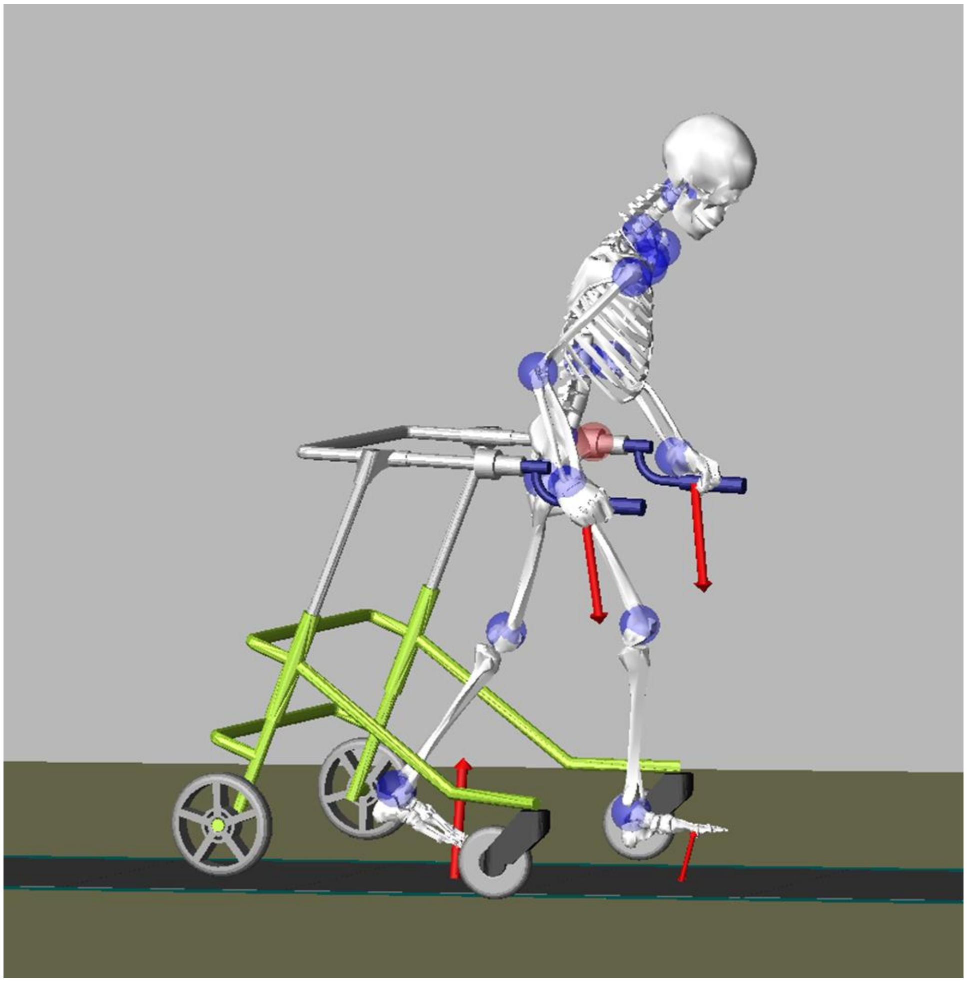

Figure 4.3: Sample frame from dynamic model simulation of walker-assisted gait. Red arrows indicate contact forces applied to the body.

4.2.4 Kinematics and Kinetics

Joint kinematics and kinetics were output from the model for the whole gait cycle. The model outputs for joint angles were used to quantify posture by looking at the knee, hip, and trunk angles along with the pelvis segment orientation. The traditional method to examine posture uses four time points of the gait cycle to represent a range of angles through the gait cycle: initial heel strike, midstance, pre-swing, and mid-swing (Logan, Byers-Hinkley, et al., 1990). Midstance and mid-swing times were defined as half the time between heel strike and toe off and vice versa (Gibson, Jeffery, & Bakheit, 2006). The novel method for examining the kinematics and kinetics of walker use utilized statistical parametric mapping to measure the significance of changes in the knee, hip, and trunk over the whole gait cycle. A better posture would be seen by a more upright position of the three joints and a lower variance in angles in the trunk. TD and CP group baseline gait was compared to observe the pathology of the CP group. To avoid confounding statistical variables four subjects from the CP group were used as they did not require crutches or other assistance during their baseline gait.

Joint kinematics and kinetics over the gait cycle were compared between baseline and walker-assisted gait for the TD group and the CP group. The group of CP subjects who used crutches was compared in their baseline and walker-assisted gait to the gait of the unassisted group during their walker-assisted gait. The effect of the walker on model outputs provided potential reasons for the spatiotemporal differences noted in Chapter 3. It was important to keep these changes in mind as the model outputs were examined over single gait cycles that were normalized by stride time. Since velocity was lower in walker-assisted trials, some of the model outputs were expected to reach lower peaks over the gait cycle (Schwartz et al., 2008).

4.2.5 Statistical Parametric Mapping

Baseline and walker trial outputs were compared using one-dimensional statistical parametric mapping (SPM) originally designed for neurological testing (Friston et al., 2011). SPM has been shown to be a valuable statistical test for biomechanical comparisons by Pataky in a series of studies and SPM1D software interface in Python and Matlab (Pataky, 2010, 2012). SPM analysis performs better over time periods than individual tests of each point because it considers the surrounding data points in calculating the p-value in post hoc analysis (Pataky et al., 2016). SPM allowed for the comparison of joint kinematics and kinetics over the gait cycle and identified time periods where walker-assisted gait deviated from baseline gait.

4.3 Results

4.3.1 Subjects

Subject anthropometrics were recorded during collection and are presented in Table 4.1. Body Mass Index was calculated based on subject mass, height, and age. Two groups were created to compare outcomes, one group of children with a diagnosis of cerebral palsy (CP) and another of typically developed (TD) peers as a control group. A range of children and young adults participated in the study for both groups. Through t-test comparison, weight ($p=0.019$) and height ($p=0.049$) were different between the groups. This highlights the differences between the populations, but for the purposes of this work the populations were considered similar as significance was not found in age or body mass index (BMI).

Table 4.1: Subject Anthropometrics. Significance between TD and CP groups is marked as * (p<.05), ** (p<0.01), *** (p<0.001).

Group	Age (years)	Weight (kg)	Height (cm)	BMI
TD	16±5.4	57.4±20.4	160.6±19.7	21.2±4.8
CP	13.5±2.4	36.2±5.6 *	144.0±6.6 *	17.2±2.1

4.3.2 Walker Effect on Posture

The effect of the walker on the posture of the user was quantified by calculating the knee, hip, and trunk angles at four time points for baseline and walker trials (Table 4.2). The pelvis flexion angles are defined by a negative value indicating a forward rotation. The model's neutral pelvic angle was not equivalent to anatomical neutral. Static orientation of the model made the lumbar takeoff angle -18 degrees. For the CP group, greater negative values resulted because the model did not account for pelvic tilt of the subject. The effect of this is carried over into the lumbar joint by subtracting from the flexion angle. Statistical comparisons between baseline and walker trials aided in defining the difference between a child with CP walking with and without a walker. For the CP group, lumbar extension was greater at mid-stance and mid-swing between baseline and walker trials ($p<0.05$). The CP group had an inherent pelvic tilt quantified by the difference between the pelvis segment angle and the lumbar. The pelvic flexion angle was more neutral in the walker-assisted trials for the children with CP. Upper torso posture was better in the TD group with a walker as thoracic joint flexion was lower and more neutral.

Table 4.2: Flexion angles for the pelvis segment and thoracic, lumbar, hip, and knee joints at four significant time points of gait cycle. Presented as average ± standard deviation. T-test significance for p<0.05 (*), p<0.01 (**), and p<0.001 (***).

Pelvis	**Trial Type**	**Initial Contact**	**Mid-Stance**	**Pre-Swing**	**Mid-Swing**
TD	Baseline	-14.98±2.31	-15.62±2.24	-15.36±2.71	-15.51±2.10
	Walker	-13.96±2.87	-15.28±2.98	-14.92±3.27	-15.57±2.93
CP	Baseline	-17.15±4.62	-13.72±4.37	-13.70±4.08	-13.41±5.25
	Walker	-14.05±1.31	-9.74±0.55	-9.38±3.82	-9.63±1.07
Thoracic					
TD	Baseline	1.90±3.67	4.71±3.75	4.71±3.56	5.17±3.71
	Walker	-2.63±4.18*	-1.17±4.25**	-0.97±4.52**	-0.90±4.17**
CP	Baseline	6.35±8.84	12.62±7.59	13.95±9.82	12.57±6.38
	Walker	6.54±11.50	11.58±12.35	9.75±13.34	10.88±11.86
Lumbar					
TD	Baseline	-18.48±2.57	-20.25±2.71	-19.55±3.26	-20.49±2.62
	Walker	-17.70±3.06	-19.84±3.73	-18.50±4.46	-20.09±4.21
CP	Baseline	-28.29±3.57	-26.71±2.85	-27.71±3.30	-26.60±2.93
	Walker	-32.56±2.88	-32.22±2.60*	-31.54±3.03	-32.61±2.68*
Hip					
TD	Baseline	35.06±3.61	8.66±3.68	0.50±6.26	35.64±3.74
	Walker	33.41±5.18	8.44±6.30	3.60±7.11	34.68±2.99
CP	Baseline	40.40±5.37	14.75±14.43	3.38±11.10	31.03±6.02
	Walker	37.85±5.42	19.88±8.71	4.09±7.41	28.92±8.96
Knee					
TD	Baseline	4.90±2.01	7.28±2.48	37.87±5.40	52.43±6.75
	Walker	3.04±2.59	6.07±4.01	38.66±6.18	50.07±5.48
CP	Baseline	29.03±5.89	19.17±15.75	41.18±9.04	54.54±5.45
	Walker	29.06±6.29	15.42±17.00	31.38±11.00	49.02±10.19

4.3.3 Model Outputs

Dynamic models were successfully created for three trials of each condition of walking for each subject. The model outputs included joint angles and moments for the defined joints of the model. TD and CP group baseline gait was compared initially to examine the difference in gait between the groups (Figures 4.4-4.6). Four subjects comprised the CP group as the rest of the group used crutches or a walker during baseline gait and would have been a confounding variable in the gait of the whole group. Within group differences were marked with shaded boxes on the figures where the walker trials were significantly different than the baseline. Pelvic segment tilt, obliquity, and rotation angles were included to show the inherent neutral angle of the model, as it was not equivalent to anatomical neutral. Pelvic tilt angles were offset from neutral by values between -10 and -18 degrees, while obliquity and rotation angles fluctuated around 0 degrees.

4.3.3.1 TD Group vs CP Group Baseline Gait

In the lower body, the CP group had a lower range of motion and increased loads on the joints over their baseline gait than the TD group highlighting the pathology of their gait (Figure 4.4). Based on gait kinematics, the CP group walked with crouch gait and reciprocal effects on joint kinetics were observed. The ankle was in sustained dorsiflexion and reduced internal rotation and inversion early in swing phase. The knee was in flexion throughout the gait cycle and never reached an extended, zero-degree, angle. The hip had greater flexion in CP gait and more internal rotation. Ankle plantarflexion moment was characteristic of a lack of heel strike as the observed initial peak of load bearing did not occur in the TD group. Knee extension moment was greater in late stance phase and hip extension moment was greater through stance phase.

In the upper body, CP gait was characterized by worse posture than TD gait (Figure 4.5). Pelvic flexion offset was around 15 degrees on average through the gait cycle for both TD and CP groups. The CP group had greater extension in the lumbar joint and greater range of motion and peak flexion in the thoracic joint. The TD group laterally rotated in the lumbar joint towards the stance foot with two peaks during stance phase of both feet. The CP group had a reduced second peak in the lumbar during terminal stance and terminal swing phase of the gait cycle. The CP group axially rotated less than the TD group in the lumbar joint with lower peaks around heel strike. The CP group had a greater range of lateral motion in the thoracic joint than the TD group with cyclical peaks characterizing bending towards the stance foot. Axial rotation of the thoracic joint was delayed in the CP group around contralateral and ipsilateral mid-swing compared to the TD group. The lumbar flexion moment was greater in CP gait but not significant. The lateral trunk moments of both lumbar and thoracic joints were greater in the CP group than the TD group around mid-stance and early swing. Axial rotational moment in the lumbar was opposite in sign in the CP group compared to the TD group around toe off times. Shoulder flexion was decreased in CP gait characterizing an overall lower range of motion in the arms by the CP group (Figure 4.6). Children with CP held their arms up to the side and out in front of them particularly during mid-stance.

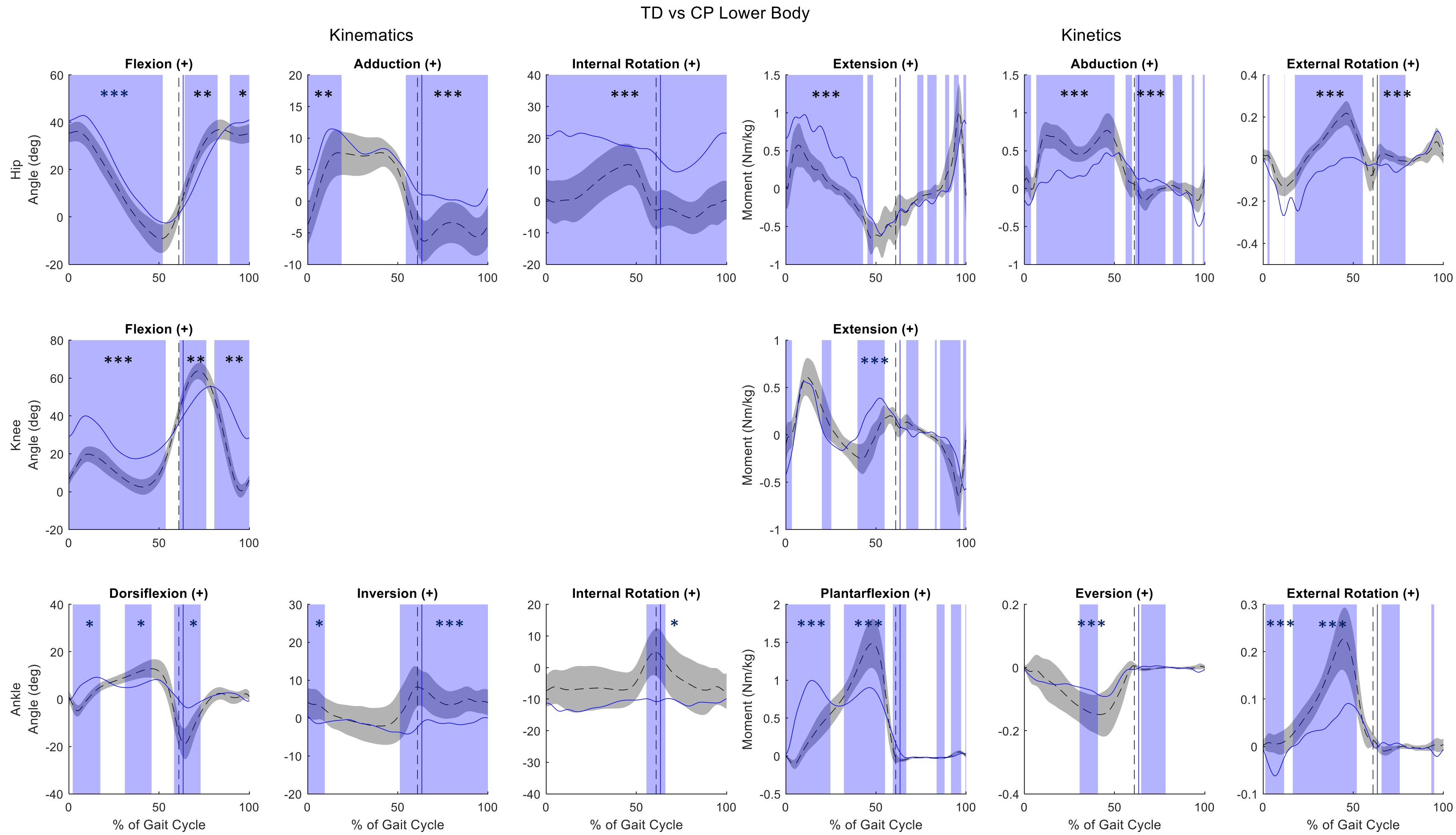

Figure 4.4: Lower body kinematics (left) and kinetics (right) for TD group (black) compared to CP group (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

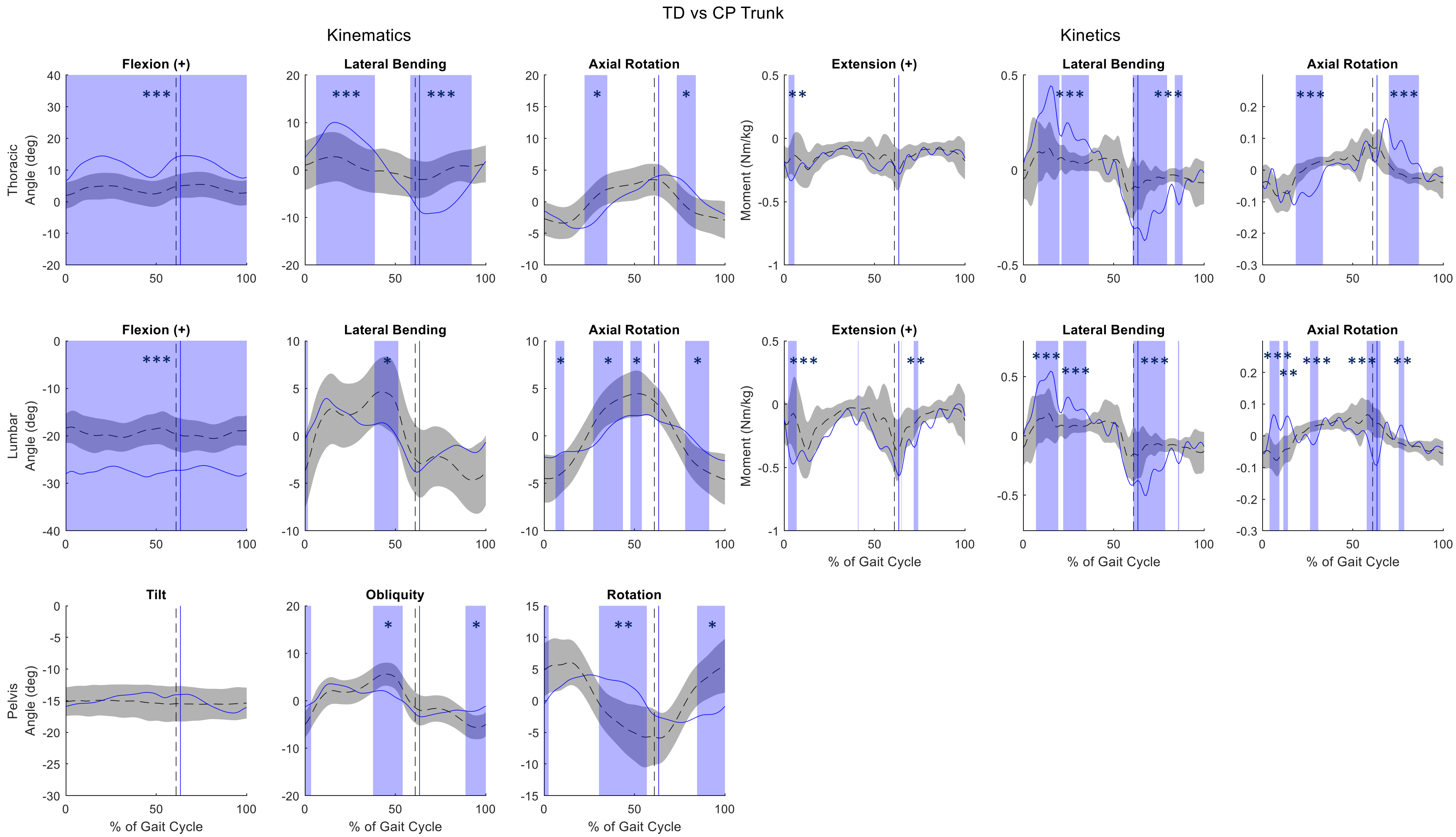

Figure 4.5: Trunk kinematics (left) and kinetics (right) and pelvic segment kinematics (bottom) for TD group (black) compared to CP group (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Flexion was characterized by bending forwards and extension by bending backwards from the reference frame on the pelvis. Positive lateral bending and axial rotation was in the direction of rotation towards the stance foot. Pelvic tilt is negative for forward rotation, obliquity is positive for rotation away from stance foot rotation is positive for vertical-axis rotation away from stance foot.

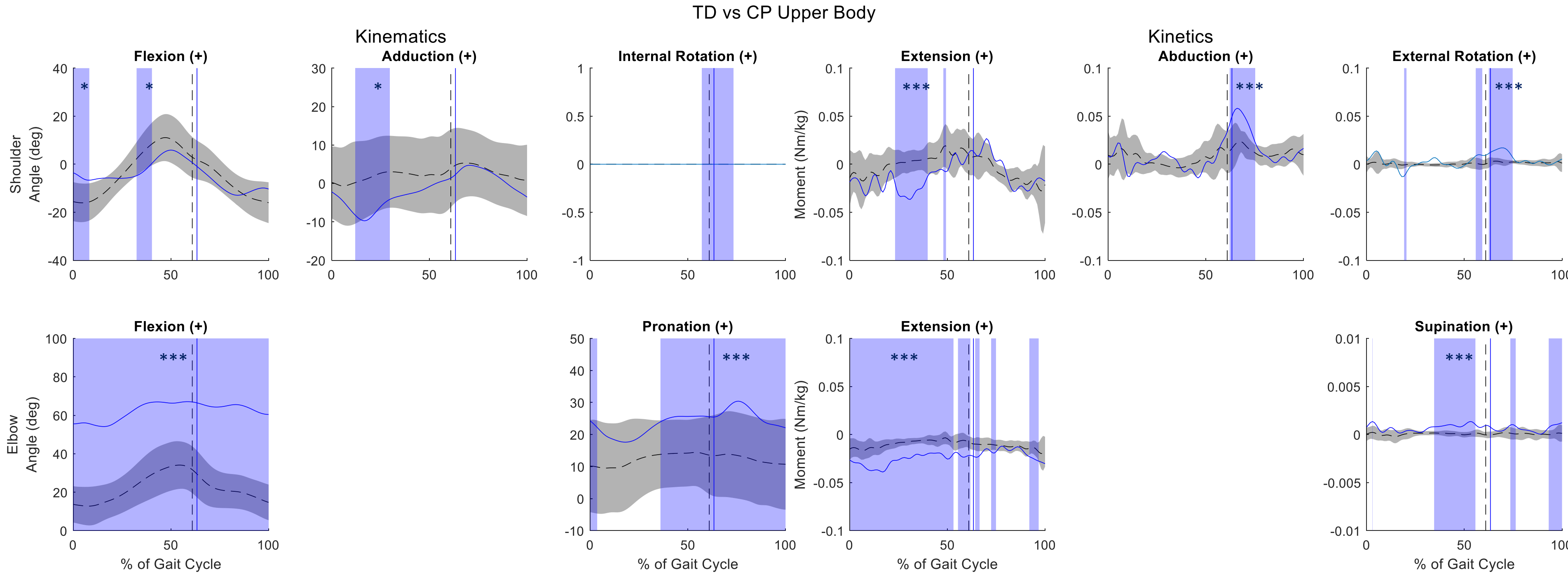

Figure 4.6: Upper body kinematics (left) and kinetics (right) for TD group (black) compared to CP group (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

4.3.3.2 TD Group Baseline vs Walker-Assisted Gait

Baseline and walker trial model outputs were compared using one-dimensional statistical parametric mapping (SPM) analysis. In the TD group, ankle kinematics and kinetics were no different in walker-assisted gait compared to baseline (Figure 4.7). Peak knee flexion was lower during initial loading and a reciprocal decrease in extension moment occurred in walker-assisted gait. The TD group also had lower abduction angles in the hip around toe off and heel strike times of walker-assisted gait. In walker-assisted gait the TD group had lower flexion angles in the thoracic joint and a greater flexion moment during swing phases of gait for the lumbar and thoracic joints (Figure 4.8). Axial rotation range of motion was decreased in the lumbar and thoracic joints during late stance phase and toe off. Axial rotational moment was decreased in lumbar and thoracic joints during peaks near toe off. Walker interaction during gait caused shoulder extension throughout TD gait and greater elbow flexion compared to baseline (Figure 4.9). Shoulder flexion moment was lower in walker-assisted gait during late swing phase approaching heel strike.

