10-2011

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Journal of Experimental & Clinical Medicine, Vol. 3, No. 5 (October 2011): 213-217. DOI.

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Abstract

Background

Many children with spastic diplegic cerebral palsy (CP) use anterior or posterior walkers to aid ambulation. Prolonged use may lead to upper extremity (UE) pathology later in life, including arthritis and joint contractures.

Purpose

This study analyzes the dynamics (kinematics and kinetics) of the shoulder (glenohumeral), elbow, and wrist joints during anterior and posterior walker use. It also examines the dynamic effects of adjusting handle height and grip rotation.

Methods

Ten children with CP underwent motion analysis with upper and lower extremity marker sets and six-degree-of-freedom instrumented walker handles, while using both anterior and posterior walkers. One child underwent the same analysis, with added trials for wrist derotation (adjusted axial grip rotation) and wrist plus elbow derotation (adjusted handle height). A validated kinematic and kinetic model was applied to calculate UE joint angles, joint reaction forces (JRFs), and joint reaction moments (JRMs).

Results

Surprisingly, no statistically significant differences in UE angles, JRFs, or JRMs were observed between anterior and posterior walkers. Wrist derotation, however, decreased the flexion JRM seen at the wrist, and elbow derotation decreased the flexion JRM seen at the elbow.

Conclusion

Anterior and posterior walkers produce similar UE motion and peak loading values. Wrist and elbow joint derotation alters the dynamic effects experienced by the UEs. UE motion analysis during aided gait can be useful for optimizing UE loading conditions to limit pathology later in life.

Key words

cerebral palsy, joint dynamics, motion analysis, upper extremity, walkers

1. Introduction

Cerebral palsy (CP) is a neurological condition that affects muscle tone and coordination. It is the result of pathologic brain development, insult, or injury and generally appears in infancy or early childhood. The causes of CP remain unclear, but risk factors include low birth weight, intrauterine infection, and multiple gestations.[1] The overall prevalence of CP in the United States is between three and four per 1000 live births,[2, 3] or approximately 10,000 to 17,000 babies each year.[4] CP is the largest diagnostic group treated in pediatric rehabilitation.[5]

Spastic diplegia, the most common distribution of involvement, accounts for an estimated 75–87% of patients with CP.[6] Spasticity is defined as a velocity-dependent increase in muscle tone. This decreases the patient's motor control and balance, thereby impairing their ability to walk and perform other functional activities. Individuals with spastic diplegia have greater involvement in the lower extremities (LEs) than in the upper extremities (UEs). Assistive devices, including walkers, allow the UEs to aid in the weight-bearing, transfer, and stability aspects of ambulation not fully provided by the LEs.[7, 8] Clinicians usually prescribe one of two primary types of walkers to children with spastic diplegic CP: anterior or posterior (Figure 1).


Figure 1. L: Anterior and R: posterior walker.

Many children with spastic diplegic CP also have some degree of involvement in their UEs, including spasticity, muscle contractures or imbalances, and weakness. While the clinical manifestations of CP are highly variable, due to the locus of specific brain lesions, a common pathologic UE pattern consists of shoulder adduction and internal rotation, elbow flexion, forearm pronation, and wrist flexion.[9]

Few clinical tools objectively evaluate the UE impairments of patients with CP during assisted gait. Motion analysis has been used extensively to evaluate the LEs in many pathologic conditions, including CP.[10, 11, 12] Some studies, including those from our group,[13, 14, 15, 16] have begun to use this technology to investigate the UEs during assisted gait.[17, 18] Although UE motion is difficult to compare across populations and studies, it is an important tool for quantifying changes over time, changes secondary to clinical/surgical intervention, and changes resulting from alterations in the walker configuration.

In addition to kinematics, it is important to study the internal joint reaction forces (JRFs) and joint reaction moments (JRMs) that result from walker use. The resulting gait pattern is quadrupedal, which places varied demands upon the UEs for balance, support, and gait progression.[16, 17, 18, 19, 20] Joint load-related pathology, including carpal tunnel syndrome[21] and shoulder injury and arthritis later in life, have been linked to the prolonged use of walking aids and wheelchairs.[22, 23] It is important to allow the LEs to bear as much weight as possible during walker use to re-establish a bipedal pattern while avoiding UE overuse injury.

