Original Research

# Predictors of Walking Efficiency in Children With Cerebral Palsy: Lower-Body Joint Angles, Moments, and Power

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**Background.** People with cerebral palsy (CP) experience increased muscle stiffness, muscle weakness, and reduced joint range of motion. This can lead to an abnormal pattern of gait, which can increase the energy cost of walking and contribute to reduced participation in physical activity.

**Objective.** The aim of the study was to examine associations between lower-body joint angles, moments, power, and walking efficiency in adolescents with CP.

**Design.** This was a cross-sectional study.

**Methods.** Sixty-four adolescents aged 10 to 19 years with CP were recruited. Walking efficiency was measured as the net nondimensional oxygen cost (NNcost) during 6 minutes of overground walking at self-selected speed. Lower-body kinematics and kinetics during walking were collected with 3-dimensional motion analysis, synchronized with a treadmill with integrated force plates. The associations between the kinematics, kinetics, and NNcost were examined with multivariable linear regression.

**Results.** After adjusting for age, sex, and Gross Motor Function Classification System level, maximum knee extension angle ($\beta = -0.006$), hip angle at midstance ($\beta = -0.007$), and maximum hip extension ($\beta = -0.008$) were associated with NNcost. Age was a significant modifier of the association between the NNcost and a number of kinematic variables.

**Limitations.** This study examined kinetic and kinematic variables in the sagittal plane only. A high interindividual variation in gait pattern could have influenced the results.

**Conclusions.** Reduced knee and hip joint extension are associated with gait inefficiency in adolescents with CP. Age is a significant factor influencing associations between ankle, knee, and hip joint kinematics and gait efficiency. Therapeutic interventions should investigate ways to increase knee and hip joint extension in adolescents with CP.

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Cerebral palsy (CP) arises due to damage to the developing brain.[1] Children with CP experience muscle weakness and decreased joint range of motion (ROM), which can alter gait patterns and lead to a greater energy cost of walking compared with their typically developing peers.[2,3] This inefficient pattern of walking in CP can lead to increased fatigue[4] and contribute to reduced participation in activities of daily living.[5] The association between abnormal gait patterns and a greater energy cost of walking in CP has not been well established. Identifying these associations could have important clinical implications and help to inform evidence-based therapeutic interventions.

Abnormal strength and joint ROM are key features of gait inefficiency in CP.[6–8] For example, muscle and joint contracture, contributing to a reduced joint ROM, is mainly caused by an imbalance between the agonist and the antagonist muscles in CP, and this has been shown to contribute to a higher oxygen cost of walking.[9,10] However, research concerning the associations between lower limb joint ROM and strength is sparse and demonstrates conflicting relations between muscle strength, gait parameters, and walking efficiency. For example, Kramer and MacPhail[11] reported that energy expenditure indices were modestly correlated with knee extensor strength but not knee flexor strength in adolescents with CP. This contrasts with more recent studies showing that reduced ankle ROM and reduced knee flexor strength are associated with a higher energy expenditure index and poor gait efficiency in those with spastic CP.[7,8]

Eek et al[6] found that although children with CP exhibited weakness in all lower limb muscles, this was only associated with reduced push-off moments and power around the ankle during gait. This suggests that weakness of other lower limb muscle groups, such as those at the knee and hip, could also affect the output at the ankle. These findings were supported by Pouliot-Laforte et al,[12] who found that only ankle joint power was correlated with energy expenditure index whereas knee and hip power were not.

This current body of research highlights the need to investigate the associations between all lower limb joints and walking efficiency in children and adolescents with CP. There are currently few data on the relationships between joint angles, moments, and joint power at the ankle, knee, and hip and walking efficiency in CP. Studies that have investigated these associations tend to be limited by small sample sizes and the use of energy expenditure index as a surrogate measure of efficiency. Therefore, the aim of this cross-sectional study was to examine the associations between lower-body joint angles, moments, power, and walking efficiency. The results could inform therapeutic interventions that increase gait efficiency among people with CP.

# Methods

## Participants

Sixty-four adolescents with CP were recruited from the National Health Service, schools, and organizations for disability in England. Inclusion criteria were: (1) diagnosis of spastic CP and age 10 to 19 years; (2) able to walk independently with or without a mobility aid (ie, Gross Motor Function Classification System [GMFCS] levels I–III); and (3) able to activate the ankle plantarflexors. Exclusion criteria were: (1) orthopedic surgery of the lower extremities in the past 12 months; (2) botulinum toxin type A injections in the past 6 months; (3) serial casting in the past 6 months; and (4) insufficient cognition to comply with the assessment procedures. The assessments of the present study were undertaken as part of the baseline assessment for a randomized controlled trial to determine the feasibility, acceptability, and efficacy of resistance training for adolescents with cerebral palsy.[13] All participants aged 16 years or older provided written informed consent. Participants less than 16 years old provided written informed assent, and parents/guardians additionally provided written consent. This study had ethical approval from Brunel University London's Department of Clinical Sciences' Research Ethics Committee and a National Research Ethics Service Committee London-Surrey Border.

