



# Kinetic comparison of walking on a treadmill versus over ground in children with cerebral palsy



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## ABSTRACT

Kinetic outcomes are an essential part of clinical gait analysis, and can be collected for many consecutive strides using instrumented treadmills. However, the validity of treadmill kinetic outcomes has not been demonstrated for children with cerebral palsy (CP). In this study we compared ground reaction forces (GRF), center of pressure, and hip, knee and ankle moments, powers and work, between overground (OG) and self-paced treadmill (TM) walking for 11 typically developing (TD) children and 9 children with spastic CP. Considerable differences were found in several outcome parameters. In TM, subjects demonstrated lower ankle power generation and more absorption, and increased hip moments and work. This shift from ankle to hip strategy was likely due to a more backward positioning of the hip and a slightly more forward trunk lean. In mediolateral direction, GRF and hip and knee joint moments were increased in TM due to wider step width. These findings indicate that kinetic data collected on a TM cannot be readily compared with OG data in TD children and children with CP, and that treadmill-specific normative data sets should be used when performing kinetic gait analysis on a treadmill.

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## 1. Introduction

Kinetic outcomes are an essential part of clinical gait analysis. While kinematics are used to quantitatively describe the abnormalities of movement patterns on the level of joint and segment angles, kinetics give an indication of the causes of these motions and the relation with underlying muscle function. Kinetic outcomes of gait analysis typically contain the hip, knee, and ankle joint moments as well as their powers. Joint moments describe the net internal moments delivered by all muscles and ligamentous tissue around the joint, thereby giving an indication of the minimum force level that muscles need to produce at any instant during the gait cycle. Joint powers describe the rate, amount, and timing of energy generation and dissipation around a joint.

Children with cerebral palsy (CP) typically present abnormal patterns of joint moments and powers during gait. For instance, in a crouched gait pattern abnormally high moments can occur around the hip, knee, and ankle, requiring excessively high muscle forces (Lin et al., 2000). Abnormally high or low powers are also typically seen in these patients, in combination with aberrant and inefficient timing. A toe-walking gait pattern for instance can coincide with high power dissipation and generation peaks in

early and mid-stance (Svehlik et al., 2010), which do not contribute to efficient propulsion. In contrast, ankle power during push-off is typically diminished (Riad et al., 2008; Svehlik et al., 2010), which may lead to an inefficient gait pattern (Donelan et al., 2002). For a thorough understanding of a patient's gait pattern, it is important to accurately describe the joint moments and powers in combination with the kinematics.

Kinetic data are typically collected using ground-embedded force plates, and a single complete foot contact is needed per plate for correct calculation of joint moments and powers during a stride. This can make it cumbersome and time-consuming to collect only a few good strides. The introduction of instrumented split-belt treadmills with built-in force plates allows for kinetic data collection of many consecutive strides. However, there are several technical challenges inherent of treadmill-embedded force plates, such as increased compliance of the large plates and more low-frequency vibrations compared with ground-mounted force plates (Sloot et al., 2015b). This is expected to increase the noise and decrease the accuracy of the forces and center of pressure, which could negatively affect joint moment and power calculations. Before utilizing instrumented treadmills for kinetic gait analysis in research and clinical practice, it is thus important to critically assess the measured moments and powers.

A few studies have compared treadmill kinetics to overground data. Riley et al. (2007) found that in healthy adults joint moments, powers, and GRF peaks were generally smaller during treadmill walking compared with overground, for the same

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**Table 1**  
Spatiotemporal and kinetic outcome parameters.

