

Dynamic Resources Used in Ambulation by Children With Spastic Hemiplegic Cerebral Palsy: Relationship to Kinematics, Energetics, and Asymmetries

Background and Purpose. The atypical walking pattern in children with spastic cerebral palsy is assumed to involve kinematic and morphological adaptations that allow them to move. The purpose of this study was to explore how the requirements of the task and the energy-generating and energy-conserving capabilities of children with cerebral palsy relate to kinematic and mechanical energy patterns of walking. **Subjects.** Six children with hemiplegic cerebral palsy and a matched group of typically developing children participated in the study. **Methods.** Kinematic data were collected at 5 different walking speeds. Vertical stiffness, mechanical energy parameters, and landing angle were measured during the stance phase. **Results.** The affected side of the children with cerebral palsy showed greater vertical stiffness, a greater ratio of kinetic forward energy to potential energy, and a smaller landing angle when compared with those of the nonaffected lower extremity and with those of typically developing children. **Discussion and Conclusion.** Previous research has shown that children with cerebral palsy assumed a gait similar to an inverted pendulum on the nonaffected limb and a pogo stick on the affected limb. Our results indicate that asymmetries between lower extremities and differences from typically developing children in the landing angle of the lower extremity, vertical lower-extremity stiffness, and kinetic and potential energy profiles support the claim that walking patterns in children with spastic hemiplegic cerebral palsy emerge as a function of the resources available to them. [Fonseca ST, Holt KG, Fetters L, Saltzman E. Dynamic resources used in ambulation by children with spastic hemiplegic cerebral palsy: relationship to kinematics, energetics, and asymmetries. *Phys Ther.* 2004;84:344–358.]

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Typical walking can be characterized as being similar to the motion of an inverted pendulum, with exchanges between potential and kinetic energy.^{1,2} As in an inverted pendulum, the body's center of mass reaches its highest point at mid-stance. The kinetic energy (related to speed) is lowest at this point, while potential energy (related to height) reaches its maximum. As the center of mass descends, kinetic energy increases while the potential energy decreases, demonstrating the transference from one form of energy to another. This mechanism is believed to be an optimal way to conserve energy during walking. Part of the energy in each walking cycle, however, is lost due to internal tissue friction and impact during foot contact. In order to keep the system moving, therefore, it is essential to generate energy by muscle contraction at the appropriate moment in each cycle.

Children with spastic cerebral palsy cannot provide enough power through the ankle plantar flexors during

the push-off phase of walking.^{3,4} Without some form of energy being generated or conserved during this phase of walking, a pendular (typical) gait pattern is nearly impossible. To compensate for the energy losses during walking, therefore, alternate energy generation and conservation mechanisms must emerge. Locomotor patterns and morphological changes seen in children with spastic cerebral palsy, such as muscle co-contraction,^{5,6} a plantar-flexed foot at initial contact,⁷ and increased connective tissue within the muscle,^{8,9} may result from an attempt to walk in the absence of normal muscle force. According to this view, the remaining energy-generating and energy-conserving capabilities of these children should be considered the available "dynamic resources" or "dynamic action capabilities," which allow the emergence of an atypical, but functional, walking pattern.^{3,10,11}

The energy-generating dynamic resource is the appropriately timed force produced through muscle contrac-

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All authors provided concept/idea/research design and writing. Dr Fonseca provided data collection and fund procurement. Dr Fonseca and Dr Holt provided data analysis and project management. Dr Holt and Dr Saltzman provided consultation (including review of manuscript before submission).

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tions during walking (impulsive forcing). The energy-conserving resources are the elastic energy conservation (energy return) and the exchanges between potential and kinetic energies.^{3,10} The elastic energy return is provided by tissue stiffness that depends on neuromuscular mechanisms and soft tissue elastic properties. The exchange between potential and kinetic energies is produced by the energy transfers through the pendulum-like properties of limbs and segments operating in a gravitational environment. Thus, the properties of the person—such as body mass, length of the lower extremity, stiffness, force (eg, power produced by the gastrocnemius muscle during push-off)—working in the gravitational environment act to constrain the system and allow for the emergence (or not) of functional behaviors. The gait patterns of children with spastic cerebral palsy may not be typical or “normal,” but, nonetheless, they are often functional in the sense that they achieve the goal of locomotion through the use of available dynamic resources.^{3,11,12}

Although the patterns these children use are functional, they also come at a cost in terms of metabolic energy^{13,14} and the potential for long-term musculoskeletal injury.^{15,16} Rose et al¹³ observed that, during treadmill walking, mean energy expenditure indexes were higher for children with cerebral palsy than for typically developing children. This result is in agreement with reports of high mechanical energy cost and low total energy exchange measured in children with cerebral palsy.¹⁷ These higher costs may reflect the increased vertical displacement of the center of gravity seen when these children walk¹⁸ and the presence of co-contraction during walking. Although the electromyographic profile of typically developing children demonstrates phasic, well-defined bursts of muscle activity in the lower extremity, children with spastic cerebral palsy show a low-level, ongoing muscle activity with marked coactivation of agonist and antagonist muscles.¹⁹

Several investigators^{20–22} have applied biomechanical models in an attempt to understand the dynamics involved in activities such as walking and running. In previous studies examining walking behavior of children with hemiplegic cerebral palsy, we modeled walking as an inverted pendulum with springs that are maintained by a periodically applied force (escapement-driven, damped, inverted pendulum/mass-spring system).^{3,11} The results of these studies demonstrated that children with spastic hemiplegic cerebral palsy showed decreased force production and increased global stiffness in the affected limb compared with the nonaffected limb and compared with the limbs of typically developing children.

In our study, further analyses were performed on the data collected by Fonseca et al.³ We assumed that the

kinematic patterns and asymmetries seen in the gait patterns of children with cerebral palsy are the result of the presence of different dynamic resources (sometimes called “dynamical resources”). These differences in resources, we believe, should be reflected in the child’s kinematics and energetics during walking. Findings from the previous study suggested that the walking of children with cerebral palsy has characteristics of running.³ Possibly, by using greater elastic energy return that comes from a gait such as running, children with spastic cerebral palsy are able to walk at different speeds. Appropriate models for walking gait of children with cerebral palsy, therefore, may be similar to those used in running studies.^{2,23} Although walking relies on exchanges between potential and kinetic energies, running uses elastic energy as the primary means of energy conservation (elastic energy return).²⁰ A stiffer system (as observed in running) is well adapted to conserve elastic energy. The increased stiffness due to increased reflexes, co-contraction,^{5,6} and changes in the mechanical properties of connective and muscular tissues^{8,9} help provide the exchange between elastic and potential (gravitational) energies that characterizes this motion. Because the motion of the center of mass in children with cerebral palsy has a greater excursion in the vertical direction,¹⁸ the stiffness observed in the plantar flexors can be considered to produce the effect of a vertical spring. In addition, the persistent plantar-flexed foot⁷ seems to be compatible with the idea of a vertical spring, because it aids the loading of the elastic tissues of the plantar flexors by the energy transferred from the non-affected lower extremity. Based on our assumptions and on our previous findings, we predicted that the affected lower extremity would show greater vertical stiffness compared with the lower extremities of typically developing children. The use of elastic energy return in the vertical direction, therefore, could be an efficient means of conserving the energy needed for motion in the absence of sufficient muscle force.

