Equinus Deformity as a Compensatory Mechanism for Ankle Plantarflexor Weakness in Cerebral Palsy

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A theory for equinus gait in cerebral palsy (CP) is that the strong plantarflexors prevent the weak dorsiflexors from achieving dorsiflexion, thereby causing the ankle to be in a plantarflexed position. Recent work has indicated that both the ankle dorsiflexors and plantarflexors are weak. The purpose of this research was to theoretically and experimentally demonstrate that equinus deformity gait could be a compensatory strategy for plantarflexor weakness. It was hypothesized that children with CP utilize an equinus position during gait as a consequence of their weakness. A two-dimensional, sagittal plane model estimating plantarflexor forces through the Achilles tendon was developed. Five able-bodied (AB) children were tested utilizing heel-toe and progressively increasing toe walking strategies. Four children with CP were tested as they walked using their equinus gait. Results demonstrated that AB children assuming the toe walking stance progressively reduced the plantarflexor force when compared to their heel-toe walking trials. However, their toe walking strategy could not reduce the plantarflexor force level to that of the children with CP during the gait cycle. It was concluded that the equinus deformity posture complemented the CP children’s plantarflexor weakness. Therefore, by implementing a concomitant strategy to maintain a reduced force state, equinus deformity could be used as a compensatory mechanism for individuals with plantarflexor weakness.

Key Words: strength, plantarflexors, model, distribution

Introduction

Cerebral palsy (CP) is a nonprogressive disorder characterized by impairment of motor function secondary to injury of the immature brain (Ingram, 1984). Equinus deformity is the most common problem in children with spastic CP (Banks &
Green, 1958) and the most common indication for surgery (Gage, 1991; Simon & Ryan, 1992). One general theory for equinus gait is that the strong plantarflexors prevent the weak dorsiflexors from achieving dorsiflexion, thereby causing the ankle to be in a plantarflexed position at initial contact and throughout support (Gage, 1991; Simon & Ryan, 1992). The strong plantarflexors could generate a torque under both voluntary control of the patient or involuntary control often associated with spasticity, including excessive tone (Gage, 1991), dynamic contractures, clonus (Simon & Ryan, 1992), or dynamic spasticity.

Little objective data exist to support the theory of strong plantarflexors. Recent publications indicate that both the ankle dorsiflexors and plantarflexors are weak (Engsberg, Ross, & Park, 1999; Engsberg, Ross, Olree, & Park, 2000; Ross & Engsberg, 2002). Based on a group of 60 children with cerebral palsy and 50 able-bodied children, it was reported that the children with CP could only produce a maximum plantarflexion torque 40% of that of able-bodied (AB) children, and a dorsiflexion torque 56% of that of AB. The ankle spasticity associated with this group varied greatly, with no consistent relationship between spasticity and strength. Additionally, it has been shown that the torque produced around the ankle decreased with increased plantarflexion (Sale, Quinlan, Marsh, McComas, & Belangen, 1982). Using a group of 20 AB males, they analyzed plantarflexor torque through radiographic analysis of the moment arms around the ankle joint and collected data during a series of voluntary and externally stimulated plantarflexor contractions. It was demonstrated that when assuming a plantarflexed position, less torque could be generated.

The association between weak plantarflexors and the utilization of an equinus gait has been put forth by Kerrigan and co-workers (Kerrigan, Riley, Rogan, & Burke, 2000). Testing 10 AB participants during toe walking and normal heel-toe walking, they demonstrated that toe walking required less plantarflexor force and power than normal walking. Kerrigan et al. concluded that toe walking could have compensatory advantages for patients with upper motor neuron injury and lower extremity weakness. On the other hand, it was demonstrated that AB persons walking on their toes had an increase in muscle activity as compared with heel-toe walking (Burnfield, Gronley, Mulroy, & Perry, 2001). Also using a group of 10 AB persons participating in a series of heel-toe and toe walking trials, it was demonstrated that toe walking generated higher plantarflexor moments during loading response and midstance as compared to normal heel-toe gait. It was surmised that to achieve the same moment as found in heel-toe gait, twice the relative plantarflexor effort was required.