Parameter	Unit	TD		CP		p-values		
		OG	TM	OG	TM	CON	GRP	Inter
Walking speed	m/s	1.34 ± 0.15	1.28 ± 0.20	1.12 ± 0.17	1.04 ± 0.31	0.229	0.009	0.901
GRF vert peak1	N/kg	11.03 ± 1.34	11.29 ± 0.79	11.73 ± 1.59	12.56 ± 1.02	0.134	0.031	0.428
GRF vert peak2	N/kg	10.46 ± 1.52	10.93 ± 0.57	10.24 ± 1.07	10.32 ± 0.60	0.254	0.310	0.418
GRF ap brake peak	N/kg	1.82 ± 0.45	1.98 ± 0.29	1.57 ± 0.30	1.80 ± 0.59	0.100	0.180	0.728
GRF ap propul peak	N/kg	1.90 ± 0.33	1.92 ± 0.43	1.53 ± 0.37	1.66 ± 0.63	0.569	0.061	0.638
GRF ml peak	N/kg	0.45 ± 0.12	0.88 ± 0.19	0.56 ± 0.17	1.09 ± 0.22	0.000	0.017	0.314
CoP ap mean	%	81.26 ± 11.34	71.03 ± 10.76	87.77 ± 22.78	100.48 ± 20.12	0.564	0.021	0.000
CoP ml mean	%	44.76 ± 5.71	25.18 ± 12.39	45.73 ± 10.77	45.46 ± 20.24	0.002	0.055	0.002
Trunk lean fw mean	deg	4.58 ± 4.82	7.99 ± 4.70	4.64 ± 5.81	6.68 ± 7.75	0.037	0.982	0.649
<b>Moment</b>								
Hip extension M peak	Nm/kg	0.79 ± 0.20	1.12 ± 0.22	0.79 ± 0.23	1.02 ± 0.28	0.000	0.558	0.458
Hip flexion M range	Nm/kg	1.54 ± 0.40	1.71 ± 0.26	1.53 ± 0.47	1.52 ± 0.43	0.403	0.505	0.369
Hip abduction M peak	Nm/kg	0.60 ± 0.20	0.83 ± 0.26	0.42 ± 0.08	0.55 ± 0.22	0.001	0.012	0.238
Knee extension M peak	Nm/kg	0.56 ± 0.27	0.52 ± 0.25	0.50 ± 0.21	0.38 ± 0.27	0.140	0.341	0.452
Knee extension M mean	Nm/kg	0.00 ± 0.08	-0.02 ± 0.06	0.00 ± 0.08	-0.09 ± 0.10	0.002	0.231	0.070
Knee abduction M peak	Nm/kg	0.33 ± 0.10	0.49 ± 0.17	0.29 ± 0.13	0.36 ± 0.26	0.008	0.220	0.212
Ankle extension M peak	Nm/kg	1.19 ± 0.20	1.11 ± 0.17	1.17 ± 0.26	1.27 ± 0.26	0.780	0.469	0.039
Ankle extension M DBI	-	0.14 ± 0.20	0.02 ± 0.23	0.45 ± 0.42	0.70 ± 0.25	0.162	0.001	0.000
<b>Power/work</b>								
Hip work generated S	J/kg	0.43 ± 0.13	0.46 ± 0.19	0.46 ± 0.14	0.53 ± 0.10	0.104	0.434	0.509
Hip work absorbed S	J/kg	0.12 ± 0.07	0.12 ± 0.06	0.12 ± 0.06	0.08 ± 0.08	0.357	0.379	0.504
Knee work generated S	J/kg	0.15 ± 0.08	0.18 ± 0.08	0.11 ± 0.06	0.11 ± 0.04	0.394	0.034	0.439
Knee work absorbed S	J/kg	0.51 ± 0.16	0.42 ± 0.14	0.39 ± 0.18	0.34 ± 0.14	0.035	0.136	0.464
Ankle power peak S	W/kg	2.01 ± 0.52	1.64 ± 0.67	1.16 ± 0.58	1.01 ± 0.59	0.079	0.004	0.444
Ankle work generated S	J/kg	0.18 ± 0.05	0.12 ± 0.05	0.13 ± 0.08	0.10 ± 0.05	0.005	0.131	0.306
Ankle work absorbed S	J/kg	0.15 ± 0.04	0.17 ± 0.06	0.16 ± 0.06	0.20 ± 0.10	0.006	0.575	0.248
Total net hip work	J/kg	0.31 ± 0.13	0.34 ± 0.22	0.34 ± 0.11	0.45 ± 0.09	0.090	0.223	0.395
Total net knee work	J/kg	-0.36 ± 0.11	-0.24 ± 0.14	-0.28 ± 0.16	-0.23 ± 0.14	0.023	0.413	0.279
Total net ankle work	J/kg	0.02 ± 0.07	-0.05 ± 0.09	-0.03 ± 0.07	-0.10 ± 0.08	0.000	0.121	0.985
Total net work	J/kg	-0.02 ± 0.12	0.05 ± 0.22	0.03 ± 0.13	0.11 ± 0.15	0.114	0.320	0.974