A number of other phenomena might be expected of a gait pattern that uses elastic energy return. For example, to adequately load the spring leg (affected lower extremity), children with cerebral palsy could raise the center of mass on the nonaffected lower extremity to increase the amount of potential energy that would be transferred to elastic energy on the affected lower extremity (Fig. 1). By raising the center of mass rather than projecting the body through a “flatter” trajectory (as observed in typical walking), we would predict that the ratio of kinetic forward energy (KE_f) to potential energy (PE) would be reduced in the nonaffected limb.

Another adaptive asymmetry in gaits of children with cerebral palsy that might make an important contribution to the successful accomplishment of walking is the

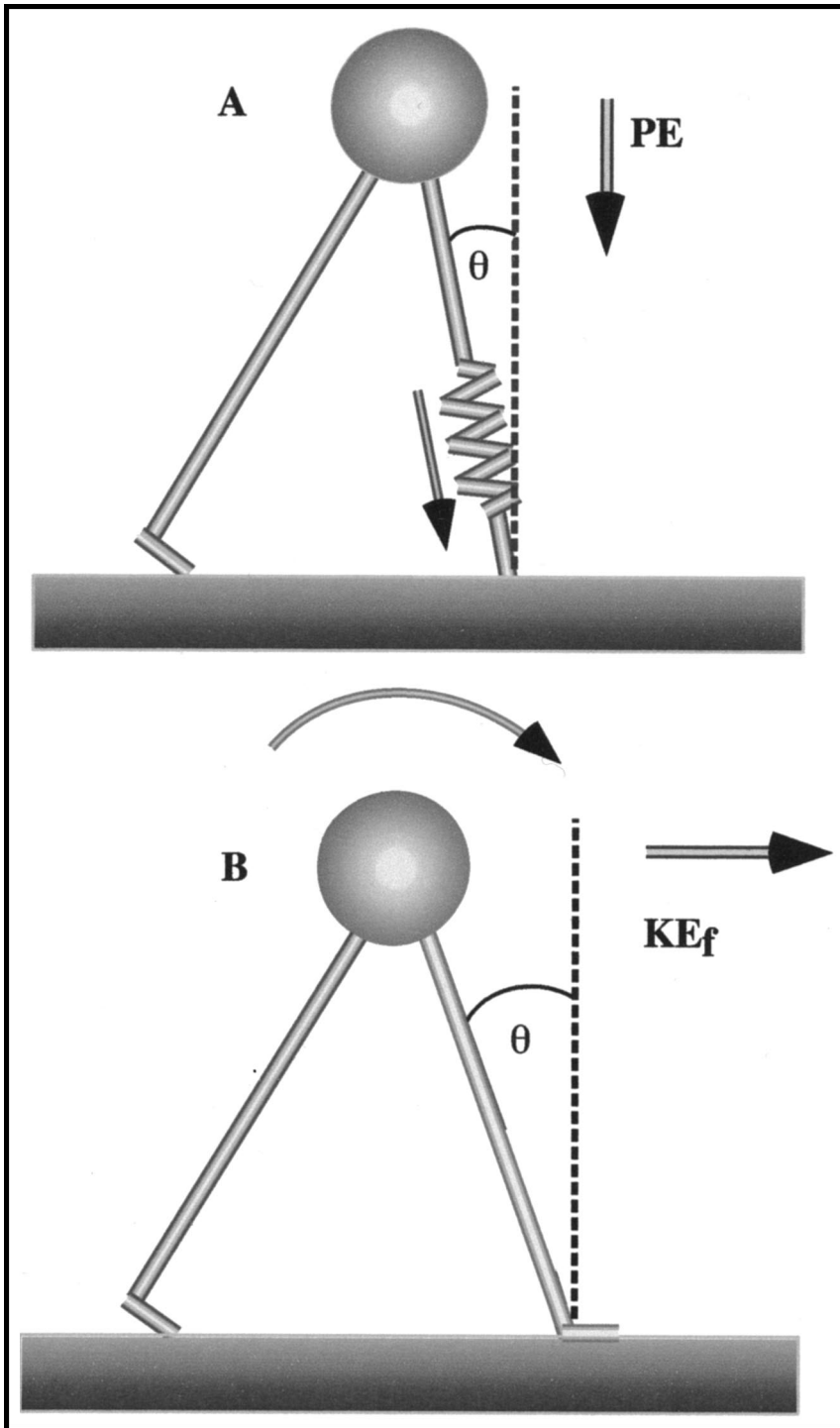


Figure 1. Schematic representation of locomotion in (A) children with spastic hemiplegic cerebral palsy and (B) typically developing children. KE_f =kinetic forward energy, PE =potential energy, θ =landing angle.

presence of side differences in step length. During ground contact, the energy of the stance lower extremity is transferred to the landing lower extremity. The effect of this energy on the landing lower extremity will vary according to the angle at which it is transferred. The

angle formed between a vector connecting the center of mass to the ankle joint and another vector representing the vertical (Y axis) defines the landing angle (Fig. 1). This angle is closely related to step length. If the body is to be used as a vertical spring, we expect that the landing angle of the affected lower extremity may be closer to the vertical, producing a shorter step on the affected side of children with cerebral palsy.

The changes in dynamic resources caused by the neurological insult in children with cerebral palsy are expected to have a predictable impact on their observed walking patterns. Thus, the purpose of our current study was to investigate how the dynamic resources of children with cerebral palsy and the requirements of the task relate to kinematic and mechanical energy patterns of their atypical gait. Based on our previous research and the arguments presented above, the following hypotheses are proposed:

Hypothesis 1: Children with hemiplegic cerebral palsy will show greater vertical stiffness during the stance phase of walking on their affected side compared with their nonaffected side and compared with typically developing children.

Hypothesis 2: Children with hemiplegic cerebral palsy will have lower mean ratios of KE_f to PE of the center of mass during stance phase on their nonaffected sides than on their affected side or than those of typically developing children.

Hypothesis 3: Children with hemiplegic cerebral palsy will have smaller landing angles for their center of mass about the ankle joint on the affected side compared with the nonaffected side and with those of typically developing children.

Method

Subjects

Six children with spastic hemiplegic cerebral palsy and 6 typically developing children participated in this study. The children with hemiplegic cerebral palsy ranged

from 8 to 15 years of age (\bar{X} =10.5, SD =2.8) and the typically developing children ranged from 9 to 12 years of age (\bar{X} =10.67, SD =1.2). Three male and 3 female subjects participated in each group. In the group of children with cerebral palsy, 3 subjects had lesions affecting the left side, and 3 subjects had lesions affecting the right side. Groups were matched as close as possible for age (within 3 years), sex, body weight (within 25%), and height (within 10%). The mean height and weight for the group of children with cerebral palsy were 1.45 m (SD =0.14) and 37.38 kg (SD =13.66), respectively. The mean height and weight for the group of typically developing children were 1.44 m (SD =0.09) and 37.70 kg (SD =10.23), respectively. The children in both groups were recruited from schools, clinics, and hospitals in the Boston area. Detailed data on the children's characteristics were presented elsewhere.³

All children with cerebral palsy had a physician's diagnosis of spastic hemiplegia and were capable of independent walking without crutches, walkers, or braces. In addition, these participants had no history of cardiovascular disease and did not have surgery in the 24 months prior to the study. All participants in the group of typically developing children had no prior history of cardiovascular, neurological, or orthopedic disease. All children were paid for their participation in this study. The rights of the participants were secured, and the children, their parents, or their guardians signed a consent form in accordance with the policies of Boston University Institutional Review Board.