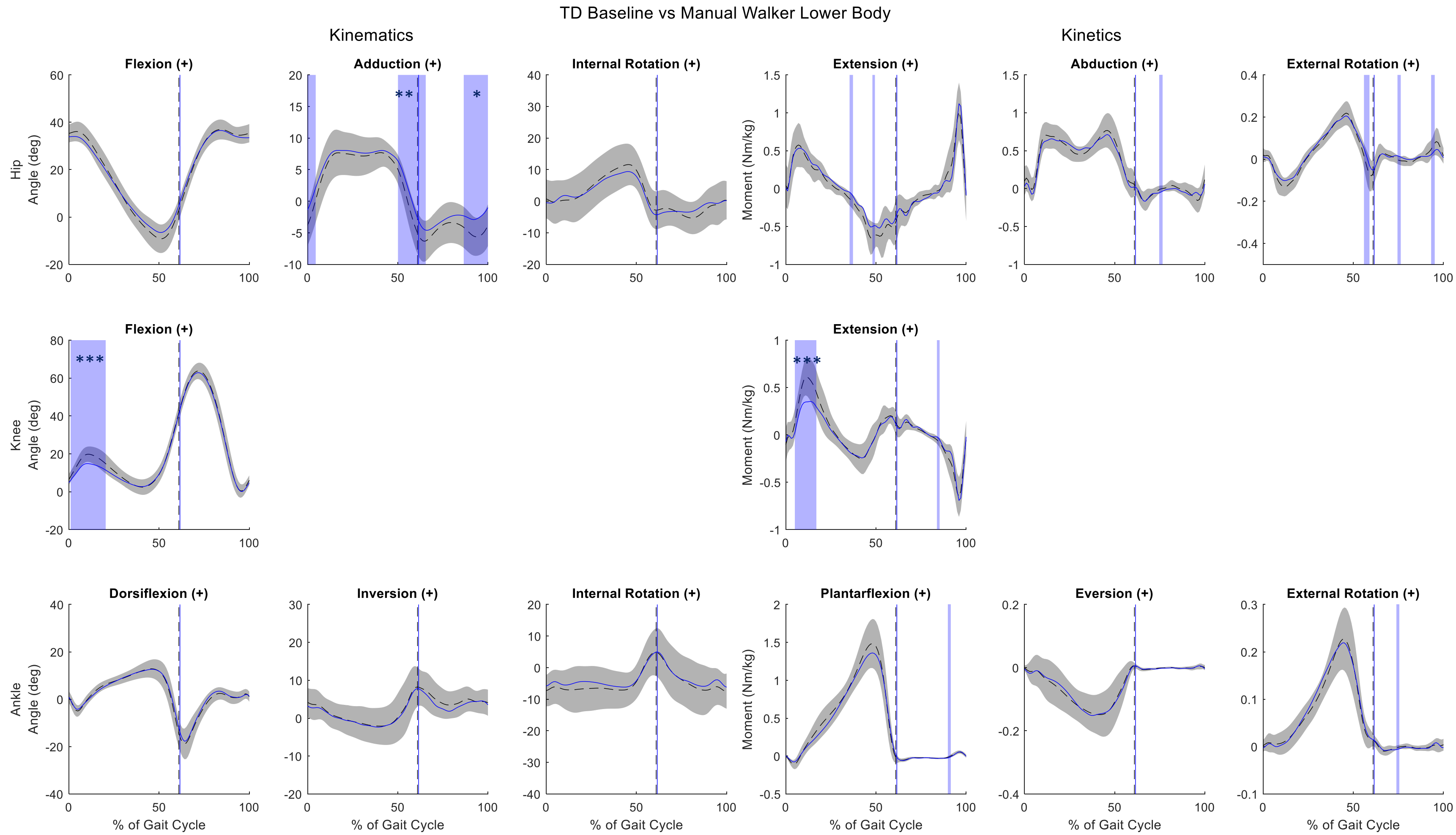

Figure 4.7: Lower body joint kinematics (left) and kinetics (right) for the TD group baseline gait (black) compared to walker-assisted gait (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

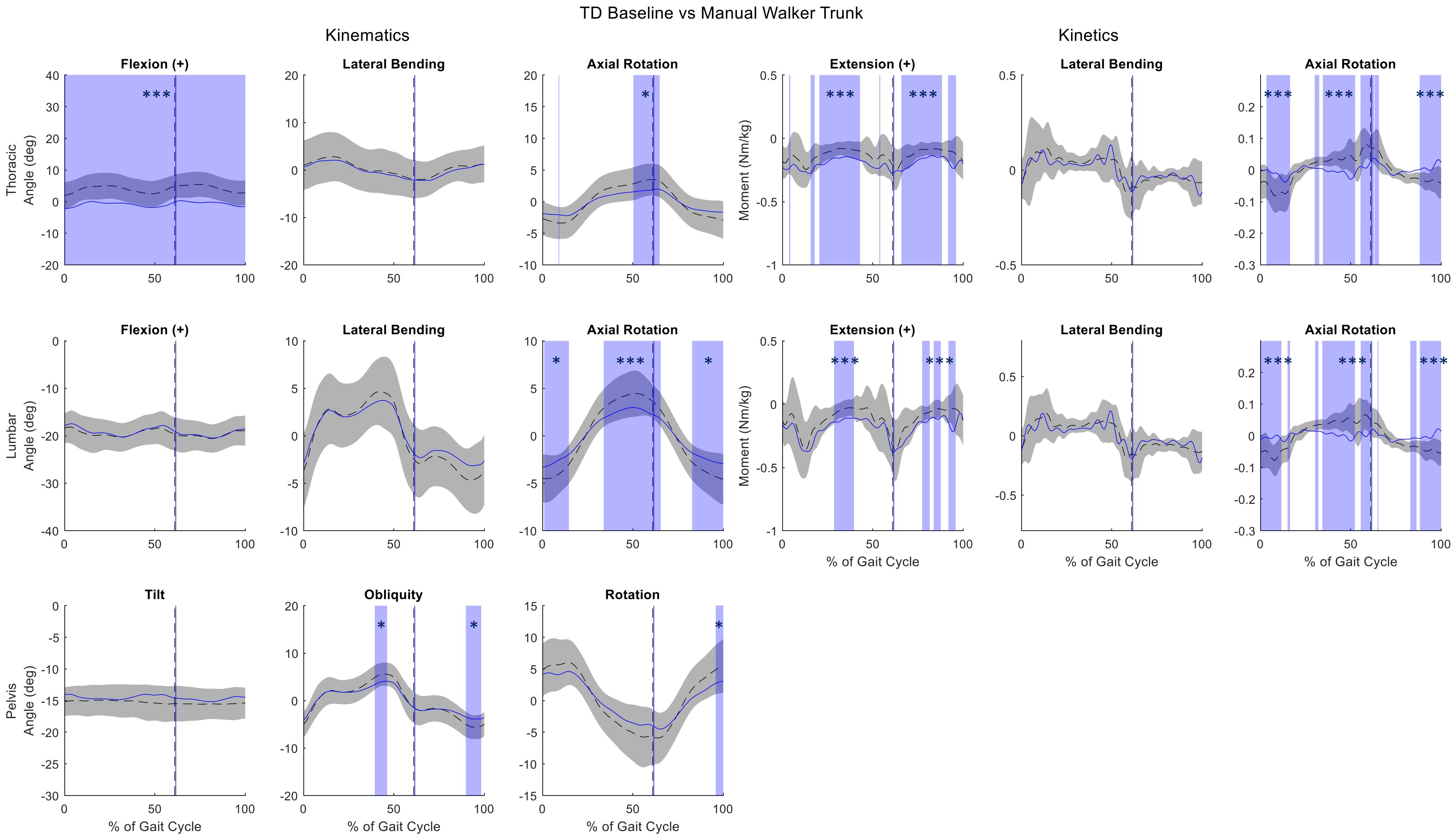

Figure 4.8: Trunk kinematics (left) and kinetics (right) and pelvic segment kinematics (bottom) for TD group baseline gait (black) compared to walker-assisted gait (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Flexion was characterized by bending forwards and extension by bending backwards based off of pelvic tilt. Positive lateral bending and axial rotation was in the direction of rotation towards the stance foot. Pelvic tilt is negative for forward rotation, obliquity is positive for rotation away from stance foot rotation is positive for vertical-axis rotation away from stance foot.

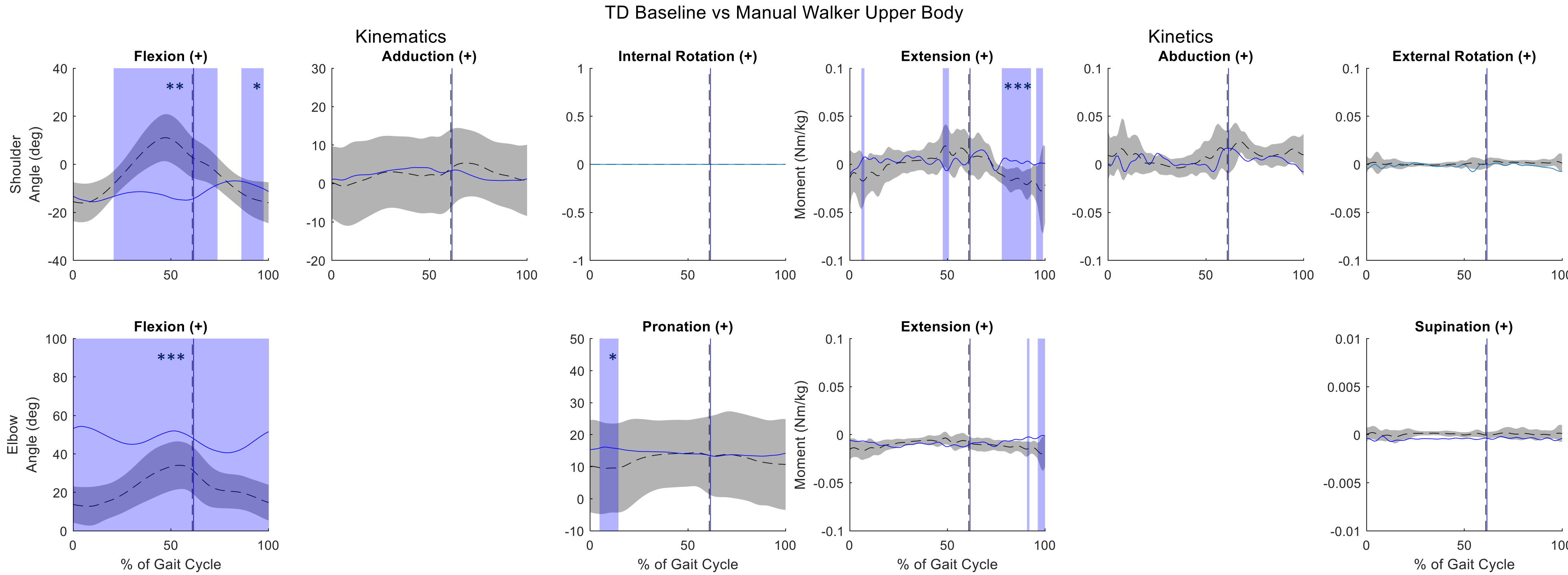

Figure 4.9: Upper body kinematics (left) and kinetics (right) for TD group baseline gait (black) compared to walker-assisted gait (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

4.3.3.3 CP Group Baseline vs Walker-Assisted Gait

In the CP group, posture was better and a decreased load on the lower body was observed (Figure 4.10). The knee was more extended particularly during swing phase and early in stance. Lower extension moments during initial loading occurred in the hip, knee, and ankle. Lumbar extension angles were greater in walker assisted gait throughout the gait cycle (Figure 4.11). Thoracic lateral bending was decreased over the gait cycle in walker-assisted gait compared to baseline gait during single support times. Decreased lateral bending moments in the thoracic and lumbar joints near toe off occurred in walker-assisted gait. Shoulder abduction angle in early stance phase was minimized in walker-assisted trials, the CP group typically adducted their shoulders throughout the gait cycle (Figure 4.12). Additionally, the CP group had a greater shoulder adduction moment throughout the gait cycle which was significant during swing phase. Shoulder extension moment was increased in early stance phase. Elbow flexion was decreased in late stance and throughout swing in walker assisted gait. Extension moment in the elbow was greater in walker assisted gait throughout the gait cycle and pronation moment was greater around the time of heel strikes.

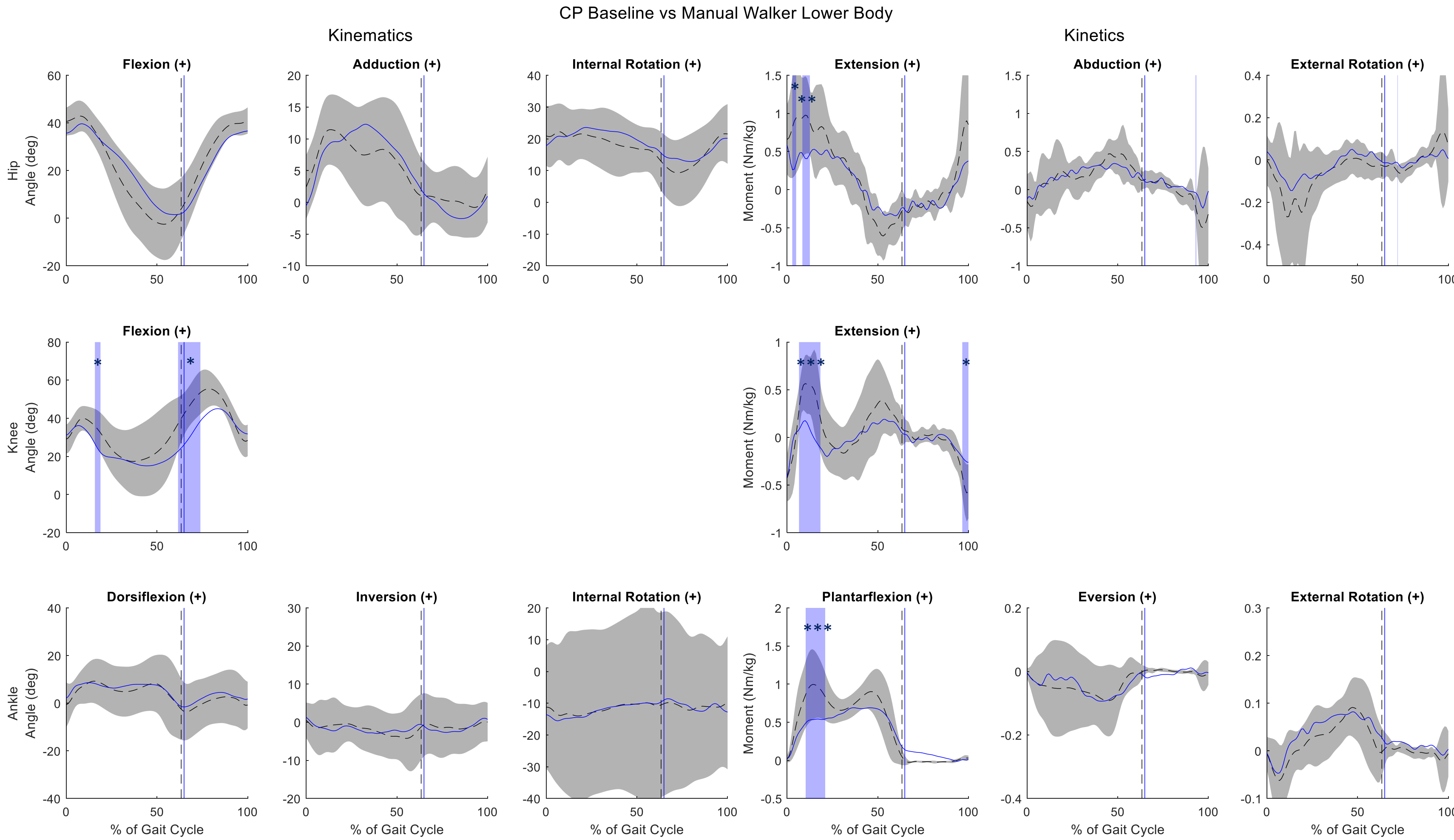

Figure 4.10: Lower body joint kinematics (left) and kinetics (right) for the CP group baseline gait (black) compared to walker-assisted gait (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

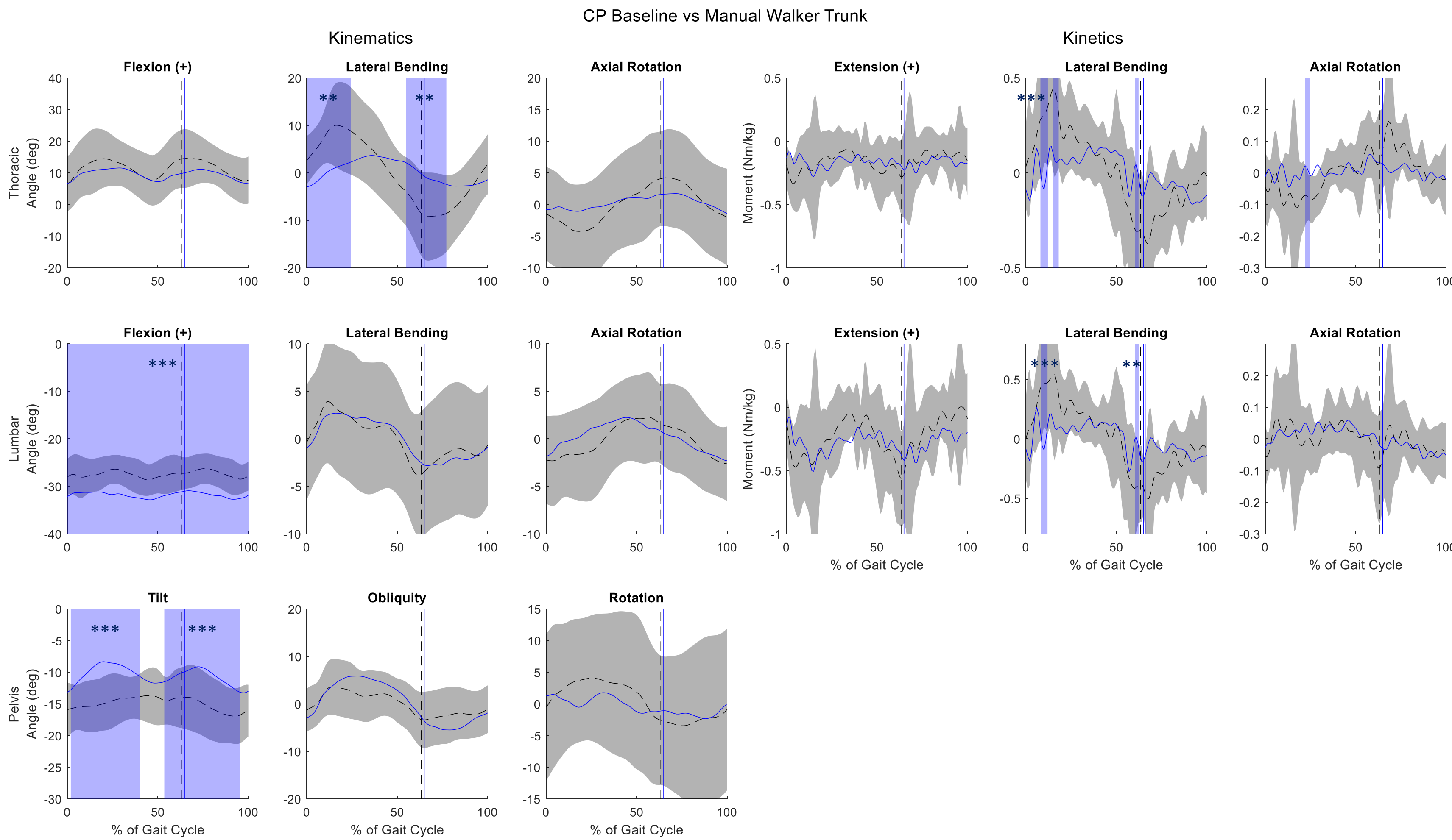

Figure 4.11: Trunk kinematics (left) and kinetics (right) and pelvic segment kinematics (bottom) for CP group baseline gait (black) compared to walker-assisted gait (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Flexion was characterized by bending forwards and extension by bending backwards based off of pelvic tilt. Positive lateral bending and axial rotation was in the direction of rotation towards the stance foot. Pelvic tilt is negative for forward rotation, obliquity is positive for rotation away from stance foot rotation is positive for vertical-axis rotation away from stance foot.

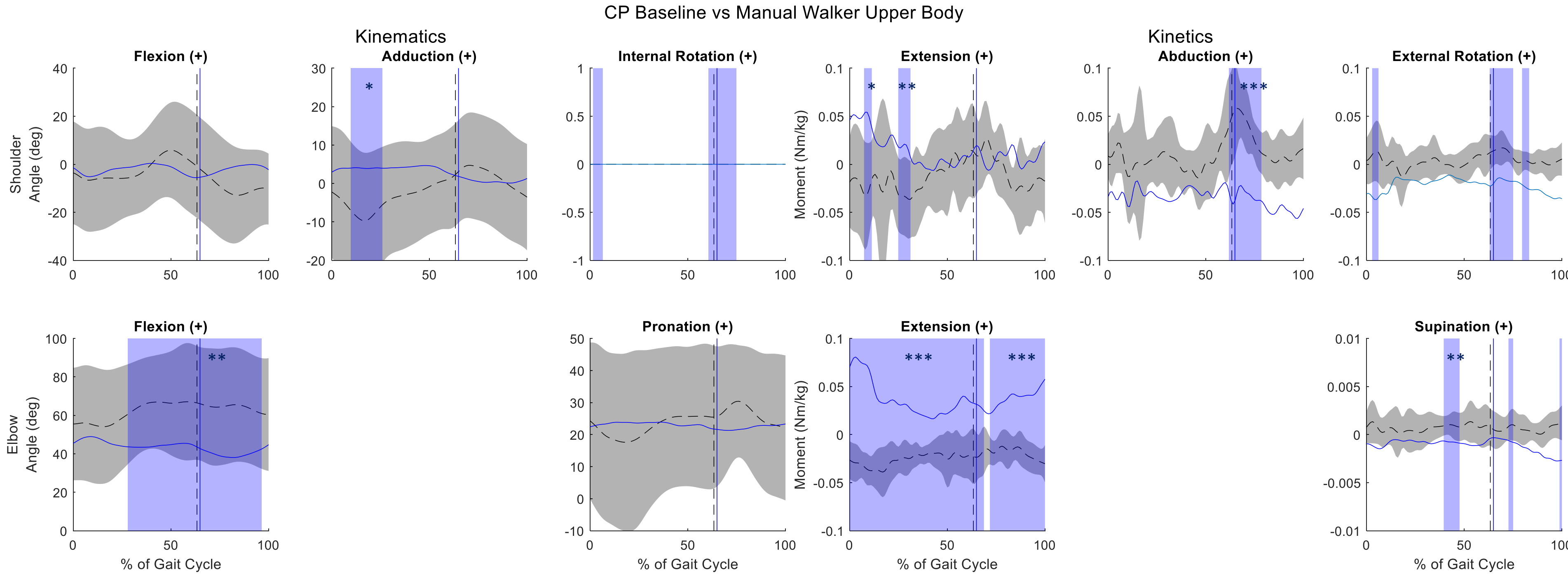

Figure 4.12: Upper body kinematics (left) and kinetics (right) for CP group baseline gait (black) compared to walker-assisted gait (blue). SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

4.3.3.4 CP Group vs CP Crutch-User Gait

The CP group which walked without assistive devices (NA) in their baseline trial was compared to the CP group who used crutches (CA) during baseline trials. The manual walker trials for the NA group (black) were compared to the baseline (blue) and manual walker (red) trials for the CA group using SPM (Figures 4.13-4.15). Kinematics for all trials were considered but kinetics for baseline, crutch-assisted trials were not included as the crutches were not instrumented like the walker. The CA group walked with increased pathologic gait in the ankle joint during baseline, crutch-assisted gait (Figure 4.13). Increased dorsiflexion, inversion and internal rotation of the ankle characterized the CA group in crutch- and walker-assisted trials. Increased plantarflexion and inversion moment occurred in CA group compared to NA group walker-assisted trials. Hip and knee flexion were similar to the NA group but hip adduction was decreased in the CA group at mid-stance and internal rotation was lower through the gait cycle compared to walker-assisted gait of the NA group. Hip extension moment was greater in the CA group at contralateral toe off than the NA group.

The CA group was less stable in the torso in crutch-assisted gait compared to the NA group (Figure 4.14). Decreased extension in the lumbar and extension in the thoracic joint were observed in the baseline gait of the CA group compared to walker-assisted gait in the NA group. In walker-assisted gait, the CA group had increased extension in the lumbar joint similar to eh NA group. The CA group walked with greater peak lateral bending in crutch-assisted gait and had decreased lateral bending in walker-assisted gait. The CA group had greater flexion moments in the lumbar and thoracic joints throughout the gait cycle compared to the NA group.

The CA group showed a different pattern of usage of assistive devices than the NA group based on upper body kinematics (Figure 4.15). Elbow flexion was increased in CA group crutch-and walker-assisted trials. Elbow pronation was greater in crutch-assisted gait but the same in walker-assisted gait as the NA group. The shoulder was extended near heel strike for CA group crutch-assisted trials and for the entire gait cycle during walker-assisted trials. Shoulder adduction was increased in the CA group in early swing for both crutch- and walker-assisted gait. In walker-assisted gait, the CA group had increased, cyclic flexion moment in the shoulder and extension moment in the elbow compared to the NA group. Peak moment occurred at contralateral and ipsilateral heel strike times and troughs occurred around mid-stance.

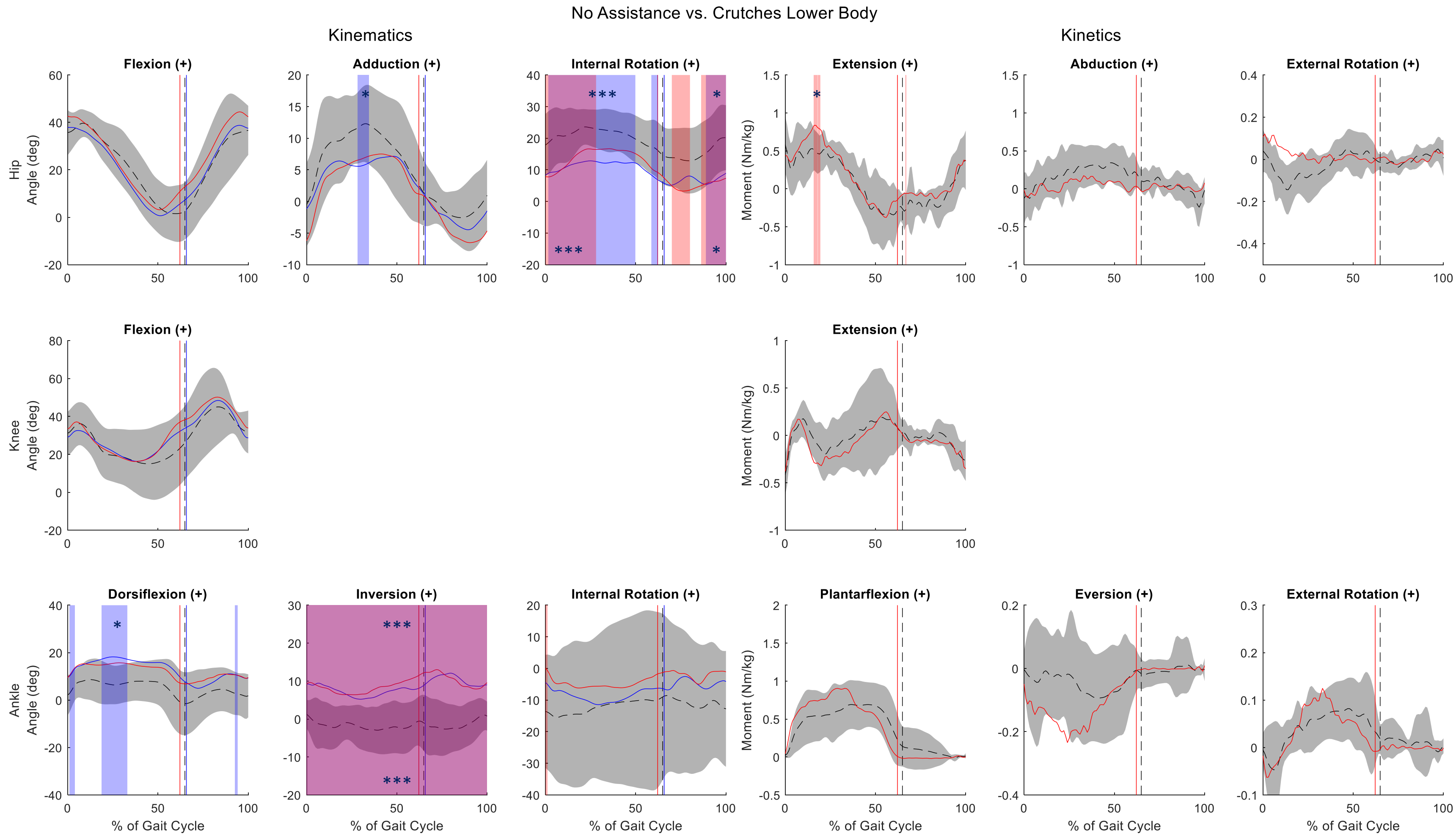

Figure 4.13: Lower body joint kinematics (left) and kinetics (right) for manual walker gait in the NA group (black) compared to crutch-assisted gait in the CA group (blue) and walker-assisted gait in the CA group (red) with significance from SPM analysis marked with shaded boxes in respective colors and with p<0.05 (*), p<0.01 (**), and p<0.001 (***) for crutch-assisted gait (top) and walker-assisted gait (bottom).

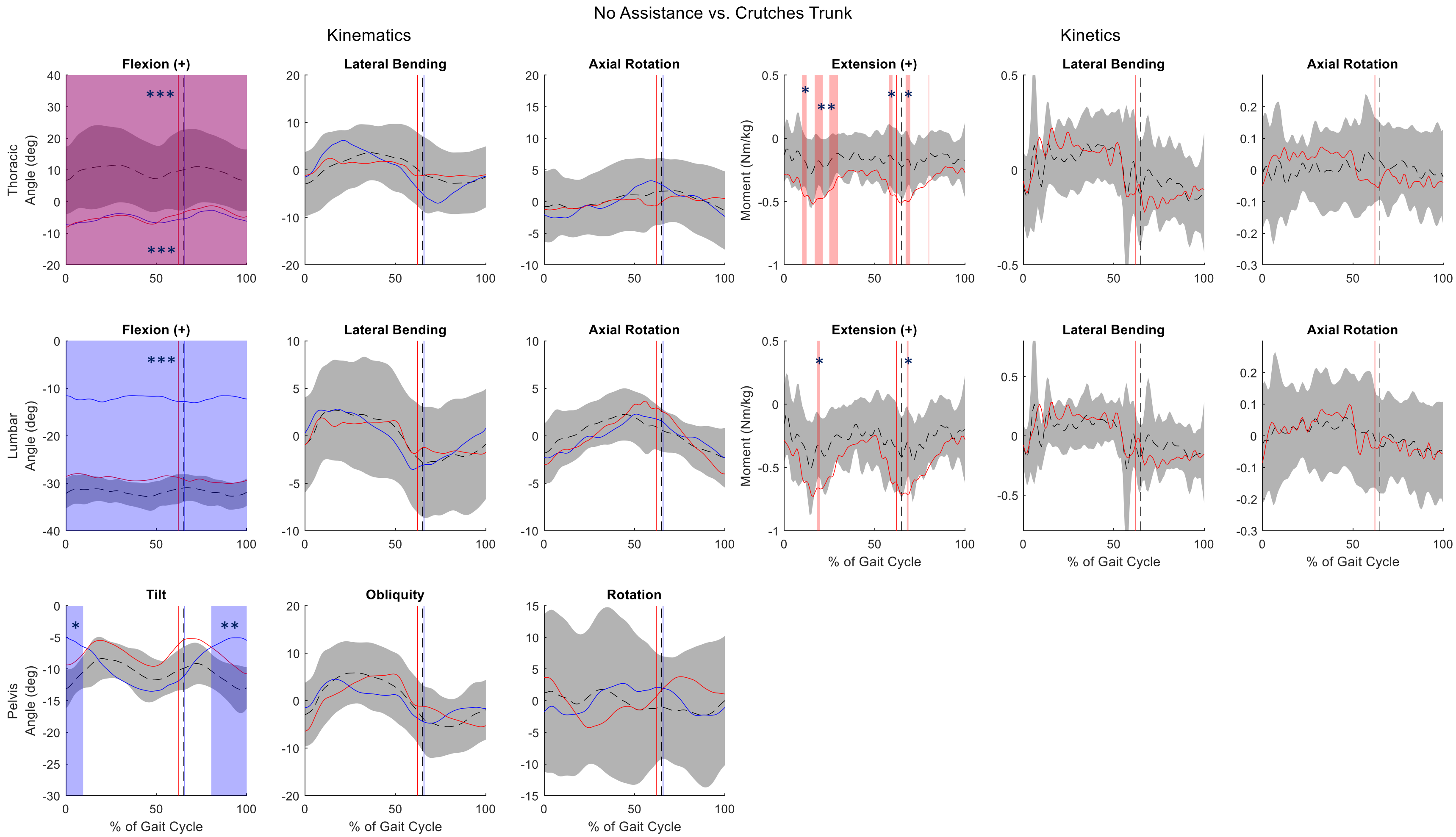

Figure 4.14: Trunk joint kinematics (left) and kinetics (right) and pelvic segment kinematics (bottom) for manual walker gait in the NA group (black) compared to crutch-assisted gait in the CA group (blue) and walker-assisted gait in the CA group (red) with significance from SPM analysis marked with shaded boxes in respective colors and with p<0.05 (*), p<0.01 (**), and p<0.001 (***) for crutch-assisted gait (top) and walker-assisted gait (bottom). Flexion was characterized by bending forwards and extension by bending backwards based off of pelvic tilt. Positive lateral bending and axial rotation was in the direction of rotation towards the stance foot. Pelvic tilt is negative for forward rotation, obliquity is positive for rotation away from stance foot rotation is positive for vertical-axis rotation away from stance foot.