## Measures

**Gait efficiency.** Gait efficiency was measured using a protocol outlined by Schwartz.[14] Participants' body mass to the nearest 0.1 kg and height to the nearest 0.1 cm were measured. Oxygen consumption was recorded using a portable metabolic system (K5; Cosmed, Rome, Italy) during a 10-minute rest period in a semirecumbent position and then during 6 minutes of overground walking at a self-selected speed. Participants' walking speed (m/s) was measured in 1-minute intervals. Prior to each test, the Cosmed K5 was calibrated for oxygen and carbon dioxide concentrations and turbine flow according to the manufacturer's recommendations. Participants were asked to fast for 4 hours prior to the test and wore their usual footwear and orthotic devices. Where required, use of a walking aid was permitted.

Gait efficiency was reported as net nondimensional oxygen cost (NNcost) because this has been shown to be the most appropriate measure of gait efficiency in children with CP.[14–16] Two minutes of steady-state data were used in the calculation of NNcost. The first 2 minutes of rest and walking, respectively, were removed. Using the remaining data a 2-minute period that demonstrated least variation in oxygen uptake ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$), and with a respiratory exchange ratio (RER) of less than 1, was identified for rest and walking, respectively.[16] The average oxygen uptake values ($\dot{V}O_2$, mL/min) and RERs were computed over the steady-state periods. Equation 1 was used to convert average $\dot{V}O_2$

during resting and walking from milliliters per minute to joules per second:

$$[(4.90 \times \text{RER} + 16.040) \times \dot{V}\text{o}_2(\text{mL/ min})]/60. \quad (1)$$

NNcost was calculated by using Equation 2:

$$\text{NNcost} = [(\dot{V}\text{o}_{2\text{walk}} - \dot{V}\text{o}_{2\text{rest}})/v] \cdot (1/mg)\,, \quad (2)$$

where $\dot{V}o_2$walk is oxygen consumption during walking (J/s), $\dot{V}o_2$rest is oxygen consumption during resting (J/s), $v$ is walking speed (m/s), $m$ is body weight (kg), and $g$ is acceleration due to gravity (9.81 m/s).

**Lower-body walking kinetics and kinematics.** Kinematic and kinetic data during treadmill walking were collected using a computerized motion capture system (Motion Analysis, Motion Analysis Corporation, Santa Rosa, CA, USA) with 8 infrared cameras and a treadmill with dual integrated force plates capable of capturing X, Y, and Z force components (Bertec Fully Instrumented Treadmill, Bertec Corporation, Columbus, OH, USA). The Motion Analysis software Cortex was used for the synchronized capture of kinematic data (150 Hz) and force data (2100 Hz).

Joint kinematic and kinetic data were measured from a full-body motion analysis marker set using the Conventional Gait Model. For this purpose, 31 reflective markers were fixed bilaterally on the skin at the following bony landmarks: orbit and occiput, C7 spinous process and proximal sternum, bilaterally at the distal end of each clavicle, anterior superior iliac spine, sacrum, the greater trochanters, midthigh, medial and lateral femoral epicondyles, on each tibia (midway between the ankle and knee), the medial and lateral malleoli, the heads of the first and fifth metatarsals, and the calcanei. A calibration trial was conducted prior to testing. Each participant was asked to adopt the upright standing position on the treadmill and a neutral position (baseline) was recorded while the participant stood upright.

Prior to recording experimental trials, participants performed a familiarization session to ensure they were comfortable walking on the treadmill, and to establish a comfortable preferred walking speed. During the walking trials, participants were barefoot and instructed to lightly hold onto the handrail to maintain balance; however, they were requested to maintain their normal upright posture and have full body weight on their feet during standing and walking. Each participant began by walking at a relatively slow speed, and treadmill speed was slowly increased until the participant reported that the current speed was faster than the preferred speed. The treadmill speed was then slowly decreased until the participant reported that the current speed was slower than preferred. This procedure was repeated 3 times and the average of the 3 "faster" and 3 "slower" than preferred speeds was taken as the participant's preferred walking speed.[17] This procedure also allowed treadmill familiarization and participant warm-up. Participants were asked to walk on the treadmill at their preferred walking speed for 2 minutes while kinematic and kinetic data were recorded.

All kinematic and kinetic calculations were performed using Visual3D v.5 software (C-Motion, Inc, Germantown, MD, USA) and the C-Motion Product Documentation guidelines (http://www.c-motion.com). The kinematic and kinetic variables were calculated using the V3D default calculations (the Conventional Gait Model). Marker coordinates and force data were filtered with a zero-lag, fourth-order Butterworth filter with a cutoff frequency of 6 Hz. A 6-degrees-of-freedom Visual 3D–linked model, including 9 body segments (head, thorax/abdomen, pelvis, right and left thigh, right and left shank, and right and left foot), was assigned to motion data. Joint angles were computed as the angles between the proximal and distal segment of the relevant joint. The line connecting the segments was calculated by finding the midpoint between medial and lateral markers. Specifically, ankle angle was measured as the angle between the line connecting the malleoli and the metatarsals (ie, foot segment), and the line connecting the malleoli and the epicondyles (ie, lower leg segment). Knee joint angle was measured as the angle between the line connecting the malleoli and the epicondyles (ie, lower leg segment), and the line connecting the epicondyles and the hip joint (ie, thigh segment). Hip joint angle was measured as an angle between the line connecting the epicondyles and the hip joint (ie, thigh segment), and the line connecting the hip joints and the acromions (ie, trunk segment). Net joint moments were computed as the proximal moment on the segment at the relevant joint. Joint powers were computed as scalars by multiplying joint moment by angular velocity. Joint angles, moments, and power were calculated for ankle, knee, and hip joints. The gait events were identified based on the ground reaction forces as "foot on," "foot off," and "midstance." Foot initial contact ("foot on") was calculated as the first frame at which ground reaction force exceeded 20 N (selected because of the noise that moving treadmill belts introduce into the vertical GRF). "Foot off" was calculated as the first frame at which GRFs were less than 20 N. The "midstance" was calculated as the frame corresponding to a midpoint between the "foot-on" and "foot-off" events.

