Original article

Predictors of walking efficiency in children with cerebral palsy: lower body joint angles, moment and power

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Abstract

Background

People with cerebral palsy (CP) experience increased muscle stiffness, muscle weakness and reduced joint range of motion (ROM). This can lead to an abnormal pattern of gait, which may 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

Cross-sectional study.

Methods

Sixty-four adolescents aged 10-19 y with CP were recruited. Walking efficiency was measured as the net non-dimensional oxygen cost (NNcost) during 6-min over ground walking at self-selected speed. Lower body kinematics and kinetics during walking were collected with 3D 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 GMFCS level, maximum knee extension angle (β=-0.006), hip angle at mid-stance (β = -0.007) and maximum hip extension (β=-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 inter-individual variation in gait pattern may 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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Introduction

Cerebral palsy (CP) arises as a result of damage to the developing brain[1]. Children with CP experience muscle weakness and decreased joint range of motion (ROM), which may alter gait patterns and lead to a greater energy cost of walking compared to their typically developing peers.[2, 3] This inefficient pattern of walking in CP may 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 may help to inform evidence-based therapeutic interventions.

Abnormal strength and joint range of motion are key features of gait inefficiency in CP.[6,7,8] 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 are sparse and demonstrate conflicting relationships between muscle strength, gait parameters and walking efficiency. For example, Kramer et al. (1994)[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. (2011)[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, may also affect the output at the ankle.

These findings were supported by Pouliot-Laforte (2014)[12] who found that only ankle joint power was correlated with energy expenditure index but 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 is currently a paucity of data on the relationships between joint angles, moments and joint power at the ankle, knee and hip and walking efficiency in CP. Those studies, which 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 may 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 organisations for disability in England. Inclusion criteria were (1) spastic CP aged 10–19 years, (2) the ability to walk independently with or without a mobility aid (i.e., Gross Motor Function Classification System [GMFCS] levels I–III), and (3) the ability to activate the ankle plantarflexors. Exclusion criteria were (1) orthopaedic 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 cognitive understanding to comply with the assessment procedures. The assessments of the present study were undertaken as part of the baseline assessment for a randomised controlled trial to determine the feasibility, acceptability and efficacy of resistance training for adolescents with cerebral palsy.[13] All participants aged over 16 years provided written informed consent. Participants under 16 years provided written informed assent and parents/guardians additionally provided written consent. This study has ethical approval from Brunel University London's Department of Clinical Sciences' Research Ethics Committee and a National Research Ethics Service (NRES) 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 was measured. Oxygen consumption was recorded using a portable metabolic system (Cosmed K5, Rome, Italy) during a 10 min rest period in a semi-recumbent position and then during 6 min of over ground walking at a self-selected speed. Participants' walking speed ($m.s^{-1}$) was measured in 1 min intervals. Prior to each test, the Cosmed K5 was calibrated for oxygen and carbon dioxide concentration 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 non-dimensional oxygen cost (NNcost) as 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 two minutes of rest and walking, respectively, were removed. Using the remaining data a two minute period that demonstrated least variation in VO_2 and VCO_2, and with respiratory exchange ratio (RER) of less than 1, was identified for rest and walking, respectively.[16] The average oxygen uptake values (VO_2, in $ml.min^{-1}$) and RERs were computed over the steady state periods. The following equation was used to convert average VO_2 during resting and walking from $ml.min^{-1}$ to $J.s^{-1}$:

$$[(4.90 \times RER + 16.040) \times VO_2\,(ml.\,min^{-1})]/60$$

Equation 1

NNcost was calculated by (equation 2):

$$NNcost = [(VO_{2walk} - VO_{2rest})/v]\cdot(1/mg)$$

where VO_{2walk} is oxygen consumption during walking ($J.s^{-1}$), VO_{2rest} is oxygen consumption during resting ($J.s^{-1}$), v is walking speed ($m.s^{-1}$), m is body weight (kg), g is acceleration due to gravity (9.81 $m.s^{-2}$)