The purpose of this study was to theoretically and experimentally demonstrate that equinus deformity gait could be a compensatory strategy for plantarflexor weakness. It was hypothesized that children with cerebral palsy utilize an equinus position during gait as a consequence of their weakness.

**Methods**

A simple two-dimensional model for sagittal plane gait was derived to examine the relationships between plantarflexor force, vertical ground reaction force, and the orientation of the foot and tibia with respect to the ground. The model’s development consisted of a two-step process: (1) an inverse dynamics calculation to experimentally determine the resultant ankle joint forces and moments (Figure 1a.
Equinus Deformity in CP

Figure 1 — (a) 2-D free body diagram of the foot used in the inverse dynamics calculation. Ground reaction forces (F_{GRFx} and F_{GRFy}) and moment (M_{Gz}) act at the point of application, G. The force exerted by the foot, F_{foot}, acts through the foot’s center of mass. Through these known values, the resultant joint forces (F_{Ay} and F_{Ax}) and moment (M_{Az}) at the ankle, Point A, can be calculated. Point T is the Achilles tendon insertion. (b) 2-D foot-ankle rigid body model showing geometry and projected moment arm distances D_{1X}, D_{2X}, and D_{3X}.

<table>
<thead>
<tr>
<th>Variable and Definition</th>
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<tbody>
<tr>
<td>f_{pf}: Plantarflexor force generated by gastrocnemius and soleus, acting parallel to the shaft of the tibia</td>
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<tr>
<td>F_{pf}: Perpendicular component of f_{pf}</td>
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<tr>
<td>F_{GRFy}: Ground reaction force (GRF); forces generated by participant’s weight and associated forces due to body’s momentum and accelerations over the force plate</td>
</tr>
<tr>
<td>D_1: Distance between 5th metatarso-phalangeal joint center and lateral malleolus</td>
</tr>
<tr>
<td>D_2: Distance between 5th metatarso-phalangeal joint center and an approx. gastrocnemius/soleus muscle insertion point on posterior calcaneus</td>
</tr>
<tr>
<td>D_3: Distance between posterior calcaneus and lateral malleolus</td>
</tr>
<tr>
<td>Θ_{inc}: Angle between the ground and D_2</td>
</tr>
<tr>
<td>Θ_{abs}: Angle between shaft of tibia and the plane parallel to floor bisecting the talocrural joint center</td>
</tr>
<tr>
<td>α: Offset angle between D_1 and D_2</td>
</tr>
<tr>
<td>D_{1X}: Projected distance of D_1 onto the x-z plane</td>
</tr>
<tr>
<td>D_{2X}: Projected distance of D_2 onto the x-z plane</td>
</tr>
<tr>
<td>D_{3X}: Difference between D_{1X} and D_{2X}</td>
</tr>
</tbody>
</table>
and 1b; also see Appendix), and (2) a set of assumptions that would permit the theoretical distribution of the resultant joint moment to the load-carrying structures crossing the ankle joint. The following assumptions were made:

1. The model focused on the lower leg, the foot, and the vertical forces (ground reaction force and plantarflexor force) encountered during gait. These segments, forces, and their respective points of action could be resolved using a 2-D sagittal plane analysis, therefore a 3-D analysis would be unnecessarily complex and was not used.

2. The foot’s structure was represented as a single 2-D rigid body (Triangle AGT, Figure 1b), acting at Point G fixed to the ground. Joint and soft tissue compression and the effects of the foot’s intrinsic muscles on the dimensions of the triangle were assumed to be small.

3. Horizontal forces acting at Points G and A were considered small as compared with the vertical ground reaction force and plantarflexor force, and thus were not considered.