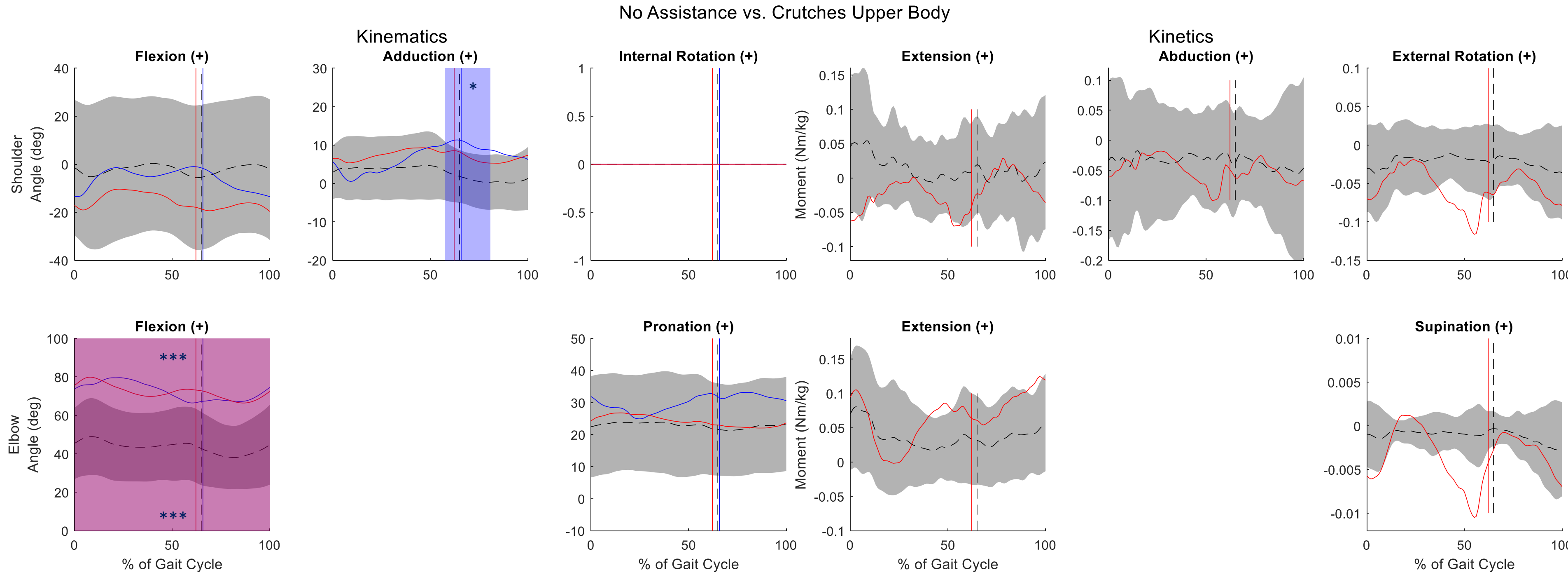

Figure 4.15: Upper body joint kinematics (left) and kinetics (right) for manual walker gait in the NA group (black) compared to crutch-assisted gait in the CA group (blue) and walker-assisted gait in the CA group (red) with significance from SPM analysis marked with shaded boxes in respective colors and with p<0.05 (*), p<0.01 (**), and p<0.001 (***) for crutch-assisted gait (top) and walker-assisted gait (bottom). Kinetics of the shoulder and elbow are on larger scales than other group plots.

4.4 Discussion

The dynamic model created for baseline and walker trials provided additional data for secondary analysis. Further evidence of the pathologic gait of CP was found by comparing the kinematics and kinetics of baseline gait of the TD and CP groups. The model outputs for kinematics and kinetics revealed more about the impact of the posterior walker on the gait of the user. TD and CP groups interacted with the walker in different ways and the lower body outcomes agreed with the assumption that the groups would alter their gait to incorporate the posterior walker. The posterior walker limited the TD group motion in the knee during initial loading and the thoracic joint posture was more vertical. The posterior walker reduced the load on the lower body joints in the CP group during early stance phase. The CP group was more stable in walker assisted gait when comparing to baseline unassisted and crutch-assisted gait as lateral motion and moment in the torso was reduced. In Chapter 3, positive effects of the posterior walker were only seen in the GMFCS III group but better posture and reduced load in the lower body are seen in the whole CP group when walking with a walker. So far, the utility of a posterior walker has been dependent upon what metric was used as the factor for analysis but the model kinematics and kinetics offer a more complete analysis of the walker-assisted gait cycle.

The joint kinematics for the CP group had a lower range of motion than in the TD group. Previous work has shown a decreased range of motion in the ankle, knee, and hip joints for children with CP (Krautwurst et al., 2016a). This was particularly characterized by less extension in the hip around 40-55% of the gait cycle, more flexion in the knee throughout the gait cycle, and less plantarflexion in the ankle around toe off. In this work, the ankle angles typically did not vary outside of the 0- to 20-degree range and remained relatively constant through the gait cycle quantifying a lower engagement of the ankle joint through plantarflexion motion. The lack of ankle movement was caused by the prescribed ankle foot orthoses worn by the children with CP. The CP group walked in slight dorsiflexion and engaged a plantarflexion moment throughout their stance phase representing a slight toe-walking gait. The knees and hips spent more time in flexion than the typical gait which is characteristic of apparent equinus and crouch gait (Armand et al., 2016a). The CP group was less stable than the TD group because CP lumbar and thoracic joints were aligned at worse postures in the sagittal plane and the lumbar joint had a lower range of motion in the frontal and transverse planes. The thoracic joint in the CP group had a greater range of motion in lateral bending than the TD group which indicated a use of the upper torso for control of stability in addition to the lower torso. Higher moment in the torso highlighted the need for support in their gait as in baseline trials the CP group required more moment generation about their torso to walk and maintain their posture. Shoulder and elbow range of motion was lower in the TD group while moment was higher. The CP group moved their upper body less than the TD group in baseline trials but required greater joint moments to remain stable in their gait. Further analysis on stability of gait should be done examining the angular momentum of the body segments to quantify whole-body stability during gait. The incorporation of a posterior walker was expected to increase the stability and promote better posture of the user.

Within the TD group, the addition of the walker lowered the peak flexion angle of the knee during stance phase, which was not confirmed by the traditional individual time point analysis. These results are indicative of the walker limiting the range of motion of the TD group. The walker lowered the mechanical load on the knee by lowering the peak extension moment in stance phase. This occurred while also decreasing the flexion range of motion at this time period in the knee. The TD group had lower range of motion in axial rotation of the lumbar and thoracic joints which is one factor contributing to the lower stride length in walker-assisted gait found in Chapter 3 (S. D. Russell, Bennett, Kerrigan, & Abel, 2007). Posterior walkers have been shown to positively affect posture in children with CP (Logan, Byers-Hinkley, et al., 1990), and TD group posture was altered as well. The thoracic joint flexion angle was lower throughout the gait cycle, operating at a more neutral position and a lower angular displacement on average. Lumbar and thoracic moments were greater in flexion and lower in axial rotation. The load of the walker was taken on by activation in the torso for the TD group while through the rest of the body other moments and kinematics were lowered. The lumbar and thoracic joint flexion moments were greater during both ipsilateral and contralateral single support times. As expected the shoulder and elbow joints had a lower range of motion as the arms were interacting with the walker and remained relatively constant through the gait cycle. The shoulder joint moment was lower during swing phase. In baseline trials the flexion moment of the shoulder was in preparation for the swing of the shoulder during stance phase, but in the walker trials the shoulder joint range of motion was lower.

The CP group saw greater lumbar extension at mid-swing and mid-stance in walker-assisted gait. SPM analysis also found the lumbar to be more extended during walker-assisted gait. The CP group was largely characterized by an anterior pelvic tilt so the increase in trunk extension angle resulted in a more upright position for the user. Posterior walkers have been shown to be most effective in increasing posture for children with CP (Brooke M. Greiner et al., 1993). On average, the walker affected the upright position and the stability of the user positively in the CP group. The CP group had reduced lateral bending in the thoracic joint and slightly less range of motion in the shoulder and elbow when walking with the walker. There was a large standard deviation in this group, so it is likely that more data and a refined subject group would increase the power of the findings presented and potentially reveal more about the population. For the CP group, flexion moment for hip, knee and ankle joints in the walker trials typically reached lower magnitude at the local maxima in mid-stance. Similar to the TD group, the walker decreased the load on the lower body joints particularly in stance phase but to a greater degree in the CP group. This is due to the CP group relying more on the walker as a population than the TD group. Elbow flexion angle was lower from mid-stance through swing phase. The CP group walked with their elbows highly flexed in baseline trials, walking with the walker necessitated the extension of the elbows to hold on to the walker for support and to pull the walker along. Children with CP used the walker for support during gait and the shoulder and elbow seem to be the primary loading joints for this support. Increased upper body load occurred alongside lower body joint load reductions which could allow children with CP to walk for longer periods throughout the day. The increased adduction moment of the shoulder in the walker trials show the reliance on and functional use of the walker by the CP group. Increased extension moment in the elbow highlights the vertical force being applied to the walker. The greatest elbow extension moment occurred after heel strike during double support. Cyclical loading revealed the CP group relying on the walker progressively more through swing reaching a maximum during double support after heel strike and a decrease in reliance during double support. While the TD group absorbed the walker's load through flexion moment in the torso joints, the CP group did not. By sharing the load of support during their gait cycle over four limbs, children with CP were able to walk better than in baseline trials. With a posterior walker, users are able to walk longer during the day train their muscles more to avoid atrophy as they grow. The CP group bore the weight of pulling the walker in the shoulder and elbow joints while also relying more heavily on the walker for support through their gait.

The CA engaged with the walker more than the NA group and had greater positive effects on posture by using the walker than while using their prescribed crutches. Lower body kinematics remained relatively unchanged from baseline to walker-assisted gait in the CA group. Compared to the NA group, the CA group displayed increased pathology in the lower body through greater ankle dorsiflexion throughout the gait cycle. The CA group had increased posture when walking with the walker as the lumbar was more extended to a similar value of the NA group. Lateral bending in the thoracic joint was decreased when walking with the walker as well compared to walking with crutches. Compared to the NA group, the CA group relied on the walker more for assistance. Lower body kinetics were similar except hip extension moment was greater in early stance and ankle moments were greater during stance phase. Flexion moments were greater over the whole gait cycle in CA walker-assisted gait compared to the NA group and shoulder and elbow joint kinetics were greater and more cyclic in usage in the CA group. Peak upper body loads occurred after heel strike and torso loads peaked following toe off. The increased usage of the walker by the CA group was expected as the group was not able to walk without assistance and thus were more affected by CP.

4.5 Conclusions

The dynamic model performed adequately to give accurate results for joint kinematics and kinetics. The positive impact of a posterior walker on gait was observed in both groups. In walker-assisted gait, both groups experienced decreased lower body load early in stance phase. The inclusion of the walker in the model saw a more upright positioning of the torso in the CP group indicating a more stable positioning of the body. The TD and CP group incorporated the walker into their gait in different ways. While decreasing knee moment, the TD group pulled the walker along through torso flexion and axial moment. The CP group pulled the walker along and loaded the walker for support through increased moments in the arms. The mechanical work of the body will be investigated in Chapter 5 to observe the effect of the walker on energetic cost on joints and the whole body.

Reference List

Armand, S., Decoulon, G., & Bonnefoy-Mazure, A. (2016). Gait analysis in children with cerebral palsy. EFORT Open Reviews, 1(12), 448–460. https://doi.org/10.1302/2058-5241.1.000052

Baker, R., Leboeuf, F., Reay, J., & Sangeux, M. (2017). The Conventional Gait Model - Success and Limitations. In Handbook of Human Motion (pp. 1–19). Cham: Springer International Publishing. https://doi.org/10.1055/s-0028-1087518

Baughman, L. D. (1983). Development of an Interactive Computer Program to Produce Body Description Data. Dayton, OH. Retrieved from http://www.dtic.mil/docs/citations/ADA133720

Cheng, H., Obergefell, L., & Rizer, A. (1994). Generator of Body Data (GEBOD) , Manual. Dayton, OH. Retrieved from http://www.dtic.mil/docs/citations/ADA289721

Friston, K. J., Ashburner, J. T., Kiebel, S. J., Nichols, T. E., & Penny, W. D. (2011). Statistical parametric mapping: the analysis of functional brain images. Amsterdam: Elsevier.

Gibson, T., Jeffery, R. S., & Bakheit, A. M. O. (2006). Comparison of three definitions of the mid-stance and mid-swing events of the gait cycle in children. Disability and Rehabilitation. https://doi.org/10.1191/0959683606hl959rp

Greiner, B. M., Czerniecki, J. M., & Deitz, J. C. (1993). Gait Parameters of Children With Spastic Diplegia: A Comparison of Effects of Posterior and Anterior Walkers. Archives of Physical Medical Rehabilitation, 74, 381–385. https://doi.org/0003-9993/93/7404-0174

Konop, K. A., Strifling, K. M. B., Wang, M., Cao, K., Eastwood, D., Jackson, S., … Harris, G. F. (2009). Upper extremity kinetics and energy expenditure during walker-assisted gait in children with cerebral palsy. Acta Orthopaedica et Traumatologica Turcica, 43(2), 156–164. https://doi.org/10.3944/AOTT.2009.156

Krautwurst, B. K., Dreher, T., & Wolf, S. I. (2016). The impact of walking devices on kinematics in patients with spastic bilateral cerebral palsy. Gait & Posture, 46, 184–187. https://doi.org/10.1016/j.gaitpost.2016.03.014

Logan, L., Byers-Hinkley, K., & Ciccone, C. D. (1990). Anterior Versus Posterior Walkers: A Gait Analysis Study. Developmental Medicine & Child Neurology, 32(12), 1044–1048. Retrieved from http://doi.wiley.com/10.1111/j.1469-8749.1990.tb08521.x

Logan, L., Byers-Hinkley, K., & Ciccone, C. D. (1990). Anterior Versus Posterior Walkers: A Gait Analysis Study. Developmental Medicine & Child Neurology, 32(12), 1044–1048. https://doi.org/10.1111/j.1469-8749.1990.tb08521.x

Pataky, T. C. (2010). Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. Journal of Biomechanics. https://doi.org/10.1016/j.jbiomech.2010.03.008

Pataky, T. C. (2012). One-dimensional statistical parametric mapping in Python. Computer Methods in Biomechanics and Biomedical Engineering, 15(3), 295–301. https://doi.org/10.1080/10255842.2010.527837

Pataky, T. C., Robinson, M. A., & Vanrenterghem, J. (2016). Region-of-interest analyses of one-dimensional biomechanical trajectories: bridging 0D and 1D theory, augmenting statistical power. PeerJ, 4, e2652. https://doi.org/10.7717/peerj.2652

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011). The gait of children with and without cerebral palsy: Work, energy, and angular momentum. Journal of Applied Biomechanics, 27(2), 99–107.

Russell, S. D., Bennett, B. C., Kerrigan, D. C., & Abel, M. F. (2007). Determinants of gait as applied to children with cerebral palsy. Gait and Posture, 26(2), 295–300. https://doi.org/10.1016/j.gaitpost.2006.09.079

Schwartz, M. H., Rozumalski, A., & Trost, J. P. (2008). The effect of walking speed on the gait of typically developing children. Journal of Biomechanics, 41(8), 1639–1650. https://doi.org/10.1016/J.JBIOMECH.2008.03.015

Seth, A., Sherman, M., Reinbolt, J. A., & Delp, S. L. (2011). OpenSim: a musculoskeletal modeling and simulation framework for in silico investigations and exchange. Procedia IUTAM, 2, 212–232. https://doi.org/10.1016/j.piutam.2011.04.021

Posterior Walker-Assisted Gait in Children With and Without Cerebral Palsy: Mechanical Work

5.1 Introduction

It is generally believed that typically developing children walk with a cyclic gait pattern which employs movement methods that are the most energetically efficient to achieve the desired motion (Cavagna, Thys, Zamboni, Vagna, & Thiys, 1976). Bipedal walking is comparable to a simple pendulum system where the tradeoff between potential and kinetic energy drives the system. Decreases in energetic efficiency occur in bipedal walking through muscle actuation to overcome energy loss and to maintain stability of the system. Cerebral Palsy (CP) is a non-progressive motor development disorder initiated by a lesion to the brain around the time of birth. CP can inhibit the growth, neurological control, and function of ambulatory muscles often causing different visual gait patterns depending on how it manifests itself in the patient. For persons with CP, gait efficiency can be significantly lower than their typically developed peers due to excess motion used for balance and co-contraction of muscles (Ries & Schwartz, 2018; S. Russell et al., 2011a). This excess energy cost is important for clinicians to keep in mind as they consider interventions to assist with gait. While CP is non-progressive, children with CP are observed as they grow because the effectiveness of their musculoskeletal structure in gait may decline as they age (Damiano, 2006; Hanna et al., 2009; Richards & Malouin, 2013). This can cause further motor impairment and make ambulation more inefficient if not impossible and force the young adult into a wheelchair. Clinical intervention often includes the prescription of the use of an assistive device which acts to stabilize the user during walking, like a posterior walker. This work theorized that the posterior walker would add an energetic cost associated with pulling it along and increase the energy requirements of the users' gait. The energetics of walker-assisted gait have not been examined in this way to define the cost of the walker in previous literature.

Many studies have shown that the gait of children with CP requires more energy use compared to their typically developed peers through oxygen consumption and heart rate measures (Norman et al., 2004; Unnithan et al., 1996). Studies have also shown that as children with CP age, their energy use during gait increases (Bolster, Balemans, Brehm, Buizer, & Dallmeijer, 2017b). These methods of measuring energy cost on the individual cannot correlate an increase in energy cost to a specific time in the gait cycle or part of the body as they are calculated for the whole body over long periods of activity. Mechanical energy metrics can be applied to the whole body while still providing individual time points of data for different segments and joints of the body. In pediatric CP gait mechanical work is higher than in the gait of their typically developed peers (Van de Walle et al., 2012; van den Hecke et al., 2007). For mechanical energy outcomes to be accurate they must incorporate the entire body for analysis otherwise the minimum energetic cost would be biased by the excluded segments.

The energetic cost of using an assistive device for a child with CP has not been adequately addressed in literature. Previous work examining walker use made mechanical energy calculations for upper body segments and has not looked at the whole body. One study by Konop et al. (Konop et al., 2009) looked at upper body kinetics during walker use. They found preliminary data suggesting a relationship between heart rate and the force used to move the walker forward. Other studies have only looked at the kinematics of the gait and did not instrument the device to model kinetics (Krautwurst, Dreher, & Wolf, 2016b; Paul et al., 2007; Protas et al., 2007). This work computed the load of the walker in pediatric gait by measuring the forces applied to the walker and calculating mechanical joint work over the gait cycle. Lower body, torso, and upper body segment work were calculated to examine the effect of walker-assisted gait on different joints in the body. It was hypothesized that the total mechanical work for baseline gait would show similar trends as previously published metabolic data comparing CP and TD groups. Additionally, the incorporation of the walker into a dynamic model was expected to increase the accuracy of the model outcomes for walker-assisted gait and reveal the dynamic cost of the walker on gait. By pulling the walker along, the CP group was expected to spend more energy in walker-assisted gait than baseline for the benefit of using the walker for support.

5.2 Methods

5.2.1 Subjects and Model

The kinematic and kinetic data for the same 17 children collected in the Motion Analysis and Motor Performance laboratory at the University of Virginia from Chapter 4 were used. Data exported from Vicon Nexus software was filtered and prepared in Matlab then modelled in MSC.Adams with the LifeMod plugin. A model with 19 segments and 18 joints was created for each subject based on individual anthropometric data. Trials for baseline walking and walking with a posterior pediatric walker were modelled. Inverse dynamic simulations were performed and joint kinematics and kinetics were exported as model outputs. Individual outputs from baseline and walker trials were exported for analysis.

5.2.2 Walker Interaction

The forces on the walker during a gait cycle were examined to quantify the incorporation of the walker into the user's gait. Directional forces were summed over the gait cycle (gc):

$$F_X = \int_{gc} f_x \, dx \qquad F_Y = \int_{gc} f_y \, dx \qquad F_Z = \int_{gc} f_z \, dx$$

The force in the direction of travel (F_X) indicates how much force was used to move the walker. Anterior force was normalized by body mass to quantify the force applied to move the walker by each subject. The vertical (F_Z) and lateral (F_Y) forces measure the degree to which the walker was used for stability. Vertical and lateral forces applied to the handles were normalized via body mass to quantify the stabilizing forces applied by each test subject.

The work done on the walker was quantified to examine the energetic cost of pulling the walker through the subject's gait. Work done to move the walker was calculated by multiplying the movement of the walker by the force applied to the load cells over each time point in the gait cycle (gc):

$$W_{walker} = \int_{gc} |f * \Delta d|$$

The work on the walker was normalized by the net excursion of the subject in the direction of travel over the gait cycle. Negative and positive work on the walker required action by the user and contributed to the cost of the walker for movement. Total work was calculated as the absolute value of the work performed on the walker.

5.2.3 Total Mechanical Work

Gait efficiency is an important outcome measure in assessing the motor function of children with cerebral palsy (Bolster et al., 2017b; Norman et al., 2004). Mechanical work estimates the energy cost on an individual by quantifying the mechanical energy cost of movement. Using the model outputs, the mechanical work was calculated like that presented by Chen et al. (Chen, Kuo, & Andriacchi, 1997). The mechanical work at the joint *j* is calculated as the sum over the gait cycle of the product of the joint moment (τ_j) and change in joint angle ($\Delta\theta_j$):

$$W_j^T = \int_{gc} |\tau_j * \Delta\theta_j|$$

Mechanical work calculates the minimum work required for the system's movement. Co-contraction and metabolic costs are not included, each joint's contribution to the total cost is included. Individual joint totals were summed up over the gait cycle to calculate full body, lower body, and upper body mechanical work.

5.2.4 Joint Work

Calculating mechanical work of joints gives the ability to break the body down into individual segments and identify the joints with the greatest moments produced and the specific time points during the gait cycle where the work was greatest or least:

$$W_j = |\tau_j * \Delta\theta_j|$$

The body was split up into three parts: lower body joints including ankles, knees, and hips; upper body joints including wrists, elbows, shoulders, and scapulae; torso joints including lumbar, thoracic, and neck. The work generated at these segments was examined in relation to the gait cycle. Both positive and negative work require muscle forces and metabolic work to complete the actions of the body. The absolute value of the joint work was calculated so that when summing up into different segments there would be no cancellation between positive and negative values of work for different joints.

5.3 Results

5.3.1 Subjects

Subject anthropometrics were recorded during collection and are presented in Table 5.1. Body Mass Index was calculated based on subject mass, height, and age. Two groups were created to compare outcomes, one group of children with a diagnosis of cerebral palsy (CP) and another of typically developed (TD) peers as a control group. A range of children and young adults participated in the study for both groups. Through t-test comparison, weight (p=0.019) and height (p=0.049) were different between the groups. This highlights the differences between the populations, but for the purposes of this work the populations were considered similar as significance was not found in age or body mass index (BMI).

Table 5.1: Subject Anthropometrics. Significance between TD and CP groups is marked as * (p<.05), ** (p<0.01), *** (p<0.001).

Group	Age (years)	Weight (kg)	Height (cm)	BMI
TD	16±5.4	57.4±20.4	160.6±19.7	21.2±4.8
CP	13.5±2.4	36.2±5.6 *	144.0±6.6 *	17.2±2.1

5.3.2 Walker Interactions

The average forces exerted on the walker over a gait cycle were separated into anterior, lateral, and vertical directions for each subject and by groups, as presented in Table 5.2. Between groups, significance was seen in anterior/posterior and vertical forces ($p<0.001$). Large standard deviations within the CP group lower the power of these results and point to differences in walker use between subjects in the CP group. The work done to move the walker was greater in the CP group (Table 5.3). While both TD and CP groups applied similar negative work values, the CP group required more positive work to pull the walker for the necessary support they needed during gait.

Table 5.2: Average forces on the walker over gait cycle in the anterior (+)/posterior (-), lateral (+)/medial (-), and down (+)/up (-) directions normalized by mass of subject. Significance between TD and CP groups marked as * ($p<0.05$), ** ($p<0.01$), *** ($p<0.001$).

	Anterior/Posterior Force (N)	Medial/Lateral Force (N)	Vertical Force (N)
TD	5.933±4.198	2.026±2.587	17.45±14.27
CP	13.31±6.716*	0.450±11.08	82.18±54.56*

Table 5.3: Average total, positive, and negative work done on the walker over the gait cycle normalized by excursion of subject over gait cycle. Significance between TD and CP groups marked as * ($p<0.05$), ** ($p<0.01$), *** ($p<0.001$).

	Total Work on Walker (J/m)	Positive Work on Walker (J/m)	Negative Work on Walker (J/m)
TD	7.08±2.59	6.57±3.39	-0.51±1.09
CP	10.62±3.77	10.18±3.85	-0.44±0.73

5.3.3 Total Mechanical Work

Mechanical joint work was calculated for each joint then summed for the full body, lower body, upper body and torso and presented in Table 5.4. Total mechanical joint work for the whole body was not significantly different for the TD group when comparing the walker trials to the baseline trials. Reductions in knee joint work were offset by increases in shoulder, elbow, and wrist work.

The CP group was comprised of four subjects who were able to ambulate without walker or crutch assistance in their baseline gait. The CP group had reduced total work in walker-assisted gait compared to baseline gait which was not expected as the walker had to be pulled along to be used for support in their gait. The load of pulling the walker increased the upper body activation. The upper body joint work for the CP group was greater in the shoulder, elbow, and wrist in walker-assisted gait. Large decreases in the lower body and torso joints from baseline gait to walker-assisted gait highlight a decreased load on the lower body and more control of the torso.

Additionally, the CP group walker-assisted gait was compared to the walker-assisted gait of the subgroup of CP subjects who used crutches in their baseline gait. Larger variance in the smaller groups reduced the power of the results but the full body work was similar between the two groups of CP subjects. The group that walked unassisted in baseline was not as accustomed to using assistive devices and a larger portion of the full body work was performed by the lower body in walker-assisted gait. The crutch-assisted group had larger upper body work than the group of CP subjects who could walk without assistance in their baseline.

Table 5.4: Average total mechanical work for individual joints summed over a gait cycle. Significance between baseline and walker trials for TD group and Unassisted CP group is marked as * (p<0.05), ** (p<0.01), *** (p<0.001). Walker assisted trials of crutch and unassisted subgroup are compared and marked following the same method.

