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Article Title: The Gait of Children with and without Cerebral Palsy: Work, Energy, and Angular Momentum

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The Gait of Children with and without Cerebral Palsy:

Work, Energy, and Angular Momentum

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ABSTRACT

This paper describes a method to characterize gait pathologies like cerebral palsy using work, energy, and angular momentum. For a group of 24 children, 16 with spastic diplegic cerebral palsy and 8 typically developed, kinematic data was collected at the subjects self selected comfortable walking speed. From the kinematics, the work-internal, external, and whole body; energy-rotational and relative linear; and finally the angular momentum were calculated. Our findings suggest that internal work represents 53% and 40% respectively of the whole body work in gait for typically developed children and children with cerebral palsy. Analysis of the angular momentum of the whole body, and other sub groupings of body segments, revealed a relationship between increased angular momentum and increased internal work. This relationship allows one to use angular momentum to assist in determining the kinetics and kinematics of gait which contribute to increased internal work. Thus offering insight to interventions which can be applied to increase the efficiency of bipedal locomotion, by reducing internal work which has no direct contribution to center of mass motion, in both normal and pathologic populations.

Keywords: Pathological gait, Biomechanical Modeling, Inverse Dynamics, Gait Analysis, Internal External Work, cerebral palsy, pediatrics.
INTRODUCTION

Spastic cerebral palsy has been qualified as a motor disorder characterized by a velocity dependent increase of the stretch reflexes (Lance, 1980) resulting from a non-progressive lesion of the developing brain. While the gait of children with cerebral palsy (CP) has been found to be more heterogeneous than that of typical developed (TD) children, there are certain features that are stereotypical patterns of gait. Children with CP typically have shorter strides, slower preferred walking speed (Pierson et al., 2008; Bernardi et al., 1999; Tuzson et al., 2003), reduced range of motion of the joints, and altered ankle moment patterns as a result of the lack of a true heel strike (Gage, 1991; Perry, 1992).

The alterations in the gait patterns of children with CP commonly result in altered energetics in walking compared to typically developed children. The gait of children with diplegic CP typically requires two to three times greater metabolic energy and mechanical work (Waters & Mulroy, 1999; Unnithan et al., 1999), to travel the same distance and is less pendular with reduced energy recovery, 45% (Bennett et al., 2005). In general the examination of mechanical work has focused on the total or external work; defined as the work to move the body center of mass, with little attention to the internal work; defined as work associated with motion relative to the center of mass. The one exception is the work of van den Hecke et al (2007b) that reported increased internal work in a population with “spasticity and hemiplegic CP” relative to controls.

External work only explains approximately 50% of the work done in walking in typically developed adults (Winter, 1990; Saibene, 1990; Donelan et al., 2002).
et al (1995) included internal work, the work done to move body segments about the center of mass, in his analysis of walking and found that for normal adults the internal work done in walking equaled the external work at self selected comfortable walking speed. Van den Hecke reported that internal work during gait in children with hemiplegia was approximately 86% of the corresponding external work magnitude (van den Hecke et al., 2007a).

The computation of the mechanical work of walking remains a controversial topic. The computation of total work as the sum of the internal and external work has been disputed on theoretical grounds (Zatsiorsky, 1998; Aleshinsky, 1986). However, no current method accounts for the actual effort associated with ambulation given the issues concerning energy transfer between segments and limbs, elastic energy storage and return, co-contraction, or the efforts of biarticulate muscles (Sutherland, 2005). Despite the shortcomings, there remains interest in internal and external work, especially in groups with walking disabilities, (Detrembleur et al., 2003; Detrembleur et al., 2005; Frost et al., 2002; Unnithan et al., 1999; Bennett et al., 2005) as it can help to identify the phases of gait where disabilities affect efficiency. In the present study the use of internal work, work relative to the center of mass, is especially appropriate because the angular momentum is typically also computed relative to the center of mass.