Instrumentation

Two Optotrak 3020 System* position sensors were used to collect 3-dimensional (3D) kinematic data. The sensors were used to capture the position of the infrared-light-emitting markers placed on specific locations on the participant's body. One position sensor was connected in series with the other sensor and then to an IBM-compatible computer through the Optotrak System Control Unit.

Footswitches were constructed using commercially available force-sensing resistors[†] that were linked in series and fitted within custom-made shoe insoles. The footswitches were linked to an electrical circuit²⁴ that produced 3.5 V when the shoe insoles were loaded by the children's weight. The voltage output of the circuit was fed to the Optotrak Analog to Digital Unit that was also connected to the computer through the Optotrak System Control Unit. This setup allowed the synchronization of the kinematic and analog data and provided data

for event identification. Both data were recorded at the rate of 100 Hz.

Photoelectric cells connected to a digital timer provided information about the participant's speed. These photoelectric cells were placed 2 m apart before the beginning of the testing area and were used to monitor the subjects' speed during the test procedures.

Protocol

The protocol of the present study was similar to the one described previously.³ The 2 Optotrak infrared-light position sensors were leveled in the frontal and sagittal planes and placed vertically, 12 m apart, on opposite sides of a walkway. Thus, the position sensors were located to monitor the sagittal-plane motions of the participants bilaterally. This placement allowed a viewing volume of 3.54 m wide \times 2.36 m high. The viewing volume was calibrated by moving a 20-marker rigid body frame in front of the position sensors.

Infrared-light-emitting markers were placed bilaterally on the following locations: head (zygomatic process), shoulder (acromial process), hip (greater trochanter), knee (lateral condyle), ankle (lateral malleolus), and foot (fifth metatarsal head). A physical therapist took measurements of foot, shank, thigh, and total length of the lower extremity using the markers as a reference for the location of the landmarks. In addition, body mass and height were measured in all subjects. The footswitches were inserted into the participants' shoes, and the subjects were asked to walk with them for 1 minute to ensure proper fitting and comfort.

All subjects were initially asked to walk at their comfortable speed on a 20-m walkway in the Department of Physical Therapy at Boston University. The comfortable (preferred) walking speed was the one that the subjects consistently selected on 3 consecutive trials. The subjects were then asked to walk at speeds considered to be fast, very fast (fastest), slow, and very slow (slowest). The actual speed was measured by the photoelectric cells, and the information was used to give feedback to the subjects during the experiment. Each child walked on the walkway at each of the 5 different speeds for at least 10 successful trials. A successful trial was defined as the one in which a full stride (2 successive heel-strikes of the same lower extremity) was captured for each side. The sequence of the testing conditions was randomly assigned during the experiment. Kinematic data were collected during all trials.

Data Reduction

Kinematic data were converted into 3D data by means of the Optotrak system software, using the calibration information for analysis. Filtering consisted of a second-order

* Northern Digital Inc, 103 Randall Dr, Waterloo, Ontario, Canada N2V 1C5.

† Interlink Electronics, 546 Flynn Rd, Camarillo, CA 93012.

Butterworth filter with a cutoff frequency of 6 Hz. After filtering, missing markers were interpolated using a second-order polynomial interpolation. No more than 15 consecutive missing frames were interpolated. Whenever more than 15 frames were missing, a custom-written program was used to apply a cubic spline interpolation, up to a maximum of 30 missing frames. Our previous study³ indicated that the mean errors associated with this interpolation procedure were of 1.4 mm in the Y dimension, 0.7 mm in the X dimension, and 0.5 mm in the Z dimension. Trials with more than 30 missing frames were not included in the analyses.

Gait events such as foot contact and toe-off were automatically identified from the analog footswitch data by a custom-written software program, which was written by the first author (STF) using MATLAB software.[‡] The program located the frames in which an abrupt change in voltage occurred as a result of foot loading (foot contact) and unloading (toe-off). The 3D kinematic data were edited, based on the information provided by the footswitches, to include only full strides for each side. Another custom-written program[‡] was used to estimate the 3D location of the center of mass of the body for each frame. This calculation was based on an 8-segment model with 2 feet, 2 shanks, 2 thighs, head and neck, and trunk. Thus, the presence of any length discrepancy in the lower extremities was considered in this procedure. Anthropometric information from children was used for the calculation of the body's center of mass.²⁵

The filtered and edited files were processed through Northern Digital's DAP software,* generating data for each subject containing the calculated 3D linear displacement, velocity and acceleration of each marker, and center of mass. A custom-written program, written by the first author (STF),[‡] was used to calculate the angle formed between the vector connecting the body's center of mass to the ankle joint and another vector representing the vertical. This angle measured at the moment of foot contact was defined as the extremity's *landing angle* (Fig. 1).

Mechanical energy of the center of mass in the sagittal plane (PE and KE_f) was calculated as described by Winter.²⁶ The analyses used the velocity (*v*) and displacement data for the center of mass during the stance phase. Kinetic forward energy was calculated as the model's mass (*m*) multiplied by the square of the center of mass velocity in the horizontal direction: KE_f=*mv*². The mass of the model (mass of the center of mass moving around the ankle) was calculated as the body mass minus the mass of the stance foot. Potential energy was calculated as the model's mass (*m*) multiplied by the height of the center of mass (*h*) and the gravitational constant (*g*): PE=*mgh*. The height of the

center of mass was calculated as the instantaneous center of mass position in the Y dimension minus the lowest Y position value for that stride. Because the amount of energy depends on the subject's mass, KE_f needs to be normalized relative to the amount of PE to allow comparisons between different individuals (KE_f/PE ratio). Smaller ratios indicate an increased amount of PE during walking.

Vertical stiffness of the center of mass during the step phase was calculated according to Hooke's law (*F=ky*), where *F* represents force and *k* the stiffness and *y* the displacement of the center of mass in the vertical direction. According to Newton's second law, force can be replaced by the product of the system's mass (*m*) by its acceleration (*a*), rendering *ma=ky*. Because the individual's mass remained constant throughout the motion, the vertical stiffness (*k_v*) can be calculated as *k_v=a/y*. Therefore, the slope of a linear regression between the vertical acceleration (*a*) of the center of mass (dependent variable) and the vertical displacement (*y*) of the center of mass (independent variable) during the stance phase was taken as the model's vertical stiffness.

Because of the previously found speed differences between the groups,³ vertical stiffness, KE_f/PE ratio, and landing angle were normalized by the subject's actual walking speed. The subject's walking speed was calculated as the product of stride length and stride frequency.