Work (J/kg*m)	TD Group		CP Group		
			Unassisted subgroup		Crutch-assisted subgroup
Joints	Baseline	Walker	Baseline	Walker	Walker
Ankle	0.385 ± 0.065	0.387 ± 0.058	0.489 ± 0.140	0.357 ± 0.088	0.338 ± 0.147
Knee	0.382 ± 0.055	0.320 ± 0.064 *	0.494 ± 0.061	0.368 ± 0.106	0.344 ± 0.138
Hip	0.593 ± 0.125	0.557 ± 0.081	1.013 ± 0.197	0.918 ± 0.099	0.785 ± 0.182
LOWER BODY	**2.72 ± 0.31**	**2.53 ± 0.27**	**3.99 ± 0.74**	**3.29 ± 0.52**	**2.93 ± 0.89**
Lumbar	0.097 ± 0.044	0.109 ± 0.054	0.310 ± 0.082	0.248 ± 0.076	0.254 ± 0.136
Thoracic	0.070 ± 0.045	0.082 ± 0.043	0.452 ± 0.185	0.284 ± 0.138	0.346 ± 0.285
Neck	0.015 ± 0.006	0.018 ± 0.014	0.066 ± 0.021	0.072 ± 0.044	0.076 ± 0.059
TORSO	**0.18 ± 0.09**	**0.21 ± 0.10**	**0.83 ± 0.25**	**0.60 ± 0.25**	**0.68 ± 0.47**
Scapular	0.026 ± 0.010	0.023 ± 0.014	0.083 ± 0.017	0.101 ± 0.047	0.146 ± 0.108
Shoulder	0.017 ± 0.009	0.024 ± 0.015	0.078 ± 0.022	0.110 ± 0.030	0.141 ± 0.052
Elbow	0.008 ± 0.003	0.012 ± 0.008	0.032 ± 0.006	0.063 ± 0.030	0.124 ± 0.108
Wrist	0.001 ± 0.000	0.003 ± 0.002 *	0.004 ± 0.002	0.014 ± 0.007 *	0.041 ± 0.037
UPPER BODY	**0.10 ± 0.04**	**0.12 ± 0.07**	**0.39 ± 0.08**	**0.58 ± 0.21**	**0.90 ± 0.61**
TOTAL	**3.01 ± 0.38**	**2.86 ± 0.33**	**5.21 ± 0.83**	**4.47 ± 0.89**	**4.51 ± 1.84**

5.3.4 Mechanical Work of the Body

Mechanical work over the gait cycle was calculated for each joint over the gait cycle and the results were summed into different body segments: full body, lower body, torso, and upper body work. Baseline gait was compared between TD and CP groups and walker-assisted gait was compared to baseline gait within groups. Walker-assisted trials were compared to baseline trials within groups and the crutch-assisted group was compared to the unassisted group in walker-assisted trials.

5.3.3.1 TD vs CP Mechanical Work

TD group mechanical work was compared to the CP group using SPM statistical analysis (Figure 5.1). Total work in TD baseline gait was characterized by small peak work contributions in early stance and late swing phases of gait. Additionally, large peak work occurred in late stance in preparation for toe off. The majority of the work was done by the lower body in TD baseline gait where the same patterns were observed as in the full body. Sagittal plane work mirrored the summation of all the planes with peaks in early and late stance and late swing. While frontal plane work had two peaks in early and late stance and transverse plane work had a peak in late stance.

The CP group required a greater mechanical work to move through their gait successfully than the TD group. The CP group required a larger net work in early stance, mid-stance, and through swing compared to the TD group. In late stance phase, the CP group had a similar peak work to the TD group. Lower body sagittal plane work contributed to increased work in CP during early and mid-stance phase and sagittal work in the lower body was lower in late stance in CP. Upper body and torso segment sagittal work were also increased in CP throughout the gait cycle. Frontal plane increases occurred in the torso and upper body segments while the lower body in CP was similar to that in the TD group.

The CP group used their torso more for balance and stability than the TD group which had reduced work in the torso and upper body. Cyclic patterns in all three planes of motion in the torso with greater magnitudes highlight the larger activation of the torso in CP. The upper body also had greater use in CP with patterned usage in total work.

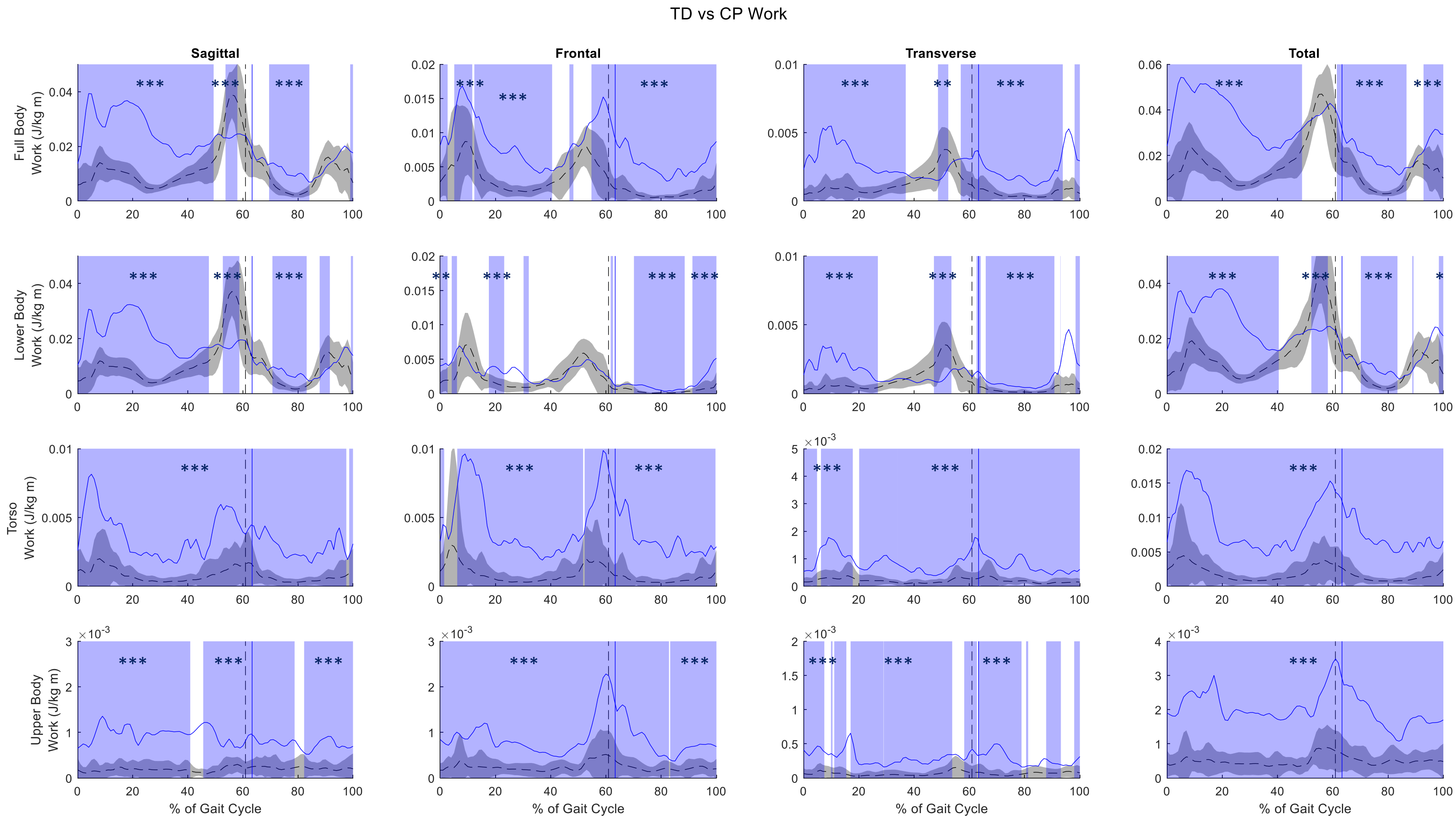

Figure 5.1: Mechanical work over the gait cycle for the full body, lower body, torso, and upper body segments. Baseline gait for TD group (black) and CP group (blue) work is compared using SPM analysis. SPM significance is highlighted by blue shaded boxes and marked with p<0.05 (*), p<0.01 (**), and p<0.001 (***).

5.3.3.2 TD Baseline vs Walker-Assisted Gait Mechanical Work

The TD group gait required slightly less peak mechanical work when walking with a walker but was not different from baseline at times of peak work (Figure 5.2). In early and late stance and late swing, the lower body had lower peak work in walker-assisted gait than in baseline. The upper body had increased work from late swing to early stance in walker-assisted gait. Minima of mechanical work in baseline gait occurring at mid-stance and mid-swing increased in walker-assisted gait in the full body, torso, and lower body.

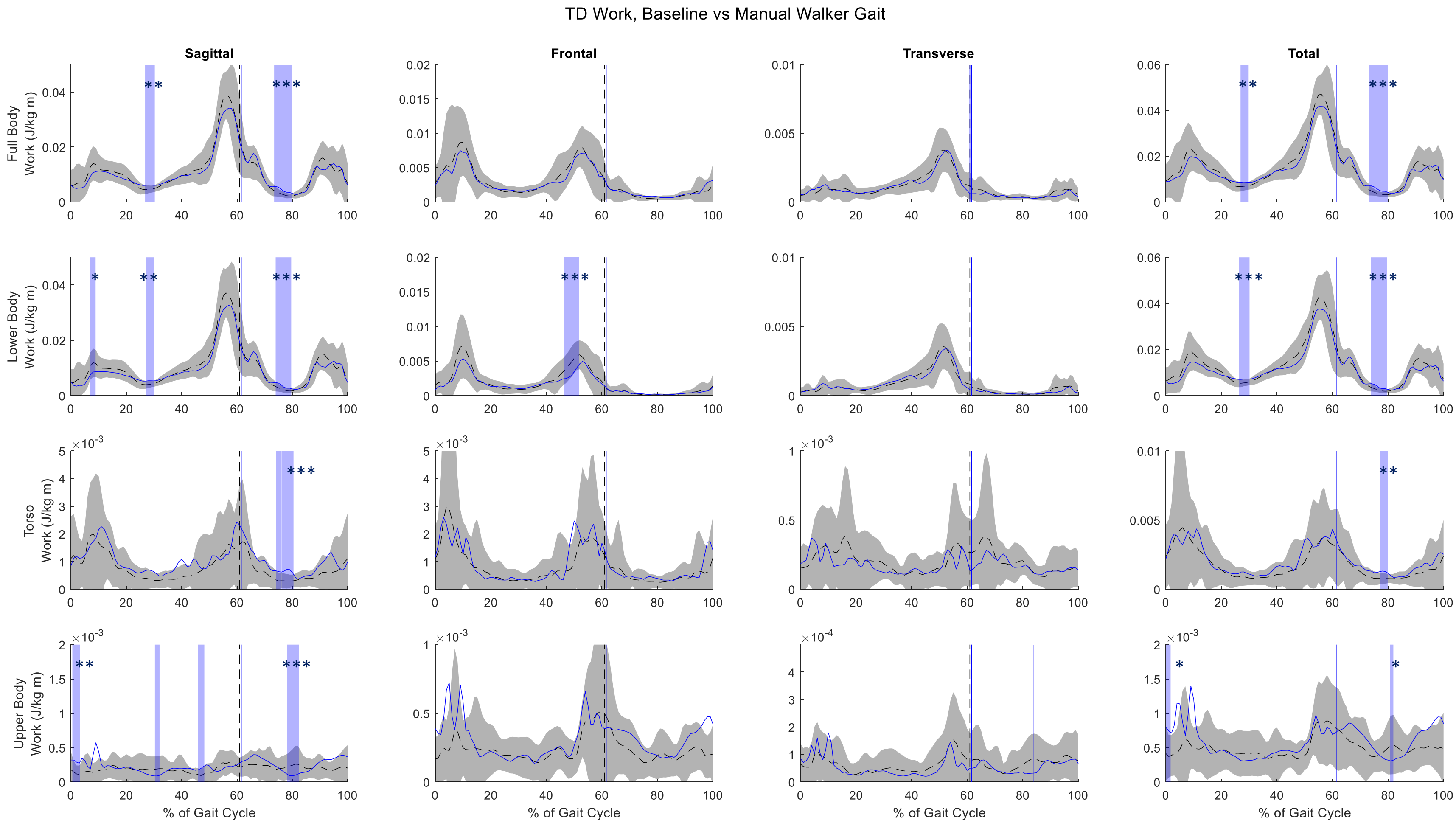

Figure 5.2: Mechanical work over the gait cycle for the full body, lower body, torso, and upper body segments. TD group baseline gait (black) compared to manual walker gait (blue) work is compared using SPM analysis. SPM significance is highlighted by blue shaded boxes and marked with $p<0.05$ (*), $p<0.01$ (**), and $p<0.001$ (***).

5.3.3.3 CP Baseline vs Walker-Assisted Gait Mechanical Work

The CP group had significantly reduced mechanical work during stance phase in walker-assisted gait compared to their baseline (Figure 5.3). Sagittal plane work was reduced in the lower body particularly in early stance. The cyclic pattern from baseline gait in the frontal plane did not occur in the upper body and was minimized in the torso during walker-assisted gait. Peak upper body work changed from occurring at toe off in baseline gait to heel strike and early stance in walker assisted gait.

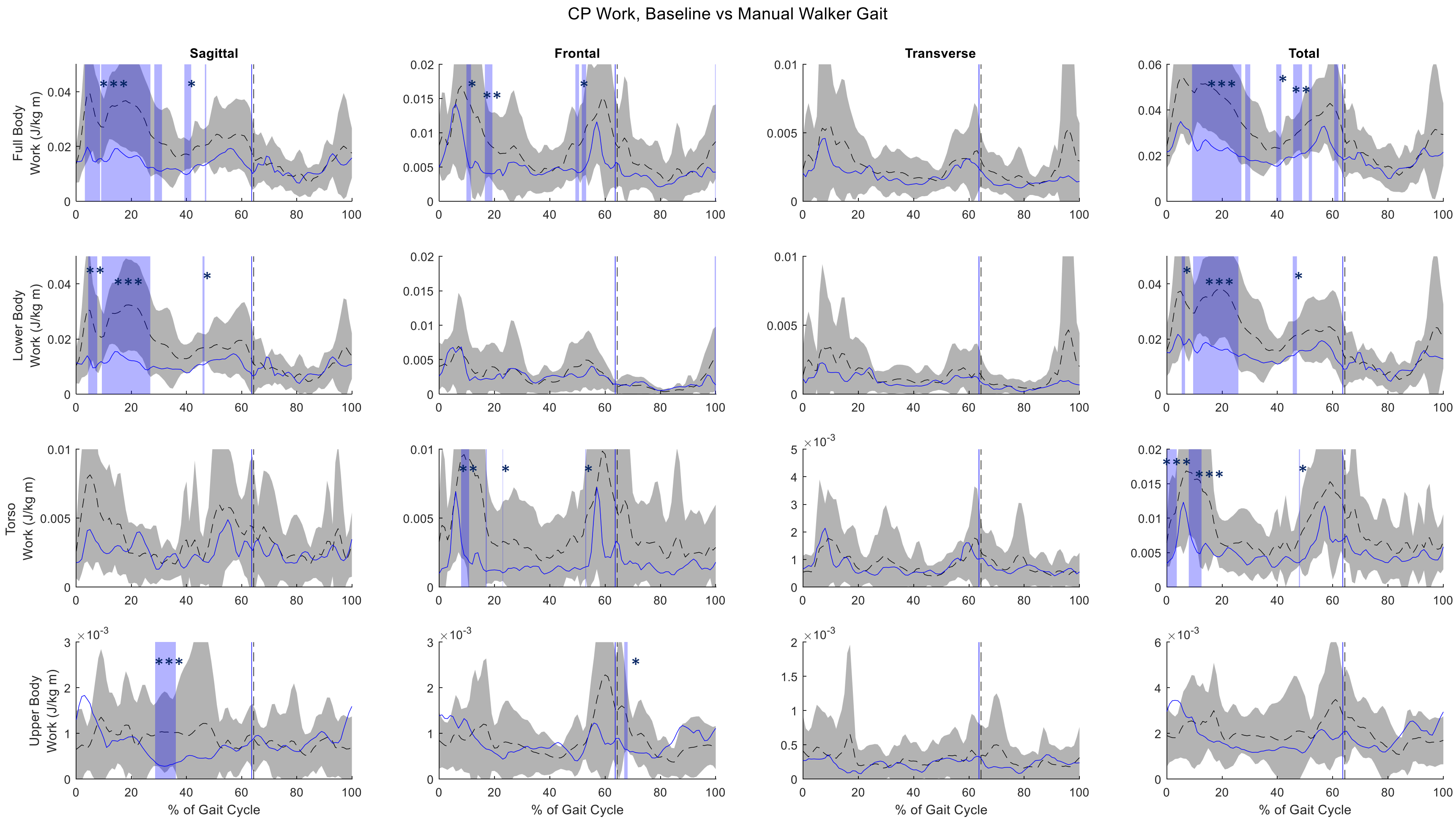

Figure 5.3: Mechanical work over the gait cycle for the full body, lower body, torso, and upper body segments. CP group baseline gait (black) and manual walker gait (blue) work is compared using SPM analysis. SPM significance is highlighted by blue shaded boxes and marked with $p<0.05$ (*), $p<0.01$ (**), and $p<0.001$ (***).

5.3.3.4 Mechanical Work in Walker-Assisted Gait for Un- vs Crutch-Assisted Baseline Groups

In examining the full body, the crutch-assisted (CA) group did less work than the unassisted (NA) group when walking with a walker (Figure 5.4). In the lower body, the sagittal plane work was greater in early stance and lower in late stance in the CA group. Torso segment peak work was reduced in early and late stance in the CA group compared to the NA group. Little significance occurred in the upper body comparisons but the CA group had greater peaks and thus greater work over the gait cycle than the NA group in the upper body segments.

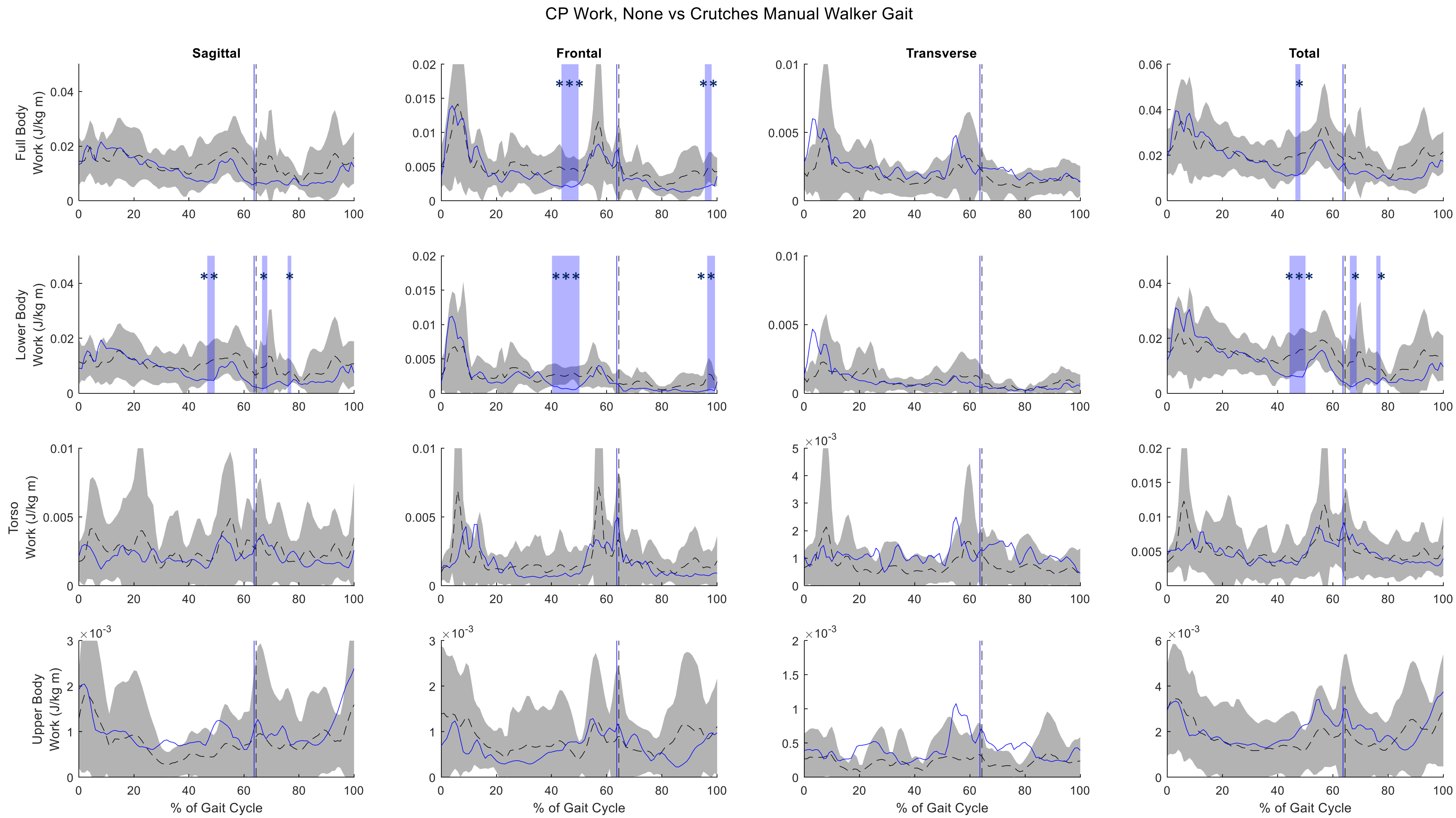

Figure 5.4: Mechanical work over the gait cycle for the full body, lower body, torso, and upper body segments. Manual walker gait for NA group (black) and CA group (blue) work is compared using SPM analysis. SPM significance is highlighted by blue shaded boxes and marked with p<0.05 (*), p<0.01 (**), and p<0.001 (***).

5.4 Discussion

This study employed a method for calculating full body and individual joint mechanical work while walking with an instrumented pediatric posterior walker. Along with the spatiotemporal measures, kinematics, and kinetics from Chapters 3 and 4, a more complete picture of the pathology of CP and the effect of a posterior walker on gait was found by comparing the mechanical work of walker-assisted to baseline gait and quantifying the user's interaction with the walker. Overall, the walker reduced the total work necessary for the mechanical motion of the body during gait in both the TD and CP groups. The TD and CP groups experienced reduced total mechanical work over the gait cycle in the lower body. In addition, the CP group also had reduced work in the torso and an increased work in the upper body. In examining the work over the gait cycle the same effects can be observed but the increase in work in the upper body was not clear as it was normalized to the length of the gait cycle. Reduced velocity observed in the walker-assisted trials may contribute to the reduced joint mechanical work. The posterior walker reduced the mechanical cost of gait with little side effects on the mechanical work of the body. Further comparisons of these outcomes to oxygen consumption and heart rate metrics would further inform this outcome. The results of this work suggest that the benefit of the walker outweighs its detriment for the CP group when considering the mechanical work of the body over the gait cycle.

The mechanical work for the TD and CP group baseline trials showed similar relationships to previously published data. Russell et al. calculated a 59% greater work in CP gait compared to TD gait (S. Russell et al., 2011a). In this work, a more involved group of CP subjects averaged 73% greater total body work compared to the TD group during baseline gait. As was expected, the CP group showed significantly greater work being done in the lower body, torso, and upper body segments. In the CP group, the upper body accounted for a larger portion of the full body work than in the TD group. This increase in work was due to the increase in upper body motion for angular momentum generation to aid in the balance and stability of gait. In TD baseline gait there was very little upper body motion observed. At toe off times the CP group needed greater work generation in the upper body and torso to remain stable in their baseline, unassisted gait compared to the TD group. While the lower body frontal plane work was similar in CP to TD group gait, the torso and upper body increases for stability affected the full body outcomes. Increases in the sagittal plane work for the lower body highlighted the increased cost of movement for the CP group particularly through stance phase. The incorporation of a walker into the CP group gait reduced the excessive motion in the upper body and was expected to add an energetic cost for pulling the walker around.

The mechanical work required for gait in the TD group was mostly unaffected by the inclusion of the walker. The TD group did not apply a great amount of force to the walker compared to the CP group. It was expected that the walker inclusion in their gait would not change or slightly increase the total mechanical work of the body as the walker was not used for stability by the TD group. Compared to the baseline trials, the full body work in the walker trials was lower but not significant ($p=0.178$). In Chapter 4, the knee and hip joints had a lower range of motion in the TD group which reduced the mechanical work of the knee. While minor changes were observed throughout the gait cycle in these segments, the full body work was slightly reduced in early and late stance and increased in mid-stance and mid-swing.

Pathologic CP gait showed a greater significance when comparing between baseline and walker-assisted trials. From Chapters 3 and 4 the CP group was more stable in walker-assisted gait, this was corroborated by the larger vertical and lateral forces on the walker. These forces indicate a reliance on the walker for a point of contact with the ground to increase balance. The larger forces and work on the walker in the direction of travel also highlight the greater effort required to move the walker compared to the TD group. Reductions in the total lower body and torso work were not offset by increases in the upper body to move the walker forward. In walker-assisted gait the lower body joints produced lower moments to achieve the motion in the model which reduced the work in the lower body. Peak work around toe off in the upper body and torso were also reduced in walker-assisted gait as the walker enabled a more stable gait and posture characterized by less motion and joint moment generation in the upper body. The full body work calculation did not show the effect of the walker energy cost between baseline and walker trials.

The full body total work was lower for the walker trials compared to baseline which was counter to the outcome expected. Previous studies have found lower velocity of walking gait typically accompanies decreased joint moment and power (de David, Carpes, & Stefanyshyn, 2015; Schwartz et al., 2008). The reduced velocity in walker-assisted trials could be a contributing factor to the reduced work calculated in the lower body and torso as this was not a controlled factor in this study. Larger reductions in work were observed in the CP group than the TD group while velocity decreased for both groups to similar levels of significance. Further work should examine the effect of velocity of gait on mechanical work of the joints and how it relates between walker-assisted and unassisted gait. The computation of body angular momentum would also be a valuable assessment of stability of the system for future work to consider (S. D. Russell et al., 2007).

The walker-assisted trials of the subgroup of children with CP which walked with crutches (CS) in their baseline gait were compared to the CP group which did not use crutches (NA) to observe the effect of CP on the walker use for different groups of children with CP. The children who used crutches in their typical gait were unable to ambulate without them and so were more reliant on an assistive device to move. With a smaller group who used crutches, significance of the results was low but the CS group trended towards a greater total mechanical work in walker-assisted trials than the NA group. Lower body segment work was reduced in the CS group but torso and upper body work were greater. Particularly, the elbow and wrist work were significantly increased in the CS group indicating a greater reliance on the walker for support. As this population was unable to walk without assistance in their baseline gait, their reliance on the walker was expected to be greater than the NA group.

5.5 Conclusion

Total mechanical work for the whole body, torso, and lower and upper body segments was calculated for TD and CP group baseline and walker trials. Mechanical work was lower in the TD group than the CP group. The TD group saw little difference in any summed segment work totals when using the walker compared to baseline. The CP group used the walker for support as opposed to the TD group and had to apply greater force in the anterior direction over a gait cycle to pull the walker. While they required more work to move the walker, the CP group walked better with a posterior walker and with reduced mechanical work of the body through their gait cycle. The mechanical work may have been effected by the velocity of gait and further analysis of the effect of velocity on mechanical work should be performed for this population. Future work to compare these outcomes to metabolic cost measures like oxygen consumption and heart rate would be valuable to compare to the outcomes in this work. Additionally, the stability of the system could be measured through angular momentum for further understanding of the effects of a posterior walker in CP gait.

Reference List

Bolster, E. A. M., Balemans, A. C. J., Brehm, M. A., Buizer, A., & Dallmeijer, A. J. (2017). Energy cost during walking in association with age and body height in children and young adults with cerebral palsy. Gait and Posture, 54, 119–126. https://doi.org/10.1016/j.gaitpost.2017.02.026

Cavagna, G. A., Thys, H., Zamboni, A., Vagna, G. A. C., & Thiys, H. (1976). THE SOURCES OF EXTERNAL WORK IN LEVEL WALKING AND RUNNING. J. Physiol (Vol. 262).