The angular momentum of a person during locomotion has received little attention. Recent work has suggested that angular momentum, like energy is “conserved to a large extent” and could be a control parameter in walking, similar to the zero moment point (ZMP) control often exploited to generate stable gait patterns in robotics.
(Hirai et al., 1998; Komura et al., 2005). Yet, there has been little data presented to validate this claim and no one has explored angular momenta or the implications with respect to biomechanics, motor control, and energy of gait in typically developed children or children with CP. Simoneau and Krebs (Simoneau & Krebs, 2000) reported similar total body angular momenta about the center of mass for an entire gait cycle between elderly fallers and non-fallers, but had a total of only five subjects. Kaya et al. (Kaya et al., 1998) also studied elderly subjects and reported peak angular momentum values for the whole body and HAT (head, arms and torso). Kaya et al. (Kaya et al., 1998) found differences in the maximum HAT angular momenta in the sagittal and lateral planes between healthy elders and elders with bilateral vestibular hypofunction. Both studies found the whole body angular momentum normalized by body mass (m^2 rad/s) was small (<0.03).

Recently, Gu (Gu J W, 2003) showed that whole-body angular momentum, in relation to its center of mass, is highly regulated throughout the walking gait cycle in healthy adults (N=5). Popovic et al. (Popovic et al., 2004; Popovic & Englehart, 2004) extended that work. The most complete analysis, to date, of angular momentum in walking was performed by Herr and Popovic (2008). They used a 16 segment model to describe the angular momentum of 10 individuals walking at their self-selected speed. For the most part they confirmed their earlier work that angular momenta be highly regulated and with a small absolute value.

The purpose of this study was to investigate the use of angular momentum as a quantitative metric for gait analysis. We hypothesized that the internal work represents a
significant portion of the overall work of both normal and pathologic gait. We further hypothesized that total body angular momenta would be greater in children with compared to typically developed children. Finally we hypothesized that the angular momentum of the body can be used as an indicator to quantify the contribution of individual limbs and segments to the total internal work done over a gait cycle. Our results reveal that angular momentum offers new insights into the dynamic motion of walking in children with CP.
METHODS

Subjects

Kinematic data on a convenience sample of 24 children were collected and analyzed. This group of children consisted of representative samples of two populations. The first group of age-matched controls was comprised of 8 children without known musculoskeletal, neurological, cardiac, or pulmonary pathology, and included 2 females and 6 males averaging 10.3±1.9 years of age, 141.8±12.9 cm in height, and 39.8±14.7 kg in mass. The second group consisted of 16 children diagnosed with spastic diplegic CP. These subjects were community ambulators, GMFCS level I-II with a mean score of 92±7% on the Gross Motor Function Measure (Russell et al., 1993) and walked without aids. They included 7 females and 9 males averaging 10.1±4.1 years of age, 135.5±29.4 cm in height, and 34.4±16.9 kg in mass. None of the subjects had undergone surgery or other significant treatments within the 12 months prior to being tested and all subjects with CP walked with some degree of equinus gait, i.e. they had no true heel strike. For both groups no intra-gender differences were noted. All tests were conducted in the Motion Analysis and Motor Performance Laboratory at the University of Virginia. Subject assent and parental consent was approved by the University of Virginia’s Human Investigation Committee and was obtained for all subjects.

Subjects were instructed to walk barefoot along the 10m laboratory walkway at their self selected comfortable walking speed (CWS). Three-dimensional kinematic data were collected using a six camera Vicon Motion Analysis System (Oxford Metrics, UK) at 120 Hz, and full body 38 marker set. At least three trials were performed with two to
five strides in each trial. Of the captured trials, the trials with an average walking speed within 10% of the subject’s mean self selected comfortable walking speed were averaged and then analyzed, for all subjects a minimum of 5 strides were averaged and used for analysis.