Statistical Analysis

Univariate analyses of variance (ANOVAs) with 1 between-subject effect (groups) and 2 within-subject effects (extremity and speed condition) were used to test the study hypotheses. The following dependent variables were used: (1) vertical stiffness, (2) KE_f/PE ratio, and (3) landing angle. The level of significance was set at $\alpha=.05$. Whenever a main effect or interaction effect was found to be significant, focused contrasts were performed to locate significant differences according to our proposed hypotheses. Bonferroni correction for multiple comparisons in terms of speed conditions was applied to contrast results, changing the significance level from .05 to .01. Coefficients of variation (CV) were used to describe the intersubject variability of the dependent variables measured in this study.²⁷

Results

Speed Normalized Vertical Stiffness (Hypothesis 1)

Analysis for normalized vertical stiffness demonstrated group ($F=10.12$; $df=1,50$; $P=.0025$), side ($F=6.10$; $df=1,50$; $P=.0165$), and side-by-group interaction ($F=13.85$; $df=1,50$; $P=.0005$) effects. Planned contrast analyses showed that subjects with spastic hemiplegic cerebral palsy had greater normalized vertical stiffness on the affected side than on the nonaffected side in the

[‡] The MathWorks Inc, 3 Apple Hill Dr, Natick, MA 01760-2098.

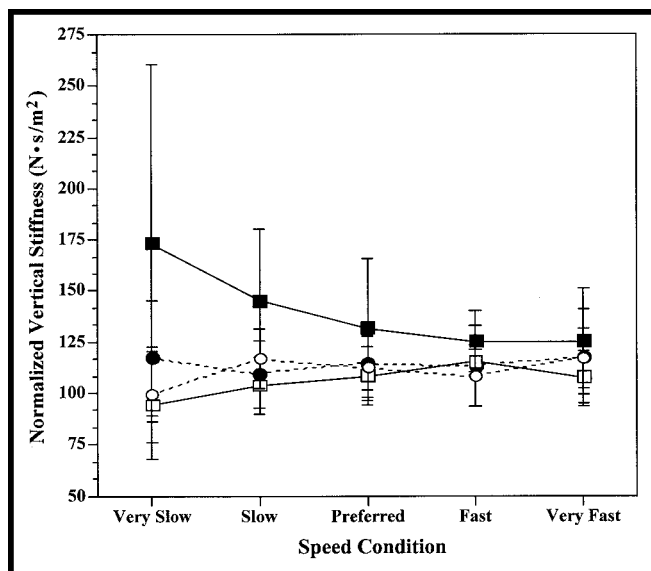


Figure 2. Mean and standard deviation of vertical stiffness normalized for speed for all groups according to speed condition. Data are reported in newton-seconds per square meter ($[(N/m)/[m/s]] = N \cdot s/m^2$). Groups are: (■) the affected lower extremity of children with spastic hemiplegic cerebral palsy, (●) the nonaffected lower extremity of children with spastic hemiplegic cerebral palsy, (●) the nonaffected lower extremity of children with spastic hemiplegic cerebral palsy, (□) the lower extremity of typically developing children that corresponds to the affected (matched) lower extremity of children with spastic hemiplegic cerebral palsy, and (○) the lower extremity of typically developing children that corresponds to the nonaffected (matched) lower extremity of children with spastic hemiplegic cerebral palsy.

very slow ($\bar{X}_{\text{affected}} = 173.32 \text{ N} \cdot \text{s}/\text{m}^2$, $\bar{X}_{\text{nonaffected}} = 117.28 \text{ N} \cdot \text{s}/\text{m}^2$; $P < .001$) and slow ($\bar{X}_{\text{affected}} = 145.11 \text{ N} \cdot \text{s}/\text{m}^2$, $\bar{X}_{\text{nonaffected}} = 109.40 \text{ N} \cdot \text{s}/\text{m}^2$; $P = .009$) conditions. Comparison between the affected side of the subjects with spastic hemiplegic cerebral palsy and typically developing subjects demonstrated higher mean vertical stiffness at the very slow speed ($\bar{X}_{\text{affected}} = 173.32 \text{ N} \cdot \text{s}/\text{m}^2$, $\bar{X}_{\text{typically developing}} = 96.71 \text{ N} \cdot \text{s}/\text{m}^2$; $P < .0001$) and slow speed ($\bar{X}_{\text{affected}} = 145.11 \text{ N} \cdot \text{s}/\text{m}^2$, $\bar{X}_{\text{typically developing}} = 110.26 \text{ N} \cdot \text{s}/\text{m}^2$; $P = .0035$).

The normalized vertical stiffness at the preferred speed condition was not different between subjects with spastic hemiplegic cerebral palsy and typically developing subjects ($P = .06$). No other statistically significant difference was observed between groups and lower extremities. The hypothesis that children with hemiplegic cerebral palsy would have greater vertical stiffness on their affected side compared with the nonaffected side and compared with typically developing children was supported by the results only at slower speed conditions. Figure 2 shows the plot of mean and standard deviations for all groups and conditions.

Coefficients of variation indicated that the intersubject variability in normalized vertical stiffness depended on the speed of walking and the group analyzed. The CV

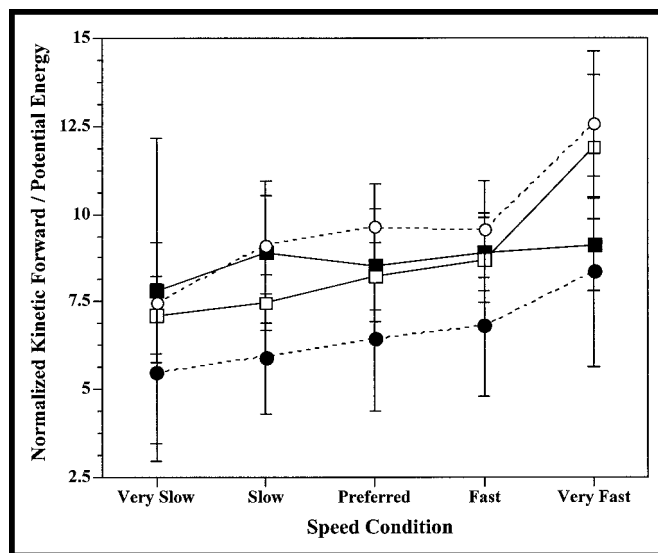


Figure 3. Mean and standard deviation of the ratio of kinetic forward energy to potential energy normalized for speed for all groups according to speed condition. Groups are: (■) the affected lower extremity of children with spastic hemiplegic cerebral palsy, (●) the nonaffected lower extremity of children with spastic hemiplegic cerebral palsy, (□) the lower extremity of typically developing children that corresponds to the affected (matched) lower extremity of children with spastic hemiplegic cerebral palsy, and (○) the lower extremity of typically developing children that corresponds to the nonaffected (matched) lower extremity of children with spastic hemiplegic cerebral palsy.

varied from 50.1% at the very slow speed to 11.8% at the faster speed conditions for the affected side of subjects with spastic hemiplegic cerebral palsy and from 23.8% to 11.7% for the nonaffected side of subjects with spastic hemiplegic cerebral palsy. In the group of typically developing subjects, the CV of the normalized vertical stiffness varied from 28% at the very slow speed to 11.7% at the preferred speed.