Chen, I. H., Kuo, K. N., & Andriacchi, T. P. (1997). The influence of walking speed on mechanical joint power during gait. Gait & Posture, 6(3), 171–176. https://doi.org/10.1016/S0966-6362(97)00009-X

Damiano, D. L. (2006). Activity, Activity, Activity: Rethinking Our Physical Therapy Approach to Cerebral Palsy. Physical Therapy, 86(11), 1534–1540.

de David, A. C., Carpes, F. P., & Stefanyshyn, D. (2015). Effects of changing speed on knee and ankle joint load during walking and running. Journal of Sports Sciences. https://doi.org/10.1080/02640414.2014.946074

Hanna, S. E., Rosenbaum, P. L., Bartlett, D. J., Palisano, R. J., Walter, S. D., Avery, L., & Russell, D. J. (2009). Stability and decline in gross motor function among children and youth with cerebral palsy aged 2 to 21 years. Developmental Medicine and Child Neurology. https://doi.org/10.1111/j.1469-8749.2008.03196.x

Konop, K. A., Strifling, K. M. B., Wang, M., Cao, K., Eastwood, D., Jackson, S., … Harris, G. F. (2009). Upper extremity kinetics and energy expenditure during walker-assisted gait in children with cerebral palsy. Acta Orthopaedica et Traumatologica Turcica, 43(2), 156–164. https://doi.org/10.3944/AOTT.2009.156

Krautwurst, B. K., Dreher, T., & Wolf, S. I. (2016). The impact of walking devices on kinematics in patients with spastic bilateral cerebral palsy. Gait & Posture, 46, 184–187. https://doi.org/10.1016/j.gaitpost.2016.03.014

Norman, J. F., Bossman, S., Gardner, P., & Moen, C. (2004). Comparison of the Energy Expenditure Index and Oxygen Consumption Index During Self-Paced Walking in Children with Spastic Diplegia Cerebral Palsy and Children Without Physical Disabilities. Pediatric Physical Therapy, 16(4), 206–211. https://doi.org/10.1097/01.PEP.0000145930.84009.23

Paul, S. M., Siegel, K. L., Malley, J., & Jaeger, R. J. (2007). Evaluating interventions to improve gait in cerebral palsy: a meta-analysis of spatiotemporal measures. Developmental Medicine & Child Neurology, 49(7), 542–549. https://doi.org/10.1111/j.1469-8749.2007.00542.x

Protas, E. J., Raines, M. L., & Tissier, S. (2007). Comparison of Spatiotemporal and Energy Cost of the Use of 3 Different Walkers and Unassisted Walking in Older Adults. Archives of

Physical Medicine and Rehabilitation, 88(6), 768–773. https://doi.org/10.1016/j.apmr.2007.03.001

Richards, C. L., & Malouin, F. (2013). Cerebral palsy. In Pediatric Neurology Part I (Vol. 111, pp. 183–195). https://doi.org/10.1016/B978-0-444-52891-9.00018-X

Ries, A. J., & Schwartz, M. H. (2018). Low gait efficiency is the primary reason for the increased metabolic demand during gait in children with cerebral palsy. Human Movement Science, 57, 426–433. https://doi.org/10.1016/j.humov.2017.10.004

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011). The gait of children with and without cerebral palsy: work, energy, and angular momentum. J Appl Biomech, 27(2), 99–107. https://doi.org/10.1123/jab.27.2.99

Russell, S. D., Bennett, B. C., Kerrigan, D. C., & Abel, M. F. (2007). Determinants of gait as applied to children with cerebral palsy. Gait and Posture, 26(2), 295–300. https://doi.org/10.1016/j.gaitpost.2006.09.079

Schwartz, M. H., Rozumalski, A., & Trost, J. P. (2008). The effect of walking speed on the gait of typically developing children. Journal of Biomechanics, 41(8), 1639–1650. https://doi.org/10.1016/J.JBIOMECH.2008.03.015

Unnithan, V. B., Dowling, J. J., Frost, G., & Bar-Or, O. (1996). Role of cocontraction in the O2 cost of walking in children with cerebral palsy. Medicine and Science in Sports and Exercise, 28(12), 1498–1504. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8970144

Van de Walle, P., Hallemans, A., Schwartz, M., Truijen, S., Gosselink, R., & Desloovere, K. (2012). Mechanical energy estimation during walking: Validity and sensitivity in typical gait and in children with cerebral palsy. Gait & Posture, 35(2), 231–237. https://doi.org/10.1016/j.gaitpost.2011.09.012

van den Hecke, A., Malghem, C., Renders, A., Detrembleur, C., Palumbo, S., & Lejeune, T. M. (2007). Mechanical Work, Energetic Cost, and Gait Efficiency in Children With Cerebral Palsy. Journal of Pediatric Orthopaedics, 27(6), 643–647. https://doi.org/10.1097/BPO.0b013e318093f4c3

Bibliography

Armand, S., Decoulon, G., & Bonnefoy-Mazure, A. (2016a). Gait analysis in children with cerebral palsy. *EFORT Open Reviews*, *1*(12), 448–460. https://doi.org/10.1302/2058-5241.1.000052

Armand, S., Decoulon, G., & Bonnefoy-Mazure, A. (2016b). Gait analysis in children with cerebral palsy. *EFORT Open Reviews*, *1*(12), 448–460. https://doi.org/10.1302/2058-5241.1.000052

Bachschmidt, R., Harris, G. F., Ackman, J., Hassani, S., Carter, M., Caudill, A., … Klein, J. (2000). Quantitative study of walker-assisted gait in children with cerebral palsy: anterior versus posterior walkers. In *Pediatric Gait: A New Millennium in Clinical Care and Motion Analysis Technology* (pp. 217–223). IEEE. https://doi.org/10.1109/PG.2000.858901

Baker, R., Leboeuf, F., Reay, J., & Sangeux, M. (2017). The Conventional Gait Model - Success and Limitations. In *Handbook of Human Motion* (pp. 1–19). Cham: Springer International Publishing. https://doi.org/10.1055/s-0028-1087518

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: Benefits, demands, and adverse consequences. *Archives of Physical Medicine and Rehabilitation*, *86*(1), 134–145. https://doi.org/10.1016/j.apmr.2004.04.023

Baughman, L. D. (1983). *Development of an Interactive Computer Program to Produce Body Description Data*. Dayton, OH. Retrieved from http://www.dtic.mil/docs/citations/ADA133720

Bolster, E. A. M., Balemans, A. C. J., Brehm, M.-A., Buizer, A. I., & Dallmeijer, A. J. (2017a). Energy cost during walking in association with age and body height in children and young adults with cerebral palsy. *Gait & Posture*, *54*, 119–126. https://doi.org/10.1016/j.gaitpost.2017.02.026

Bolster, E. A. M., Balemans, A. C. J., Brehm, M. A., Buizer, A., & Dallmeijer, A. J. (2017b). Energy cost during walking in association with age and body height in children and young adults with cerebral palsy. *Gait and Posture*, *54*, 119–126. https://doi.org/10.1016/j.gaitpost.2017.02.026

Brégou Bourgeois, A., Mariani, B., Aminian, K., Zambelli, P. Y., & Newman, C. J. (2014). Spatio-temporal gait analysis in children with cerebral palsy using, foot-worn inertial sensors. *Gait and Posture*. https://doi.org/10.1016/j.gaitpost.2013.08.029

Burns, F., Stewart, R., Reddihough, D., Scheinberg, A., Ooi, K., & Graham, H. K. (2014). The cerebral palsy transition clinic: administrative chore, clinical responsibility, or opportunity for audit and clinical research? *Journal of Children's Orthopaedics*. https://doi.org/10.1007/s11832-014-0569-0

Cans, C. (2000). Surveillance of cerebral palsy in Europe: a collaboration of cerebral palsy surveys and registers. *Developmental Medicine & Child Neurology*, *42*(12), 816–824. https://doi.org/10.1111/j.1469-8749.2000.tb00695.x

Cavagna, G. A., Thys, H., Zamboni, A., Vagna, G. A. C., & Thiys, H. (1976). *THE SOURCES OF EXTERNAL WORK IN LEVEL WALKING AND RUNNING*. *J. Physiol* (Vol. 262).

Chen, I. H., Kuo, K. N., & Andriacchi, T. P. (1997). The influence of walking speed on mechanical joint power during gait. *Gait & Posture*, *6*(3), 171–176. https://doi.org/10.1016/S0966-6362(97)00009-X

Cheng, H., Obergefell, L., & Rizer, A. (1994). *Generator of Body Data (GEBOD) , Manual.* Dayton, OH. Retrieved from http://www.dtic.mil/docs/citations/ADA289721

Colver, A., Rapp, M., Eisemann, N., Ehlinger, V., Thyen, U., Dickinson, H. O., … Arnaud, C. (2015). Self-reported quality of life of adolescents with cerebral palsy: a cross-sectional and longitudinal analysis. *The Lancet*, *385*(9969), 705–716. https://doi.org/10.1016/S0140-6736(14)61229-0

Damiano, D. L. (2006). Activity, Activity, Activity: Rethinking Our Physical Therapy Approach to Cerebral Palsy. *Physical Therapy*, *86*(11), 1534–1540.

de David, A. C., Carpes, F. P., & Stefanyshyn, D. (2015). Effects of changing speed on knee and ankle joint load during walking and running. *Journal of Sports Sciences*. https://doi.org/10.1080/02640414.2014.946074

Delp, S. L., Anderson, F. C., Arnold, A. S., Loan, P., Habib, A., John, C. T., … Thelen, D. G. (2007). OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement. *IEEE TRANSACTIONS ON BIOMEDICAL ENGINEERING*, *54*(11). https://doi.org/10.1109/TBME.2007.901024

DeLuca, P. A. (1991). Gait analysis in the treatment of the ambulatory child with cerebral palsy. *Clinical Orthopaedics and Related Research*, (264), 65–75. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/1997253

Dziuba, A. K., Tylkowska, M., & Jaroszczuk, S. (2014). Index of mechanical work in gait of children with cerebral palsy. *Acta of Bioengineering and Biomechanics / Wroclaw University of Technology*, *16*(3), 77–87. https://doi.org/10.5277/abb140309

Fast, A., Wang, F. S., Adrezin, R. S., Cordaro, M. A., Ramis, J., & Sosner, J. (1995). The instrumented walker: Usage patterns and forces. *Archives of Physical Medicine and Rehabilitation*, *76*(5), 484–491. https://doi.org/10.1016/S0003-9993(95)80584-2

Friston, K. J., Ashburner, J. T., Kiebel, S. J., Nichols, T. E., & Penny, W. D. (2011). *Statistical parametric mapping: the analysis of functional brain images*. Amsterdam: Elsevier.

Frost, G., Dowling, J., Bar-Or, O., & Dyson, K. (1997). Ability of mechanical power estimations to explain differences in metabolic cost of walking and running among children. *Gait & Posture*, *5*(2), 120–127. https://doi.org/10.1016/S0966-6362(96)01108-3

Gage, J. R., DeLuca, P. A., & Renshaw, T. S. (1996). Gait analysis: principle and applications with emphasis on its use in cerebral palsy. *Instructional Course Lectures*, *45*(10), 491–507. Retrieved from http://jbjs.org/content/77/10/1607

Gibson, T., Jeffery, R. S., & Bakheit, A. M. O. (2006). Comparison of three definitions of the mid-stance and mid-swing events of the gait cycle in children. *Disability and*

Rehabilitation. https://doi.org/10.1191/0959683606hl959rp

Gough, M., Eve, L. C., Robinson, R. O., & Shortland, A. P. (2004). Short-term outcome of multilevel surgical intervention in spastic diplegic cerebral palsy compared with the natural history. *Developmental Medicine & Child Neurology*, *46*(02), 91–97. https://doi.org/10.1017/S0012162204000192

Greiner, B. M., Czerniecki, J. M., & Deitz, J. C. (1993). Gait parameters of children with spastic diplegia: a comparison of effects of posterior and anterior walkers. *Archives of Physical Medicine and Rehabilitation*, *74*(4), 381–385. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8466419

Greiner, B. M., Czerniecki, J. M., & Deitz, J. C. (1993). Gait Parameters of Children With Spastic Diplegia: A Comparison of Effects of Posterior and Anterior Walkers. *Archives of Physical Medical Rehabilitation*, *74*, 381–385. https://doi.org/0003-9993/93/7404-0174

Hanna, S. E., Rosenbaum, P. L., Bartlett, D. J., Palisano, R. J., Walter, S. D., Avery, L., & Russell, D. J. (2009). Stability and decline in gross motor function among children and youth with cerebral palsy aged 2 to 21 years. *Developmental Medicine and Child Neurology*. https://doi.org/10.1111/j.1469-8749.2008.03196.x

Holder, C. G., Haskvitz, E. M., & Weltman, A. (1993). The Effects of Assistive Devices on the Oxygen Cost, Cardiovascular Stress, and Perception of Nonweight-Bearing Ambulation. *Journal of Orthopaedic & Sports Physical Therapy*, *18*(4), 537–542. https://doi.org/10.2519/jospt.1993.18.4.537

Honeycutt, A., Dunlap, L., Chen, H., al Homsi, G., Grosse, S., & Schendel, D. (2004). Economic Costs Associated with Mental Retardation, Cerebral Palsy, Hearing Loss, and Vision Impairment --- United States, 2003. Retrieved April 17, 2018, from https://www.cdc.gov/mmwr/preview/mmwrhtml/mm5303a4.htm

Johnson, D. C., Damiano, D. L., & Abel, M. F. (1997). The Evolution of Gait in Childhood and Adolescent Cerebral Palsy. *Journal of Pediatric Orthopaedics*, *17*(3), 392–396. Retrieved from https://ovidsp.tx.ovid.com/sp-3.30.0b/ovidweb.cgi

Johnston, T. E., Moore, S. E., Quinn, L. T., & Smith, B. T. (2004). Energy cost of walking in children with cerebral palsy-- relation to the Gross Motor Function Classification System. *Developmental Medicine & Child Neurology*, *46*(1), 34–38. https://doi.org/10.1111/j.1469-8749.2004.tb00431.x

Ju Kim, C., & min Son, S. (2014). Comparison of Spatiotemporal Gait Parameters between Children with Normal Development and Children with Diplegic Cerebral Palsy. *Journal of Physical Therapy Science*, *26*(9), 1317–1319. Retrieved from https://www.jstage.jst.go.jp/article/jpts/26/9/26_jpts-2014-008/_pdf/-char/ja

Kienast, G., Bachmann, D., Steinwender, G., Zwick, E. B., & Saraph, V. (1999). Determination of Gait Patterns in Children with Cerebral Palsy Using Cluster Analaysis. *Gait and Posture*, *10*, 57. Retrieved from http://www.gaitposture.com/article/S0966-6362(99)90401-0/pdf

Konop, K. A., Strifling, K. M. B., Wang, M., Cao, K., Eastwood, D., Jackson, S., … Harris, G. F. (2009). Upper extremity kinetics and energy expenditure during walker-assisted gait in children with cerebral palsy. *Acta Orthopaedica et Traumatologica Turcica*, *43*(2), 156–164. https://doi.org/10.3944/AOTT.2009.156

Krautwurst, B. K., Dreher, T., & Wolf, S. I. (2016a). The impact of walking devices on kinematics in patients with spastic bilateral cerebral palsy. *Gait & Posture*, *46*, 184–187. https://doi.org/10.1016/j.gaitpost.2016.03.014

Krautwurst, B. K., Dreher, T., & Wolf, S. I. (2016b). The impact of walking devices on kinematics in patients with spastic bilateral cerebral palsy. *Gait & Posture*, *46*, 184–187. https://doi.org/10.1016/j.gaitpost.2016.03.014

Lephart, K., Utsey, C., Wild, D. L., & Fisher, S. R. (2014). Estimating Energy Expenditure for Different Assistive Devices in the School Setting. *Pediatric Physical Therapy*, *26*(3), 354–359. https://doi.org/10.1097/PEP.0000000000000047

Logan, L., Byers-Hinkley, K., & Ciccone, C. D. (1990). Anterior Versus Posterior Walkers: A Gait Analysis Study. *Developmental Medicine & Child Neurology*, *32*(12), 1044–1048. Retrieved from http://doi.wiley.com/10.1111/j.1469-8749.1990.tb08521.x

Logan, L., Byers-Hinkley, K., & Ciccone, C. D. (1990). Anterior Versus Posterior Walkers: A Gait Analysis Study. *Developmental Medicine & Child Neurology*, *32*(12), 1044–1048. https://doi.org/10.1111/j.1469-8749.1990.tb08521.x

Norman, J. F., Bossman, S., Gardner, P., & Moen, C. (2004). Comparison of the Energy Expenditure Index and Oxygen Consumption Index During Self-Paced Walking in Children with Spastic Diplegia Cerebral Palsy and Children Without Physical Disabilities. *Pediatric Physical Therapy*, *16*(4), 206–211. https://doi.org/10.1097/01.PEP.0000145930.84009.23

Palisano, R. J., Rosenbaum, P., Bartlett, D., & Livingston, M. H. (2008). Content validity of the expanded and revised Gross Motor Function Classification System. *Developmental Medicine & Child Neurology*, *50*(10), 744–750. https://doi.org/10.1111/j.1469-8749.2008.03089.x

Park, E. S., Park, C. I., & Kim, J. Y. (2001). Comparison of anterior and posterior walkers with respect to gait parameters and energy expenditure of children with spastic diplegic cerebral palsy. *Yonsei Medical Journal*, *42*(2), 180–184. https://doi.org/10.3349/ymj.2001.42.2.180

Pataky, T. C. (2010). Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. *Journal of Biomechanics*. https://doi.org/10.1016/j.jbiomech.2010.03.008

Pataky, T. C. (2012). One-dimensional statistical parametric mapping in Python. *Computer Methods in Biomechanics and Biomedical Engineering*, *15*(3), 295–301. https://doi.org/10.1080/10255842.2010.527837

Pataky, T. C., Robinson, M. A., & Vanrenterghem, J. (2016). Region-of-interest analyses of one-dimensional biomechanical trajectories: bridging 0D and 1D theory, augmenting statistical power. *PeerJ*, *4*, e2652. https://doi.org/10.7717/peerj.2652

Pauk, J., Ihnatouski, M., Daunoraviciene, K., Laskhousky, U., & Griskevicius, J. (2016). Research of the spatial-temporal gait parameters and pressure characteristic in spastic diplegia children. *Acta of Bioengineering and Biomechanics*, *18*(2), 121–129.

https://doi.org/10.5277/ABB-00393-2015-02

Paul, S. M., Siegel, K. L., Malley, J., & Jaeger, R. J. (2007). Evaluating interventions to improve gait in cerebral palsy: a meta-analysis of spatiotemporal measures. *Developmental Medicine & Child Neurology*, *49*(7), 542–549. https://doi.org/10.1111/j.1469-8749.2007.00542.x

Pontonnier, C., & Dumont, G. (2009). Inverse dynamics method using optimization techniques for the estimation of muscles forces involved in the elbow motion. *International Journal on Interactive Design and Manufacturing (IJIDeM)*, *3*(4), 227–236. https://doi.org/10.1007/s12008-009-0078-4

Prosser, L. A., Lauer, R. T., VanSant, A. F., Barbe, M. F., & Lee, S. C. K. (2010). Variability and symmetry of gait in early walkers with and without bilateral cerebral palsy. *Gait & Posture*, *31*(4), 522–526. https://doi.org/10.1016/j.gaitpost.2010.03.001

Protas, E. J., Raines, M. L., & Tissier, S. (2007). Comparison of Spatiotemporal and Energy Cost of the Use of 3 Different Walkers and Unassisted Walking in Older Adults. *Archives of Physical Medicine and Rehabilitation*, *88*(6), 768–773. https://doi.org/10.1016/j.apmr.2007.03.001

Raja, K., Joseph, B., Benjamin, S., Minocha, V., & Rana, B. (2007). Physiological cost index in cerebral palsy: its role in evaluating the efficiency of ambulation. *Journal of Pediatric Orthopedics*, *27*(2), 130–136. https://doi.org/10.1097/01.bpb.0000242440.96434.26

Reddihough, D. S., & Collins, K. J. (2003). The epidemiology and causes of cerebral palsy. *Australian Journal of Physiotherapy*, *49*(1), 7–12. https://doi.org/10.1016/S0004-9514(14)60183-5

Richards, C. L., & Malouin, F. (2013). Cerebral palsy. In *Pediatric Neurology Part I* (Vol. 111, pp. 183–195). https://doi.org/10.1016/B978-0-444-52891-9.00018-X

Ries, A. J., & Schwartz, M. H. (2018). Low gait efficiency is the primary reason for the increased metabolic demand during gait in children with cerebral palsy. *Human Movement Science*, *57*, 426–433. https://doi.org/10.1016/j.humov.2017.10.004

Rodda, J., & Graham, H. K. (2001). Classification of gait patterns in spastic hemiplegia and spastic diplegia: a basis for a management algorithm. *European Journal of Neurology*, *8*(s5), 98–108. https://doi.org/10.1046/j.1468-1331.2001.00042.x

Rosenbaum, P., Paneth, N., Leviton, A., Goldstein, M., Bax, M., Damiano, D., … Jacobsson, B. (2007). A report: the definition and classification of cerebral palsy April 2006. *Developmental Medicine & Child Neurology*, *49*(SUPPL.109), 8–14. https://doi.org/10.1111/j.1469-8749.2007.tb12610.x

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011a). The gait of children with and without cerebral palsy: work, energy, and angular momentum. *J Appl Biomech*, *27*(2), 99–107. https://doi.org/10.1123/jab.27.2.99

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011b). The gait of children with and without cerebral palsy: Work, energy, and angular momentum. *Journal of Applied Biomechanics*, *27*(2), 99–107.

Russell, S. D., Bennett, B. C., Kerrigan, D. C., & Abel, M. F. (2007). Determinants of gait as applied to children with cerebral palsy. *Gait and Posture*, *26*(2), 295–300. https://doi.org/10.1016/j.gaitpost.2006.09.079

Schwartz, M. H., Rozumalski, A., & Trost, J. P. (2008). The effect of walking speed on the gait of typically developing children. *Journal of Biomechanics*, *41*(8), 1639–1650. https://doi.org/10.1016/J.JBIOMECH.2008.03.015

Seth, A., Sherman, M., Reinbolt, J. A., & Delp, S. L. (2011). OpenSim: a musculoskeletal modeling and simulation framework for in silico investigations and exchange. *Procedia IUTAM*, *2*, 212–232. https://doi.org/10.1016/j.piutam.2011.04.021

Stambolian, D., Asfour, S., & Eltoukhy, M. (2014). Using vicon bodybuilder and plug-in-gait to generate L5/S1 angles, forces and moments. In *2014 IEEE Aerospace Conference* (pp. 1–7). IEEE. https://doi.org/10.1109/AERO.2014.6836487

Thys, H., Willems, P. A., & Saels, P. (1996). Energy Cost, Mechanical Work and Muscular Efficiency in Swing-Through Gait with Elbow Crutches. *Journal of Biomechanics*, *29*(11), 1473–1482. Retrieved from https://ac.els-cdn.com/002192909684543X/1-s2.0-002192909684543X-main.pdf?_tid=c556cdc2-c711-4ddb-8055-89990e0db1fe&acdnat=1520428515_718582a0cd09299825d02b5e9e888582

Todorov, E., Erez, T., & Tassa, Y. (2012). MuJoCo: A physics engine for model-based control. In *2012 IEEE/RSJ International Conference on Intelligent Robots and Systems* (pp. 5026–5033). IEEE. https://doi.org/10.1109/IROS.2012.6386109

Unnithan, V. B., Dowling, J. J., Frost, G., & Bar-Or, O. (1996). Role of cocontraction in the O2 cost of walking in children with cerebral palsy. *Medicine and Science in Sports and Exercise*, *28*(12), 1498–1504. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8970144

Van de Walle, P., Hallemans, A., Schwartz, M., Truijen, S., Gosselink, R., & Desloovere, K. (2012). Mechanical energy estimation during walking: Validity and sensitivity in typical gait and in children with cerebral palsy. *Gait & Posture*, *35*(2), 231–237. https://doi.org/10.1016/j.gaitpost.2011.09.012

van den Hecke, A., Malghem, C., Renders, A., Detrembleur, C., Palumbo, S., & Lejeune, T. M. (2007). Mechanical Work, Energetic Cost, and Gait Efficiency in Children With Cerebral Palsy. *Journal of Pediatric Orthopaedics*, *27*(6), 643–647. https://doi.org/10.1097/BPO.0b013e318093f4c3

Winter, S., Autry, A., Boyle, C., & Yeargin-Allsopp, M. (2002). Trends in the Prevalence of Cerebral Palsy in a Population-Based Study. *Pediatrics*, *110*(6), 1220–1225. https://doi.org/10.1542/peds.110.6.1220

Yeargin-Allsopp, M., Van Naarden Braun, K., Doernberg, N. S., Benedict, R. E., Kirby, R. S., & Durkin, M. S. (2008). Prevalence of cerebral palsy in 8-year-old children in three areas of the United States in 2002: a multisite collaboration. *Pediatrics*, *121*(3), 547–554. https://doi.org/10.1542/peds.2007-1270

Yepremian, D., Desailly, E., Lejeune, L., Hareb, F., Jarrige, M., Bouchakour, D., & Khouri, N.

(2009). The effects of walkers on lower limb kinematics of children with cerebral palsy. *Gait & Posture*, *30*, S90. https://doi.org/10.1016/j.gaitpost.2009.08.134

Appendix A: Walker Model

The posterior walker model (Figure A.1) was designed using AutoCAD software to be able to ensure the elements were made to scale and the parts had accurate weights and material properties. The walker was a modified Nurmi Neo Posterior Pediatric Walker made by Otto-Bock (Figure A.2). Individual parts were created and then connected with necessary joints to allow for flexibility in how the walker was adjusted for each subject. Revolute joints were made for all four wheels and the front wheel casters. These joints allowed the walker to move as it did in the trial. Defined contact between the floor and the wheels created friction which drove the wheels to rotate.

The walker was modified for each subject’s use and the model incorporated these changes. The walker was adjusted for each subject so that the handles were aligned with their ulnar styloid process of the wrist. Translational joints were created in the model for the height adjustors on the posterior frame and for the connection between the top bar and the posterior frame. The translational joints were free to move for the initial simulation to the first frame of the trial so that the height and position of the walker were accurate for the trial. After the walker reached the initial position and was checked to be correct the translational joints were fixed. For taller subjects the handles of the walker were flipped vertically to align correctly with the wrist. Revolute joints were created for the handles of the walker. For all subjects, the handles were either down or up, and the joints were updated in the model to reflect their position in the trial and locked in that position. This ensured the correct application of the force from the load cells to the subject hand center of mass.

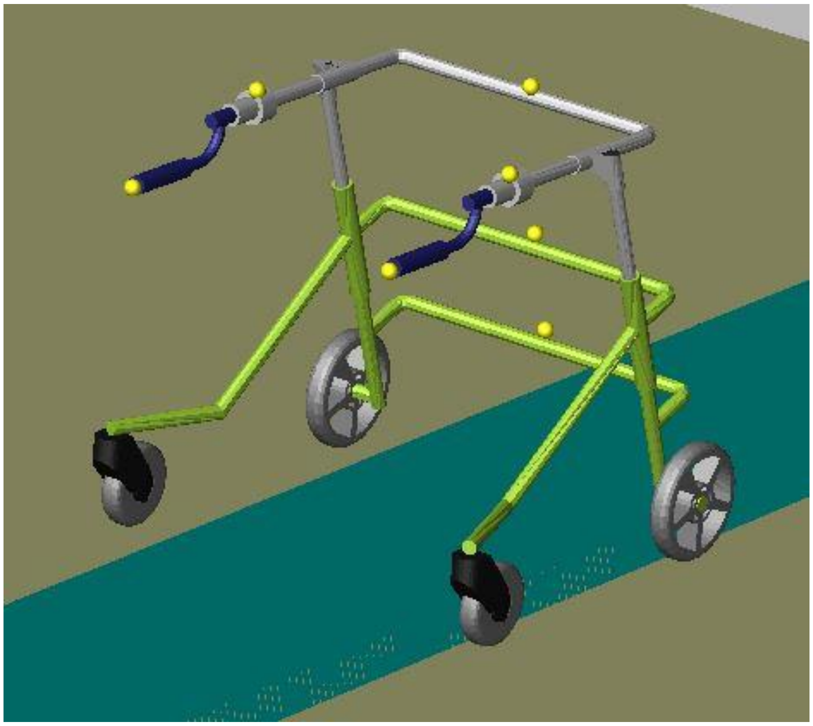

Figure A.1: Walker model in MSC Adams with kinematic markers used to drive the motion of the walker in simulations.

Figure A.2: Modified Nurmi Neo Posterior Pediatric Walker used for all manual walker trials with attached load cells for collection of handle reaction force data.

The computational models for each trial were constructed and simulations performed using the following protocol in MSC.ADAMS with the LifeMod plugin.