Subject Specific Model

A 17 segment 16 joint patient specific model was created for each subject in MSC.Adams, using the LifeMod plug-in developed by Biomechanics Research Group (San Clemente, CA), from individual anthropometric data (age, weight, height, and gender). The 17 model segments were the: head, neck, upper torso, central torso, lower torso, upper arms (2), lower arms (2), hands (2), upper legs (2), lower legs (2), and feet (2). The segments’ physical properties (mass, length, and moment of inertia) were defined using the Generator of Body Data (GeBOD) database (Cheng et al., 1994), GeBOD includes anthropometric databases for adults, males and female, and also children based on height, weight, and age. The 16 joints were each specified as tri-axis hinge joint arrangements, however, the elbow, and wrist joints are reduced to two axis and the knee joints to only one. The model was developed in a global coordinate system that had the X axis positive in the direction of travel; the Y axis positive in the vertical, up, direction; and the Z axis positive to the models right. Measured marker positions were exported from VICON to the LifeMod model, and kinematic analysis were performed with the model motion developed to match the measured motion. Kinematic data were broken into individual gait cycles, beginning and ending on left foot ground contact, and low pass filtered at 4 Hz with a two way butterworth filter. Stride data were then splined
to 100 data points, representing 100% of the gait cycle, using a cubic spline (Matlab, The Mathworks Natick MA).

**Angular Momentum**

The angular momentum of a body is a conserved vector quantity that represents an object’s motion about either a fixed point in space, or associated systems center of mass. Angular momentum of limb, with moment of inertia tensor $I$ and rotation velocity vector $\omega$, about its local center of mass is defined as:

$$L_{LocalCoM} = I\omega$$

(1)

The moment of momenta resulting from the segments motion relative to the total body center of mass was calculated as:

$$L_{MomentOfMomentum} = r_{i/g} \times m_i v_{i/g}$$

$$r_{i/g} = \text{vector from CoM to segment}$$

$$v_{i/g} = \text{velocity of segment relative to CoM}$$

$$m_i = \text{mass of segment} i$$

(3)

The total vector angular momentum about the center of mass is:

$$L_{Total} = L_{LocalCoM} + L_{MomentOfMomentum}$$

(4)

Note that because this is a tensor we are able to look at the component about each global axis independently. Angular momentum data were made unitless via normalizing by, body mass, height, and average walking speed.

**Work/Energy**
The total energy of a system in each of the three cardinal axes were calculated independently and can be described via the following method developed by Willems (Willems et al., 1995):

\[
E_{Total} = MgH + \frac{1}{2} MV_g^2 + \sum_{i=1}^{n} \left( \frac{1}{2} m_i v_{g,i}^2 + \frac{1}{2} I_i \omega_i^2 \right)
\]

for segments \( i = 1 \rightarrow n \)

The first two terms represent the “external energy” or the energy state of the center of mass as it moves through space and is based only on the motion of the composite total body center of mass. The two other terms represent “internal energy” or the energy of the limbs as they move about the center of mass and are based on the rotational motion of each segment about its own center of mass and the linear velocity of the segments center of mass relative to the composite center of mass. For the purpose of analysis the internal energy resulting from the relative linear velocity of each segment center of mass with respect to the total body center of mass will be referred to as the relative energy.

Work is the total change in energy, and was calculated following the method of Willems (Willems et al., 1995). Intra-limb energy transfer was allowed between both adjacent and nonadjacent segments of a limb, i.e. the thigh, shank, and foot of a leg, with no transfer between limbs, no energy transfer was allowed across planes.

\[
W_{\text{wo/transfer}} = |\Delta PE| + |\Delta KE|
\]
\[
W_{\text{w/transfer}} = |\Delta PE + KE|
\]

**Moment, work, energy relation**

Maintaining the directionality of work/energy by not allowing cross planar energy transfer facilitated the development of a statistical relationship between momentum and
work/energy. For this analysis relative energy curves were used since they represent over 90% of the internal energy and can be split into principle directions for more direct comparison to the tensor momenta values.