2. Purpose

The objective of the current study was to characterize wrist, elbow, and shoulder (glenohumeral) joint dynamics (position, JRFs, and JRMs) during anterior and posterior walker use. We also examined the potential for UE load reduction through walker modification, a novel concept not explored in current literature. A quantitative approach was taken to assess the effects of handle height and grip rotation modifications during both anterior and posterior walker usage.

3. Methods

3.1. Subjects

Ten participants with spastic diplegic CP were tested (mean age 12.1 years, height 1.3 m, weight 35.6 kg). The inclusion criteria required routine walker use for at least 1 month, an Ashworth score of 2 or less (slight increase in tone) at the elbow joint, no botulinum toxin type A treatment within the past 6 months, and no orthopedic surgery within the past year. An additional participant, with the same inclusion criteria, was tested with an adjustable walker (adaptive therapeutic walker, ATW).

3.2. Protocol

All participants underwent motion analysis, with reflective surface markers placed on anatomical locations of the UEs and LEs. Kinetic data were collected using custom designed walker handles instrumented with 6-axis strain gage-based load cells (AMTI, Watertown, MA, USA). Motion data were collected at 60 Hz with a 12-camera system (Vicon, Oxford, UK), and kinetic data were simultaneously collected at 1500 Hz. The marker set, motion capture system, walker handles, and dynamic model have been described and validated previously by our group.[24, 25] Figure 2 describes the data collection and calculation process.

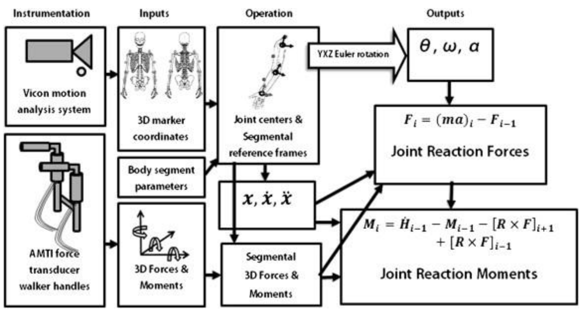

Figure 2. Data collection and analysis process. $x, \dot{x}, \ddot{x}$ are center of mass position, velocity, and acceleration, respectively; θ, ω, α are joint angle, angular velocity, and angular acceleration, respectively. In the equations, *F* = force, *m* = mass, *a* = acceleration, $\dot{H}$ = mass moment of inertia, *R* = moment arm; *i* refers to the current segment, *i − 1* to the distal segment, and *i + 1* to the proximal segment.

The walkers (Sunrise Medical, Longmont, CO, USA; Kaye Products, Inc., Hillsborough, NC, USA) were height-adjusted so that the handles were even with the ulnar styloid process with arms at the sides. Data were first collected with the subjects' usual walker type (posterior, in all cases). At least three acceptable gait cycles at a self-selected pace and walking style were obtained. For the 10 participants, a 30-day acclimation period was given before testing was repeated using the alternate walker type (anterior).

The participant undergoing the ATW protocol, performed testing with both walker types during the same visit. An adjustable walker modified with Velcro handle grips was used, while the participant wore snug-fitting Velcro gloves. The walker height was initially set as described above. A physical therapist then rotated the handle grips so that the wrists were de-rotated to a more neutral position (0 degrees flexion). The final adjustment involved altering the handle heights to move the elbows into 30 degrees flexion, while the wrist derotation remained intact. Data were collected as described above for each of the three configurations. The entire process was repeated for the alternate walker type, for a total of six data sets (original anterior and posterior walker, adjusted wrist anterior and posterior walker, adjusted wrist and elbow anterior and posterior walker).