Import Body and Attach Recording Joints (once for each subject)

1. Put the Static C3D and SLF file for the body in the static trial folder
2. Open MSC.ADAMS, in the upper right dialog box:
    a. First dropdown → '_Xchange'
    b. Sceond dropdown → 'Import SLF Model'
    c. Right click in gray box → 'Browse' → Select static SLF file of the body
    d. Uncheck box for 'Ground Reaction Force'
    e. Click Apply (bottom right of window) and allow to process
3. In upper right dialog box:
    a. First dropdown → 'Joints'
    b. Second dropdown → 'Create Base Joint Set'
    c. Click light bulb next to 'Prepare Model with Recording Joints'
    d. Click 'Select All'
    e. Click Apply, wait for teal-colored balls to appear at joints of model
4. Save the Body model
    a. File (top left of GUI) → Save database as → 00_Body

Make Static Model (once for each subject)

1. Import initial static kinematic position of subject in the Static trial
    a. First dropdown → 'Motion'
    b. Second dropdown → 'Import Motion Capture Data'
    c. Right click gray box → Browse → Select the static body SLF
    d. Data prefix → type 'static'
    e. Apply
2. Create spring-damper connections between motion capture and model elements
    a. Second dropdown → 'Create Base Motion Agent Set'
    b. Set marker set (dropdown in top middle) → 'Plug in Gait Marker Set'
    c. Data prefix → type 'static'
    d. At top of GUI, SDR → Baseball → Skeleton → Set MA weights
    e. Apply
3. Move model to static position
    a. First dropdown → '_Analyze'
    b. Second dropdown → 'Dynamics' (should already be here)
    c. Check box for 'Freeze Motion Agents for Equilibrium Analysis'
    d. At top of GUI, SDR → Baseball → Ankle Constraints → Lock Ankles
    e. Click 'Analyze' button, left of stop sign in dialog box
4. Save the file
    a. Check model for errors in positioning

   b. Click ‘Update Model Posture with Equilibrium Results’
   c. Click ‘Synchronize Body Marker Location with Data Locations’
   d. At top of GUI, SDR → Baseball → Ankle Constraints → Unlock Ankles
   e. File → Save database as → Name it ‘00_Static’
5. Create left- and right-facing static files to match direction of gait for trials
   a. Define hinge joint at center of mass of model to rotate it
      i. In file explorer, find .cmd file for ‘rotate_xxxx’ and edit in notepad
      ii. Ctrl-F to change the subject identifier to the correct subject ID for current model
      iii. Save as, ‘rotate_xxxx.cmd’
      iv. In MSC.ADAMS, File → Import → Choose ‘rotate_xxxx.cmd’
   b. Drive motion of hinge joint to rotate body
      i. Right click hinge joint in GUI
      ii. Menu: ‘Joint: rotateStatic’ → Modify
      iii. Click ‘Impose Motion(s)’
      iv. Change Rot Z” dropdown to ‘disp(time)=’
      v. In function box, type ‘pi/3 * time’
      vi. In top right dialog box, First dropdown → ‘_Analyze’
      vii. Second dropdown → ‘Dynamics’ (should already be here)
      viii. Check box for ‘Disable Motion Agents’
      ix. Click ‘Analyze’
      x. Click ‘Update Model Posture with Equilibrium Results’
      xi. Delete hinge joint, right click joint
      xii. Menu: ‘Joint: rotateStatic’ → Delete, delete all dependents
   c. File → Save database as → 00_Static_R2L
   d. Repeat steps (a)-(c) from ‘00_Static’
      i. Change ‘pi/3’ to ‘–pi/3’
      ii. Save resulting database as 00_Static_L2R
6. Copy the corresponding static file into all trial folders based on direction of walk for each trial.

*** = <u>Walker Model only, not performed for baseline trials</u>
General organization for folders follows: Home → Models → (Subject) → (Trial)
→ Code → Modeling
Models are made in Subject → Trial folders, corresponding code used for model creation is found in the Code → Modeling folder.

Make Inverse Model for Trial

1. Each trial folder should contain trial data, static model files, and MSC.ADAMS shortcut with ‘aviewAS’ and ‘aviewBS’ cmd files
2. In MSC.ADAMS, File → Open Database (Don’t save changes to SDR_Utility) → select ‘00_Static_L2R’ or ‘00_Static_R2L’, whichever applies for this trial
3. *** Load Static Walker and floor for model
   a. File → Import
   b. File type dropdown → Adams/View Command File (*.cmd)

    c. Right click 'File to Read' → Browse → select '01_Create Ottobock Walker – No Motor.cmd' in Code → Modeling folder
    d. Click 'Ok'

4. *** Update position for handles of walker
    a. Check if handles are up or down in trial, if up then do step (b) otherwise skip to (5)
    b. Drive motion of hinge joints on handles
        i. Right click left and right handle joints
        ii. Click modify
        iii. Click 'Impose Motion'
        iv. Drive displacement of joint to be 'pi * time'
        v. In top right dialog box, check box to 'Disable Motion Agents'
        vi. Run analysis for one second
        vii. Click 'Update Model Posture with Equilibrium Results'
5. *** Fix Top Bar of Walker as adjustment for this portion is done
    a. File → Import
    b. File Type → Adams/View Command File (*.cmd)
    c. Right click 'File to Read' → Browse select '02_Fix Top Bar.cmd' in Code → Modeling folder
    d. Click 'Ok'
    e. Click 'Delete All' if an error is thrown (this will happen twice or not at all depending on if the handle orientation was changed in step 4)
6. *** Import Static Data for Walker
    a. File → Import
    b. File Type → Test Data
    c. Select 'Create Splines'
    d. Right click 'File to Read' → Browse → select TXT file of static walker data
    e. Independent column index → '1'
    f. Click 'Ok'
7. *** Make walker kinematic markers
    a. File → Import
    b. File type → Adams/View Command File (*.cmd)
    c. Right click 'File to Read' → Browse → select '03_makeWalkerPoints_newVicon.cmd' in Code → Modeling folder
    d. Click 'Ok'
8. *** Make spring connections from kinematic markers to walker model
    a. File → Import
    b. File type → Adams/View Command File (*.cmd)
    c. Right click 'File to Read' → Browse → select '04_PointForceSpring.cmd' in Code → Modeling folder
    d. Click 'Ok'
9. Import Motion Capture Data for Body
    a. First dropdown → 'Motion'
    b. Second dropdown → 'Import Motion Capture Data'
    c. Right click gray box → Browse → Select the body marker SLF
    d. Data prefix → type 'body'

   e. Apply
   f. Wait for Status to display: 'Data Set Read Complete'
10. Create spring damper connection from kinematic markers to body model
   a. Second dropdown → 'Create Base Motion Agent Set'
   b. At top of GUI, SDR → Baseball → Skeleton → Number Markers
   c. SDR → Baseball → Skeleton → Set MA weights
   d. Check notes for trial to see if any spring-damper weights should be set to 0 because they were missing during the trial
   e. Set marker set (dropdown in top middle) → 'Plug in Gait Marker Set'
   f. Data prefix → type 'body'
   g. Apply
      i. Should yield red dots and two sets of yellow dots in GUI
   h. SDR → Baseball → Skeleton → Sync New MA
11. Perform Simulation to bring body, and walker if applicable, to initial frame of trial
   a. First dropdown → '_Analyze'
   b. SDR → Baseball → Skeleton → Clear Old MA
      i. Should see one set each of red and yellow dots now
   c. Check box for 'Freeze Motion Agents for Equilibrium Analysis'
   d. Click 'Analyze'
   e. Check for correct kinematic location of body and walker
   f. Click 'Update Model Posture with Equilibrium Results'
12. *** Delete walker spring connections
   a. File → Import
   b. File type → Adams/View Command File (*.cmd)
   c. Right click 'File to Read' → Browse → select '04_PostStatic_Test.cmd' in Code → Modeling folder
   d. Click 'Ok'
13. File → Save database as → '01_*trial*_FirstFrame.bin'
   a. *Trial* = trial name
   b. *Ex: '01_UVAWalker01_FirstFrame.bin'*
14. Perform inverse kinematic simulation for body
   a. First dropdown → '_Analyze'
   b. SDR → Baseball → Skeleton → Clear Old MA
   c. Uncheck box for 'Freeze Motion Agents for Equilibrium Analysis'
   d. Click 'Set End Time to Motion Data End'
   e. Click 'Analyze'
   f. Watch simulation to see if it worked
      i. Click icon, , in the main tool box (left)
      ii. Click play forward to watch trial
      iii. This analysis is just for body so do not worry about the walker if it is present
      iv. Check to make sure the body kinematics look correct
   g. Click 'Save Analysis' in upper right dialog box
      i. Name → inv
      ii. Click 'Ok'
   h. File → Save database as → '01_*trial*_Inv.bin'

Make Dynamic Body Models

1. Continue from database from step 14
2. Load floor if not done already (Walker trials can skip this step)
    a. File → Import
    b. File type → Adams/View Command File (*.cmd)
    c. Right click 'File to Read' → Browse → select 'newLabFloor.cmd in Code → Modeling folder
    d. Click 'Ok'
3. Load ground reaction force plate data
    a. File → Import
    b. File type → Test Data
    c. Select 'Create Splines'
    d. Right click gray box → Browse → select TXT file of force plate data
    e. Independent column index → '1'
    f. Select 'Ok'
4. Apply ground reaction forces
    a. File → Import
    b. File type → Adams/View Command File (*.cmd)
    c. Right click 'File to Read' → Browse → select trial-specific cmd file for FP application
    d. Click 'Ok'
        i. In GUI, red arrows for moments and forces should appear on ankles of subject
5. Create PD controllers for joints based on joint kinematics from inverse simulation
    a. First dropdown → 'Joints'
    b. Second dropdown → 'Training'
    c. Click the lightbulb in the 'Forward Dynamics' row
    d. Click 'Apply'
        i. Joints should turn dark blue/purple
6. Create COM Tracker Force to make COM follow inverse kinematic motion
    a. First dropdown → 'Motion'
    b. Second dropdown → 'Create Tracker Agent'
    c. Check box for 'Create Track Force Data Request'
    d. Add two zeroes to both stiffness and both damping values
    e. All '_dof' should be set to driven
    f. Click 'Apply'
        i. Finished when Status reads 'Synchronize MA Locations'
7. Perform dynamic simulation for body
    a. First dropdown → '_Analyze'
    b. Check box for gravity → click '-Y' button
    c. Click 'Set End Time to Motion Data End'
    d. Check box for 'Disable Motion Agents'
    e. Click 'Analyze'
    f. Watch simulation to see if it worked

        i. Click icon, , in the main tool box (left)
        ii. Click play forward to watch trial
        iii. This analysis is just for body so do not worry about the walker if it is present
        iv. Check to make sure the body kinematics and kinetics look correct
    g. Save Analysis
        i. Name → fwd
        ii. Click ‘Ok’
8. File → Save database as → 02_*trial*_Fwd.bin
9. Export Data
    a. Press F3 to open command window (sometimes it takes two presses)
    b. In command line, type: var cre var=.world.trial str=‘trial’
        i. *Trial* is the current trial, eg UVAWalker01
    c. Got to PostProcessor View. Click the icon, , in main tool box (left)
    d. File → Import → Command File
    e. Right click ‘File to Read’ → Browse → select ‘PlotAngleTorqueForces_fixedDirections_fwd.cmd’ in Code → Adams Post-Processing folder
10. File → Save database
    a. Don’t create a backup copy

Make Dynamic Walker Models

1. ***Import walker motion capture data
    a. File → Import
    b. File type → Test Data
    c. Select ‘Create Splines’
    d. Right click ‘File to Read’ → Browse → select TXT file of moving walker data
    e. Independent Column Index → ‘1’
    f. Click ‘Ok’
2. ***Create walker kinematic markers in model
    a. File → Import
    b. File type → Adams/View Command File (*.cmd)
    c. Right click ‘File to Read’ → Browse → select ‘WalkerMotionWalking.cmd’
    d. Click ‘Ok’
3. ***Make forces to drive motion of walker
    a. File → Import
    b. File type → Adams/View Command File (*.cmd)
    c. Right click ‘File to Read’ → Browse → select ‘MakeBushingMovements.cmd’
    d. Click ‘Ok’
4. ***Load walker handle reaction force data
    a. File → Import
    b. File type → Test Data
    c. Select ‘Create Splines’
    d. Right click ‘File to Read’ → Browse → select TXT file of load cell data
    e. Independent Column Index → ‘1’

   f. Click ‘Ok’
5. ***Apply walker handle reaction force data to model
   a. File → Import
   b. File type → Adams/View Command File (*.cmd)
   c. Right click ‘File to Read’ → Browse → select trial-specific load cell cmd file
   d. Click ‘Ok’
      i. To view the handle reaction forces it will help to increase the transparency of the hands to ~70
      ii. To do this, right click the ‘Part: Right Hand’ and select ‘Appearance’, then modify ‘Transparency’
6. ***Run dynamic walker model simulation
   a. First dropdown → ‘_Analyze’
   b. Check box for gravity → click ‘-Y’ button
   c. Click ‘Set End Time to Motion Data End’
   d. Check box for ‘Disable Motion Agents’
   e. Click ‘Analyze’
   f. Watch simulation to see if it worked
      i. Click icon, , in the main tool box (left)
      ii. Click play forward to watch trial
      iii. Check to make sure the body and walker kinematics and kinetics look correct
   g. Save Analysis
      i. Name → fwdWalker
      ii. Click ‘Ok’
7. ***File → Save database as → 02_*trial*_FwdWalker.bin
8. ***Export Data
   a. Got to PostProcessor View. Click the icon, , in main tool box (left)
   b. File → Import → Command File
   c. Right click ‘File to Read’ → Browse → select ‘PlotAngleTorqueForces_fixedDirections_fwdWalker.cmd’ in Code → Adams Post-Processing folder
9. ***File → Save database
   a. Don’t create a backup copy

Repeat for all trials until done

Appendix B: Future Considerations

Automated-Walker Assisted Gait in Children with and without Cerebral Palsy: Kinematics, Kinetics, and Work

B.1 Introduction

Cerebral Palsy (CP) is a non-progressive motor development disorder initiated by a lesion to the brain around the time of birth. CP inhibits the growth and function of ambulatory muscles in the lower legs which causes different visual gait patterns depending on how it manifests itself in the patient. Bipedal walking at a self-selected speed is typically characterized by getting from one position to the next in an efficient manner. For persons with CP, gait efficiency is significantly lower than their typically developed peers due to excess motion used for balance and cocontraction of muscles (Ries & Schwartz, 2018; S. Russell et al., 2011a). This excess energy cost is important for clinicians to keep in mind as they consider interventions to assist with gait. While CP is non-progressive, children with CP are observed as they grow because their muscles do not grow linearly with their bones. This can cause further motor impairment and make ambulation more inefficient. Clinical intervention often includes the prescription of the use of an assistive device which acts to stabilize the user during walking, like a posterior walker.

Many studies have shown that the gait of children with CP requires more energy use compared to their typically developed peers through oxygen consumption and heart rate measures (Norman et al., 2004; Unnithan et al., 1996). Studies have also shown that as children with CP age their energy use increases (Bolster et al., 2017b). These estimates of energy cost on the individual cannot correlate an increase in cost to a specific time in the gait cycle as they are calculated for whole periods of activity. Mechanical energy metrics can be applied to the whole body while still providing individual time points of data for different segments of the body. In pediatric CP gait mechanical work is higher than in the gait of their typically developed peers (Van de Walle et al., 2012; van den Hecke et al., 2007). For mechanical energy outcomes to be accurate they must incorporate the entire body for analysis otherwise the minimum energetic cost would be biased by the excluded segments.

A motorized walker has the potential to decrease the work required by the user for their gait, while still providing the positive outcomes of stability and support for the CP population. It is important to decrease the cost of the walker on the body but not to entirely remove the energetic cost of gait. Children with CP need to continue walking and training their muscles to avoid muscular atrophy, which would result in them using a wheelchair later in life. The control for motorizing a walker used in this work used an infrared sensor to calculate the distance of the user from the back of the walker. As the subject moved from their initialized position, torque was applied to the wheels to keep the subject's distance from the walker the same or less than the initialized position. In preparing the control, TD subjects walked with the walker and the control was tuned to where the subject could walk without pulling the walker along and steering it when necessary.

Additionally, the interaction of assistive devices and a child with CP has not been given a lot of attention in literature. Previous work examining walker use made mechanical energy calculations for upper body segments of the body and have not looked at the whole body. One study by Konop et al. (Konop et al., 2009) looked at upper body kinetics during walker use. They found preliminary data suggesting a relationship between heart rate and the force used to move the walker forward. Other studies have only looked at the kinematics of the gait and did not instrument the device to model kinetics (Krautwurst et al., 2016b; Paul et al., 2007; Protas et al., 2007). This work calculated whole-body mechanical joint work over the gait cycle. Calculations were made for the exclusion and inclusion of the walker interaction to be able to define the importance of including the assistive device reaction forces in the kinetic model. It was hypothesized that the total body mechanical work would show similar trends as previously published metabolic data and that the incorporation of the walker would increase the accuracy of the model outcomes.

B.2 Methods

B.2.1 Automated Control of a Walker

A pediatric posterior walker was instrumented with an infrared (IR) sensor that detected the distance of the torso away from the back of the walker. The control mechanism for accelerating the walker used the output from the IR sensor to apply torque to the wheels to move the walker as the user moved forward. Through experimentation with the weights of the PID control in TD subjects the walker followed along with a subjects' gait without noticeable need for pulling and only required steering in the event of a turn.

B.2.2 Subjects and Model

The anthropometric data for the same 17 children collected in the Motion Analysis and Motor Performance laboratory at the University of Virginia from Chapter 3 was used. Data exported from Vicon Nexus software was filtered and prepared in Matlab then modelled in MSC.Adams with the LifeMod plugin. A model with 19 segments and 18 joints was created for each subject based on individual anthropometric data. Trials for walking with a manual and motorized posterior pediatric walker and were modelled. Inverse dynamic simulations were performed and joint kinematics and kinetics were exported as model outputs. Individual outputs from walker trials were exported for analysis.

Kinetic analysis was performed on four subjects in the CP group. The motorized walker had different handles in earlier trials and the forces at the handles of the walker were not collected for those subjects.

B.2.3 Spatiotemporal Analysis

Spatiotemporal analysis was conducted for trial outputs as presented in previous work from kinematic data (Johnson et al., 1997; Ju Kim & min Son, 2014; Pauk et al., 2016). Stride length for the left and right leg stride were calculated as the distance travelled by the respective foot between the first and second heel strike of the same foot. The two locations of the foot were calculated as the average of the ankle, heel, and toe markers at the heel strike time points. Stride length was normalized by the length of the respective leg so that the results could be compared across different subjects. Step length for the left and right step was calculated as the distance from the contralateral to the ipsilateral foot at heel strikes of the respective feet. The location of the feet was the average of the three foot markers on the foot. Step length was normalized by the length of the respective leg for which the step was taken.

Stride time was calculated as the period from initial heel strike to terminal heel strike and reported in seconds (Figure B.1). The cadence was calculated as the inverse of the stride time and converted to steps per minute. Velocity was calculated by dividing the non-normalized stride length by the stride time.

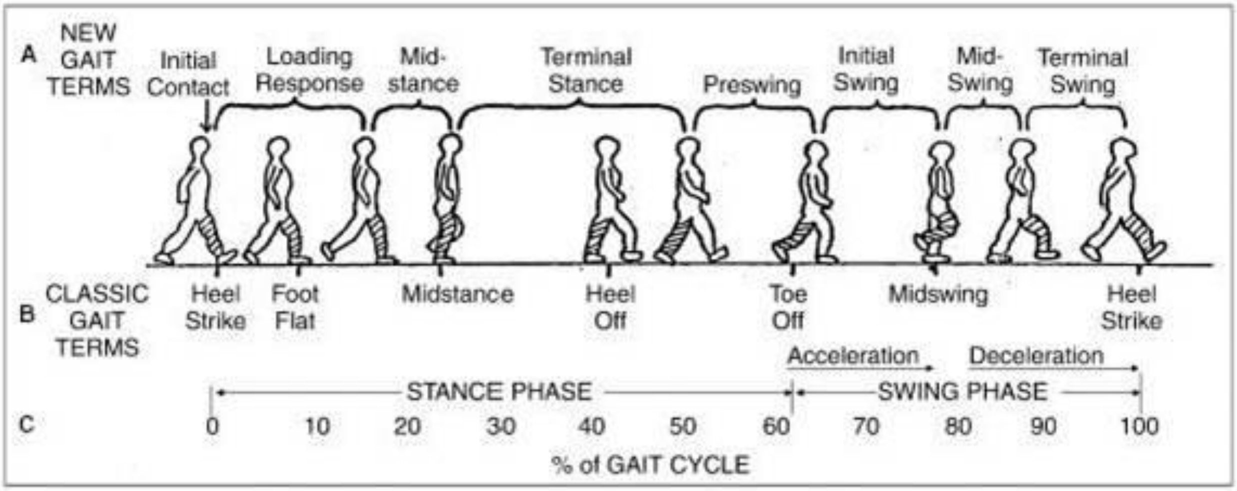

Figure B.1: Gait cycle events and terminology.

Contralateral foot off and contact and ipsilateral toe off events were calculated as percentage of gait cycle from initial to terminal heel strike. Two values for each parameter were generated for each trial as left and right strides were used for this analysis. Single and double support time for both left and right strides were calculated using events labelled from force plate data. Single support time was calculated from adding the time period between contralateral toe off and contralateral foot strike and the time period between ipsilateral toe off and ipsilateral foot strike. Double support time was calculated by adding the time period between initial foot strike and contralateral toe off and the time period between contralateral heel strike and ipsilateral toe off. Single and double support were calculated as percentage of the gait cycle by dividing the time of support by the respective stride time for the left and right stride.

Parameters were compared using statistical t-tests where the samples were assumed to have equal variance for comparing within groups and unequal variances for comparisons between groups.

B.2.4 Reaction Forces

The ground and walker handle reaction forces were recorded for walker-assisted gait trials with the manual and control-automated walkers. Posterior walkers reduce the load on the lower body joints in populations which rely on the walker for ambulation (Bateni & Maki, 2005; Fast et al., 1995). The GRF was examined so that the reliance on the walker could be quantified at different times over the gait cycle. A greater reliance on the walker would present as a lower magnitude GRF over the gait cycle. Handle reaction forces were examined to define the dependence on the walker and the applied force to move the assistive device by each group. The

CP group was expected to have a higher reliance on the walker and use more force to move the walker along through their gait cycle.

B.2.5 Total Mechanical Work

Gait efficiency is an important outcome measure in assessing the motor function of children with cerebral palsy (Bolster et al., 2017b; Norman et al., 2004). Mechanical work estimates the energy cost on an individual by quantifying the mechanical energy cost of movement. Using the model outputs, the mechanical work was calculated like that presented by Chen et al. (Chen et al., 1997). The mechanical work at the joint *j* is calculated as the sum over the gait cycle of the product of the joint torque (T_j) and joint angle (A_j):

$$W_j^T = \int_{gc} \left| T_j * A_j \right| dt$$

Mechanical work calculates the minimum work required for the system's movement. Cocontraction and metabolic costs are not included, each joint's contribution to the total cost is included. Individual joint totals were summed up over the gait cycle to calculate full body, lower body, and upper body mechanical work.

B.2.6 Joint Work

Calculating mechanical work of joints gives the ability to break the body down into individual segments and identify the joints with the greatest moments produced and the specific timepoints during the gait cycle where the work was greatest or least:

$$W_j = \left| T_j * A_j \right|$$

The body was split up into three parts: lower body joints including ankles, knees, and hips; upper body joints including wrists, elbows, shoulders, and scapulas; torso joints including lumbar, thoracic, and neck. The work generated at these segments was examined in relation to the gait cycle. The absolute value of the joint work was calculated so that when summing up into different segments there would be no cancellation between positive and negative values of work for different joints.

B.3 Results

B.3.1 Subjects

The same subject group used in earlier chapters was used for this study. Subject anthropometrics were recorded during collection. Two groups were created to compare outcomes within the groups of how they used the manual and automated walkers. One group was comprised of children with a diagnosis of cerebral palsy (CP) and another of typically developed (TD) peers as a control group. The automated walker was only instrumented for two of the TD group and four of the CP group. Kinetics were examined in these subjects while spatiotemporal and kinematic measures were examined in all subjects.

B.3.2 Spatiotemporal Gait Parameters

Spatiotemporal metrics of gait were compared between manual walker and automated walker gait within the TD and CP groups (Table B.1). In the TD group, automated walker gait was not significantly different than manual walker gait in spatiotemporal metrics. The TD group trended towards having shorter strides and lower velocity in automated walker gait, but no other differences were observed. The CP group was slightly more affected when walking with the automated walker compared to the manual walker. Shorter time in single support was observed but no other significance was found. Although not significant, changes in spatiotemporal metrics indicate the CP group walked better with the automated walker. Stride length remained constant from manual to automated walker gait while velocity and cadence increased. Percentage of time spent in double support increased as well.

Table B.1: Spatiotemporal Parameters. Significance within groups between manual and automated walker trials is marked as * (p<0.05), ** (p<0.01), *** (p<0.001).

Parameter Type	**TD**		**CP**	
Parameter	**Manual**	**Automated**	**Manual**	**Automated**
Spatial				
Stride Length (m)	1.17±0.17	1.13±0.14	0.88±0.09	0.88±0.09
Normalized Stride Length	1.39±0.12	1.35±0.12	1.15±0.14	1.15±0.12
Step Length (m)	0.58±0.08	0.55±0.07	0.42±0.04	0.43±0.04
Normalized Step Length	0.69±0.06	0.66±0.05	0.56±0.06	0.56±0.06
Velocity (m/s)	1.07±0.14	1.02±0.11	0.68±0.16	0.74±0.20
Temporal (gc = gait cycle)				
Single Support (% of gc)	77.3±3.28	77.6±3.85	72.7±6.24	71.3±1.04
Double Support (% of gc)	22.7±3.28	22.4±3.85	27.3±6.24	28.7±1.04
Stride Time (sec)	1.10±0.08	1.11±0.10	1.36±0.32	1.25±0.20
Cadence (step/min)	110.3±8.4	109.1±10.1	92.6±20.5	99.0±18.0
Single Support (sec)	0.85±0.05	0.86±0.06	0.98±0.21	0.89±0.15*
Double Support (sec)	0.25±0.05	0.25±0.06	0.38±0.14	0.36±0.05
Opposite Foot Off (% of gc)	11.2±1.78	11.0±1.97	13.8±3.26	14.2±0.79
Opposite Foot Contact (% of gc)	50.0±0.26	50.2±0.56	50.2±0.80	49.7±0.78
Toe Off (% of gc)	61.6±1.46	61.6±2.21	63.7±3.37	64.2±0.82

Splitting the CP group into subgroups yielded more significance between walker trials and revealed the effect of the automated walker on different users of the walker (Table B.2). The GMFCS II group walked faster and with more stability with the automated walker than the manual walker. The GMFCS II subgroup walked with a greater velocity and higher cadence with an automated walker and a greater percentage of time was spent in single support. The GMFCS III subgroup was less affected by the walker and walked with less stability in the automated walker. Percentage of time spent in double support was increased in the automated walker trials. Velocity and cadence of walk trended towards decreasing in automated walker gait for the GMFCS III subgroup. The crutches subgroup was a further division of the GMFCS III subgroup and was similarly affected by the automated walker compared to manual walker. Percentage of time in double support increased indicating a lack of stability in gait. The crutches subgroup walked better as velocity and cadence increased in automated walker trials.

Table B.2: Spatiotemporal Parameters for subgroups of CP group based on GMFCS level. Significance between manual and automated walker trials is marked as * (p<0.05), ** (p<0.01), *** (p<0.001).