Equation 2

Lower body walking kinetics and kinematics

Kinematic and kinetic data during treadmill walking were collected using a computerised motion capture system (Motion Analysis, Motion Analysis Corporation, Santa Rosa, California, USA) with eight infrared cameras and a treadmill with dual integrated force plates capable of capturing X, Y and Z force components (Bertec Corporation, Columbus, OH, USA). The Motion Analysis software Cortex (Motion Analysis Corporation, Santa Rosa, California, USA) was used for the synchronised capture of kinematic data (150 Hz) and force data (2100 Hz).

Joint kinematic and kinetic data was measured from a full body motion analysis marker set using the Conventional Gait Model. For this purpose, thirty-one 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, mid-thigh, 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. Participants were asked to stand on the treadmill in the upright standing position and a neutral position (baseline) was recorded whilst the participant stood upright.

Prior to recording experimental trials, participants performed a familiarisation 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 having full body weight on their feet during standing and walking. Participants begun by walking at a relatively slow speed and treadmill speed was slowly increased until the participant reported the current speed was faster than their preferred speed. The treadmill speed was then slowly decreased until the participant reported the current speed to be slower than preferred. This procedure was repeated three times and the average of the three 'faster' and three 'slower' than preferred speeds were taken as the participant's preferred walking speed.[17] This procedure also allowed for treadmill familiarisation and participant warm-up. Participants were asked to walk on the treadmill at their preferred walking speed for 2 min whilst 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 their provided 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). Both marker coordinates and force data were filtered with a zero lag, fourth order Butterworth filter with a cutoff frequency of 6 Hz. A 6-degree of freedom Visual 3D linked model including nine 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 (i.e. foot segment), and the line connecting the malleoli and the epicondyles (i.e. lower leg segment). Knee joint angle was measured as the angle between the line connecting the malleoli and the epicondyles (i.e. lower leg segment), and the line connecting the epicondyles and the hip joint (i.e. thigh segment). Hip joint angle was measured as an angle between the line connecting the epicondyles and the hip joint (i.e. thigh segment), and the line connecting the hip joints and the acromions (i.e. 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 “mid-stance”. 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 at the first frame were ground reaction forces were less than 20 N. The “mid-stance” was calculated as the frame corresponding to a mid-point between the “foot on” and “foot off” events.

We were specifically interested in joint angles and moments at mid-stance 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 maximal achievable joint extension angle during gait, describing the amount of joint extension lag; peak plantarflexion moment describing the ankle’s ability to propel the body forward; and 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 mid-stance (i.e. single leg support), 2) maximal ankle, knee and hip extension angles. The following variables were extracted from the kinetic analyses: 1) ankle peak plantarflexion moment (i.e. ankle push-off moment) 2) ankle, knee and hip moments at mid-stance 3) ankle, knee and hip joint peak positive (i.e. power generation) and peak negative power (i.e. power absorption) of one gait cycle. Because of a high inter-individual variation in the joint extension lag at a specific gait instance (i.e. inability to achieve anatomical knee position during the mid-stance, due to knee being flexed or hyperextended), the data has 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 three 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, standard deviations (SD), ranges, interquartile ranges, counts and frequencies, where appropriate. In order 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, as 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 the α level of 0.10, 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, USA).