The motivation for comparing the momentum and internal energy can be seen in their mathematical definitions (Eqns 1, 3, and 5). Analysis revealed that the relative linear terms ($v_{i/g}$) dominate both the energy and momentum in gait, as shown in results section. Thus the equations can be reduced to $L = r_{i/g} \times m v_{i/g}$ and $E = \frac{1}{2} m_i v_{i/g}^2$ where we see that momentum is proportional to the velocity based rate of change of energy. For comparison of work (change in energy) and momentum we calculate the absolute momentum area over a gait cycle (the integral of the momentum curve over a gait cycle). For stable walking the average angular momentum about the body center of mass must be zero about each axis. The absolute momentum area represents the average deviation of the momentum magnitude from the zero mean required for stable gait.

Statistical Analysis

Between group comparisons of the dependent measures were made with an independent single tailed t-test of differing variances with $\alpha=0.05$. Shapiro-Wilk’s W tests were conducted to verify normality of the data. Effect size, $\theta = \frac{mean_{CP} - mean_{TD}}{stdev_{CP}}$ was also computed between group means, standard deviation values from the group with cerebral palsy were used in all data presented as they resulted in more conservative (smaller) values. Multiple linear least squares regression analyses were conducted to determine any relationship between the angular momentum curves and the relative
energy curves over a full gait cycle. A cross correlation over a gait cycle was conducted to relate time dependent momentum of the swing leg in the X and Y directions with energy in the Z and X directions respectively for typically developed and CP subjects. All analyses were accomplished using the software program Statistica 5.1 with resulting confidence intervals and R² values reported.
RESULTS

When walking at their comfortable walking speed, children with CP did 59% more total work per unit mass and distance traveled than the typically developed group, 1.64 J/kg-m and 1.03 J/kg-m respectively, P<0.01, $\theta=1.61$. For subjects with CP external work (0.99 J/kg-m) was 102% greater than typically developed children (0.49 J/kg-m), P<0.01, $\theta=1.40$. This external work represented 60% of this total work in children with CP while only representing 47% of the total work for typically developed children. The balance of the total work was the internal work or the work associated with the motion of body segments about the body center of mass. While walking, children with CP tended to do 21% more internal work than typically developed children, P=0.053, $\theta=0.70$. For children with CP this resulted in 0.66 J/kg-m of internal work being done vs. 0.54 J/kg-m for typically developed children, 40% and 53% of the total work respectively.

For both population samples, segment momentum about the body center of mass was dominated by the relative linear velocity term (Eqn 3). Similarly the motion of a segment relative to the body center of mass accounted for more than 92% of the change in total internal energy over a gait period for both groups (Figure 1). It was also determined that children with CP walked at a higher (21%) mean normalized total body relative energy level (Eqn 5) than typically developed children, 0.132 J/kg-m and 0.109 J/kg-m respectively. Since relative linear energy accounts for most of the internal energy and can be represented as the sum of three directional values and thus related to the vector angular momentum quantities, energy results following will be reported as relative linear energy.
For both groups the average angular momenta were small for the entire gait cycle. The angular momenta of the total body about the center of mass had similar general trajectories trends for children with CP and the controls (Figure 2, column 1). However, there were noteworthy differences. About the Z axis, the absolute area under the curve, momentum area, was 73% larger in the group with CP, P<0.01, θ=1.00 (Figure 2, column 1). The momenta curve of the children with CP lagged that of the typically developed children by 31°, p<0.01, θ=0.87. There was large variation of the momentum area about the X axis, reflecting the side to side leaning motions of the body, and it tended to be larger in the group with CP by 50%, p=0.058, θ=0.49 (Figure 2, column 1). The extrema occurred just before the swing foot left the ground and were about twice as large in the children with CP. The momenta about the vertical, Y, axis were small and similar for the two groups.