We were specifically interested in joint angles and moments at midstance where the body mass is supported by a single leg demonstrating lower limb ability to support the body and extend in an antigravity position. We were also interested in: (1) maximal achievable joint extension angle during gait, describing the amount of joint extension lag; (2) peak plantarflexion moment, describing the ankle's ability to propel the body forward; and (3) joint peak positive and negative power, as a measure of energy generation or absorption at the joint. The following variables were extracted from the kinematic analyses: (1) ankle, knee, and hip joint angles at midstance (ie, single leg support); and (2) maximal ankle, knee, and hip extension angles. The following variables were extracted from the kinetic analyses: (1) ankle peak plantarflexion moment (ie, ankle push-off moment); (2) ankle, knee, and hip moments at midstance; (3) ankle, knee, and hip joint peak positive power (ie, power generation) and peak negative power (ie, power absorption) of 1 gait cycle. Because of a high interindividual variation in the joint extension lag at a specific gait instance (ie, inability to achieve anatomical knee position during the midstance, due to knee being flexed or hyperextended), the data have either a positive (dorsiflexion, knee and hip hyperextension) or a negative (plantarflexion, knee and hip flexion) sign. A value of "0" represents the anatomical position in standing. All variables were extracted for the more-impaired leg; joint angles and moments are reported in a sagittal plane about a frontal axis (flexion-extension). The averaged values of 3 gait cycles are reported for all variables. The moment and power values were normalized to participants' body mass.

### Statistical Analysis

The distribution of data was explored using Q-Q plots, histograms, and cross-tabulations. Age was treated as a continuous variable in all analyses. Variables were described using means, SDs, ranges, interquartile ranges, counts, and frequencies, where appropriate. To examine the unadjusted associations between NNcost, kinematics, and kinetics, respectively, Pearson product-moment correlation coefficients were calculated. Following this, multivariable linear regression models were fitted to explore the association between kinematics and kinetics (independent variables) and NNcost (dependent variable), after adjusting for age, sex, and GMFCS level. Separate models were fitted for each kinematic and kinetic variable, respectively. We chose to adjust for age, sex, and GMFCS level, by including these as independent variables in each of the models, because age, sex, and GMFCS level are likely confounders of associations between kinematics and NNcost, and between kinetics and NNcost, respectively. We further explored age and sex as potential effect modifiers by including interaction terms between age and each kinematic or kinetic variable, and between sex and each kinematic or kinetic variable, respectively, in each model. Finally, to investigate which variables best predicted NNcost, we included variables that were associated with NNcost at $\alpha = .10$, as identified from Pearson correlation coefficients, in a multivariable linear regression model. Assumptions of linear regression, namely a linear association between independent and dependent variables, homoscedasticity, and normally distributed errors, were explored using scatter plots, Q-Q plots, and histograms. All analyses were performed in Stata version 15.0 (StataCorp LLC, College Station, TX, USA).


### Role of the Funding Source

The funding source had no role in the design of this study, its execution, analyses, interpretation of the data, or decision to submit results.


## Results

### Participant Characteristics

Of the 64 participants recruited to the study, 6 did not have data for NNcost. Descriptive statistics are reported for the remaining 58 participants, including 28 (48.3%) participants at GMFCS level I, 23 (39.7%) at level II, and 7 (12.1%) at level III. Participants were aged 10 to 19 years (mean [SD] age = 13.7 [2.6] years); mean height was 154.3 [13.1] cm and mean body mass was 49.3 [13.6] kg. Twenty-five (43.1%) of the study participants were female, 31 (53.5%) participants had unilateral CP, and 27 (46.5%) had bilateral CP. The number of participants with data on other variables ranged from 43 to 58. NNcost, gait speed during treadmill walking, and kinematics and kinetics are reported in Table 1.

### Unadjusted Associations Between NNcost, Kinematics, and Kinetics

Correlation coefficients between NNcost, kinematics, and kinetics showed that NNcost was associated with knee and hip angles at midstance and maximum knee and hip extension angles so that NNcost decreased with more extended knee and hip angles ($r = -0.52$ to $-0.59$; $P < .001$; Tab. 2). There were no other associations between NNcost, kinematics, and kinetics, respectively.

### Associations Between NNcost, Kinematics, and Kinetics Adjusted for Age, Sex, and GMFCS Level

Multiple linear regression analyses showed that after adjusting for age, sex, and GMFCS level, maximum knee extension angle ($\beta = -0.006$; 95% CI = $-0.012$ to $-0.0005$; $P = .034$), hip angle at midstance ($\beta = -0.007$; 95% CI = $-0.013$ to $-0.001$; $P = .028$), and maximum hip extension ($\beta = -0.008$; 95% CI = $-0.014$ to $-0.003$; $P = .003$) were associated with NNcost (Tab. 3). There was no evidence that any other kinematic or kinetic variables were associated with NNcost.