4. The model was only applicable from mid- and terminal stance to the early stages of preswing, 0.35% and 0.52% of the gait cycle. The forces generated around the ankle during this period were solely due to the plantarflexors (Perry, 1992, 1993). Actions outside this range were not considered.

5. The overall contributions due to inertial effects, forces associated with the weight of the foot, horizontal ground reaction forces, and dorsiflexion moments were negligible as compared with those resulting from the ground reaction force and the participant’s mass (Clauser, McConville, & Young, 1969; Hinrichs, 1990).

With these assumptions, the model described a triangular segment, AGT, rotating about Point G with a vertical ground reaction force acting at that point, and a plantarflexor force acting at Point T, parallel to the long axis of the tibia. The sole load-carrying structure was the Achilles tendon.

The anatomical points (posterior calcaneous, 5th metatarso-phalangeal joint, and lateral malleolus) represented the triangle’s vertices (Figure 1b). Using these points, we determined $D_1$, $D_2$, and $D_3$. In order to describe the spatial orientations of the model during the gait cycle, we derived three angles (Figure 1b). The first, $\Theta_{inc}$, the foot’s inclination angle, was defined as the angle between the ground and segment $D_2$. The second, $\Theta_{tibia}$, the tibia’s absolute angle, was defined as the angle between the shaft of the tibia and a horizontal line within the sagittal plane through the ankle joint. The third angle, $\alpha$, derived from the foot’s actual geometry, was the angle formed by Segments $D_1$ and $D_2$.

Using Assumption 5, the moments due to the model’s rotational motion were zero, therefore a quasi-static analysis equating the torques generated around the ankle were evaluated and simplified to solve for $f_{pf}$, the plantarflexor force was directed through the Achilles tendon (see Appendix). Distances $D_{1X}$ and $D_{3X}$ were the projected lengths of $D_1$ and $D_3$ onto the ground. The ground reaction force, $F_{GRFy}$, acted upon $D_{1X}$, and $F_{PF}$; the vertical component of $f_{pf}$, acted upon $D_{3X}$.

Data were collected from 5 able-bodied participants (1 M, 4 F; ages 28.2 ± 6.9 yrs) recruited from the human performance laboratory and 3 CP participants (4 F; age 5 ± 0) who had participated in previous laboratory testings. All CP participants were independent ambulators who walked on their toes. Since AB persons
did not possess the neurological or physical attributes associated with cerebral palsy and could perform the toe-walking task without assistance, they could act as the study’s control. Our study was concerned with contrasting toe walking in the pediatric population with a fully developed heel-toe walking pattern; therefore the use of adult controls was appropriate and necessary.

To prepare each participant for testing, we measured foot segments D_1, D_2, and D_3. Retro-reflective markers were placed on the lateral fibular head, the lateral malleolus, the 5th metatarsal, and the posterior calcaneus. Regarding the measurement of D_1 and D_2, even though the 1st metatarso-phalangeal joint bears most of the body’s weight, the additional longitudinal distance between the 1st and 5th metatarso-phalangeal joints was experimentally determined to be negligible, therefore the joint associated with the marker placement was used. Only one side of the body was studied.

Kinematic data were captured at 60 Hz using a 6-camera HiRes motion analysis system (Motion Analysis Corp., Santa Rosa, CA) while the participants walked barefoot along a 9-meter walkway. Eight to 10 trials were collected per CP patient. Able bodied participants, mimicking the CP toe walking pattern, performed the same tasks; 18 trials were collected per AB participant. Through each subsequent trial, the AB increased their foot’s inclination angle, Θ_{inc}, and decreased their tibial angle, Θ_{tibia}, assuming a more crouched posture, thus further emulating equinus deformity of increasing severity (Figure 2).