Other differences between the CP and typically developed groups were found when comparing angular momentum properties of the upper and lower body (Figure 3, columns 1-2 respectively). The CP group’s upper body momenta area were larger about both the X and Z axes, P<0.05, θ=0.64 and P<0.01, θ=0.93 respectively (Figure 3). This corresponded to a trend of 52% larger mean work associated with the motion of the upper body for the CP group, 0.0986 J/kg-m compared to the typically developed group, 0.0646 J/kg-m, though the difference was not statistically significant, P=0.18, θ=0.42. Also of note was the phase difference about the Z axes, where the CP group led the typically developed group by 75° about the Z axis, P<0.01, θ=1.12. The lower body data revealed that subjects with CP had a greater momenta area about the Y and Z axis that the
typically developed group, both axis P<0.01, θ=1.09, θ=0.91 respectively. This reflected the trend towards 16% greater work associated with the lower body motion of the CP group, 0.562 J/kg-m compared with typically developed mean work of 0.481 J/kg-m, P=0.06, θ=0.59. The lower body momentum phase was different in all three axis, X, Y, and Z, with the CP population, lagged 18° in the X axis, P<0.01, θ=0.65, and lead 7° in the Y axis, P<0.05, θ=0.58, and lead 18° about the Z axis, P<0.05, θ=0.48. For both groups the angular momenta area was larger for the lower body, relative to the upper body, about both the Y and Z axis, P<0.001, θ>1.82 for all cases, while there was no difference for the CP group about the X axis, the momenta was slightly larger about the X axis for the typically developed group, P<0.05, θ=0.81 (Figure 3).

Narrowing the focus of the analysis to single limbs and limb segments offers more insight into elementary spatial movements. The swing leg of children with CP had larger angular momentum about all three axes (X P<0.01, Y P<0.01, and Z P<0.05) but particularly about the X and Y axis during swing, i.e. 10-55% gait (Figure 4). However, the associated internal work of the swing leg over an entire gait cycle showed little overall difference between groups (CP 0.263 J/kg-m, TD 0.234 J/kg-m, P=0.15). Analysis of the angular momentum of the body trunk, calculated as the sum of the individual contribution of the head, neck, and torso, resulted in differing angular momentum area and internal work between groups. The momentum area was larger in the CP group about all three axis (X P=0.027 θ=1.07, Y P<0.022 θ=0.42, Z P<0.01 θ=0.59) and the CP group required more work per stride, 0.0299 J/kg-m vs 0.0110 J/kg-m.
The angular momenta and internal work of the arms were similar for both groups.

Multiple regression analysis revealed a correlation between angular momentum and work in walking for both children with CP and typically developed children. In total body analysis the relative energy in the X and Z directions correlated with the associated momentum for both groups, TD: $R^2=0.75$ and 0.95 respectively, $P<0.0001$, and CP: $R^2=0.88$ and 0.72 respectively, $P<0.0001$ (Figure 2). There was a poor correlation of the relative energy in the Y direction, TD: $R^2=0.33$ $P<0.0001$ and CP: $R^2=0.09$ $P<0.005$.

Results for the swing leg were similar to the whole body results (Figure 4). Analysis of the relative energy in the X and Z axes correlated to the associated momentum for both groups, TD $R^2=0.91$ and 0.82 respectively, $P<0.0001$, and CP $R^2=0.95$ and 0.78 respectively, $P<0.0001$ with a poor correlation for the relative energy in the Y direction, TD: $R^2=0.33$ $P<0.0001$ and CP: $R^2=0.01$ $P<0.05$. The cross correlation analysis revealed that for typically developed children the momentum about the X axis and energy in the Z direction had a correlation coefficient of $C=0.8334$ and $C=0.7943$ for momenta about the Y axis and energy in the X direction. Children with CP showed correlations $C=0.7871$ for momenta about X and energy in Z direction and $C=0.9196$ for momenta about the Y axis and energy in the X direction.
DISCUSSION

This study developed the use of internal and external work along with angular momentum as methods of analyzing gait in typically developed children and children with cerebral palsy. Furthermore it laid the groundwork for the use of angular momentum to quantify the contribution of body segments to the total internal work required for locomotion. The angular momentum methods described in this paper are often implemented in the analysis and control of robotics (Hirai et al., 1998; Komura et al., 2005), their practical application to the analysis of human movement have just recently begun to be investigated.