Results

Participant characteristics

Of the 64 participants recruited to the study, six did not have data for NNcost. Descriptive statistics are reported for the remaining 58 participants, including 28 participants at GMFCS Level I (48.3%), 23 at Level II (39.7%) and 7 at Level III (12.1%). Participants were aged 10 to 19 years (mean±SD age: 13.7 ± 2.6 y; mean±SD height: 154.3 ± 13.1 cm; mean±SD mass: 49.3 ± 13.6 kg) Twenty-five of the study participants were females (43.1%), 31 (53.5%) participants had unilateral CP and 27 (46.5%) had bilateral CP. The number of people 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, $p < 0.001$; Table 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 (β = -0.006, 95% CI -0.012 to -0.0005, p = 0.034), hip angle at mid-stance (β = -0.007, 95% CI -0.013 to -0.001, p = 0.028), and maximum hip extension (β = -0.008, 95% CI -0.014 to -0.003, p = 0.003) were associated with NNcost (Table 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 ($p > 0.05$). 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 mid-stance (β = 0.003, 95% CI 0.001 to 0.006, $p = 0.017$), (2) maximal ankle plantarflexion (β = 0.003, 95% CI 0.001 to 0.005, $p = 0.013$), (3) knee angle at mid-stance (β = -0.002, 95% CI -0.004 to -0.001, $p = 0.002$), (4) maximum knee extension angle (β = -0.002, 95% CI -0.004 to -0.0003, $p = 0.022$), (5) hip angle at mid-stance (β = -0.002, 95% CI -0.004 to -0.0004, $p = 0.013$), and (6) maximum hip extension angle (β = -0.002, 95% CI -0.003 to -0.001, $p = 0.009$).

The results indicate that for every year increase in age, the association between ankle angle at mid-stance, maximal ankle plantarflexion angle and NNcost, increases by 0.003. Similarly, the association between knee angle at mid-stance, maximum knee extension angle hip angle at mid-stance, maximum hip extension angle and NNcost, respectively, decreases by 0.002 for every 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 (β = 0.004, 95% CI -0.006 to 0.013, p=0.451) but a negative association between maximum knee extension angle and NNcost at age 19 years (β = -0.014, 95% CI -0.023 to 0.006, p=0.002). Table 4 shows associations between NNcost and ankle angle at mid-stance, maximal ankle plantarflexion angle, knee angle at mid-stance, maximum knee extension angle hip angle at mid-stance 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 (i.e. knee angle at mid-stance, maximum knee extension angle, hip angle at mid-stance, and maximum hip extension angle), showed that none of these variables were independent predictors of NNcost (Table 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, and (3) no significant associations were found for other joint moments or joint powers. 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 for adolescents with CP to adequately extend the knee and hip joint during walking could be caused either 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 may influence the ability of the lower limb muscles to maintain a near isometric contraction during the single support phase of gait – a factor which has been shown to occur in typically developing children to maintain efficient gait.[19,20] However, in children with CP, Kalsi et al. (2016)[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 may therefore influence the degree of eccentric lengthening in the lower limbs muscles and explain why knee and hip joint kinematics were found to be related to gait efficiency in our study.

As a result of reduced knee and hip extension, reduced moments and power in joints of the lower limbs may have also 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. (2017)[25] found that despite crouch gait being one of the most exhausting gait patterns encountered by adolescents with CP, knee flexion angle during gait explained only 5–20% of the variability in oxygen consumption. This highlights the need to explore patient-specific factors that contribute to increased gait inefficiency in this population.

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 complaint in children with CP[26], may contribute to the production of joint moments and power during gait, by storing and releasing elastic energy, whilst 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. (2016)[27] showed that hip range of motion and strength in the coronal, but not sagittal plane were fairly to moderately correlated with gait data of children with CP and may explain why joint moments and power were not significantly associated with gait efficiency in this study. Similarly, ankle kinetics are difficult to measure given the multitude of impairments and deformities affecting ankle joint function in those with CP. It is possible that the lack of associations between ankle kinetics and NNcost in the present study were not able to be captured in a single plane. A further exploration of ankle kinetics in multiple planes may have provided more meaningful data on the associations of ankle kinetics to NNcost.