### Effect of Age and Sex on Associations Between NNcost, Kinematics, and Kinetics

There was no evidence that the association between any kinematic or kinetic variable and NNcost differed according to sex ($P > .05$). However, there was strong evidence that age modified the association between NNcost and a number of kinematic variables. The following regression coefficients were identified for interaction terms between age and (1) ankle angle at midstance ($\beta = 0.003$; 95% CI = 0.001–0.006; $P = .017$), (2) maximal ankle plantarflexion ($\beta = 0.003$; 95% CI = 0.001–0.005; $P = .013$), (3) knee angle at midstance ($\beta = -0.002$; 95% CI = $-0.004$ to $-0.001$; $P = .002$), (4) maximum knee extension angle ($\beta = -0.002$; 95%

**Table 1.**
Descriptive Statistics for Gait Efficiency, and Joint Angles, Moments, and Power[a]

| Variable | Mean [SD] | n |
|---|---|---|
| **NNcost** | 0.53 [0.28] | 58 |
| **Treadmill gait speed, m/s** | 0.44 [0.15] | 55 |
| **Joint angles (°)** | | |
| Ankle angle at midstance | − 14.08 [8.56] | 48 |
| Max. plantarflexion angle | − 28.32 [10.30] | 48 |
| Knee angle at midstance | − 8.64 [13.32] | 48 |
| Max. knee extension angle | − 4.93 [12.61] | 48 |
| Hip angle at midstance | − 9.25 [11.39] | 47 |
| Max. hip extension angle | 0.82 [12.14] | 47 |
| **Joint moment, Nm/kg** | | |
| Peak ankle moment | − 0.33 [0.39] | 44 |
| Ankle moment at midstance | 0.32 [0.38] | 44 |
| Knee moment at midstance | 0.64 [0.45] | 44 |
| Hip moment at midstance | − 0.71 [0.54] | 44 |
| **Joint power, Nm/s/kg** | | |
| Ankle generating power | 0.67 [0.84] | 44 |
| Ankle absorbing power | − 0.55 [0.88] | 44 |
| Knee generating power | 1.28 [1.47] | 44 |
| Knee absorbing power | − 1.06 [0.84] | 44 |
| Hip generating power | 0.82 [0.84] | 44 |
| Hip absorbing power | − 1.23 [0.86] | 44 |

[a]NNcost = net nondimensional oxygen cost.

CI = −0.004 to −0.0003; $P$ = .022), (5) hip angle at midstance ($\beta$ = −0.002; 95% CI = −0.004 to −0.0004; $P$ = .013), and (6) maximum hip extension angle ($\beta$ = −0.002; 95% CI = −0.003 to −0.001; $P$ = .009).

The results indicate that for every 1-year increase in age, the association between ankle angle at midstance, maximal ankle plantarflexion angle, and NNcost increases by 0.003 ($\beta$ coefficient). Similarly, the association between knee angle at midstance, maximum knee extension angle, hip angle at midstance, maximum hip extension angle, and NNcost, respectively, decreases by 0.002 ($\beta$ coefficient) for every 1-year increase in age. This means that the strength of the associations between these variables and NNcost differs according to age. This results in, for example, no association between maximum knee extension angle and NNcost at age 10 years ($\beta$ = 0.004; 95% CI = −0.006 to 0.013; $P$ = .451) but a negative association between maximum knee extension angle and NNcost at age 19 years ($\beta$ = −0.014; 95% CI = −0.023 to 0.006; $P$ = .002). Table 4 shows associations between NNcost and ankle angle at midstance, maximal ankle plantarflexion angle, knee angle at midstance, maximum knee extension angle, hip angle at midstance, and maximum hip extension angle, respectively, for a person aged 10 years and a person aged 19 years.

### Prediction of NNcost

Finally, the multivariable model that included all variables associated with NNcost at the 10% level (ie, knee angle at midstance, maximum knee extension angle, hip angle at midstance, and maximum hip extension angle), showed that none of these variables were independent predictors of NNcost (Tab. 5). In combination, these variables explained only 40% of the variation in NNcost ($R^2$ = 0.40).

## Discussion

The purpose of the study was to examine the associations between lower-body joint angles, moments, power, and walking efficiency in adolescents with CP. We found that: (1) reduced knee and hip extension during walking was significantly related to less efficient gait in adolescents with CP, (2) age was a significant modifier of the associations between joint kinematics and gait efficiency,