Abbreviations: TD, typically developing; CP, cerebral palsy; OG, overground; TM, treadmill; CON, condition effect (OG versus TM); GRP, group effect (TD versus CP); Inter, interaction effect between condition and group. GRF, ground reaction force; CoP, center of pressure; vert, vertical; ap, anteroposterior; ml, mediolateral; fw, forward; M, moment; DBI, double bump index (see Section 2); S, stride. CoP ap mean and CoP ml mean are taken as percentage of footlength and footwidth respectively.

walking speed, but within normal gait variability. In healthy elderly, Watt et al. (2010) also found small reductions in the majority of moments and powers when compared to speed-matched overground walking, and Parvataneni et al. (2009) showed a decrease in the second GRF peak, associated with reduced push-off. Thus, the differences found in healthy (older) adults seemed consistent but small. Contrarily, in typically developing children, Rozumalski et al. found considerable differences between overground and treadmill running (Rozumalski et al., 2015) and walking (Rozumalski et al., 2014), due to a more anteriorly oriented ground reaction force vector on the treadmill. This indicates that different subject groups may behave differently on the treadmill, and warrants the need for further study in children with CP.

Therefore, the aim of this study was to compare kinetic data between overground and self-paced treadmill walking for TD children and children with spastic CP. We assessed hip, knee, and ankle joint moments and powers, as well as the underlying ground reaction forces (GRF) and centers of pressure (CoP).

## 2. Methods

### 2.1. Subjects

9 children with spastic CP (4 male, 5 female; age 11.6 ± 2.1 years, range 8–14; height 1.49 ± 0.13 m; weight 40.9 ± 10.3 kg) and 11 TD children similar in age, height, and weight (7 male, 4 female; age 10.6 ± 2.2 years, range 8–15; height

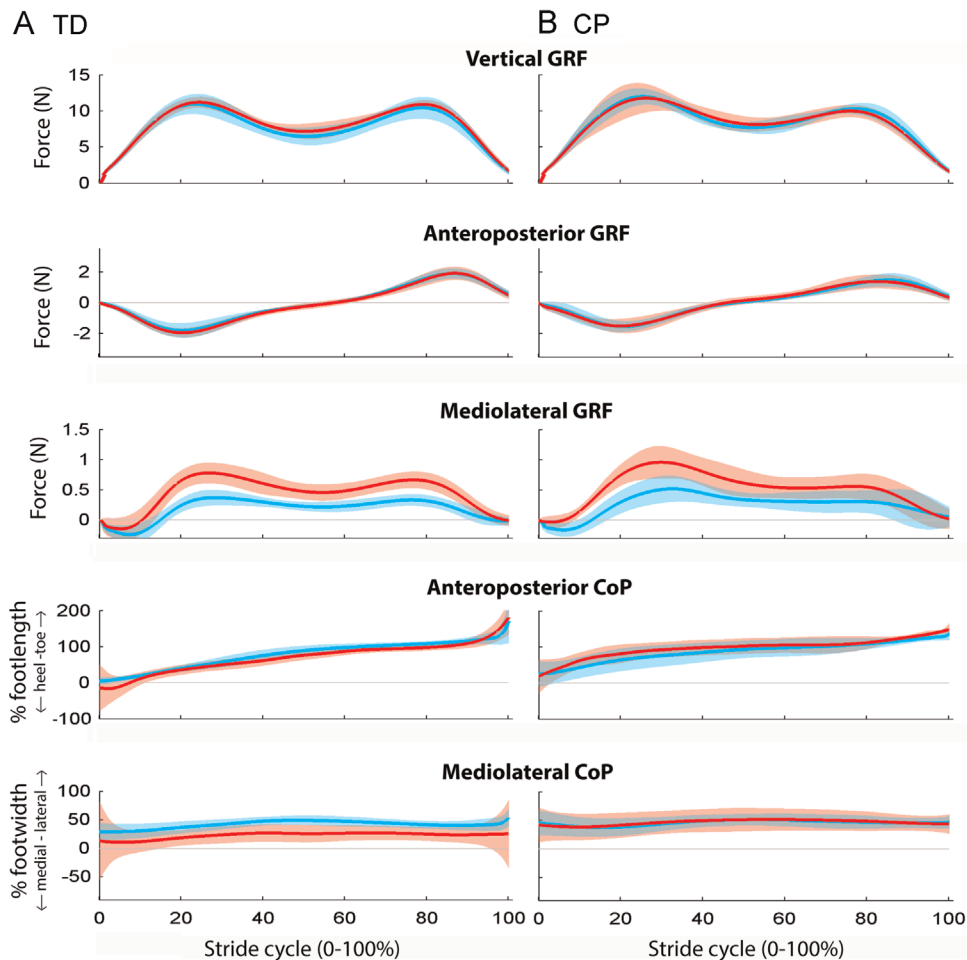
1.52 ± 0.15 m; weight 38.2 ± 10.5 kg) participated in this study. The subjects and set of experiments were the same as in our recently published kinematic comparison between overground walking, treadmill walking, and natural walking outside of a lab environment (Van der Krogt et al., 2014). The methods are briefly repeated here, with an emphasis on the kinetic measurements. Children with CP were randomly selected from our database and only included if they were able to walk independently without walking aids for at least 5 min on end and 30 min in total within two hours; were classified as level I or II on the gross motor function classification scale (GMFCS) (Palisano et al., 1997); had received no multilevel surgery, selective dorsal rhizotomy or intrathecal baclofen treatment within the last year; nor botulinum toxin A treatment within the previous 16 weeks. All parents and children aged 12 years and older provided written informed consent prior to participation. The protocol was approved by the local ethics committee of the VU University Medical Center Amsterdam.