Speed Normalized KE_f/PE Ratio (Hypothesis 2)

Results of the ANOVA for normalized KE_f/PE ratio demonstrated a group effect ($F = 12.622$; $df = 1,50$; $P = .0008$), side effect ($F = 8.435$; $df = 1,50$; $P = .0326$), and side-by-group interaction ($F = 39.941$; $df = 1,50$; $P < .0001$). Within-group comparisons demonstrated that subjects with spastic hemiplegic cerebral palsy had a lower normalized KE_f/PE ratio on the nonaffected side ($\bar{X}_{\text{very slow}} = 5.50$, $\bar{X}_{\text{slow}} = 5.92$, $\bar{X}_{\text{preferred}} = 6.44$, $\bar{X}_{\text{fast}} = 6.82$, $\bar{X}_{\text{very fast}} = 8.36$) than on the affected side ($\bar{X}_{\text{very slow}} = 7.82$, $\bar{X}_{\text{slow}} = 8.91$, $\bar{X}_{\text{preferred}} = 8.54$, $\bar{X}_{\text{fast}} = 8.90$, $\bar{X}_{\text{very fast}} = 9.12$) in all but the very fast condition ($P < .009$), indicating a greater contribution of PE on the nonaffected side.

Between-group comparisons of subjects with spastic hemiplegic cerebral palsy and typically developing subjects demonstrated that subjects with spastic hemiplegic cerebral palsy had a lower normalized KE_f/PE ratio on the affected side than typically developing

subjects only in the very fast condition ($\bar{X}_{\text{affected}}=9.12$, $\bar{X}_{\text{typically developing}}=12.3$; $P<.001$). Subjects with spastic hemiplegic cerebral palsy had lower normalized ratios on the nonaffected side ($\bar{X}_{\text{very slow}}=5.50$, $\bar{X}_{\text{slow}}=5.92$, $\bar{X}_{\text{preferred}}=6.44$, $\bar{X}_{\text{fast}}=6.82$, $\bar{X}_{\text{very fast}}=8.36$) for all conditions compared with typically developing subjects ($\bar{X}_{\text{very slow}}=7.29$, $\bar{X}_{\text{slow}}=8.30$, $\bar{X}_{\text{preferred}}=8.97$, $\bar{X}_{\text{fast}}=9.14$, $\bar{X}_{\text{very fast}}=12.23$) ($P<.009$), indicating greater PE contribution at all speeds. These results supported the hypothesis that children with hemiplegic cerebral palsy would show lower normalized KE_f/PE ratios on their nonaffected side than on their affected side and show a lower ratio than typically developing children. Figure 3 presents means and standard deviations for all groups and conditions.

Coefficients of variation indicated that the intersubject variability in normalized KE_f/PE ratios also depended on the speed of walking and group analyzed. The CV varied from 46.4% at the very slow speed to 28.7% at the fastest speed condition for the affected side of subjects with spastic hemiplegic cerebral palsy and from 55.7% to 12.5% for the nonaffected side of subjects with spastic hemiplegic cerebral palsy. In the group of typically developing children, the CV of the normalized KE_f/PE ratio varied from 22.8% at the very slow speed to 11.6% at the preferred speed.

Speed Normalized Landing Angle (Hypothesis 3)

The ANOVA results demonstrated effects according to side ($F=18.201$; $df=1,50$; $P<.0001$) and side-by-group interaction ($F=29.231$; $df=1,50$; $P<.0001$). The results showed that children with cerebral palsy had smaller normalized landing angles on their affected side ($\bar{X}_{\text{very slow}}=26.03^\circ\text{s/m}$, $\bar{X}_{\text{slow}}=13.23^\circ\text{s/m}$, $\bar{X}_{\text{preferred}}=14.44^\circ\text{s/m}$, $\bar{X}_{\text{fast}}=11.10^\circ\text{s/m}$, $\bar{X}_{\text{very fast}}=8.69^\circ\text{s/m}$) than on their nonaffected side ($\bar{X}_{\text{very slow}}=28.95^\circ\text{s/m}$, $\bar{X}_{\text{slow}}=19.87^\circ\text{s/m}$, $\bar{X}_{\text{preferred}}=17.04^\circ\text{s/m}$, $\bar{X}_{\text{fast}}=15.00^\circ\text{s/m}$, $\bar{X}_{\text{very fast}}=11.79^\circ\text{s/m}$) in all speed conditions ($P<.01$), except for the very slow speed ($P=.02$).

Comparisons between the affected side of subjects with spastic hemiplegic cerebral palsy and typically developing subjects ($\bar{X}_{\text{very slow}}=27.74^\circ\text{s/m}$, $\bar{X}_{\text{slow}}=17.87^\circ\text{s/m}$, $\bar{X}_{\text{preferred}}=17.87^\circ\text{s/m}$, $\bar{X}_{\text{fast}}=14.03^\circ\text{s/m}$, $\bar{X}_{\text{very fast}}=11.23^\circ\text{s/m}$) also demonstrated smaller mean normalized angles for the group with spastic hemiplegic cerebral palsy in all conditions ($P<.01$), except for the very slow ($P=.13$) and slow ($P=.012$) speeds. No difference was observed between the nonaffected side of subjects with spastic hemiplegic cerebral palsy and typically developing subjects ($P<.045$). Excluding the very slow speed condition, these results are in agreement with the hypothesis that children with hemiplegic cerebral palsy would demonstrate smaller landing angles of the center of mass about the ankle joint on the affected side

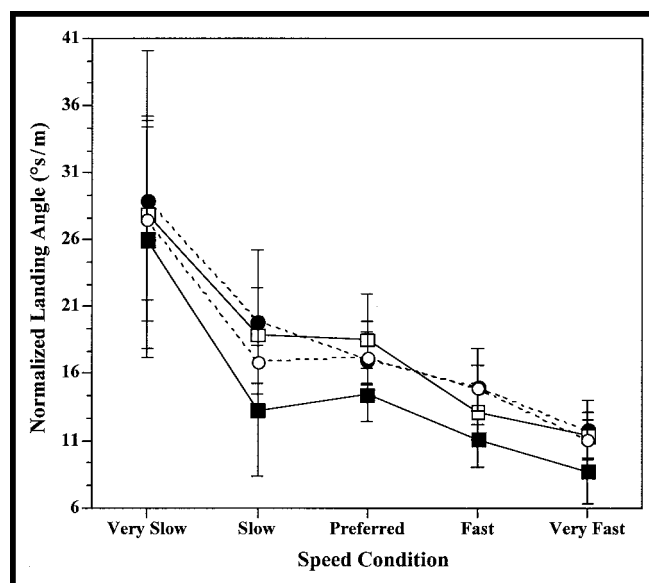


Figure 4. Mean and standard deviation of the landing angle of the center of mass about the ankle joint normalized for speed for all groups according to speed condition. Data are reported in degree-seconds per meter ($^\circ/[m/s]=^\circ\text{s/m}$). Groups are: (■) the affected lower extremity of children with spastic hemiplegic cerebral palsy, (●) the nonaffected lower extremity of children with spastic hemiplegic cerebral palsy, (□) the lower extremity of typically developing children that corresponds to the affected (matched) lower extremity of children with spastic hemiplegic cerebral palsy, and (○) the lower extremity of typically developing children that corresponds to the nonaffected (matched) lower extremity of children with spastic hemiplegic cerebral palsy.

compared with the nonaffected side and compared with typically developing children. Means and standard deviations for all groups and conditions are displayed in Figure 4.