**Table 2.**
Correlation Coefficients Between Kinematics and Kinetics, and NNcost[a]

| Variable | *R* | *P* value | n |
|---|---|---|---|
| **Joint angles (°)** | | | |
| Ankle angle at midstance | 0.05 | .766 | 46 |
| Max. plantarflexion angle | 0.17 | .257 | 46 |
| Knee angle at midstance | −0.52 | <.001 | 46 |
| Max. knee extension angle | −0.59 | <.001 | 46 |
| Hip angle at midstance | −0.56 | <.001 | 45 |
| Max. hip extension angle | −0.59 | <.001 | 45 |
| **Joint moment, Nm/kg** | | | |
| Peak ankle moment | −0.17 | .275 | 42 |
| Ankle moment at midstance | −0.16 | .305 | 42 |
| Knee moment at midstance | −0.04 | .803 | 42 |
| Hip moment at midstance | 0.17 | .271 | 42 |
| **Joint power, Nm/s/kg** | | | |
| Ankle generating power | −0.14 | .381 | 42 |
| Ankle absorbing power | 0.02 | .910 | 42 |
| Knee generating power | 0.001 | .995 | 42 |
| Knee absorbing power | 0.19 | .223 | 42 |
| Hip generating power | −0.10 | .513 | 42 |
| Hip absorbing power | 0.18 | .242 | 42 |

[a]NNcost = net nondimensional oxygen cost.

and (3) there were no significant associations for other joint moments or joint powers. To our knowledge, this is the first study to show the associations between lower-limb joint kinematics and kinetics in a large population of adolescents with CP.

## Associations Between the Walking Efficiency, Kinematics, and Kinetics

The inability of adolescents with CP to adequately extend the knee and hip joint during walking could be caused by constraints such as joint contracture, spasticity, or muscle weakness, which has been described in neurological populations to be selectively more noticeable at the joint extension end range, such as full knee and hip extension.[18] Excessive flexion at the hip and knee could influence the ability of the lower limb muscles to maintain a near isometric contraction during the single-support phase of gait—a factor that has been shown to occur in typically developing children to maintain efficient gait.[19,20] However, in children with CP, Kalsi et al[21] showed that during the single-support phase of gait, the fascicles of the ankle plantarflexors were not able to maintain an isometric contraction and were pulled into a lengthened position, which has been shown to increase the oxygen cost of walking.[22] The amount of knee and hip flexion might therefore influence the degree of eccentric lengthening in the lower limb muscles and explain why knee and hip joint kinematics were found to be related to gait efficiency in our study.

Due to reduced knee and hip extension, reduced moments and power in joints of the lower limbs also could have been expected to be associated with gait efficiency. However, this was not the case. These results support previous findings of joint extension lag in people with motor impairment.[23,24] Selective weakness at short muscle lengths, such as in crouch gait, could be associated with higher oxygen consumption. In crouch gait, excessive hip and knee flexion results in the knee and hip extensors working eccentrically at longer lengths and this could also contribute to the finding of the present study, that joint position and not strength is associated with gait inefficiency. However, Steele et al[25] found that despite crouch gait being among the most exhausting gait patterns encountered by adolescents with CP, knee flexion angle during gait explained only 5% to 20% of the variability in oxygen consumption. This highlights the need to explore patient-specific factors that contribute to increased gait inefficiency in this population.

**Table 3.**
Results From Regression Models Examining Association Between NNcost (Dependent Variable) and Kinematics and Kinetics[a]

| Model | $R^2$ | Coefficient (95% CI)[b] | *P* value | n |
|---|---|---|---|---|
| Model 1: Ankle angle at midstance | 0.5814 | − 0.002 (−0.009 to 0.006) | .656 | 46 |
| Model 2: Max. plantarflexion angle | 0.5799 | − 0.−0.001 (−0.006 to 0.007) | .808 | 46 |
| Model 3: Knee angle at midstance | 0.6070 | − 0.004 (−0.010 to 0.001) | .101 | 46 |
| Model 4: Max. knee extension angle | 0.6247 | − 0.006 (−0.012 to −0.0005) | .034 | 46 |
| Model 5: Hip angle at midstance | 0.6309 | − 0.007 (−0.013 to −0.001) | .028 | 45 |
| Model 6: Max. hip extension angle | 0.6685 | − 0.008 (−0.014 to −0.003) | .003 | 45 |
| Model 7: Peak ankle moment | 0.5817 | − 0.018 (−0.209 to 0.173) | .850 | 42 |
| Model 8: Ankle moment at midstance | 0.5894 | − 0.077 (−0.261 to 0.108) | .404 | 42 |
| Model 9: Knee moment at midstance | 0.5818 | − 0.015 (−0.166 to 0.135) | .839 | 42 |
| Model 10: Hip moment at midstance | 0.5836 | 0.029 (−0.103 to 0.161) | .658 | 42 |
| Model 11: Ankle peak positive power | 0.5813 | − 0.002 (−0.087 to 0.084) | .970 | 42 |
| Model 12: Ankle peak negative power | 0.5921 | − 0.037 (−0.114 to 0.040) | .335 | 42 |
| Model 13: Knee peak positive power | 0.5975 | 0.028 (−0.019 to 0.074) | .236 | 42 |
| Model 14: Knee peak negative power | 0.5813 | 0.0003 (−0.084 to 0.085) | .994 | 42 |
| Model 15: Hip peak positive power | 0.5820 | 0.010 (−0.073 to 0.094) | .807 | 42 |
| Model 16: Hip peak negative power | 0.5816 | − 0.007 (−0.091 to 0.078) | .877 | 42 |

[a]NNcost = net nondimensional oxygen cost.
[b]All coefficients adjusted for age, sex, and Gross Motor Function Classification System level.