The associations between NNcost and joint angles might be confounded by those 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 person'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 (10 y) it was found that greater plantarflexion at the ankle during walking, i.e. equinus foot, resulted in less efficient walking. This was contradicted by results of the oldest participants (19 y), whereby a reduced plantarflexion angle (e.g. neuromuscular flat foot) resulted in less efficient walking. In the younger participants, there was no association between knee or hip flexion angles and NNcost, whilst in the oldest participants, knee and hip flexion (i.e. crouch gait) were significantly associated with the NNcost. These findings suggest that in the younger participants, a higher energy cost and less efficient walking was related to an equinus pattern of gait, whilst in the older participants, it was related to a crouch gait. Literature supports that both gait patterns can be associated with a higher energy cost of walking as found in this study. A reduced ankle range of motion resulting from equinus or neuromuscular flat foot pattern of gait, limits ankle push-off power, which almost always leads to an increased energy cost of walking. [28,29, 30] Crouch gait found in older participants in the present study may 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, 31, 32]

It is clear from these novel findings, that age can modify the associations between joint kinematics and NNcost. There are two possible mechanisms at the level of the joint to 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 (e.g. 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 may result in crouch gait.[33] Secondly, an increase in body mass together with the progression of musculoskeletal difficulties (e.g. increasing muscle weakness, antagonist co-contraction) that increases with advancing age[38] 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 centre of mass.[34, 35, 36] As adolescents with CP age, there is a high prevalence of obesity,[37] which would further increase the body mass and require more metabolic energy to lift the centre of mass of the body. In addition, the peak motor attainment in CP is typically reached at eight to nine years 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 does not seem 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 it was also inter-individually different from the typical gait pattern. Extracting gait values from “mid-stance” (i.e. mid-point 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 was extracted as the peak value of one gait cycle, where timing could have varied considerably. Therefore, it is possible that amplitude-specific or time-specific variables may 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 inter-individual variation in gait, this may influence the associations between lower limbs joint kinetics and kinematics and NNcost. During typical walking, the largest range of motion and moment is 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 where 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, show 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. 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 adds to the 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 are needed to fully understand the complexity of walking in adolescents with CP.

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Tables

Table 1. Descriptive statistics for gait efficiency, and joint angles, moments and power

Variable	Mean (SD)	n
NNcost	0.53 (0.28)	58
Treadmill gait speed (m s^{-1})	0.44 (0.15)	55
Joint angles (°)		
Ankle angle at mid-stance	-14.08 (8.56)	48
Max plantar-flexion angle	-28.32 (10.30)	48
Knee angle at mid-stance	-8.64 (13.32)	48
Max knee extension angle	-4.93 (12.61)	48
Hip angle at mid-stance	-9.25 (11.39)	47
Max hip extension angle	0.82 (12.14)	47
Joint moment (Nm kg^{-1})		
Peak ankle moment	-0.33 (0.39)	44
Ankle moment at mid-stance	0.32 (0.38)	44
Knee moment at mid-stance	0.64 (0.45)	44
Hip moment at mid-stance	-0.71 (0.54)	44
Joint power (Nm s^{-1} kg^{-1})		
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

Table 2. Correlation coefficients (*r*) between kinematics and kinetics, and NNcost.

	r	**p value**	**n**
Joint angles (°)			
Ankle angle at mid-stance	0.05	0.766	46
Max plantar-flexion angle	0.17	0.257	46
Knee angle at mid-stance	-0.52	<0.001	46
Max knee extension angle	-0.59	<0.001	46
Hip angle at mid-stance	-0.56	<0.001	45
Max hip extension angle	-0.59	<0.001	45
Joint moment (Nm kg^{-1})			
Peak ankle moment	-0.17	0.275	42
Ankle moment at mid-stance	-0.16	0.305	42
Knee moment at mid-stance	-0.04	0.803	42
Hip moment at mid-stance	0.17	0.271	42
Joint power (Nm s^{-1} kg^{-1})			
Ankle generating power	-0.14	0.381	42
Ankle absorbing power	0.02	0.910	42
Knee generating power	0.001	0.995	42
Knee absorbing power	0.19	0.223	42
Hip generating power	-0.10	0.513	42
Hip absorbing power	0.18	0.242	42