Marker location data were converted to 3-D coordinates as a function of time using Expert Vision analysis (EVA) software (Motion Analysis Corp.); these data were further tracked, verified for continuity, and low-pass filtered to remove any high frequency components. These data were then used to determine Θ_{inc} and Θ_{tibia} (Figure 1b). The angle α was calculated using the law of cosines (Thomas, 1988). The ground reaction force data were obtained from a force plate sampling

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**Figure 2** — Able bodied equinus deformity walking trials. AB were asked to walk progressively demonstrating increasing severity of equinus deformity. (a) AB heel-toe walker; (b) through (d) demonstrate an increase in inclination angle, Θ_{inc}, and a more crouched posture, a decrease in tibial angle, Θ_{tibia}. As Θ_{inc} increased, our ankle model predicted a decrease in plantarflexor force, F_{pf}.
at 1000 Hz. These force data were processed using Kintrak software (Motion Analysis Corp.). All post-processed data were exported directly from EVa and Kintrak to the spreadsheet containing the ankle model.

CP participants were additionally tested to determine plantar- and dorsiflexion strength (Engsberg et al., 2000). They were positioned in a KinCom dynamometer with their ankle joint fixed against the ankle plantar/dorsiflexion attachment. Muscle strength was evaluated while the participant actively plantarflexed against the attachment, which moved at 10°/sec. These data were then compared against age-matched normative values. AB participants were excluded from these tests since age-matched normals already existed (Ross & Engsberg, 2002).

Using the experimental gait data, $\Theta_{\text{inc}}$, $\Theta_{\text{tibia}}$, and $F_{PF}$, we determined the plantarflexor force through the Achilles’ tendon, $f_{pf}$, (Eq. 11). All the data were normalized to the respective participant’s body weight. To help exemplify the differences between the two walking strategies, toe and heel-toe walking, we compared and contrasted the minimum and maximum normalized $f_{pf}$ ($n$-$f_{pf}$) during 0.35% and 0.52% of the gait cycle.

### Table 1  Cerebral Palsy Plantarflexor Strength

<table>
<thead>
<tr>
<th>Participant</th>
<th>Strength (% of AB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>CP #1</td>
<td>55</td>
</tr>
<tr>
<td>CP #2</td>
<td>59</td>
</tr>
<tr>
<td>CP #3</td>
<td>20</td>
</tr>
<tr>
<td>CP #4</td>
<td>65</td>
</tr>
</tbody>
</table>

*Note: Kin-Com strength results from 4 CP children demonstrated plantarflexor weakness as compared with their able-bodied peers.*

### Results

All of the CP participants possessed plantarflexor weakness (Table 1). The processed Kin-Com force results demonstrated that all CP with equinus deformity generated less plantarflexor force ($M \pm SD = 49.7\% \pm 20.2\%$ of AB) as compared with standardized able bodied data (ages 4–8).

The AB participants were asked to mimic equinus deformity toe walking by progressively increasing $\Theta_{\text{inc}}$ through a series of trials. Throughout each subsequent task, all of the AB were able to maintain an elevated $\Theta_{\text{inc}}$ as compared with their previous attempts. Figure 3 represents data from one AB participant that began with heel-toe walking and ended with a maximal plantarflexed toe walking effort; these data were contrasted against one CP walking trial. The lowest curve in the figure, the heel-toe walking result, crossed the x-axis, 0°, due to heel strike where the foot assumed an initial dorsiflexed position, a negative inclination angle, and continued toward plantarflexion, positive inclination angles, and finally toe-off. The remaining AB curves represented toe walking results. The AB’s maximal
Equinus Deformity in CP

Thor infr during gait approached and surpassed that of the CP participant, but the morphology of the AB’s curves differed. All of the AB’s toe walking results maintained a positive slope throughout the region of interest, while the CP’s effort demonstrated a plateau followed by an increase in plantarflexion.