Coefficients of variation in normalized landing angle varied from 33.8% at the very slow speed to 13.8% at the preferred speed for the affected side of subjects with spastic hemiplegic cerebral palsy and from 38.5% at the very slow speed to 18.6% at the fastest speed for the nonaffected side of subjects with spastic hemiplegic cerebral palsy. In the group of typically developing children, the CV of the normalized landing angle varied from 27.8% at the very slow speed to 11.7% at the fast speed condition.

Discussion

We examined locomotion from the perspective that the atypical walking pattern seen in children with cerebral palsy is the product of the interplay of their capabilities and environmental demands. These dynamic resources are the energy-generating and energy-conserving capabilities that a person brings to a task. The results of our study demonstrated several differences between children with cerebral palsy and typically developing children. These differences point to the use of adaptive gait patterns by children with cerebral palsy that allow them

to accomplish the task of walking. Children with cerebral palsy have multiple clinical presentations and varying levels of movement dysfunction.¹⁵ The results of our study are representative of only a small group of subjects with spastic hemiplegia who were capable of independent walking.

According to our findings, there was greater vertical stiffness in the affected lower extremity of children with spastic cerebral palsy compared with the nonaffected lower extremity and compared with the limbs of typically developing children walking at slow speeds. This finding suggests to us that the vertical stiffness in the affected lower extremity of children with cerebral palsy is provided mainly by passive structures (connective tissue) and not modulated through muscular contraction. Connective tissues (tendons and joint capsules) possess an intrinsic stiffness that cannot be reduced. Thus, the difference between the lower extremities of children with hemiplegic cerebral palsy and between groups should be more evident at slower speeds. The inherently stiffer leg allows adequate joint stability²⁸ and means of conserving energy at preferred and higher speeds in children who do not have sufficient force in their ankle plantar flexors. The fact that children with cerebral palsy show a low level of muscle activation during walking¹⁹ lends more support to this idea.

The altered mechanical properties of the connective tissues in children with cerebral palsy may offer them a functional spring that can be exploited.^{9,29} Tendons and other connective tissues are structures adapted to store and release elastic energy.²⁰ The increased tendon-to-muscle ratio seen in the muscles of children with cerebral palsy suggests that they are well adapted to use elastic energy during locomotion.^{29,30} In addition, the persistent plantar-flexed foot position (equinus foot) observed during walking in children with spastic cerebral palsy hypothetically can facilitate a vertical spring by creating a long moment arm at the foot in relation to the ankle, loading the Achilles tendon. We contend that our findings are compatible with the idea that the limb on the affected side of children with cerebral palsy acts like a pogo stick, storing and releasing elastic energy during the early stage of the stance phase of walking.

The finding that no difference in vertical stiffness was observed between children with spastic cerebral palsy and typically developing children at higher speeds suggests to us that, at these speeds, typically developing children start to use the elastic properties of their Achilles tendon in the same way that children with cerebral palsy do. This type of energy use is typical for fast running gaits.³¹ Both groups may assume a gait that exploits elastic energy storage but without the characteristic flight phase of running.²

In order to load the vertical springs of the lower extremity (Fig. 1), energy transfers from one lower extremity to another have to be made, in part, in the vertical direction. In this sense, PE from one lower extremity can be used in the form of elastic energy by the other lower extremity. Group comparisons demonstrate greater amounts of PE in relation to KE_f on the nonaffected side of children with cerebral palsy than on their affected side or than for typically developing children. This finding suggests that children with cerebral palsy take advantage of their intrinsically stiffer system by transferring the increased amount of PE from the nonaffected to the affected lower extremity to “load” the affected lower-extremity spring. Because elastic energy conservation is not directly measured by the techniques we used, this suggestion is speculative, but is compatible with the evidence we presented.

Analysis of the angle of landing of the center of mass about the ankle joint also demonstrated that children with cerebral palsy have smaller landing angles on the affected lower extremity compared with the nonaffected lower extremity and compared with typically developing children, for most conditions. By decreasing the landing angle on the affected side, children with cerebral palsy are decreasing the angular excursion of the lower extremity needed to move the center of mass in front of the ankle axis of rotation and to ensure forward progression. Simultaneously, they are able to transfer the augmented amount of PE to the affected side in a more vertically oriented direction (Fig. 1). This is done by reducing the step length on the affected lower extremity. Relying on its intrinsic vertical stiffness, the affected lower extremity, which is unable to generate the required power for push-off, needs only to return the conserved elastic energy to raise the center of mass along the long axis of the “lower extremity spring.” The nonaffected lower extremity supplies above-normal force³ (force generated by the plantar flexors) to drive the center of mass forward while standing on the affected lower extremity and to compensate for dissipative losses on both sides. The dynamics presented by children with hemiplegic spastic cerebral palsy are similar to a composite inverted pendulum and pogo stick and are compatible with the characteristics of the walking pattern observed in this population.

Despite the ability of these children to take advantage of the properties of their system, children with cerebral palsy have a greater metabolic cost of walking than typically developing children.¹³ Children with cerebral palsy have to actively raise the center of mass to increase PE on the nonaffected side, causing greater vertical motion per horizontal motion of the center of mass during walking. This mechanism results in greater metabolic energy cost per unit distance. The decreased

KE_f/PE ratios on the nonaffected side of children with cerebral palsy supports this assumption (Fig. 3).

Despite the fact that our results supported, in most part, the proposed hypotheses, some limitations of our study must be addressed. First, the small number of subjects in each group makes it difficult to show statistical differences between groups. Therefore, the study is prone to type II errors. Fortunately, the large differences among the means of the groups and the small variability found in the data (except for the very slow speed) allowed for the identification of the hypothesized differences. Second, some of the observed differences in landing angle, mechanical energy, and vertical stiffness might have been caused by differences in the available passive and active range of motion among the subjects' lower-extremity joints. The non-normalized results of the tested variables, however, varied according to the speed condition, showing a progressive increase in the values from the slow speed condition on. This variability supports the idea of exploitation of the available resources by the children rather than the presence of rigid constraints applied to the musculoskeletal system.