Parameter Type	**GMFCS II (n=2)**		**GMFCS III (n=5)**		**Crutches (n=3)**	
Parameter	**Manual**	**Automated**	**Manual**	**Automated**	**Manual**	**Automated**
Spatial						
Stride Length (m)	0.89-0.99	0.92-1.03	0.86±0.10	0.85±0.06	0.90±0.08	0.86±0.09
Normalized Stride Length	1.17-1.23	1.09-1.43	1.14±0.17	1.11±0.03	1.16±0.20	1.11±0.09
Step Length (m)	0.42-0.48	0.44-0.50	0.41±0.04	0.41±0.03	0.43±0.04	0.42±0.06
Normalized Step Length	0.57-0.58	0.53-0.69	0.55±0.08	0.54±0.02	0.55±0.10	0.54±0.07
Velocity (m/s)	0.62-0.84	0.94-1.09	0.66±0.18	0.63±0.06	0.63±0.24	0.66±0.10
Temporal (gc = gait cycle)						
Single Support (% of gc)	61.0-71.6	70.8-71.4	75.2±3.99	71.4±1.25***	75.3±5.62	70.5±2.58**
Double Support (% of gc)	28.4-39.0	28.6-29.2	24.8±3.99	28.6±1.25***	24.7±5.62	29.5±2.58**
Stride Time (sec)	1.18-1.46	0.96-0.99	1.38±0.38	1.36±0.08	1.54±0.41	1.32±0.13*
Cadence (step/min)	83.5-102.0	121.9-126.9	92.6±24.3	88.8±5.5	83.1±25.3	92.0±9.2
Single Support (sec)	0.84-0.89	0.68-0.70	1.03±0.24	0.97±0.08	1.14±0.24	0.93±0.08**
Double Support (sec)	0.34-0.57	0.28-0.29	0.35±0.14	0.39±0.01	0.40±0.18	0.39±0.06
Opposite Foot Off (% of gc)	14.5-19.8	14.3-14.9	12.4±2.11	14.0±0.86*	12.4±2.89	14.5±3.70
Opposite Foot Contact (% of gc)	51.0-51.1	49.8-50.7	49.8±0.62	49.5±0.77	49.9±0.71	49.6±5.60
Toe Off (% of gc)	65.1-70.2	64.1-65.0	62.2±1.74	64.1±0.92**	62.2±2.44	64.5±2.53*

B.3.3 Reaction Forces

Ground reaction and walker handle reaction forces for the TD group in manual and automated walker trials were compared using SPM analysis (Figure B.2). Vertical ground reaction forces were greater in manual walker trials at the second peak approaching toe off. Vertical and longitudinal handle reaction forces were lower throughout the gait cycle for the TD group in automated walker trials.

CP group reaction forces were compared using SPM analysis between manual and automated walker trials (Figure B.3). The CP group walked with more ability and activation in their lower body in automated walker trials. Longitudinal ground reaction forces were greater over mid-stance in walker trials. The CP group interacted with the walker less in automated walker trials compared to manual trials. Vertical handle reaction forces were decreased throughout the gait cycle and longitudinal forces were lower in mid-stance and mid-swing in automated walker trials.

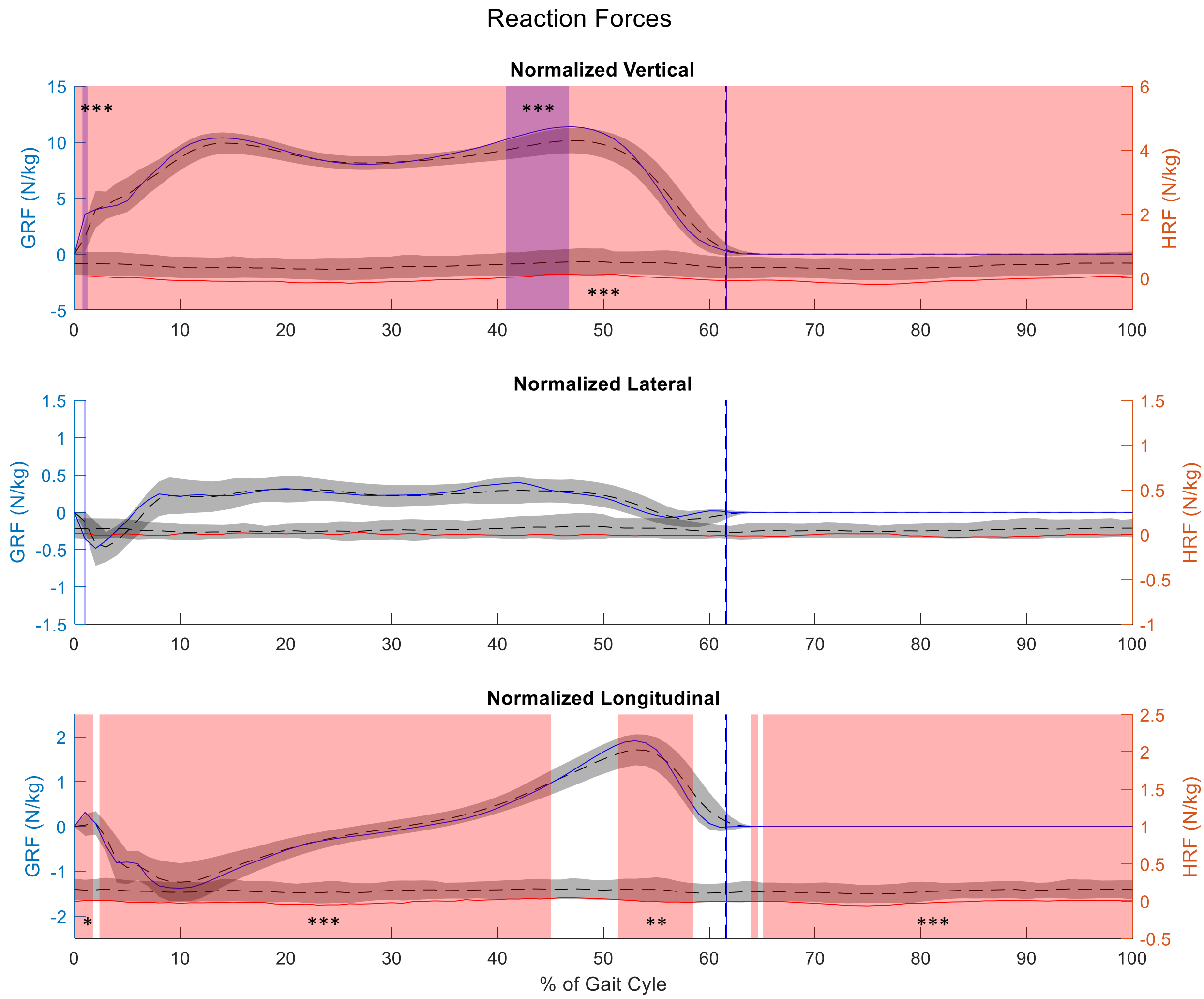

Figure B.2: TD group 3-dimensional ground and walker-handle reaction forces over the gait cycle. Significance between GRF manual (black) and automated (blue) walker-assisted trials highlighted by blue shaded boxes and HRF manual (black) and automated (blue) walker-assisted trials highlighted by red shaded boxes. SPM significance marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***) with GRF on top and HRF on bottom. Toe-off time marked by vertical lines for baseline (black) and walker-assisted (blue) trials.

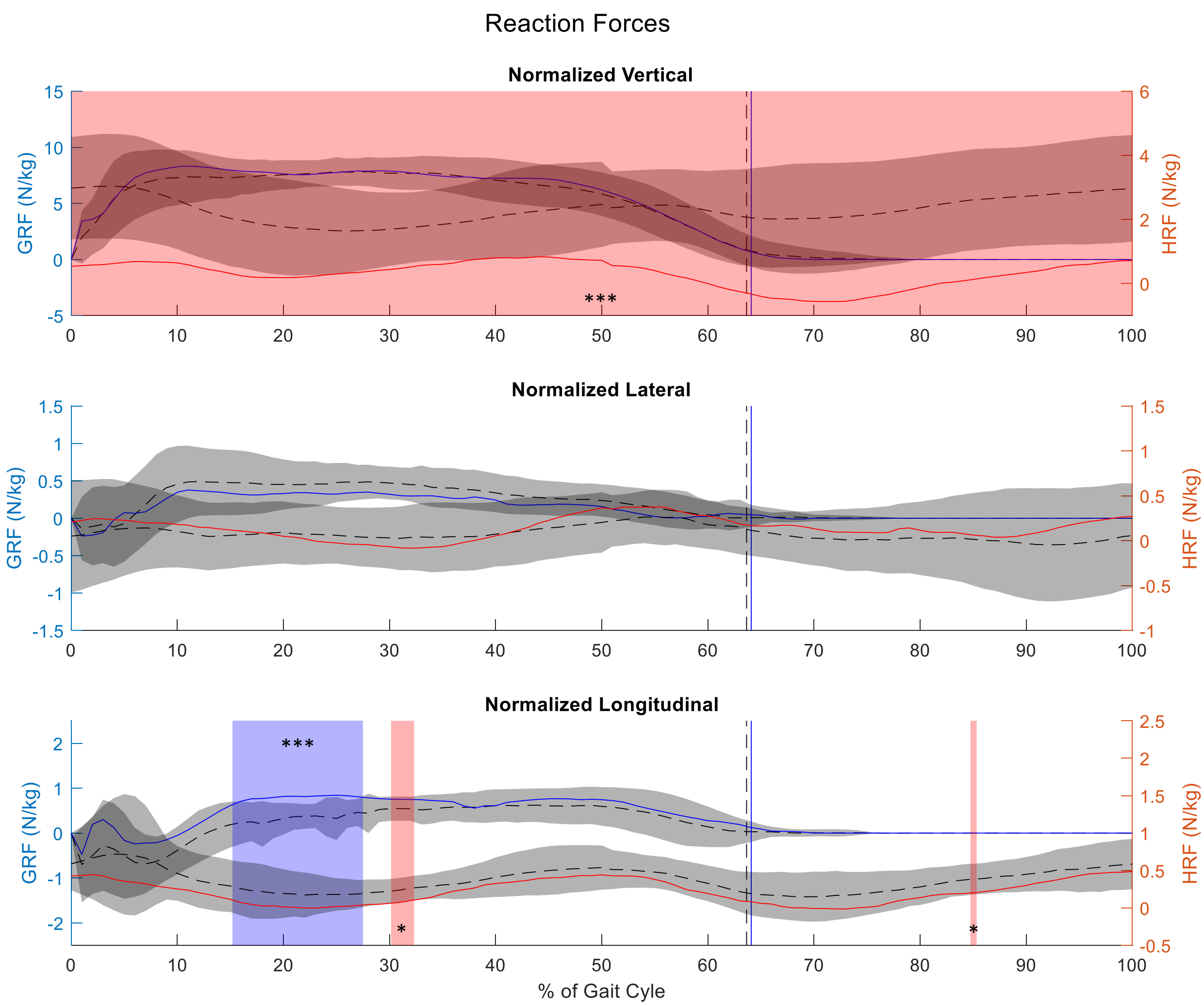

Figure B.3: CP group 3-dimensional ground and walker-handle reaction forces over the gait cycle. Significance between GRF manual (black) and automated (blue) walker-assisted trials highlighted by blue shaded boxes and HRF manual (black) and automated (blue) walker-assisted trials highlighted by red shaded boxes. SPM significance marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***) with GRF on top and HRF on bottom. Toe-off time marked by vertical lines for baseline (black) and walker-assisted (blue) trials.

B.3.4 Model Outputs

Dynamic models were created for three trials of each condition of walking for each subject. The model outputs included joint angles and moments for the defined joints of the model. TD and CP group manual and automated walker gait were compared to examine the effect of the automated control of the walker on the user's gait. The whole TD and CP groups were used in kinematic comparisons but only two TD and four CP subjects were used for joint kinetic comparisons. The automated walker was only instrumented for these two TD and four CP subjects to provide accurate kinetics to compare to manual walker trials. Within group differences were marked with shaded boxes on the figures where the automated walker trials were significantly different from manual walker-assisted gait. Pelvic segment tilt, obliquity, and rotation angles were included to show the inherent neutral angle of the model, as it was not equivalent to anatomical neutral. Pelvic tilt angles were offset from neutral by values between -10 and -18 degrees, while obliquity and rotation angles fluctuated around 0 degrees.

B.3.4.1 TD Group Manual vs Automated Walker-Assisted Gait

TD group joint kinematics for the whole group in automated walker trials were moderately different in the lower body (Figure B.4). The TD group postured themselves differently when walking with the automated walker. Greater flexion in the hip and knee at toe off reveal a decreased range of motion of these joints. Thoracic joint flexion was greater indicating a worse posture when walking with an automated walker (Figure B.5). The shoulder and elbow joints kinematics were characterized by greater abduction and greater pronation respectively throughout the gait cycle (Figure B.6). Joint kinetics were not significantly different between manual and automated walker trials for the TD group. Shoulder and elbow flexion moments were lower in automated trials which is likely a factor of the walker control decreasing the need for pulling forces on the walker.

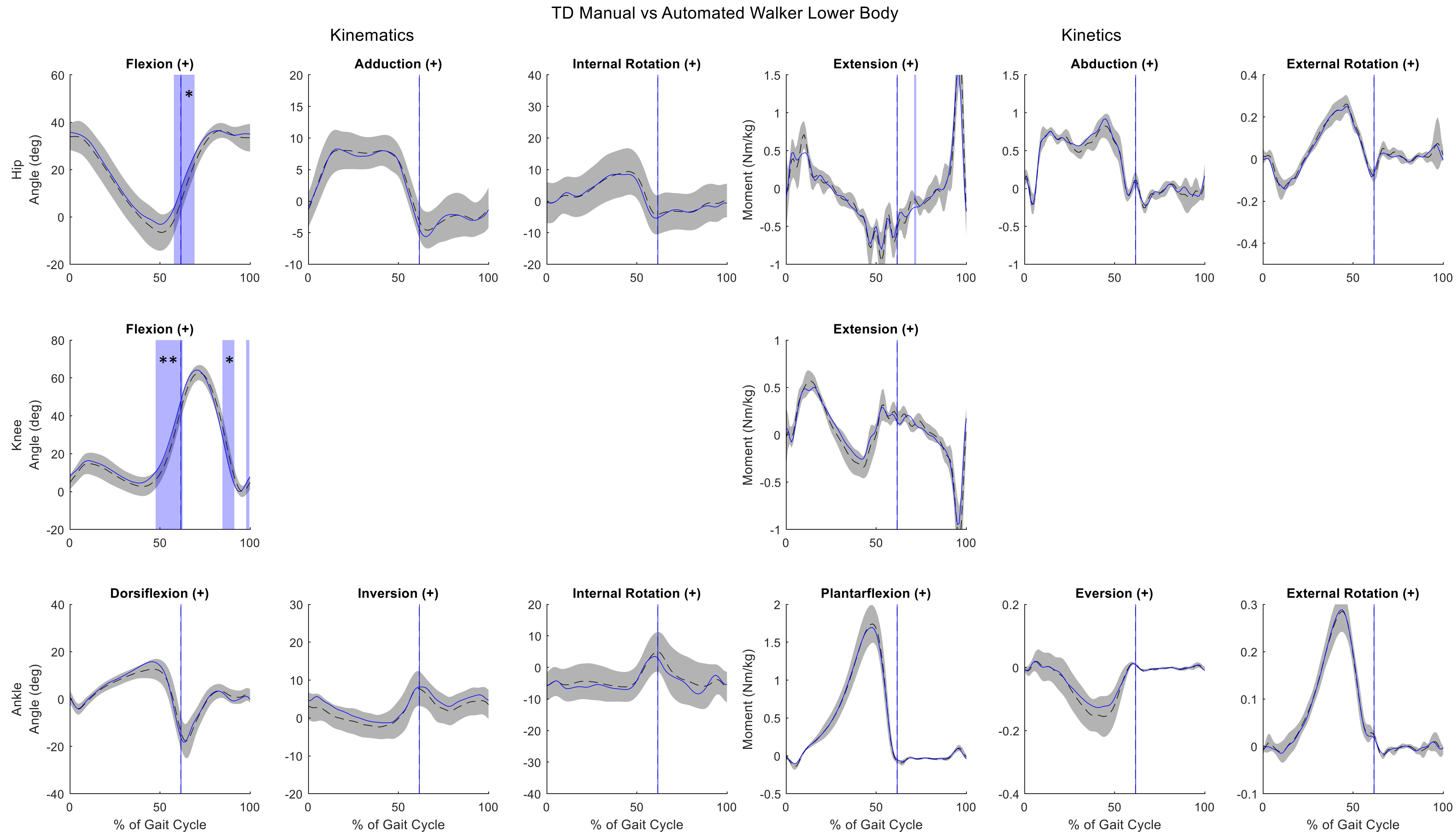

Figure B.4: Lower body joint kinematics (left) and kinetics (right) for the TD group manual (black) compared to automated (blue) walker-assisted gait. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

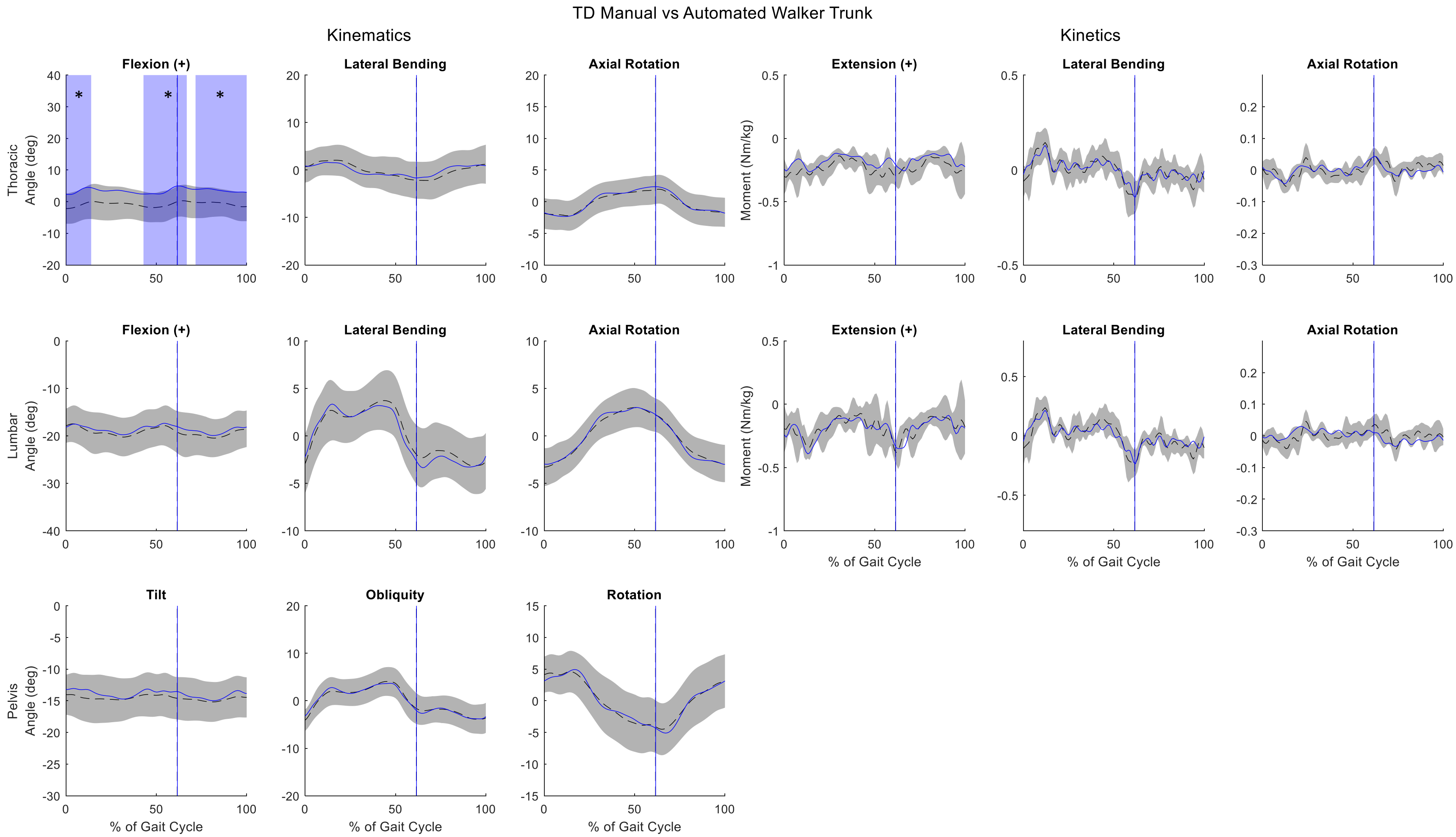

Figure B.5: Trunk kinematics (left) and kinetics (right) and pelvic segment kinematics (bottom) for TD group manual (black) compared to automated (blue) walker-assisted gait. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Trunk flexion was characterized by bending forwards and extension by bending backwards based off of pelvic tilt. Positive lateral bending and axial rotation was in the direction of rotation towards the stance foot. Pelvic tilt is negative for forward rotation, obliquity is positive for rotation away from stance foot rotation is positive for vertical-axis rotation away from stance foot.

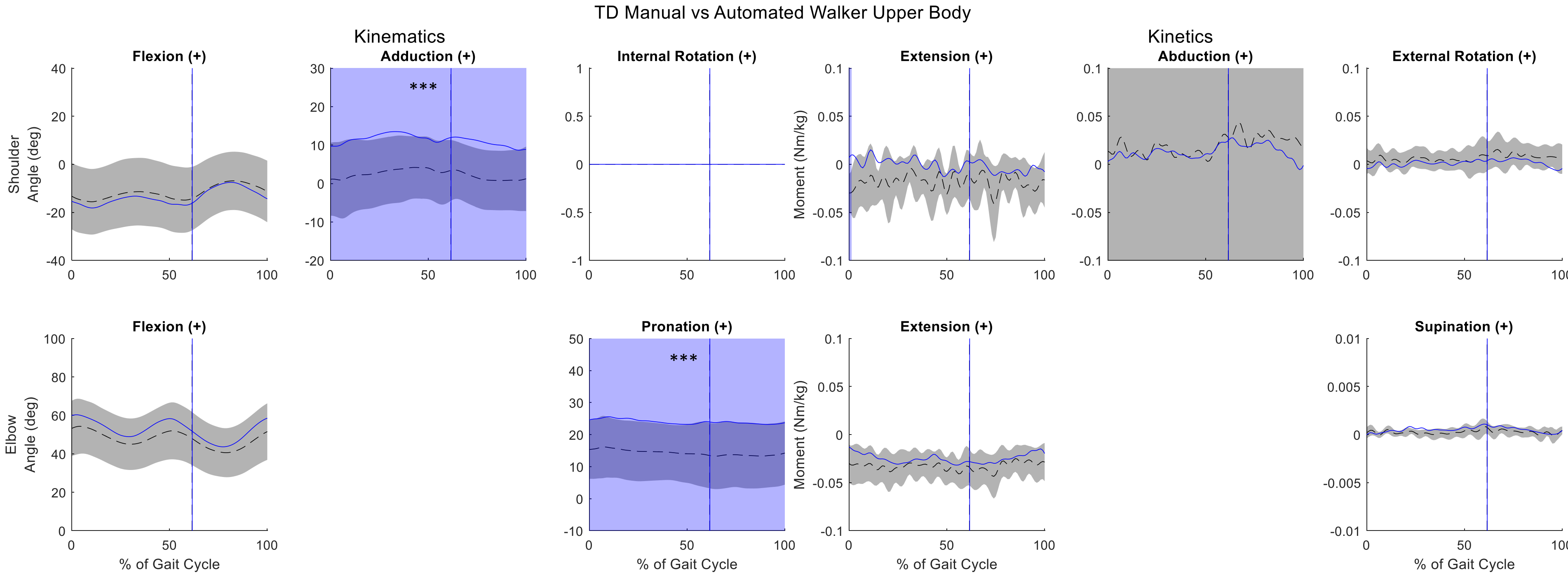

Figure B.6: Upper body kinematics (left) and kinetics (right) for TD group manual (black) compared to automated (blue) walker-assisted gait. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

B.3.4.2 CP Group Manual vs Automated Walker-Assisted Gait

The CP group joint kinematics were better in the lower body when walking with an automated walker (Figure B.7). Hip and knee joint flexion in midstance was lower and had greater range of motion through the gait cycle. Lower body kinetics were greater in automated walker trials. Hip extension moment was greater in early stance and late swing and knee flexion moment was greater in mid-stance. Torso joint kinematics and kinetics were not significantly different between manual and automated walker trials, but automated walker trials trended towards worse posture and higher load on the torso (Figure B.8). Thoracic joint lateral bending angle varied more in automated walker trials. Joint moment was greater in the thoracic and lumbar joints in the lateral direction highlighting the increased need for support of the torso through the gait cycle in automated walker-assisted gait. Upper body joint kinematics were not affected by walking with an automated walker (Figure B.9). Elbow extension moment was reduced in stance phase while shoulder extension moment was increased in late stance and swing in automated walker trials.

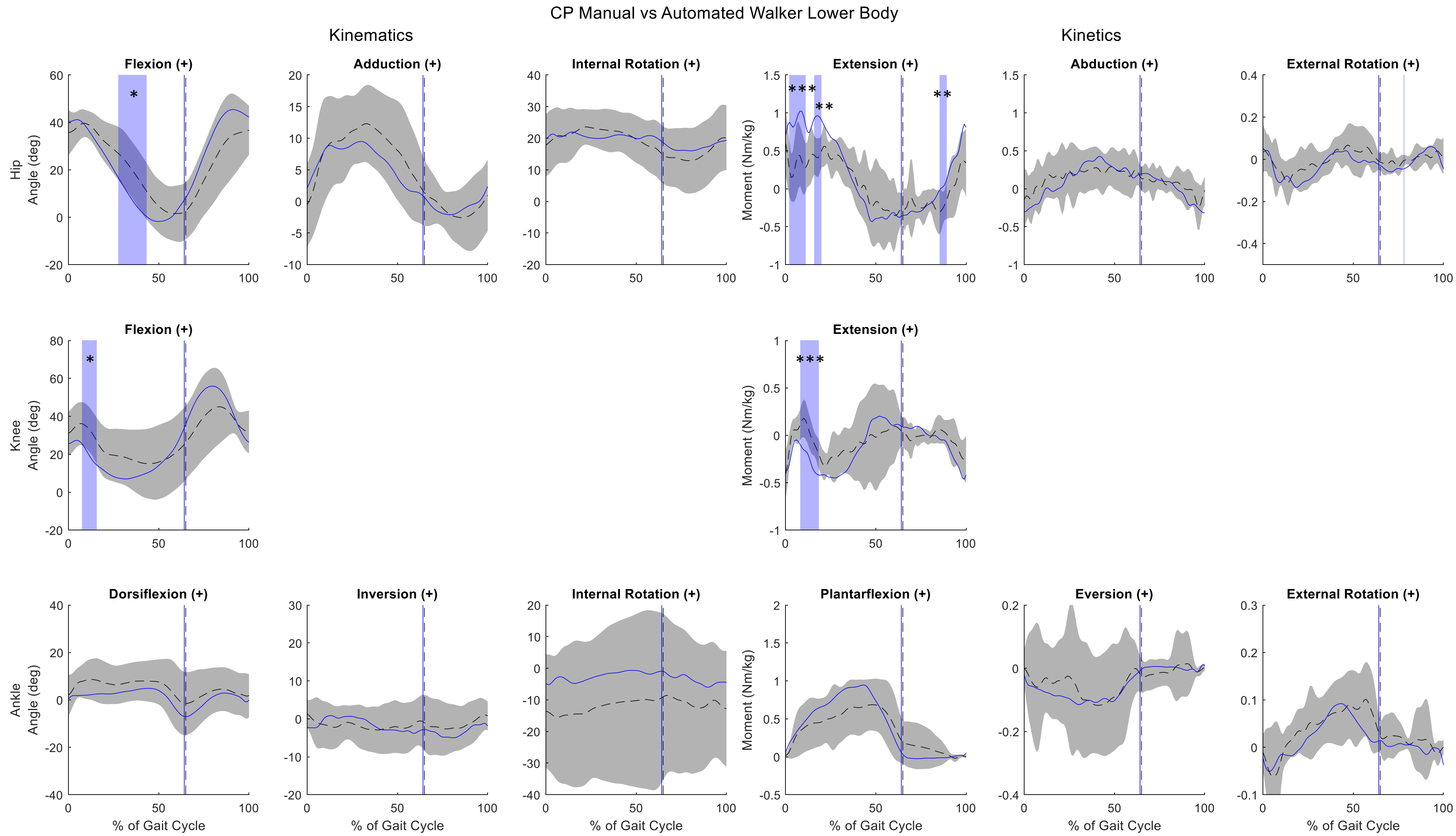

Figure B.7: Lower body joint kinematics (left) and kinetics (right) for the CP group manual (black) compared to automated (blue) walker-assisted gait. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

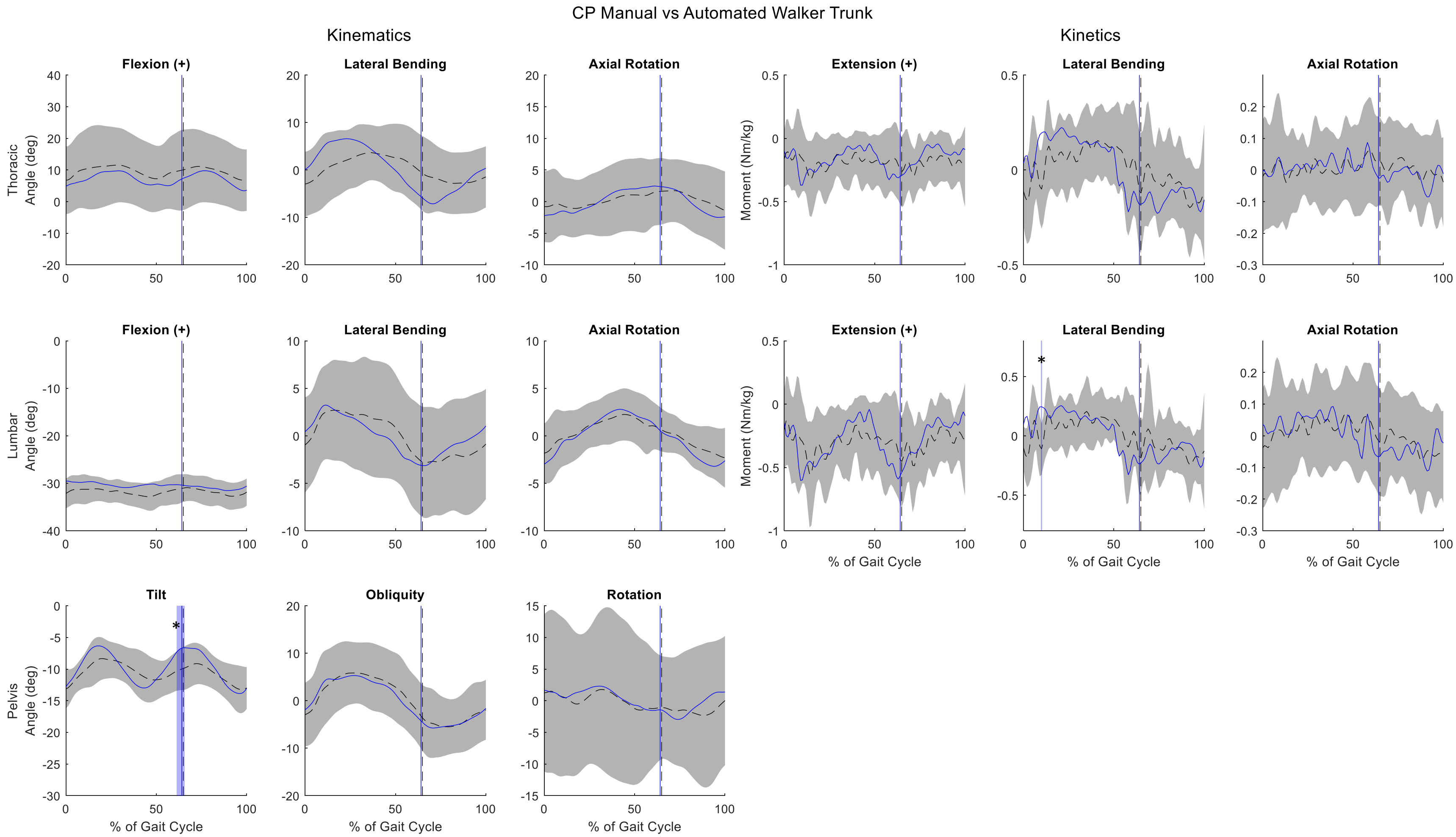

Figure B.8: Trunk kinematics (left) and kinetics (right) and pelvis segment kinematics (bottom) for CP group manual (black) compared to automated (blue) walker-assisted gait. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***). Trunk flexion was characterized by bending forwards and extension by bending backwards based off of pelvic tilt. Positive lateral bending and axial rotation was in the direction of rotation towards the stance foot. Pelvic tilt is negative for forward rotation, obliquity is positive for rotation away from stance foot rotation is positive for vertical-axis rotation away from stance foot.