Concurrent with each increase in $\Theta$ infr, the force data from the AB walking trials demonstrated a decrease in $n-f_{pf}$ (Figure 4). Heel-toe walking required the largest force. From the onset of toe walking, Trial 1, a decrease was apparent; the AB maximal effort, Trial 5, required the least plantarflexion force. When compared with the CP’s force data during gait (Figure 4), AB heel-toe walking still required the greatest $n-f_{pf}$, while CP toe walking force, the lowest curve, was less than the AB’s maximal effort, Trial 5. This result was consistent for all of our AB and CP participants; our AB participants could not re-create the CP force profile or inclination angles throughout the gait cycle.

When correlating the AB toe walking trials to their respective maximal $n-f_{pf}$, every able-bodied person demonstrated a decrease in his or her maximal $n-f_{pf}$ with each increase in $\Theta$ infr (Figure 5). Each AB participant traversed a different range of inclination angles, but the relationship between increased $\Theta$ infr and decreased $n-f_{pf}$ remained. The CP did not perform the serial walking tasks, due to their condition; however, their maximum $n-f_{pf}$ values were less than the majority of AB toe walking results. Similar values could be approached or achieved by the AB when utilizing elevated inclination angles.

Collectively the CP toe walking maximum $n-f_{pf}$ was 0.79 vs. 1.2 and 3.4 for able-bodied toe walking and heel-toe walking, respectively (Table 2). Likewise, the CP toe walking minimum $n-f_{pf}$ was 0.31 vs. 1.1 and 2.4 for able-bodied toe walking and heel-toe walking, respectively. While the natural walking strategies
Figure 4 — Normalized plantarflexor force vs. the gait cycle. As the AB participant progressed from heel-toe walking to a maximal plantarflexed effort, the peak $n-f_{pf}$ decreased. Regardless of the time within the gait cycle, the CP presented the least $n-f_{pf}$.

Figure 5 — Inclination angle, $\Theta_{inc}$, vs. normalized plantarflexor force. The AB, performing trials with subsequent increases in $\Theta_{inc}$, demonstrated a concomitant reduction in $n-f_{pf}$. They were able to achieve the CP results at similar or elevated plantarflexion as compared with the CP.
Table 2  Normalized Walking Results From 0.35 to 0.52 of Gait Cycle

<table>
<thead>
<tr>
<th></th>
<th>CP Toe walking</th>
<th>AB Toe walking</th>
<th>AB Heel-Toe walking</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minimum n-f_{pf}</td>
<td>0.31 ± 0.3</td>
<td>1.1 ± 0.5</td>
<td>2.4 ± 0.8</td>
</tr>
<tr>
<td>Maximum n-f_{pf}</td>
<td>0.79 ± 0.6</td>
<td>1.2 ± 0.5</td>
<td>3.4 ± 1.5</td>
</tr>
<tr>
<td>(% of AB heel-toe walking min.)</td>
<td>13</td>
<td>46</td>
<td>–</td>
</tr>
<tr>
<td>Maximum n-f_{pf}</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(% of AB heel-toe walking max.)</td>
<td>23</td>
<td>36</td>
<td>–</td>
</tr>
<tr>
<td>Minimum n-f_{pf}</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(% of AB toe walking min.)</td>
<td>28</td>
<td>–</td>
<td>218</td>
</tr>
<tr>
<td>Maximum n-f_{pf}</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(% of AB toe walking max.)</td>
<td>65</td>
<td>–</td>
<td>283</td>
</tr>
</tbody>
</table>

Note: CP toe walkers had the least normalized plantarflexor force; AB heel-toe walking effort had the largest. AB toe walking led to a decrease in plantarflexor force but it was not equal to that of CP. Also there was an increase between min. and max. plantarflexor force in each case, but AB toe walking effort remained in the narrowest range. AB toe walking n-f_{pf} profile during gait as a % of heel-toe walking showed a decrease in max n-f_{pf}. Conversely, the CP toe walking force profile vs. heel-toe walking showed increased max n-f_{pf} as compared to AB toe walking.