### 2.2. Design and materials

Subjects walked in random order (1) overground (OG) in a conventional gait lab and (2) on a self-paced treadmill (TM) placed in an immersive virtual environment. They wore their own shoes, which had to be low models with flat soles, and orthoses (3 CP subjects) or insoles (1 CP subject) if used on a regular basis. A safety harness was worn in both conditions, and attached loosely to the ceiling only in TM for safety reasons.

OG consisted of a 10 m walkway with two embedded force plates (AMTI, Watertown, MA USA). A target of 5 trials with correct force plate hits was collected for both legs. Subjects were not instructed to target the force plates, but only to walk at their self-selected pace across the walkway.

TM consisted of a dual-belt instrumented treadmill (R-Mill, Motekforce Link, Amsterdam, The Netherlands) in a speed-matched virtual environment projected on a 180° semi-cylindrical screen, displaying an endless, straight forest road and scenery (Gait Real-time Analysis Interactive Lab (GRAIL) system, Motekforce Link, Amsterdam, The Netherlands). The speed of the belt was real-time adjusted to



**Fig. 1.** Ground reaction force (GRF) and center of pressure (CoP) averaged over (A) all typically developing (TD) children and (B) all children with cerebral palsy (CP), for overground (cyan) and treadmill walking (red), with standard deviation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

match the subject's time-varying walking speed, by means of a self-paced (SP) speed algorithm (Sloot et al., 2014b). Subjects were instructed to walk in the middle of the treadmill, but not explicitly to place one foot on each separate belt. Between 6 and 10 min of habituation time was given to adjust to the treadmill, the virtual environment, and the SP speed algorithm. Subsequently, the last minute of a 3-min trial was used for analysis.

3D motion capture data were collected using identical systems in both labs (Optotrak, Northern Digital Inc., Waterloo, Ontario, Canada). Technical clusters of three markers were attached to the trunk, pelvis, thighs, shanks and feet. Anatomical landmarks were indicated in order to anatomically calibrate the technical cluster frames (Cappozzo et al., 1995). The markers remained attached for the entire session and the same indication of anatomical landmarks was used in both labs.

### 2.3. Data analysis

3D kinematics and kinetics were analyzed using custom-made software ([www.BodyMech.nl](http://www.BodyMech.nl), MatLab<sup>®</sup>, The Mathworks). All force data were low-pass filtered with a second order, bi-directional, Butterworth filter with a cut-off frequency of 6 Hz. Joint and segment angles were calculated following CAMARC anatomical frame definitions (Cappozzo et al., 1995). Initial contact and toe-off values were calculated from the vertical GRF, with a threshold value of 50 N. As GRF data were not available for all initial contacts in OG, in those cases instances of heel marker relative to pelvic marker velocity similar as during force-plate hit were taken. For OG, five right strides were analyzed for TD and five strides of the most affected leg for CP. For TM, the first five recorded strides with correct force data were selected from the recorded minute, for the same leg.