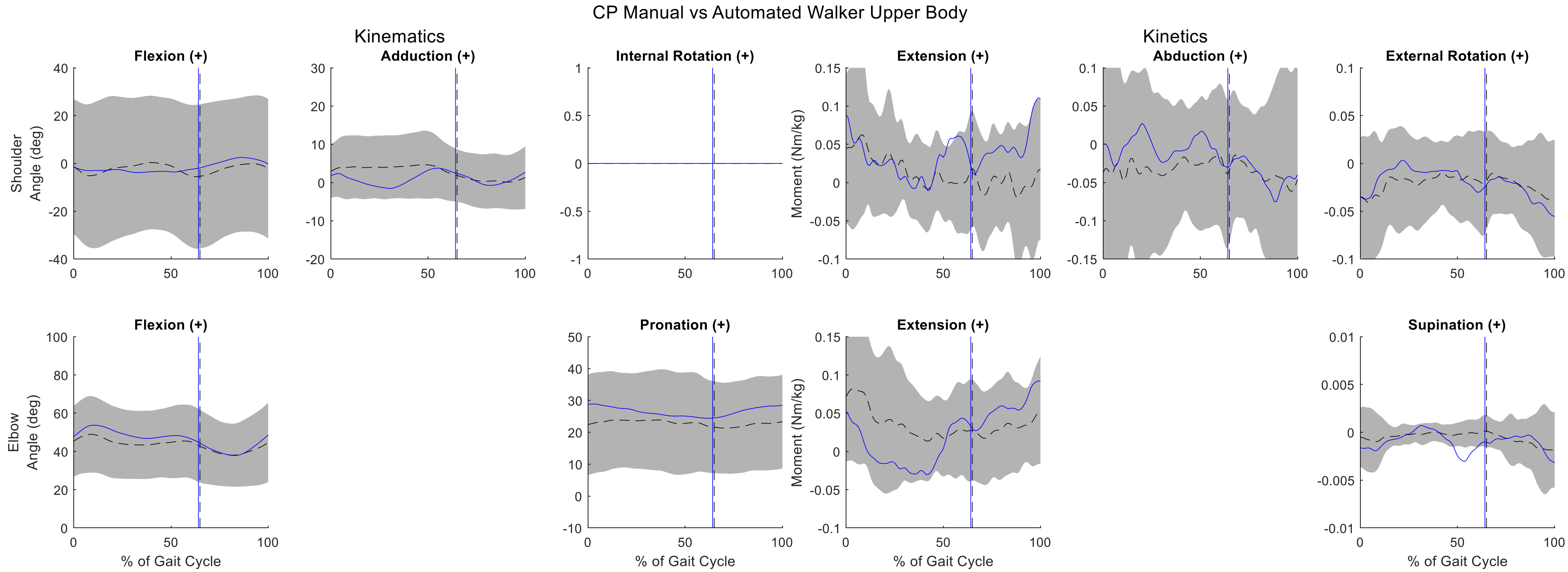

Figure B.9: Upper body kinematics (left) and kinetics (right) for CP group manual (black) compared to automated (blue) walker-assisted gait. SPM significance highlighted by blue shaded boxes and marked as p<0.05 (*), p<0.01 (**), and p<0.001 (***).

B.3.5 Total Mechanical Work

Mechanical joint work was compared between manual and automated walker-assisted trials for the TD group and two subgroups of the CP group that were unassisted and crutch-assisted in their baseline gait (Table B.6). These groups had lower members in them as the automated walker was only instrumented for some of the subjects. The total mechanical joint work for the TD group was mostly unaffected when walking with the automated walker compared to the manual walker. The full body total mechanical joint work for the TD group was not different in automated walker trials compared to manual trials. Thoracic joint work was reduced which also caused the summed segment for the torso to be decreased in automated trials.

The CP unassisted subgroup had increased total joint work in automated walker trials compared to manual walker trials. Increased lower body joint work occurred through the hip along with minor increases in the knee and ankle. Larger torso work was the result of increased joint work in the thoracic and lumbar joint work in the automated walker trials.

With only one subject, the CP crutch-assisted subgroup had large standard deviations and low significance. The subject tended towards a greater full body joint work form increases in the lower body and upper body, while the torso remained relatively consistent to the manual walker trials.

Table B.6: Average total mechanical work for individual joints summed over a gait cycle. Significance between baseline and walker trials for TD group and Unassisted CP group is marked as * ($p<0.05$), ** ($p<0.01$), *** ($p<0.001$). Walker assisted trials of crutch and unassisted subgroup are compared and marked following the same method.

Work (J/kg*m)	TD Group (n=2)		CP Group			
			Unassisted subgroup (n=3)		Crutch-assisted subgroup (n=1)	
Joints	Manual	Automated	Manual	Automated	Manual	Automated
Ankle	0.391-0.516	0.388-0.513	0.377 ± 0.120	0.401 ± 0.150	0.506 ± 0.157	0.847 ± 0.710
Knee	0.375-0.447	0.356-0.406	0.418 ± 0.138	0.539 ± 0.050	0.442 ± 0.066	0.391 ± 0.088
Hip	0.536-0.601	0.562-0.566	0.948 ± 0.215	1.208 ± 0.058**	0.940 ± 0.115	1.120 ± 0.267
LOWER BODY	**2.60-3.13**	**2.62-2.96**	**3.49 ± 0.76**	**4.29 ± 0.46****	**3.77 ± 0.51**	**4.72 ± 2.04**
Lumbar	0.059-0.071	0.056-0.077	0.256 ± 0.091	0.334 ± 0.114*	0.390 ± 0.062	0.479 ± 0.149
Thoracic	0.039-0.049	0.040-0.043	0.309 ± 0.164	0.441 ± 0.199*	0.675 ± 0.126	0.675 ± 0.105
Neck	0.010-0.011	0.008-0.015	0.077 ± 0.052	0.095 ± 0.074	0.144 ± 0.018	0.126 ± 0.020
TORSO	**0.11-0.13**	**0.11-0.13**	**0.64 ± 0.30**	**0.87 ± 0.37***	**1.21 ± 0.19**	**1.28 ± 0.25**
Scapular	0.016-0.022	0.012-0.014	0.107 ± 0.064	0.107 ± 0.044	0.268 ± 0.203	0.258 ± 0.085
Shoulder	0.010-0.012	0.010-0.014	0.103 ± 0.048	0.112 ± 0.047	0.194 ± 0.100	0.274 ± 0.109
Elbow	0.005-0.009	0.016-0.021	0.061 ± 0.053	0.069 ± 0.034	0.247 ± 0.179	0.312 ± 0.290
Wrist	0.000-0.002	0.000-0.002	0.015 ± 0.008	0.011 ± 0.005	0.083 ± 0.061	0.096 ± 0.064
UPPER BODY	**0.06-0.09**	**0.08-0.10**	**0.57 ± 0.31**	**0.60 ± 0.26**	**1.59 ± 1.06**	**1.88 ± 0.83**
TOTAL	**2.78-3.34**	**2.82-3.19**	**4.70 ± 1.21**	**5.76 ± 0.73***	**6.57 ± 1.21**	**7.87 ± 2.21**

B.3.6 Mechanical Work of the Body

Mechanical work over the gait cycle was calculated for each joint over the gait cycle and the results were summed into different body segments: full body, lower body, torso, and upper body work. TD and CP groups were compared between manual and automated walker-assisted gait. The CP group was comprised of four subjects and the TD group had two subjects as the automated walker was instrumented for these subjects.

B.3.6.1 TD Manual vs Automated Walker-Assisted Gait Mechanical Work

The TD group mechanical work in walker-assisted gait was mostly equivalent in automated and manual trials (Figure B.10). Reduced mechanical work occurred at two points during the gait cycle segments of the body. Lower body and full body sagittal plane mechanical work was lower in early stance phase which indicates a reduced initial loading on the lower body in stance. Reduced sagittal plane mechanical work in the torso at toe off highlights a lower necessary activation of the torso to enable the swing phase.

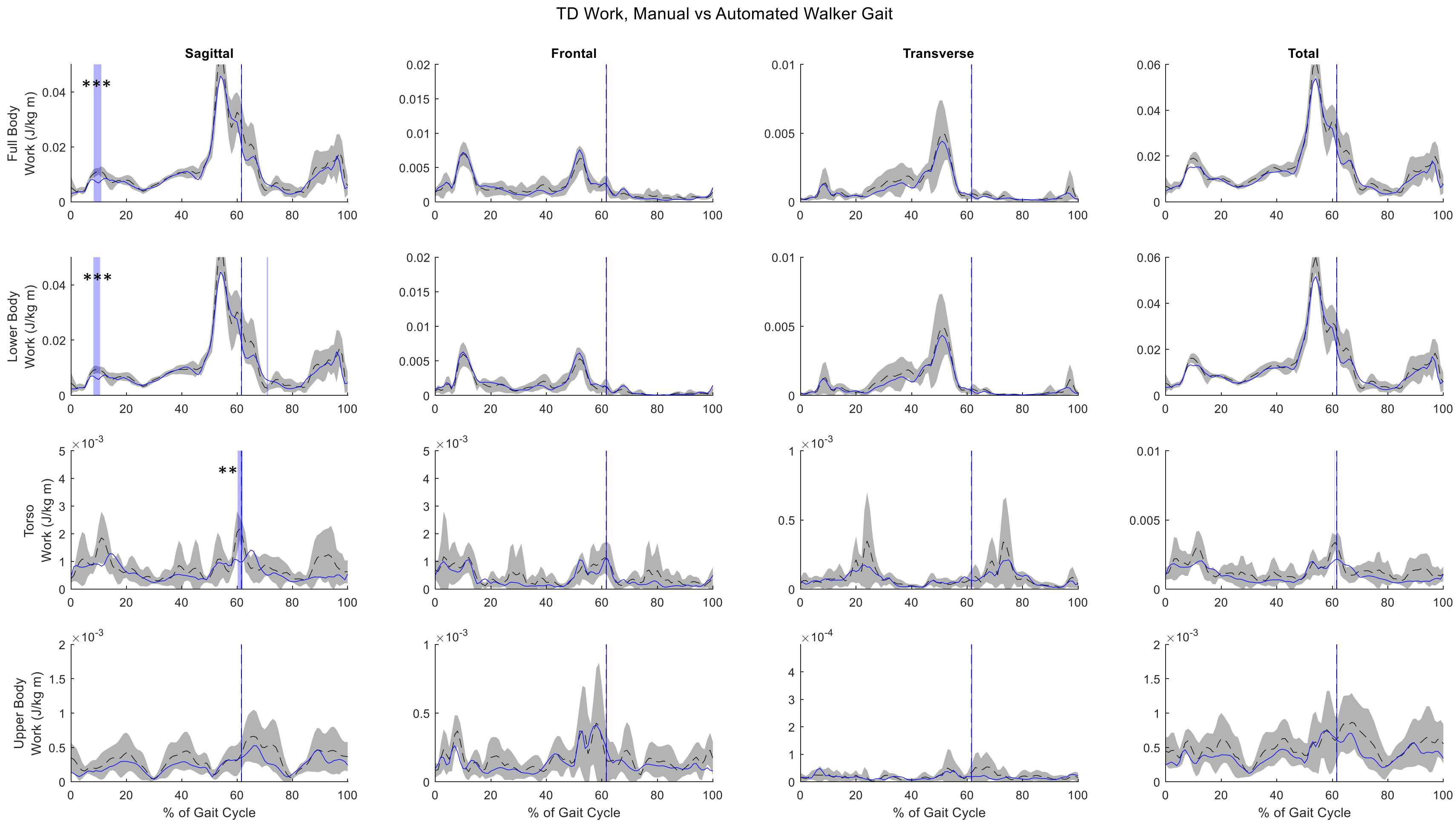

Figure B.10: Mechanical work over the gait cycle for the full body, lower body, torso, and upper body segments. TD group manual (black) compared to automated (blue) walker-assisted gait work is compared using SPM analysis. SPM significance is highlighted by blue shaded boxes and marked with p<0.05 (*), p<0.01 (**), and p<0.001 (***).

B.3.6.2 CP Manual vs Automated Walker-Assisted Gait Mechanical Work

Mechanical work was increased in the CP group automated walker trials in all segments of the body (Figure B.11). The lower body was taxed to move the body during stance phase more in automated walker trials than it was in manual walker trials. Total lower body and full body work was increased in early to mid-stance primarily from increased sagittal plane work. Torso and upper body joints were used to stabilize during transition periods of gait at toe off and heel strike and the mechanical work increased at these points in automated walker trials. Sagittal and frontal plane mechanical work of the torso was increased in automated walker trials during late stance phase. Upper body mechanical work increased during times of heel strike and toe off. The CP group walked worse with the automated walker as the mechanical work to move increased over their gait cycle.

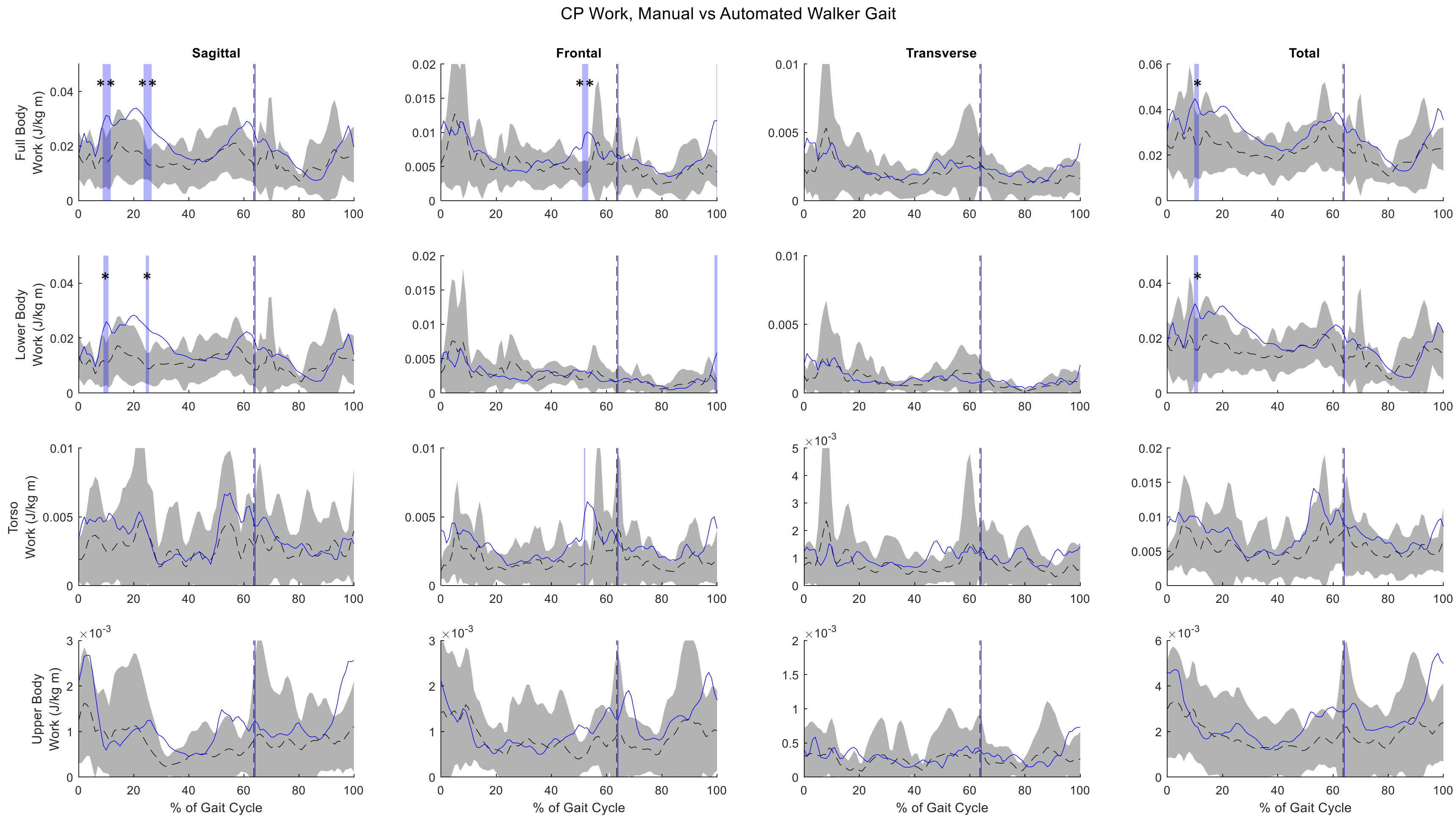

Figure B.11: Mechanical work over the gait cycle for the full body, lower body, torso, and upper body segments. CP group manual (black) and automated (blue) walker-assisted gait work is compared using SPM analysis. SPM significance is highlighted by blue shaded boxes and marked with p<0.05 (*), p<0.01 (**), and p<0.001 (***).

B.4 Discussion

This study developed a method for calculating full body mechanical work while walking with an instrumented pediatric posterior walker. Subjects walked with a manual and automated walker to examine the effect of the control on the walker on gait parameters and model outcomes. The TD group walked with slightly less mechanical work over their gait cycle while the CP group walked with more mechanical work, particularly in the lower body and torso segments when using the automated walker. While the CP group walked faster with the automated walker, the increased work required over the gait cycle indicates the control on the walker was not helpful in their gait. Further development of the control and alternate methods could prove more beneficial in an automated walker as it did allow the CP group to pull on the walker less during their gait.

For the both TD and CP groups, spatiotemporal parameters were unaffected by the automated walker compared to the manual walker, but by breaking the CP group up by GMFCS the effect of the automated control was found. The GMFCS II group walked faster and with more stability when they did not need to interact with the walker as much to pull it along. They returned to walking as they did in their baseline without the posterior walker or other support. The GMFCS III group walked worse with the automated walker, also regressing to similar spatiotemporal gait parameters as were observed in Chapter 3 in their baseline gait. Based on joint kinematic and kinetic analysis the CP group had mixed positive and negative effects on their gait when walking with the automated walker. The lower body joints had a greater range of motion and reduced load on the knee, but there was increased moment at the hip. The torso moved more laterally and greater moment generation indicated the subjects were not as stable in their gait. These effects on gait by the automated walker are opposite that which was observed by the manual walker compared to baseline gait.

Handle reaction forces were lower in the vertical direction for both groups in the automated walker trials indicating the walker was not used as much for support. This could be due to the population not adjusting to the automated walker and not trusting the motion of the walker as much as the manual one. Future work giving subjects extended time to get accustomed to the automated walker could answer if a control on the walker destabilizes the user in their gait or if they need time to get used to walking with it.

For both groups, the effects of a walker from baseline gait to using a manual walker were reversed by incorporating the control designed for the walker. The TD group benefited by from the automated walker by returning to a lower total mechanical work over their gait. In manual walker trials we saw an increase in this segment from baseline, the TD group returned to a gait that resembled their baseline gait and less work was performed by the torso. Walker-assisted gait required more energy with a control-automated walker than a manual walker in the CP group. The CP group walked with greater mechanical work generation in the lower body and torso segments, which is also where the largest changes were seen from baseline to manual walker trials.

B.5 Conclusion

TD and CP groups walked with a manual walker and an automated walker with a custom control designed to keep the walker at an initialized distance away from the subject for supportive needs. The automated walker caused users to walk similar to their baseline gait. For the TD group this was better, as energy was not expended to pull the walker along. The CP group was hindered by the automated walker as the control increased their minimum energy required for gait and decreased their stability while walking. A control for an automated walker designed around the particular use and interaction of the CP group with the walker could yield a more stable and efficient gait in the CP population as the group walked faster and with better posture than their baseline gait with the control designed in this work.

Reference List

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: Benefits, demands, and adverse consequences. Archives of Physical Medicine and Rehabilitation, 86(1), 134–145. https://doi.org/10.1016/j.apmr.2004.04.023

Bolster, E. A. M., Balemans, A. C. J., Brehm, M. A., Buizer, A., & Dallmeijer, A. J. (2017). Energy cost during walking in association with age and body height in children and young adults with cerebral palsy. Gait and Posture, 54, 119–126. https://doi.org/10.1016/j.gaitpost.2017.02.026

Chen, I. H., Kuo, K. N., & Andriacchi, T. P. (1997). The influence of walking speed on mechanical joint power during gait. Gait & Posture, 6(3), 171–176. https://doi.org/10.1016/S0966-6362(97)00009-X

Fast, A., Wang, F. S., Adrezin, R. S., Cordaro, M. A., Ramis, J., & Sosner, J. (1995). The instrumented walker: Usage patterns and forces. Archives of Physical Medicine and Rehabilitation, 76(5), 484–491. https://doi.org/10.1016/S0003-9993(95)80584-2

Johnson, D. C., Damiano, D. L., & Abel, M. F. (1997). The Evolution of Gait in Childhood and Adolescent Cerebral Palsy. Journal of Pediatric Orthopaedics, 17(3), 392–396. Retrieved from https://ovidsp.tx.ovid.com/sp-3.30.0b/ovidweb.cgi

Ju Kim, C., & min Son, S. (2014). Comparison of Spatiotemporal Gait Parameters between Children with Normal Development and Children with Diplegic Cerebral Palsy. Journal of Physical Therapy Science, 26(9), 1317–1319. Retrieved from https://www.jstage.jst.go.jp/article/jpts/26/9/26_jpts-2014-008/_pdf/-char/ja

Konop, K. A., Strifling, K. M. B., Wang, M., Cao, K., Eastwood, D., Jackson, S., … Harris, G. F. (2009). Upper extremity kinetics and energy expenditure during walker-assisted gait in children with cerebral palsy. Acta Orthopaedica et Traumatologica Turcica, 43(2), 156–164. https://doi.org/10.3944/AOTT.2009.156

Krautwurst, B. K., Dreher, T., & Wolf, S. I. (2016). The impact of walking devices on kinematics in patients with spastic bilateral cerebral palsy. Gait & Posture, 46, 184–187. https://doi.org/10.1016/j.gaitpost.2016.03.014

Norman, J. F., Bossman, S., Gardner, P., & Moen, C. (2004). Comparison of the Energy Expenditure Index and Oxygen Consumption Index During Self-Paced Walking in Children with Spastic Diplegia Cerebral Palsy and Children Without Physical Disabilities. Pediatric Physical Therapy, 16(4), 206–211. https://doi.org/10.1097/01.PEP.0000145930.84009.23

Pauk, J., Ihnatouski, M., Daunoraviciene, K., Laskhousky, U., & Griskevicius, J. (2016). Research of the spatial-temporal gait parameters and pressure characteristic in spastic diplegia children. Acta of Bioengineering and Biomechanics, 18(2), 121–129. https://doi.org/10.5277/ABB-00393-2015-02

Paul, S. M., Siegel, K. L., Malley, J., & Jaeger, R. J. (2007). Evaluating interventions to improve gait in cerebral palsy: a meta-analysis of spatiotemporal measures. Developmental

Medicine & Child Neurology, 49(7), 542–549. https://doi.org/10.1111/j.1469-8749.2007.00542.x

Protas, E. J., Raines, M. L., & Tissier, S. (2007). Comparison of Spatiotemporal and Energy Cost of the Use of 3 Different Walkers and Unassisted Walking in Older Adults. Archives of Physical Medicine and Rehabilitation, 88(6), 768–773. https://doi.org/10.1016/j.apmr.2007.03.001

Ries, A. J., & Schwartz, M. H. (2018). Low gait efficiency is the primary reason for the increased metabolic demand during gait in children with cerebral palsy. Human Movement Science, 57, 426–433. https://doi.org/10.1016/j.humov.2017.10.004

Russell, S., Bennett, B., Sheth, P., & Abel, M. (2011). The gait of children with and without cerebral palsy: work, energy, and angular momentum. J Appl Biomech, 27(2), 99–107. https://doi.org/10.1123/jab.27.2.99

Unnithan, V. B., Dowling, J. J., Frost, G., & Bar-Or, O. (1996). Role of cocontraction in the O2 cost of walking in children with cerebral palsy. Medicine and Science in Sports and Exercise, 28(12), 1498–1504. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/8970144

Van de Walle, P., Hallemans, A., Schwartz, M., Truijen, S., Gosselink, R., & Desloovere, K. (2012). Mechanical energy estimation during walking: Validity and sensitivity in typical gait and in children with cerebral palsy. Gait & Posture, 35(2), 231–237. https://doi.org/10.1016/j.gaitpost.2011.09.012

van den Hecke, A., Malghem, C., Renders, A., Detrembleur, C., Palumbo, S., & Lejeune, T. M. (2007). Mechanical Work, Energetic Cost, and Gait Efficiency in Children With Cerebral Palsy. Journal of Pediatric Orthopaedics, 27(6), 643–647. https://doi.org/10.1097/BPO.0b013e318093f4c3