Discussion

The purpose of this study was to demonstrate that equinus deformity gait could be a compensatory strategy for children with cerebral palsy who present with plantarflexor weakness. Through a 2-D inverse dynamics distributed model, it was shown
that toe walking required less plantarflexor force than heel-toe walking. The model made several assumptions: (1) a 2-D sagittal plane, vs. a 3-D force analysis, was comprehensive; (2) horizontal forces and inertial effects could be disregarded; (3) co-contraction did not occur during the region of interest; and (4) the model was only applicable during 0.35% to 0.52% of the gait cycle.

A comprehensive 2-D sagittal plane analysis was sufficient. The plantarflexor forces were assumed to act within the sagittal plane. Laterally directed forces may have been generated by each participant to correct his or her gait. When resolved to the sagittal plane, their contributions to the force vector associated with the Achilles tendon would have been small. In a study with 225 children ages 7–12, it was shown that the mediolateral-directed ground reaction force when normalized for their weight was 0.08 (Engsberg, Lee, Tedford, & Harder, 1993). This was small compared with the maximal \( n-f_{pf} \) found in heel-toe and toe walking. While 0.08 is 10% of the maximum CP plantarflexion result, a 10% increase in their \( n-f_{pf} \) would not change this study’s findings. Additionally, the force contributions due to the moments of inertia of the foot and lower leg were comparable or smaller than those associated with the mediolateral forces (Clauser et al., 1969), therefore their affects were also noted but did not contribute to the results.

The planar resolution of the lower leg and foot was a concern; retroreflective markers could not be placed on each participant in an identical manner, though efforts were made to ensure a level of uniformity. These points of interest did not lie in the same plane, therefore the physiological distances obtained by ruler, \( D_1 \), \( D_2 \), and \( D_3 \), were difficult to measure. Minor deviations in this anatomical data, \( \pm 5\% \) of the actual distance, were shown not to adversely influence our results.

The ankle model also assumed the plantarflexor group generated all forces. It has been shown that co-contraction between the flexor and extensor groups does occur in some CP patients (Brunt & Scarborough, 1988). Conversely, it has also been shown that co-contraction did not occur in a study by Engsberg et al. (2000). Given these contradictory findings, the simultaneous excitation of antagonist/protagonist pairs could neither be accepted nor refuted.

Finally, the model was only valid during plantarflexion. According to Perry, from 0.35% to 0.52% of the gait cycle forces around the ankle were due solely to the plantarflexors. The plantaris, posterior tibialis, long flexors of the toes, and peroneus longus and brevis also influence plantarflexion. (Moore, 1992). The gastrocnemius and soleus generate 80% of the torque around that ankle during plantarflexion (Murray, Guten, Baldwin, & Gardner, 1976). These muscles act along the tibia and insert into the posterior calcaneous via the Achilles tendon, hence the model represents the largest contributing factor to plantarflexion.

It was recently hypothesized that toe walking may have compensatory advantages for individuals with upper motor neuron injury and lower extremity weakness. A study by Kerrigan et al. (2000) involving ballet dancers performing a demi-pointe stance suggested that minimizing the ratio between moment arms around the ankle, \( D_{3X} \) and \( D_{1X} \), decreased the strength requirement. This theory was directly analogous to the inclination angle’s influence on \( n-f_{pl} \). From a first approximation, \( D_{1X} \) would decrease to a larger extent than \( D_{3X} \), therefore the plantarflexor force would decrease. However, the \( n-f_{pl} \) expenditure was also dependent upon its direction of action, \( \Theta_{tibia} \). The coupling of a dedicated strategy maintaining a reduction in \( n-f_{pl} \), and a bias toward an increased \( \Theta_{inc} \) and/or an
increased $\Theta_{\text{tibia}}$ favoring a reduced plantarflexor force requirement, were essential to understanding the CP toe walking pattern. Our analysis demonstrated both of these concepts.