We evaluated 3D GRF and 2D CoP data; flexion–extension moments for hip, knee and ankle; ab-adduction moments for hip and knee; and powers for hip, knee and ankle. As the foot was moving on the treadmill, CoP was expressed relative to the foot: in anteroposterior direction as a percentage of foot length (heel to toe marker, relative to heel); and in mediolateral direction as a percentage of foot width (MTP1 to MTP5 marker, relative to MTP1). As specific outcome measures for

statistical analysis we calculated a set of key peak and mean values of the GRF, CoP, moments and powers (Table 1) and the total amount of positive and negative work done for hip, knee and ankle. Furthermore, to assess the ankle moment distribution between first and second half of stance, the 'double bump index' (DBI) was introduced and calculated as the ratio of the mean ankle moment over 0–30% of the gait cycle and the mean ankle moment over 31–60% of the gait cycle, based on Van der Krogt et al. (2009). Each subject's individual difference between TM and OG was quantified by the RMSE, and expressed as a percentage of the stride-to-stride variation in OG (i.e. the mean RMSE between all individual OG strides and the OG mean for the same subject). Walking speed was reported as a potential confounding factor. For all other spatiotemporal, kinematic and subjective measures we refer to Van der Krogt et al. (2014).

### 2.4. Statistics

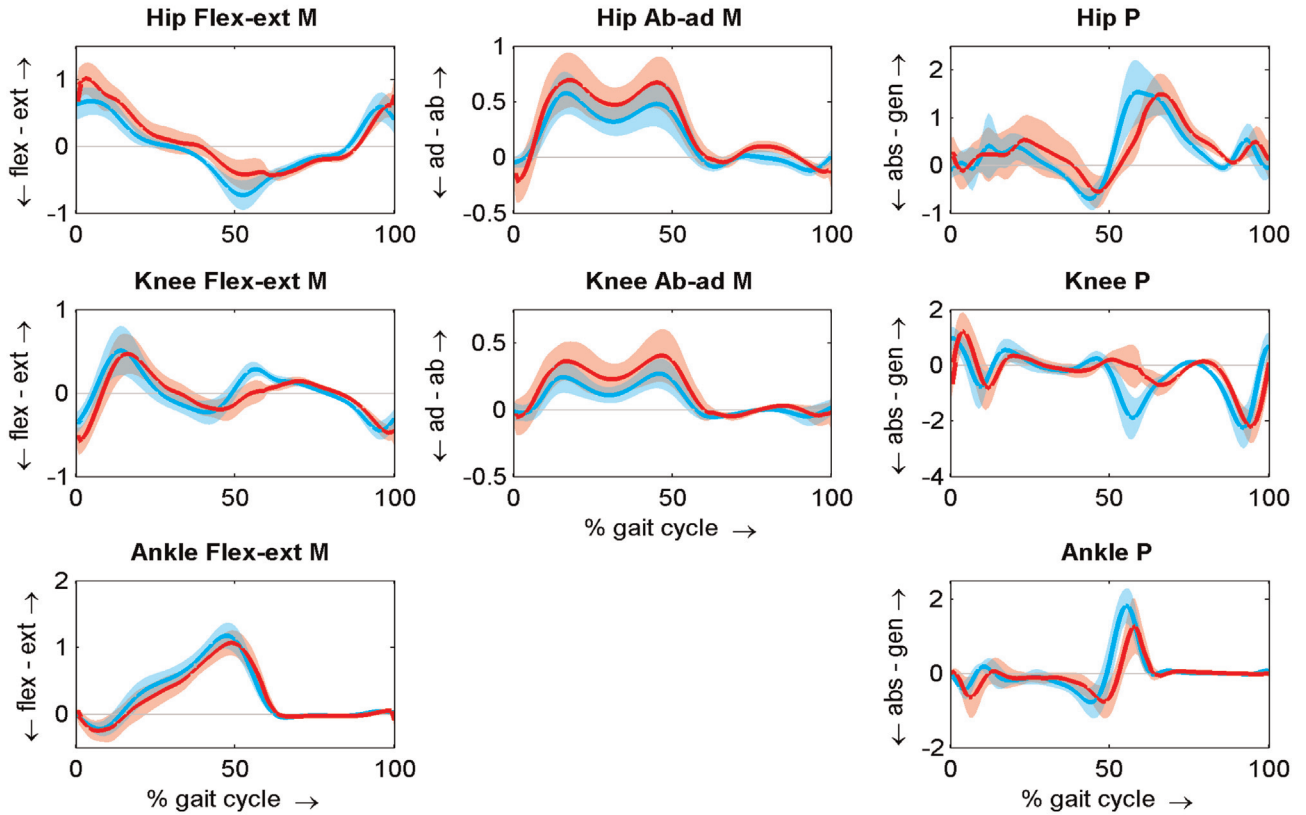
The outcome parameters were compared between OG and TM and between TD and CP using an ANOVA for repeated measurements (IBM SPSS Statistics 20, Armonk, NY, USA). *p*-values of less than 0.05 were considered statistically significant.