From a developmental perspective, children with cerebral palsy and typically developing children present similar patterns when learning to walk.³² As locomotor skills improve, typically developing children develop a walking pattern that has pendular characteristics and a phasing of forces provided by contractions of the ankle plantar flexors.^{33–35} In contrast, children with cerebral palsy retain the characteristics of their early pattern.³² Development of sufficient muscle force or neural substrates that contribute to correctly timed muscle contractions in typically developing children support the emergence of a pendular gait pattern that can take advantage of available resources.³⁶

Conversely, the failure of children with cerebral palsy to develop the typical walking pattern, or pendular motion, might be a consequence of a lack of resources necessary to induce the expected changes in space. Instead, the children with cerebral palsy capitalize on the resources available to them, and, in doing so, they develop patterns and structural adaptations to improve their effectiveness. For example, increases in elastic, noncontractile fibers within muscle tissue may facilitate elastic energy return in those muscles.

Despite the overall small variability of the analyzed data, our results did not show a stereotyped behavior in children with cerebral palsy. The observed gait patterns also might depend on the experiences allowed to the child. As the child explores the environment and learns about his or her available dynamic resources, possible movement pattern solutions emerge to allow a func-

tional interaction of the child with the physical and social environments.

The results of this study and our previous studies suggest that rehabilitation of gait disorders in children with spastic hemiplegic cerebral palsy should aim at understanding the dynamic resources available to perform the task. This approach will permit the therapist 2 possible clinical strategies: (1) improving the existing capabilities of the child to affect his or her performance within the selected pattern or (2) modifying the child's resources in order to allow the emergence of a different pattern solution. For example, a pendular (typical) gait cannot be achieved without some form of force being periodically applied through muscle contraction. In a child with weakened muscles, an alternative solution must emerge to allow him or her to move through the environment. Rehabilitation of the relevant muscles so that the child with weakened muscles is capable of generating sufficient force at the correct time (ie, in a functional manner) may be an effective way to produce a more typical pendular gait pattern. In the context of our study, the relevant muscles are the ones that can generate the required power and stability during walking (eg, ankle plantar flexors and hip flexors and extensors). At present, we are assessing the capability of functional electrical stimulation of the gastrocnemius and soleus muscles applied at the appropriate time in the stance phase of the gait cycle as a way to improve the pattern. For some children, however, the search for a "normality outcome" may not be possible when the capabilities cannot be restored or improved.

Conclusion

In this study, we demonstrated that the affected side of children with hemiplegic cerebral palsy show greater vertical stiffness at slow speeds, and smaller KE_f/PE ratios and smaller landing angles in most speed conditions during walking, when compared with their nonaffected side or with typically developing children. These differences are in agreement with our assumption that the kinematic patterns and asymmetries seen in the gait of children with cerebral palsy result from the manner in which the available dynamic resources are used. We argue that the kinematic details of walking in children with cerebral palsy are the product of the interaction among the environmental demands, the available dynamic resources, and task requirements. In this sense, changes in the children's resources are associated with changes observed in the form that the available energy is explored during walking. Asymmetries in locomotor patterns, therefore, appear to be a direct consequence of the differences in the action capabilities between the lower extremities and the musculoskeletal and neurological adaptations that support those capabilities.

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● Invited Commentary

Let me first congratulate Fonseca and colleagues on an article that advances our understanding of important features of gait in children with cerebral palsy (CP). Moreover, this article highlights the question of how these features emerge during development, which is, of course, a central issue to pediatric neurorehabilitation.

In summary, the authors found that children with hemiplegic CP produced greater vertical stiffness, greater potential energy (K/P energy, also called “PE” in the article), and a decreased landing angle during stance on their more affected extremity compared with the less affected extremity as well as with age-matched, typically developing children. The authors suggest that their data and their previous work¹ are compatible with a novel view of the underlying process by which children with CP develop independent locomotion, namely that these “impairments” are, in part, adaptations by the child to the mechanical and energetic requirements of walking.

I have focused my comments first on the authors’ “selection theory,” as I call it, followed by a few points on the method and results, several of which were touched on in the article’s “Discussion” section.

Selection Theory

Atypical movement patterns in people with pathology of the nervous system, such as those of children with CP, are often viewed as “impairments” imposed primarily by the initial neurological insult. Fonseca et al propose an interesting and important alternative view, namely that these features are adaptive and develop, at least in part, as a result of the child optimizing his or her own system in order to walk. Here the child is viewed as an active participant in a process of exploration and selection of features to match the task requirements of gait such as moving the center of gravity forward during stance.

The authors’ proposal is in line with the exploration and selection principles of modern theories of infant development²⁻⁵ and is a much-needed application of these principles to pediatric neurorehabilitation. I would submit, however, that these particular data are largely ambiguous on whether children adaptively select these features or whether these features are neural and mechanical constraints as traditionally viewed. One assumption underlying the selection theory is whether children with CP have the flexibility within these features from which to explore? For example, earlier in their walking development, did they have a sufficiently wide range of stiffness from which to explore? If so, do they, in fact, select a narrower, task-specific range of stiffness? These simple questions demand complex experiments tracking multiple behaviors in young children with CP

before they become independent walkers. This is not so much a criticism as a reflection of my own impatience with not having definitive support for such an attractive theory.

Vertical Stiffness Calculation

Although mass-spring models continue to provide useful information, there are limitations to these calculations (of which these authors are well aware). First, the authors calculated vertical stiffness based on a spring model that assumes the line of force application and line of displacement are both perpendicular to the ground. During the stance phase of a typical gait pattern, however, this is not always the case. This may be particularly relevant for the atypical gait patterns of children with CP. It would be interesting to compare these findings with other, less restrictive models of stiffness such as that proposed by McMahon and Cheng.⁶ Second, stiffness was calculated as the slope of the linear regression between vertical acceleration and displacement of the center of mass. As a result, there is an assumption of a relatively linear relationship, reflecting a simple spring. Given that the gait of children with CP is atypical, the acceleration-displacement relationship may not be linear. Nonlinearity could affect the meaning of group differences in stiffness, especially if the control group’s stiffness is relatively linear. Lastly, an obvious follow-up question of clinical interest is to what degree do hip, knee, and ankle stiffness contribute to global stiffness? It would be particularly interesting to see the distribution of joint stiffness across both the slower and faster speeds, given that only the former showed group differences.

Changes With Speed

The authors normalized stiffness, K/P energy, and landing angle to walking speed. Given that certain features of the atypical gait of children with CP may well show nonlinear changes over speeds, it would have been interesting to see the non-normalized data. This is relevant given that differences in stiffness were seen only at the slowest speeds and important group differences in K/P energy were seen only at the fastest speed. Moreover, the implications of these group differences are not completely clear, given that children with CP typically walk at their preferred walking speeds where no differences in stiffness or K/P energy were noted for the more affected lower extremity.

Role of Plantar-Flexed Ankle

Children with CP differ from typically developing children in many features of their neuromusculoskeletal system, including passive and active range of motion (ROM). It is not clear what trunk, hip, knee, and ankle ROM was present in the children with CP in this study.

As briefly mentioned by the authors, the concern here is that a lack of passive ROM, such as at the ankle, could account for the changes in global stiffness, K/P energy, and landing angle. With the small numbers of subjects, a few children lacking even a small amount of ankle dorsiflexion ROM may have skewed the group results. This may be important because very different interventions could be envisioned depending on whether the differences in stiffness, K/P energy, and landing angle were associated with a significant limitation in lower-extremity passive or active ROM or produced with full ROM.