**Table 4.**
Associations Between Joint Angles, Moment, Power, and NNcost for Variables Where Effect Modification by Age Is Present[a]

| Variable | Coefficient[b] (95% CI) for age 10 years | *P* value | Coefficient[b] (95% CI) for age 19 years | *P* value |
|---|---|---|---|---|
| Ankle angle at midstance | − 0.013 (−0.024 to −0.001) | .031 | 0.018 (0.0004 to 0.035) | .046 |
| Max. plantarflexion angle | − 0.012 (−0.024 to −0.0003) | .044 | 0.013 (0.002 to 0.024) | .026 |
| Knee angle at midstance | 0.007 (−0.002 to 0.015) | .113 | − 0.016 (−0.024 to −0.007) | .001 |
| Max. knee extension angle | 0.004 (−0.006 to 0.013) | .451 | − 0.014 (−0.023 to −0.006) | .002 |
| Hip angle at midstance | 0.002 (−0.007 to 0.012) | .578 | − 0.017 (−0.026 to −0.007) | .001 |
| Max. hip extension angle | 0.001 (−0.008 to 0.009) | .878 | − 0.017 (−0.024 to −0.009) | <.001 |

[a]NNcost = net nondimensional oxygen cost (at ages 10 years and 19 years, respectively).
[b]Adjusted for age, sex, and Gross Motor Function Classification System level.

One explanation why joint moments and power were not related to gait efficiency, is that the Achilles tendon, which has been shown to be longer and more compliant in children with CP,[26] might contribute to the production of joint moments and power during gait by storing and releasing elastic energy, while minimizing the oxygen cost of gait. A second explanation is the method used to assess joint moments and power in this study. Desloovere et al[27] showed that hip ROM and strength in the coronal but not sagittal plane were fairly to moderately correlated with gait data of children with CP, and might explain why joint moments and power were not significantly associated with gait efficiency in our study. Similarly, ankle kinetics are difficult to measure given the multitude of impairments and deformities affecting ankle joint function in people with CP. It is possible that the lack of associations between ankle kinetics and NNcost in our study were not able to be captured in a single plane. A further exploration of ankle kinetics in multiple planes might have provided more meaningful data on the associations between ankle kinetics and NNcost.

**Table 5.**
Multivariable Linear Regression Between Selected Kinematic and Kinetic Variables, and NNcost[a]

| Variable | Coefficient (95% CI) | *P* value |
|---|---|---|
| Constant | 0.506 (0.351 to 0.661) | <.001 |
| Knee angle at midstance | 0.008 (−0.011 to 0.027) | .422 |
| Max. knee extension angle | − 0.015 (−0.035 to 0.005) | .140 |
| Hip angle at midstance | − 0.002 (−0.018 to 0.015) | .853 |
| Max. hip extension angle | − 0.007 (−0.021 to 0.006) | .269 |

[a] n = 45; NNcost = net nondimensional oxygen cost.

The associations between NNcost and joint angles might be confounded by participants with a higher GMFCS level having a higher NNcost and smaller joint ROM. Thus, including GMFCS level in the regression model allowed us to explore the association between NNcost and joint angles, independent of GMFCS level. The results showed that regardless of a participant's GMFCS level, knee and hip angles were still associated with NNcost. The results in this study were also adjusted for age and sex as recommended by Bolster et al.[16] By adjusting the regression based on age, sex, and GMFCS level, we have shown that the significant associations can be attributed to actual joint kinematics or kinetics.

## Effects of Age and Sex

We also examined if age and sex influenced the associations between joint kinematics, kinetics, and gait efficiency. It was found that age, but not sex, was a significant modifier of the associations between joint angle and NNcost. Ankle angle was associated with gait efficiency in all participants. However, in the youngest participants (aged 10 years) greater plantarflexion at the ankle during walking (ie, equinus foot) resulted in less efficient walking. This contrasted with the results of the oldest participants (aged 19 years), in whom a reduced plantarflexion angle (eg, neuromuscular flat foot) resulted in less efficient walking. In the younger participants, there was no association between knee or hip flexion angles and NNcost, whereas in the oldest participants, knee and hip flexion (ie, crouch gait) were significantly associated with the NNcost. These findings suggest that in the younger participants a higher energy cost and less efficient walking were related to an equinus pattern of gait, whereas in the older participants they were related to a crouch gait. That both gait patterns can be associated with a higher energy cost of walking, as found in this study, is supported by the literature. A reduced ankle ROM resulting from equinus or neuromuscular flat-foot pattern of gait, limits ankle push-off power, which generally leads to an increased energy cost of walking.[28–30] The crouch gait found in older participants in the present study could have been required for maintaining balance and to propel the body forward, as a compensatory strategy for reduced push-off power, but at the cost of less efficient walking.[30–32]

It is clear from these novel findings that age can modify the associations between joint kinematics and NNcost. Two possible mechanisms at the level of the joint can explain these findings. First, the age-related shift from equinus gait to crouch gait as a significant predictor of NNcost, could have resulted from classic treatments targeting the Achilles tendon or plantarflexor muscles throughout adolesecence (eg, orthoses, botox injection, muscle-tendon lengthening procedures). For example, in children with spastic CP, the treatment of equinus gait with Achilles tendon–lengthening surgery is associated with overcorrection, which can result in crouch gait.[33] Secondly, an increase in body mass together with the progression of musculoskeletal difficulties (eg, increasing muscle weakness, antagonist cocontraction) with advancing age[34] could have influenced these results. It has been estimated that more than 50% of the metabolic energy expenditure at comfortable walking speed is used to lift the center of mass.[35–37] As adolescents with CP age, there is a high prevalence of obesity,[38] which would further increase the body mass and require more metabolic energy to lift the center of mass of the body. In addition, the peak motor attainment in CP is typically reached at 8 to 9 years old and tends to plateau before a decline in adolescence.[39] This increase in body mass and decline in muscle strength and motor control with advancing age is likely to be a factor that could change the relative importance of kinematics-related predictors of walking efficiency. Considering these age-related factors, it is not surprising that age can modify the predictors of walking efficiency. This, however, needs further investigation in future studies.