3.3. Data analysis

The raw data was filtered and processed with a custom UE kinematic and kinetic model to obtain joint angles, angular velocities, angular accelerations, and three-dimensional force and moment data at the wrist, elbow, and shoulder joints.[13, 16] The model complied with International Society of Biomechanics (ISB) standards[26] and computed three-dimensional Euler joint rotations comparing distal segments to proximal segments in a sagittal-coronal-transverse rotation sequence. The kinetic portion of the model used an inverse dynamics approach, similar to Vaughan et al in 1992[27] to determine the JRFs and JRMs at each UE joint. All data were time-normalized to 100% gait cycle. The magnitudes of the forces and moments were normalized to body weight (% BW) and body weight × height (% BW × H), respectively, and are reported as percentages.

3.4. Statistical analysis

The kinematic data for the initial 10 participants were compared between walker types and sides using a paired t-test with a Bonferroni adjustment.[16] Kinetic data were compared between walker types and sides using a Wilcoxon signed rank test with Benjamini-Hochberg threshold values.[13]

4. Results

4.1. Standard anterior and posterior walker

Complete results from the 10 participants have been reported previously13, 16 and are shown in Figure 3, Figure 4, Figure 5. There were no statistically significant differences between walker types (anterior and posterior) or sides (right and left, or dominant and non-dominant). The overall trends include the following.

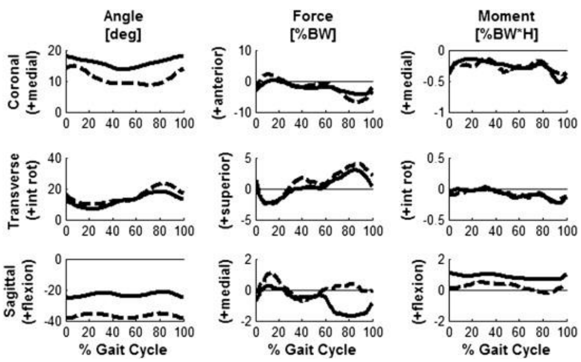

Figure 3. Average shoulder dynamics. (– Anterior Walker; - - - Posterior Walker).

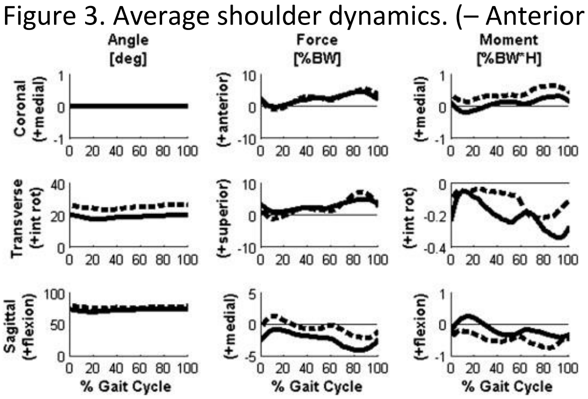

Figure 4. Average elbow dynamics. (– Anterior Walker; - - - Posterior Walker).

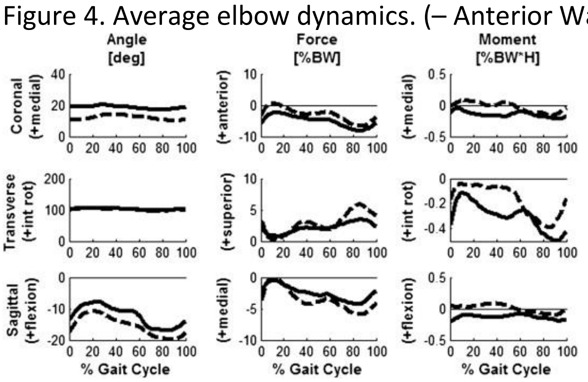

Figure 5. Average wrist dynamics. (– Anterior Walker; - - - Posterior Walker).

The shoulder (glenohumeral joint) was abducted, extended, and internally rotated with each walker type. The posterior walker produced greater shoulder extension than the anterior walker [average difference of 13.65 degrees (deg)]. The flexion JRM was greater during anterior walker use (1.67% BW × H for anterior walker, 0.97% BW × H for posterior walker).

The elbow was flexed and the forearm pronated with both walker types, but more so with the posterior walker (average difference of 4.0 deg flexion, 6.1 deg pronation). The medial JRF at the elbow was greater with posterior walker use (2.5% BW for posterior walker, 0.1% BW for anterior walker), as was the extension JRM (1.08% BW × H for posterior walker, 0.78% BW for anterior walker).