The strategy utilized by the ballet dancers was a learned skill. For the CP participants, this was a lifelong process that may have become a natural response due to the injury to their central nervous system (Davids, Foti, Dabelstein, & Bagley, 1999). Through choreographed efforts, these two groups have been able to achieve a walking pattern which significantly reduced $n-f_{pf}$. Similar to the ballet dancers, the AB participants had strong plantarflexors; however, the AB did not utilize a negative feedback response or learned mechanism which would coerce them to operate at a reduced expenditure of force.

When AB without external training were asked to assume the equinus deformity position, they could not reproduce the CP results. As shown in Table 2, they did demonstrate a decrease in $n-f_{pf}$, but the variation between the AB toe walking minimum and maximum $n-f_{pf}$ was 9%. The AB relied on their increased strength to strictly regulate their $n-f_{pf}$ within a narrow range. Conversely, the natural walking strategies, AB heel-toe and CP toe walking, presented with a greater breadth in $n-f_{pf}$, 154% for AB and 42% for CP. When the AB and CP toe walking results were contrasted against the heel-toe results, the CP had a relative increase in the maximum $n-f_{pf}$ as compared with their minimum, 13% to 23%, while AB toe walking decreased, 46% to 36%.

Analogous to the changes in force seen during CP toe walking, the CP $n-f_{pf}$ profile throughout the gait cycle mirrored the AB heel-toe walking result, while the AB toe walking did not. This demonstrated a concerted strategy, not solely encompassing an increased $\Theta_{\text{inc}}$, to remain in a posture that required decreased force. The CP could effectively adjust the orientation of each anatomical segment, complementing their degree of weakness and minimizing the degree of force and work (Olney, MacPhail, Hedden, & Boyce, 1990; Woodson, Bandy, Caris, & Baldwin, 1995). Using a strategy possessing increased confidence and balance, the novice AB may have been able to walk in a fashion analogous to the CP population. For the ballet dancers in Kerrigan et al.’s study, an increased sense of balance and force conservation was derived from either extensive training or natural instincts (Law & Harvey, 1994). The dancers’ $n-f_{pf}$ result would more closely represent the CP toe walking $n-f_{pf}$ profile during gait. Regardless of these discrepancies, the CP toe walking $n-f_{pf}$ result was less than that of the AB toe walkers throughout the gait cycle.

Children with spastic diplegia and the influence of this on demonstrated strength have been examined on numerous occasions. The present study also used CP participants who had spastic diplegia. While our focus was not directed toward finding a relationship between these factors, several conflicting theories have been proposed regarding the influence of spasticity on CP strength: (1) spastic muscles are strong (Bobath, 1985; Reimers, 1990; Rosenthal & Simon, 1992; Sienko-Thomas, Moore, Kelp-Lenane, & Norris, 1996); (2) spastic muscle weakness is related to the degree of spasticity or pyramidal tract damage (Rab, 1992); and (3) no relationship exists between spasticity and strength, and CP will generate force as needed (Guiliani, 1991; Rose & McGill, 1998). These conflicting assumptions were addressed in a recent comprehensive study on ankle spasticity and strength, and it was shown that no relationship exists between spasticity and strength.
Given these findings, it would appear that our data and results could be used without prejudice to the CP patient’s degree of spasticity.

Our study demonstrated that by assuming postures of increased $\Theta_{inc}$, the required $n_{pf}$ also decreased. Previous work has also shown that with an increase in plantarflexion angle, there was a decrease in the torque generated through both voluntary and externally stimulated muscle contractions (Sale et al., 1982). While there was agreement that increased plantarflexion produces a decrease in torque, with regard to muscle activity, Burnfield et al. (2001) have reported that toe walking requires twice the relative effort as compared to heel-toe walking. This group may have been reporting a nonproductive effort by the participants in their study.