## Study Limitations

The study has some limitations that should be considered. First, it is well known that the gait pattern of individuals with CP is highly variable. Not only did the relative timing of the gait events differ between the participants, but also the differences from the typical gait pattern varied between the participants. Extraction of gait values from "midstance" (ie, midpoint between the "foot-on" and "foot-off" events) was clear and consistent. For other variables; however, such as maximal angles, peak moment, or power, data were extracted as the peak value of 1 gait cycle, where timing could have varied considerably. Therefore, it is possible that amplitude-specific or time-specific variables might not have captured the functionally meaningful gait instance for a specific participant. Second, gait impairment was captured in a sagittal plane around the frontal axis only. Given the above-mentioned high interindividual variation in gait, this might have influenced the associations between lower-limb joint kinetics and kinematics and NNcost. During typical walking, the largest ROM and moment are expected around the frontal axis of the lower limbs. However, in a CP population, other muscle groups such as the hip abductors, adductors, or muscles responsible for ankle inversion or eversion, could have compensated for impaired flexion-extension movements at the ankle, knee, or hip joints. Therefore, the results should be interpreted with caution in individuals whose gait patterns deviate significantly from the sagittal plane.

## Conclusions

In the present study, an analysis of associations between lower-body joint angles, moments, and power and gait efficiency, showed that walking efficiency is related to knee and hip angle but not joint moments or power. Age seems to be an important factor influencing the associations between joint kinematics and NNcost. To our knowledge, this is the first study to report these clinically important variables at all lower body joints, and by adjusting the results for age, sex, and GMFCS level. This study augments evidence-based practice by showing the importance of adequate knee and hip extension during walking and will help to inform therapeutic interventions. Further research describing the associations between gait efficiency and kinetic and kinematic variables in other planes of motions is needed to fully understand the complexity of walking in adolescents with CP.

## Author Contributions

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## Ethics Approval

This study has ethical approval from the Research Ethics Committee of Brunel University London's Department of Clinical Sciences and from the National Research Ethics Service (NRES) Committee London-Surrey Border. All participants aged 16 years or older provided written informed consent. Participants younger than 16 years provided written informed assent and parents/guardians additionally provided written consent.

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## Disclosures

The authors completed the ICJME Form for Disclosure of Potential Conflicts of Interest and reported no conflicts of interest.

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### References

1 Bax M, Goldstein M, Rosenbaum P et al. Proposed definition and classification of cerebral palsy. *Devel Med Child Neurol*. 2005;8:47.

2 Brehm MA, Becher J, Harlaar J. Reproducibility evaluation of gross and net walking efficiency in children with cerebral palsy. *Dev Med Child Neurol*. 2007;49:45–48.

3 Unnithan VB, Clifford C, Bar-Or O. Evaluation by exercise testing of the child with cerebral palsy. *Sports Med*. 1998;26:239–251.

4 Russchen HA, Slaman J, Stam HJ et al. Focus on fatigue amongst young adults with spastic cerebral palsy. *J Neuroeng Rehabil*. 2014;11;161.

5 Maltais DB, Pierrynowski MR, Galea VA, Bar-Or O. Physical activity level is associated with the O2 cost of walking in cerebral palsy. *Med Sci Sports Exerc*. 2005;37:347–353.

6 Eek MN, Tranberg R, Beckung E. Muscle strength and kinetic gait pattern in children with bilateral spastic CP. *Gait Posture*. 2011;33:333–337.

7 Ballaz L, Plamondon S, Lemay M. Group aquatic training improves gait efficiency in adolescents with cerebral palsy. *Disabil Rehabil*. 2011;33:1616–1624.

8 Goh HT, Thompson M, Huang WB, Schafer S. Relationships among measures of knee musculoskeletal impairments, gross motor function, and walking efficiency in children with cerebral palsy. *Pediatr Phys Ther*. 2006;18:253–261.

9 Damiano DL, Abel MF. Functional outcomes of strength training in spastic cerebral palsy. *Arch Phys Med Rehabil*. 1998;79:119–125.

10 Zarrugh MY, Radcliffe CW. Predicting metabolic cost of level walking. *Eur J Appl Physiol Occup Physiol*. 1978;38:215–223.

11 Kramer JF, MacPhail HE. Relationships among measures of walking efficiency, gross motor ability and isometric strength in adolescents with cerebral palsy. *Pediatr Phys Ther*. 1994;6;3–8.

12 Pouliot-Laforte A, Parent A, Ballaz L. Walking efficiency in children with cerebral palsy: relation to muscular strength and gait parameters. *Comput Methods Biomech Biomed Engin*. 2014;17:104–105.