The wrist was ulnar deviated with both walker types, more so with anterior walker (average difference of 6.6 deg). Wrist extension was seen with both types. The posterior walker produced a greater anterior JRF (1.9% BW for posterior walker, -1.2% BW for anterior walker), and flexion JRM (0.27% BW × H for posterior walker, 0.09% BW × H for anterior walker).

4.2. Adaptive therapeutic walker

Results from the ATW protocol showed a decrease in wrist flexion position and a decrease in wrist extension moment when the handles were rotated. After elbow and wrist adjustments, the elbow flexion position decreased, as did the flexion moment. This is most clearly seen on the right side during posterior walker use, shown in Figure 6.

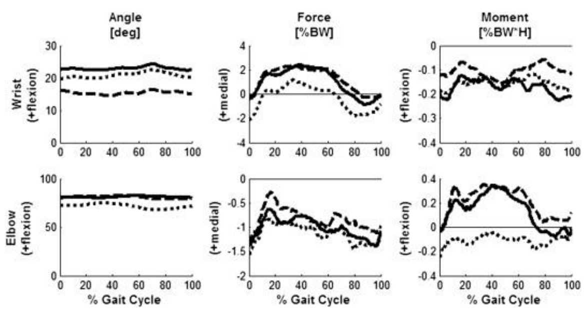

Figure 6. Right side sagittal plane wrist and elbow dynamics for posterior ATW. (– Original; - - - Adjusted wrist; ··· Adjusted wrist and elbow).

5. Discussion

Kinematic and kinetic analysis of 10 children with spastic diplegic CP showed no statistically significant differences in UE dynamics between standard anterior and posterior walkers. The ranges of motion and joint kinetics were established and found to compare favorably to values reported by our group and others in earlier studies.[17, 28, 29]

The ATW model was used to evaluate joint load effects resulting from walker handle rotation and position adjustments. Overall, results showed that moving the wrist and elbow joints to a more neutral position decreased the joint reaction moments seen at these joints. Derotating the wrist did not have a noticeable impact on the elbow or shoulder joint loads, which was a beneficial result. Wrist derotation also reduced the wrist flexion moment, another beneficial result. The limited combinations of wrist and elbow adjustment currently examined, decreased the elbow flexion JRM (a beneficial effect), but increased the wrist flexion JRM (not desired). Neither adjustment (wrist or wrist/elbow) had a noticeable impact on the shoulder flexion JRM (a beneficial effect). Continued examination of handle position and orientation adjustments as well as asymmetric adjustments remains a goal for further investigation.

Small changes to UE joint angles may affect the loading experienced by the joints. There is a complex relationship among the joints, as changing the dynamics of one joint affects the others. This is a relationship worth investigating, since optimizing joint loading conditions can have an impact both in the short term (comfort) and longer term (contractures, arthritis). While minimizing UE joint loads is an important goal, the loads that are experienced can also be used to the patient's benefit. By positioning the walker handles in such a way that the user has a more neutral wrist and elbow position (without increased shoulder loading), the UE weight-bearing activity of assisted gait could reduce/avoid later joint contractures and stiffness. Minimizing the overall UE loads while maintaining balance and stability, contributes to ambulatory function while also ensuring that the LEs bear as much weight as possible. This transition to greater LE weight bearing support also helps to increase LE strength and dynamic range while reducing UE demands and potential for later injury.

Limitations of this study relate to the sample of patients studied and the relatively limited number of cases. Because of variability within the CP population, a larger sample size would be suggested for further study and characterization of kinematic and kinetic metrics. The ATW analysis is also limited, examining a single subject. Larger populations for both protocols would enhance the results, and potentially lead to a defined methodology and clinical protocol for optimizing/reducing UE load bearing while maintaining function during walker assisted gait.

Acknowledgments

The contents of this manuscript were developed under grants from the Department of Education, NIDRR, Grant numbers H133E100007, H133P020004 and H133G010069. However, those contents do not necessarily represent the policy of the Department of Education, and you should not assume endorsement by the Federal Government.

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