Values of work, obtained in this study, compared well with published data. Willems (Willems et al., 1995), Frost (Frost et al., 1997), and Mian (Mian et al., 2006) reported total work values from 0.72-1.70 J/kg-m of total work compared to the 1.03 J/kg-m for our typically developed group. They reported external work of 0.35-0.72 J/kg-m, while we measured 0.49 J/kg-m for typically developed children. For internal work we found 0.54 J/kg-m for the typically developed children, while they found 0.35-0.50 J/kg-m for internal work for adults. Studies of children report 0.23-0.54 J/kg-m for external work (Bennett et al., 2005; van den et al., 2007). While our results lie within the range of reported values it is clear that results vary widely between studies. Part of the variation in reported values can be attributed to the varying strategies for calculating the transfer of energy; cross-plane, intra-limb, extra-limb etc, and also the models used to represent the body. These varied from simple 7 segment models to the higher dimensional 17 segment model used in this study. Models with more degrees of freedom
allow the analysis to account for more of the total body motion by not grouping segments together resulting in individual segment motion being lost. This is important in calculating the internal work.

The work values for children with CP also varied widely. For the group with CP our results fell within the published ranges for, total body work 0.8-1.67 J/kg-m (Unnithan et al., 1999; van den et al., 2007) and external work 0.46-1.31 J/kg-m (van den et al., 2007; Bennett et al., 2005). Only one other study has investigated the internal work of children with CP. Van den Heche, et al (van den et al., 2007) reported internal work values of 0.39 J/kg-m well below our results of 0.66 J/kg-m. However their data were for hemiplegics, which due to their strong side are typically considered a higher functioning group than the diplegic children in our study. They also utilized a model consisting of 7 segments, which naturally results in a lower value for internal work.

While the internal work for the CP population represents a lower percentage of the total work in gait (compared to typically developed children) it is extraneous work that does not contribute to the forward motion. As such it deserves analysis as changes to the motion kinematics associated with internal work may reduce the overall energy expenditures without impeding the forward progress desired in walking.

The angular momenta of both typically developed and CP children exhibited similar trends, both on a global scale, i.e. total, upper, or lower body, and at a more segmental scale, individual limbs, or segments. Total body momenta in both the frontal and sagittal planes of children with CP tended to exhibit greater excursion and also tended to be larger, i.e. larger moment area. This larger moment area reflects the
increased average deviation of the angular momentum from the mean of zero required for stable gait, and is indicative of the exaggerated whole body motions in both the frontal, increased lateral rocking for foot swing ground clearance, and sagittal, longer duration of double support, planes of CP gait. As the momenta analysis was refined to look at smaller groupings its representation of specific kinematics became more obvious. Cancelation of the momenta about the Y axis between the upper and lower body for both groups revealed contra lateral gait. Similarly the negative extrema of the lower body momenta about the Z axis are markers for double support and reflect the forward motion of the body CoM relative to the legs which were fixed to the ground. For children with CP the excessive momenta of the legs about both the X and Y axis (Figure 4) represent circumduction of the leg/foot during swing, a common adaptation employed to increase foot clearance in equinus gait. Angular momenta of the arms about the Y axis was timed and scaled to partially cancel the momenta generated by the lower body. Increase momentum about the X axis of the torso (Figure 4) in swing manifests itself as increased relative energy change in the Z direction. This motion represents another accommodation for equinus gait where the subject leans to one side in order to lift the hip of the swing leg to increase foot clearance during swing. Momenta demonstrate that children with CP employ the same general motion kinematics as typically developed children but can also identify where their motion trajectories differ from those of typically developed children.