## 3. Results

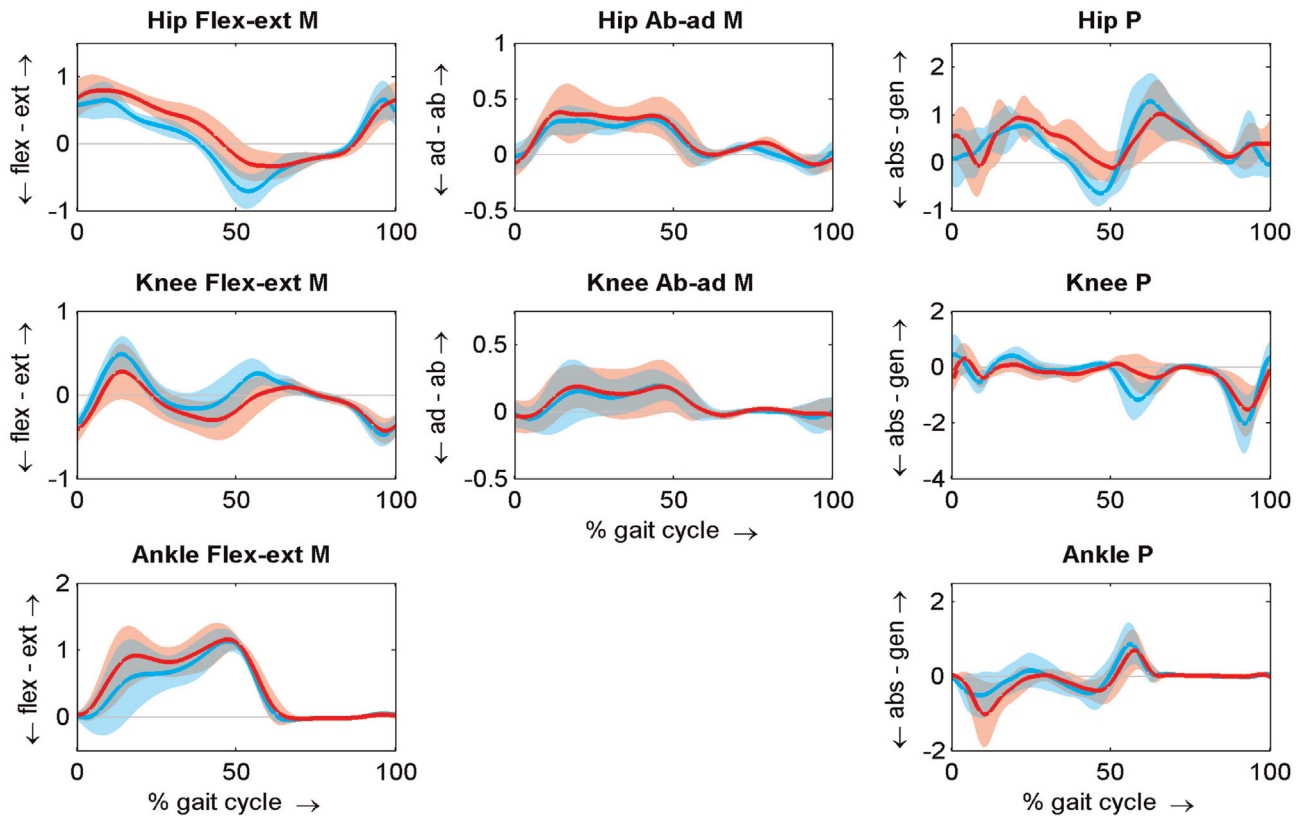
All subjects were able to complete the protocol. Despite the target of 5, the final number of correct strides in OG was  $4.6 \pm 0.5$  for TD (range 4–5), and  $3.9 \pm 1.1$  (range 2–5) for CP, due to technical and practical limitations. Walking speed was not significantly different between OG and TM, although it was slightly lower on average in TM, especially in CP (Table 1).