Table 3. Results from regression models examining association between NNcost (dependent variable) and kinematics and kinetics[a]

	R^2	Coeff[a] (95% CI)	p value	n
Model 1: Ankle angle at mid-stance	0.5814	-0.002 (-0.009 to 0.006)	0.656	46
Model 2: Max plantarflexion angle	0.5799	-0.001 (-0.006 to 0.007)	0.808	46
Model 3: Knee angle at mid-stance	0.6070	-0.004 (-0.010 to 0.001)	0.101	46
Model 4: Max knee extension angle	0.6247	-0.006 (-0.012 to -0.0005)	0.034	46
Model 5: Hip angle at mid-stance	0.6309	-0.007 (-0.013 to -0.001)	0.028	45
Model 6: Max hip extension angle	0.6685	-0.008 (-0.014 to -0.003)	0.003	45
Model 7: Peak ankle moment	0.5817	-0.018 (-0.209 to 0.173)	0.850	42
Model 8: Ankle moment at mid-stance	0.5894	-0.077 (-0.261 to 0.108)	0.404	42
Model 9: Knee moment at mid-stance	0.5818	-0.015 (-0.166 to 0.135)	0.839	42
Model 10: Hip moment at mid-stance	0.5836	0.029 (-0.103 to 0.161)	0.658	42
Model 11: Ankle peak positive power	0.5813	-0.002 (-0.087 to 0.084)	0.970	42
Model 12: Ankle peak negative power	0.5921	-0.037 (-0.114 to 0.040)	0.335	42
Model 13: Knee peak positive power	0.5975	0.028 (-0.019 to 0.074)	0.236	42
Model 14: Knee peak negative power	0.5813	0.0003 (-0.084 to 0.085)	0.994	42
Model 15: Hip peak positive power	0.5820	0.010 (-0.073 to 0.094)	0.807	42
Model 16: Hip peak negative power	0.5816	-0.007 (-0.091 to 0.078)	0.877	42

[a]all coefficients adjusted for age, sex and GMFCS level
CI: confidence interval

Table 4. Associations between joint angles, moment, power and NNcost at age 10 years and age 19 years, respectively, for variables where effect modification by age is present.

	Coeff[a] (95% CI) for age 10 years	**p value**	**Coeff[a] (95% CI) for age 19 years**	**p value**
Ankle angle at mid-stance	-0.013 (-0.024 to -0.001)	0.031	0.018 (0.0004 to 0.035)	0.046
Max plantarflexion angle	-0.012 (-0.024 to -0.0003)	0.044	0.013 (0.002 to 0.024)	0.026
Knee angle at mid-stance	0.007 (-0.002 to 0.015)	0.113	-0.016 (-0.024 to -0.007)	0.001
Max knee extension angle	0.004 (-0.006 to 0.013)	0.451	-0.014 (-0.023 to -0.006)	0.002
Hip angle at mid-stance	0.002 (-0.007 to 0.012)	0.578	-0.017 (-0.026 to -0.007)	0.001
Max hip extension angle	0.001 (-0.008 to 0.009)	0.878	-0.017(-0.024 to -0.009)	<0.001

[a]adjusted for age, sex and GMFCS level

Table 5. Multivariable linear regression between selected kinematic and kinetic variables, and NNcost (n=45)

	Coeff (95% CI)	**p value**
Constant	**0.506 (0.351 to 0.661)**	**<0.001**
Knee angle at mid-stance	0.008 (-0.011 to 0.027)	0.422
Max knee extension angle	-0.015 (-0.035 to 0.005)	0.140
Hip angle at mid-stance	-0.002 (-0.018 to 0.015)	0.853
Max hip extension angle	-0.007 (-0.021 to 0.006)	0.269

CI: confidence interval