The relationship between momenta and work were used in an effort to quantify the contribution of gait kinematics to internal work, much as the determinants of gait
Regression analysis demonstrated that for CP and typically developed children, the momenta of the leg correlated to the relative energy and facilitated the quantification of energy used in their motion. In the case of the swing leg of children with CP (Figure 4) there was excessive momentum about the X and Y axes during swing, 10-50% of gait cycle, when compared to typically developed children. The increased momenta correlated with increased relative energy in the Z and X directions respectively. In CP gait these increases in momenta were indicative of circumduction, or excessive lateral motion, of the leg for foot clearance during swing, a commonly employed technique to accommodate for equinus. In this case the presented correlation allows clinicians to associate a quantifiable amount of excess energy being used for a quantifiable amount of circumduction. This represents a new ability to relate movement associated with pathologic gait to direct increased energy costs.

Models developed with limited segments and degrees of freedom often result in limitations for simulations. A one degree of freedom joint used to represent the knee
represents such a limitation as it precludes cross plane transfer of energy, however most energy in gait occurs in the sagittal plane so cross plane transfer represents a small deviation in simulated results. In addition the relation between momenta and work/energy limits our conclusions due in part to the nature of the quantities compared i.e. scalar vs. vector quantities, the nature of energy transfer between body segments, and the cancelation of momentum between segments as they move together. Rotational motion about a point in a plane can be described as the result of two in plane orthogonal linear motions. Comparison of planer rotational properties such as momentum with linear properties such as relative energy is difficult due to linear motions being functions of rotational motion in two orthogonal planes. Despite these limitations we were still able to use momenta to quantify the specific effects gait kinematics on internal work/energy.
CONCLUSION

In analysis of gait and the metabolic cost of locomotion, internal work offers a rich area of analysis which will improve the understanding of the mechanisms underlying gait efficiency, this may then lead to improved clinical interventions to improve ambulation for increasing efficiency of walking. This is particularly relevant when looking at pathological gait such as that associated with CP since it represents motion with no direct application to forward motion. In addition, the angular momenta about the body center of mass offers insight into the dynamic motions associated with internal work. However, we see that the angular momenta are well organized within both groups, suggesting that changes in the momenta of an individual segment will result in changes in other momenta. While global analysis offers indirect insight into gait, a more comprehensive, multi-dimensional analysis of body limbs or segments offers a better cause/effect relation between gait kinematics and work.

Future work on these topics is already progressing; evaluation of the synergistic relationship of angular momentum primitives is being used to quantify the body’s regulation of angular momentum in walking (Thomas et al., 2009). Also gait modeling is adopting angular momentum as a strategy for the development of stable gait simulations, including zero moment point control, momentum feedback loops, and reduced dimension controls capitalizing on the consistency of momentum primitives. A better understanding of angular momentum as it applies to the kinematics and energy can only increase our knowledge and understanding of human gait.
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Massachusetts Institute of Technology.


Figure 1. Total body internal energy and its two components, rotational and relative, for both typically developed children and children with CP. Note that for both test groups the relative internal energy dominates the total body internal energy.
Figure 2. The angular momentum of the total body, the upper body, and the lower body for both TD children (solid line) and children with CP (dashed line) about the three principle axis, X frontal, Y transverse, and Z sagittal plane. Non-dimensional momentum was normalized by height, mass, average velocity, and is reported with respect to percent gait cycle.
Figure 3. Non-dimensional total body angular momentum about the X, Y, and Z axis over a gait cycle for both TD children (solid) and children with CP (dashed). The relative energy component of internal energy along the three orthogonal axis is presented as mJ/kg-m.
Figure 4. Non-dimensional angular momentum of the right (swing) leg about the X, Y, and Z axis over a gait cycle for both TD children (solid) and children with CP (dashed). The relative energy component of internal energy along the three orthogonal axis is presented as mJ/kg-m.