The authors state that limitations of passive ROM were probably not a primary factor in group differences in their variables because the non-speed-normalized values scaled up with speed. The relationship between features, such as stiffness and ROM, during functional movements may not be so straightforward. Stiffness, K/P energy, and landing angle are the result of the interplay of multiple joints. Thus, landing angle, for example, could theoretically change with speed even with little ankle movement due to changes at the trunk, hip, and knee. I again encourage these authors to include additional joint components, such as hip and knee stiffness, within their future work on the development of atypical gait patterns as well as the effects of intervention.

The authors propose that both typical and atypical movements emerge from the adaptive interplay of the

nervous system, the body's properties, and the environment. This is the type of rich theoretical grounding necessary to advance modern pediatric neurorehabilitation. I look forward to the authors' follow-up studies and wish them continued success.

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● Author Response

We would like to thank Galloway for his commentary and the opportunity to expand on some of the ideas advanced in the article. We have organized the response to address the specific questions raised by Galloway.

Selection Theory

The problem of selection versus constraint is indeed difficult to test experimentally, and it is confounded by the fact that there are other constraints due to the task demands (eg, speed, accuracy) and by environmental factors that surely influence the movement patterns. Our view is that rather than "selection" of a stiffness or force variable, a movement pattern *emerges* as a function of the interplay of task demands, available dynamic resources (energy-generating and energy-conserving capabilities), and environment. In our view, the child with cerebral palsy learns about (explores) his or her dynamic resources within a given environment (eg, uneven surfaces, slippery floors, underwater) and under specific task demands (eg, walking, running, jumping). On each occasion, a number of viable patterns are available. The pattern that emerges (rather than being selected) is

optimal from the perspective of stability, metabolic cost, and mechanical efficiency. Thus, although the child explores his or her mass, strength, stiffness, or body segment lengths, the "selection" occurs at the level of movement pattern. The stiffness observed in children with cerebral palsy is the product of muscle and connective tissue properties that developmentally change according to their patterns of use.^{1,2} In this sense, stiffness itself is a property that cannot serve as the basis for a selection process. Patterns of movements, however, provide enough diversity and, consequently, selection theory can be applied.

The major difference between our proposal and the traditional view of neural and mechanical constraints is that we look at the resources the child brings to the task and understand the observed movement pattern, not as an abnormal feature of a damaged central nervous system, but as a viable solution given the capabilities for force generation and conservation (dynamic resources) possessed by the child. In this view, the negative "constraint approach" is substituted by a positive perspective

that seeks to understand the observed movement pattern as an adaptive behavior that facilitates the development of, and is facilitated by, the available resources.

Vertical Stiffness Calculation

As pointed out, we calculated stiffness as the slope of the linear regression between vertical acceleration and displacement of the center of mass, assuming a linear relationship as expected in a linear spring. The regression plots of these 2 variables revealed a fairly linear relationship, with R^2 varying from .61 to .96. This relationship was stronger in children with cerebral palsy than in normally developing children. Some of the plots of vertical acceleration by vertical displacement of the center of mass demonstrated that the observed curves were typical of a vertical stiffness produced by a nonlinear "hard spring." However, even these plots had a large linear portion in the middle of the curve. Despite these possible problems, similar results were obtained with different forms of stiffness calculations. In a study of people carrying backpacks while walking at different speeds, Holt et al³ showed that knee joint vertical and global stiffness measures were equivalent.

Although the contribution of the lower-limb joints to the overall stiffness might have clinical interest, the model proposed cannot be used to identify the stiffness distribution across the joints. Due to the large variability of severity, types of impairment, and topographic distribution of altered tone observed in children with cerebral palsy, it would be very difficult to show a specific pattern of stiffness distribution across lower-limb joints. In addition, individual differences in energy-generating and energy-conserving capabilities (dynamic resources) among children with cerebral palsy may result in a different stiffness contribution of each joint, but reveal a similar overall behavior.

Changes With Speed

Data published previously on global stiffness and impulsive force demonstrated a linear relationship between these variables and speed, but with different slopes.⁴ In our study, the magnitude of the variables was a function of speed in both groups. Because there was a speed difference between groups, we believe normalization to walking speed was necessary to allow meaningful comparisons between groups or limbs. Our results varied according to the speed of walking. However, at the preferred speed, the only variable that did not show an expected difference, according to the proposed hypotheses, was vertical stiffness. This lack of difference in vertical stiffness at the preferred speed ($P=.06$) might have been influenced by the small sample size used in our study. In other studies,^{4,5} it has been shown that at the preferred speed children with cerebral palsy demonstrated greater stiffness of their affected limbs when

compared with typically developing children. In relation to differences in potential energy (K/P energy) at the preferred speed in the more affected lower extremity, it is important to note that, according to the model presented, no difference was expected between groups. A difference in K/P energy, however, was observed at all speeds in the less affected lower extremity, as we had hypothesized. Therefore, our results seem to explain the walking pattern of children with cerebral palsy at the preferred speed.

Role of Plantar-Flexed Ankle

We believe Galloway is correct when he states that differences in stiffness, K/P energy, and landing angle could be the result of changes in range of motion (ROM), stiffness, or ability to generate force at the trunk or any joint of the children's lower extremity. From a traditional therapeutic perspective, one could argue that we must treat these impairments in order to minimize the functional limitation. Many of the therapeutic measures the medical profession uses do exactly that. Examples include serial casting and tendon-lengthening surgeries. If a ROM limitation is indeed the primary cause of a gait deviation in a person with cerebral palsy, then identifying that limitation would be important. Conversely, if changes in pattern (including, for example, increased ability to generate force or power in the noninvolved limb, increasing joint stiffness, or adaptive ROM limitations) are compensations for inadequate resources (eg, lack of muscle force needed to overcome gravity or to provide joint stability), as we suspect, then treating an individual component may not be appropriate, because it is an attempt to remove an adaptation to the underlying cause (the changed dynamic resource).

As children become older, the adaptations may indeed become structural in the sense that, for example, the triceps surae muscle-tendon length ratio is reduced and there is more connective tissue embedded in the muscle.⁶⁻⁸ Nevertheless, we would argue that the benefits of treating individual components by, for example, tendon lengthening to get improved ROM at the ankle are short-lived at best and that the procedures are more likely to cause pain, weakness, or other gait deviations. Our suggestion is that early intervention at the level of dynamic resources can avoid many of the structural limitations seen after years of adaptive use of particular structures in particular patterns.

We used the term "landing angle" instead of the more easily understood term "step length" because we contend that this variable could be the result of several body motion combinations, as it depends on the behavior of the center of mass. Thus, despite small differences in the kinematic details of the walking pattern of children with cerebral palsy, a simple biomechanical model that takes

into consideration the dynamic resources available to these children can capture their overall waking behavior. In this sense, interventions should not focus only on specific limitations in lower-extremity passive or active ROM, but on providing the dynamic resources needed to better accomplish a functional behavior.

We thank Galloway for his insightful and thoughtful review of our work. We hope that the responses and arguments presented here may help clarify our understanding of the mechanisms underlying functional performance in children with cerebral palsy.

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