13 Ryan JM, Theis N, Kilbride C et al. Strength training for adolescents with cerebral palsy (STAR): study protocol of a randomised controlled trial to determine the feasibility, acceptability and efficacy of resistance training for adolescents with cerebral palsy. *BMJ Open*. 2016;6:e012839.

14 Schwartz MH. Protocol changes can improve the reliability of net oxygen cost data. *Gait Posture*. 2007;26:494–500.

**15** Thomas SS, Buckon CE, Schwartz MH et al. Walking energy expenditure in able-bodied individuals: a comparison of common measures of energy efficiency. *Gait Posture*. 2009;29:592–596.

**16** Bolster E, Balemans AC, Brehm M-A, Buizer AI, Dallmeijer AJ. Energy cost during walking in association with age and body height in children and adolescents with cerebral palsy. *Gait Posture*. 2017;54:119–126.

**17** Dingwell JB, Marin LC. Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. *J Biomech*. 2006;39:444–452.

**18** Rha D-W, Cahill-Rowley K, Young J et al. Biomechanical and clinical correlates of stance-phase knee flexion in persons with spastic cerebral palsy. *PMR*. 2016;8:11–18.

**19** Farris DJ, Sawicki GS. The mechanics and energetics of human walking and running: a joint level perspective. *J R Soc Interface*. 2012;9:110–118.

**20** Lichtwark GA, Bougoulias K, Wilson AM. Muscle fascicle and series elastic element length changes along the length of the human gastrocnemius during walking and running. *J Biomech*. 2007;40:157–164.

**21** Kalsi G, Fry NR, Shortland AP. Gastrocnemius muscle-tendon interaction during walking in typically-developing adults and children, and in children with spastic cerebral palsy. *J Biomech*. 2016;49:3194–3199.

**22** Ryschon TW, Fowler MD, Wysong RE, Anthony A, Balaban RS. Efficiency of human skeletal muscle in vivo: comparison of isometric, concentric, and eccentric muscle action. *J Appl Physiol*. 1997;83:867–874.

**23** Horstman A, Gerrits K, Beltman M et al. Muscle function of knee extensors and flexors after stroke is selectively impaired at shorter muscle lengths. *J Rehabil Med*. 2009;41:317–321.

**24** Ada L, Canning CG, Low S-L. Stroke patients have selective muscle weakness in shortened range. *Brain*. 2003;126:724–731.

**25** Steele KM, Shuman BR, Schwartz MH. Crouch severity is a poor predictor of elevated oxygen consumption in cerebral palsy. *J Biomech*. 2017;60:170–174.

**26** Barber L, Barrett R, Lichtwark G. Medial gastrocnemius muscle fascicle active torque-length and Achilles tendon properties in young adults with spastic cerebral palsy. *J Biomech*. 2012;45:2526–2530.

**27** Desloovere K, Molenaers G, Feys H et al. Do dynamic and static clinical measurements correlate with gait analysis parameters in children with cerebral palsy? *Gait Posture*. 2006;24:302–313.

**28** Ballaz L, Plamondon S, Lemay M. Ankle range of motion is key to gait efficiency in adolescents with cerebral palsy. *Clin Biomech*. 2010;25:944–948.

**29** Schwartz MH, Rozumalski A, Trost JP. The effect of walking speed on the gait of typically developing children. *J Biomech*. 2008;41:1639–1650.

**30** Collins SH, Kuo AD. Recycling energy to restore impaired ankle function during human walking. *PLoS One*. 2010;5:e9307.

**31** Steinwender G, Saraph V, Zwick EB et al. Hip locomotion mechanisms in cerebral palsy crouch gait. *Gait Posture*. 2001;13:78–85.

**32** Kuo AD, Donelan JM. Dynamic principles of gait and their clinical implications. *Phys Ther*. 2010;90:157–174.

**33** Segal LS, Thomas SES, Mazur JM et al. Calcaneal gait in spastic diplegia after heel cord lengthening—a study with gait analysis. *J Pediatr Orthop*. 1989;9:697–701.

**34** Rosenbaum P, Paneth N, Leviton A et al. A report: the definition and classification of cerebral palsy. *Dev Med Child Neurol*. 2007;49:8–14.

**35** Duff-Raffaele M, Kerrigan DC, Corcoran PJ et al. The proportional work of lifting the center of mass during walking. *Am J Phys Med Rehabil*. 1996;75:375–379.

**36** Grabowski A, Farley CT, Kram R. Independent metabolic costs of supporting body weight and accelerating body mass during walking. *J Appl Physiol*. 2005;98:579–583.

**37** Neptune RR, Zajac FE, Kautz SA. Muscle mechanical work requirements during normal walking (the energetic cost of raising the body's center-of-mass is significant). *J Biomech*. 2004;37:817–825.

**38** Pascoe J, Thomason P, Graham HK et al. Body mass index in ambulatory children with cerebral palsy: a cohort study. *J Paediatrics Child Health*. 2016;52:417–421.

**39** Hanna SE, Rosenbaum PL, Bartlett DJ et al. Stability and decline in gross motor function among children and youth with cerebral palsy aged 2 to 21 years. *Dev Med Child Neurol*. 2009;51:295–302.