Given the lower leg’s position, the physiological sarcomere lengths and overlapping thick and thin filaments may not be optimal (Guiliani, 1991; Rose & McGill, 1998). This could lead to reduced force generated but heightened muscle activity. Concomitantly, the increased biofeedback to maintain a given posture could produce elevated EMG results, giving the impression of increased effort and work. Contrary to this, it has been shown that individuals with cerebral palsy have decreased plantarflexion work as compared with AB (Engsberg et al., 2000), thus the issue of increased muscle activity relating to strength may not be valid in this case.

Our study demonstrated that by assuming a posture with increased $\Theta_{inc}$, there was a subsequent decrease in required $n_{pf}$. It has also been shown that increases in plantarflexion lead to a decreased ability to generate torque around the ankle. Therefore, assuming a posture that requires a decreased $n_{pf}$ would be beneficial for individuals with plantarflexor weakness or who are not able to generate sufficient torque in order to walk in a heel-toe fashion. Our AB could easily overcome the increased $n_{pf}$ requirements seen with a reduction in plantarflexion angle, and hence employed the learned method used by the majority of the population today, heel-toe walking.

We hypothesized that many CP patients maintain an equinus deformity stance during gait due to plantarflexor weakness. In a weakened state, the equinus position would allow the CP patient to bypass the increased strength requirements associated with a heel-toe gait and to operate within his or her reduced strength limitations. Evidence to support this hypothesis was presented in the strength results which showed that the CP patients were inherently weaker. It was also presented in the plantarflexor results during gait where it was shown that (1) assuming a toe walking posture required less plantarflexor force, and (2) CP patients must employ a dedicated strategy to guard against an increased force requirement. The incorporation of these factors makes toe walking a feasible strategy for children with cerebral palsy who present with plantarflexor weakness.

References


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**Appendix**

Immediately after heel rise (see Figure 2):

\[ F_{PF} \times D_{3x} = F_{GRFy} \times D_{1x} \]  
(1)

where,

\[ D_{1x} = D_1 \times \cos (\theta_{\text{inc}} + \alpha) \]  
(2)

\[ \alpha = \arccos \left( \frac{D_1^2 - D_2^2 - D_3^2}{2 \times D_1 \times D_2} \right) \]  
(3)

\[ D_{2x} = D_2 \times \cos (\theta_{\text{inc}}) \]  
(4)

\[ D_{3x} = D_{2x} - D_{1x} \]  
(5)

\[ F_{PF} = f_{pf} \times \sin \theta_{\text{tibia}} \]  
(6)

Balancing the moments around the ankle, Point A:

\[ M_A = I \times \omega - M_d - (r_d \times F_d) - (r_p \times F_p) + \omega \times (I \times \omega) \]  
(7)
In the quasi-static case, the magnitude of the distal, \( (r_d \times F_d) \), and proximal, \( (r_p \times F_p) \), moments with respect to the center of mass are equal and opposite. Additionally, the inertia component, \( I \), is small when compared to the forces generated.

Therefore,

\[
M_A = -M_d = -F_{PF} \times D_{3x} + F_{GRFy} \times D_{1x}
\]  

Setting, \( M_A = 0 \),

\[
F_{PF} \times D_{3x} = F_{GRFy} \times D_{1x}
\]

\[
f_{pf} \times \sin\Theta_{tibia} \times (D_{2x} - D_{1x}) = F_{GRFy} \times D_{1x}
\]

\[
f_{pf} = \left( \frac{1}{\sin\Theta_{tibia}} \right) \times F_{GRFy} \times \left( \frac{D_{1x}}{D_{2x} - D_{1x}} \right)
\]

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**Acknowledgments**

This study was supported by the National Institute of Neurological Disorders and Stroke (NINDS) of the National Institutes of Health (NIH) Grant R01 NS35830. Thank you to the BJC Human Performance Lab employees and volunteers for their instruction and patience, Suzette Madson and Gretchen Gaylor-Thomas.