Vertical and anteroposterior GRF were similar between OG and TM, but mediolateral forces were almost twice as large in TM

**A KINETICS OF TD**



**B KINETICS OF CP**



**Fig. 2.** Kinetic curves averaged over (A) all typically developing (TD) children and (B) all children with cerebral palsy (CP), for overground (cyan) and treadmill walking (red), with standard deviation. With *M* as moment and *P* as power. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

**Table 2**  
RMSE between overground and treadmill kinetic curves.

Parameter	TD RMSE (%) <sup>a</sup>		CP RMSE (%) <sup>a</sup>	
	Mean ± SD	Range	Mean ± SD	Range
Hip flexion M	4.05 ± 2.22	1.50–7.67	3.34 ± 1.76	0.95–6.31
Hip abduction M	4.69 ± 4.79	1.96–18.95	2.15 ± 0.69	1.13–3.11
Knee flexion M	3.34 ± 1.44	1.42–5.75	2.29 ± 1.21	0.95–4.26
Knee abduction M	4.88 ± 4.72	1.68–18.53	3.26 ± 2.39	0.93–8.90
Ankle flexion M	3.22 ± 1.67	1.03–6.70	2.81 ± 1.88	0.84–6.29
<b>MEAN</b>	<b>4.04 ± 2.97</b>	<b>1.52–11.52</b>	<b>2.77 ± 1.59</b>	<b>0.96–5.77</b>
Hip power	2.48 ± 1.17	0.66–4.96	2.18 ± 1.62	0.85–6.16
Knee power	3.42 ± 2.07	0.97–7.83	2.02 ± 1.25	0.81–4.83
Ankle power	2.99 ± 1.18	1.49–5.53	2.00 ± 0.93	0.93–3.80
<b>MEAN</b>	<b>2.96 ± 1.47</b>	<b>1.04–6.11</b>	<b>2.06 ± 1.27</b>	<b>0.86–4.93</b>

Abbreviations: TD, typically developing; CP, cerebral palsy; RMSE, root mean square error; M, moment; SD, standard deviation.

<sup>a</sup> All RMSE values are presented relative to the stride-to-stride variation during OG walking, i.e. the RMSE between each individual stride and the mean over all strides.

compared with OG ( $p < 0.001$ , Fig. 1, Table 1). The CoP was on average 10% more posterior on the foot in TM, while it was 13% more anterior in CP, resulting in a significant interaction effect of group and condition ( $p < 0.001$ , Table 1). The CoP was 44% more medial on the foot in TD, but almost equal between conditions in CP (interaction  $p = 0.002$ , Table 1).

Hip, knee, and ankle moments all showed significant differences between OG and TM (Fig. 2, Table 1). The hip flexion–extension moment was shifted towards more extension (35% more peak extension moment,  $p < 0.001$ ), while the range remained equal between conditions ( $p = 0.40$ ). The knee flexion–extension moment was shifted more towards flexion, especially in CP from a net zero to net negative of 0.09 N m ( $p = 0.002$ , interaction  $p = 0.07$ ). In TD, peak ankle moment was 6.7% decreased in TM compared to OG, but 8.5% increased in CP (interaction  $p = 0.039$ ). The ankle moment in CP increased especially in the first half of stance, as indicated by a significant increase in DBI in CP, which decreased in TD (interaction  $p < 0.001$ ). Hip and knee adduction moments were 35% and 36% higher respectively in TM in both groups ( $p = 0.001$  and  $p = 0.008$ ).

The total amount of net work done at the hip (area under the power curve) tended to be higher in TM compared with OG (21% on average,  $p = 0.09$ , Table 1). The amount of energy absorbed at the knee was 15% lower in TM ( $p = 0.035$ ), while 28% less energy was generated and 19% more absorbed at the ankle ( $p = 0.005$  and  $p = 0.006$ ). Peak ankle power also tended to be lower by 16% in TM in both groups ( $p = 0.08$ ).

The individual RMSE difference between OG and TM moment curves exceeded the stride-to-stride variation in OG by on average 2 to 5 times, depending on the parameter (Table 2). This effect was slightly larger in TD, due to a smaller stride-to-stride variation in OG.

#### 4. Discussion

This study compared OG and TM kinetics in children with CP and typically developing children. While our previous study (Van der Krogt et al., 2014) showed mainly minor differences in kinematics for the same group of subjects except for a wider step

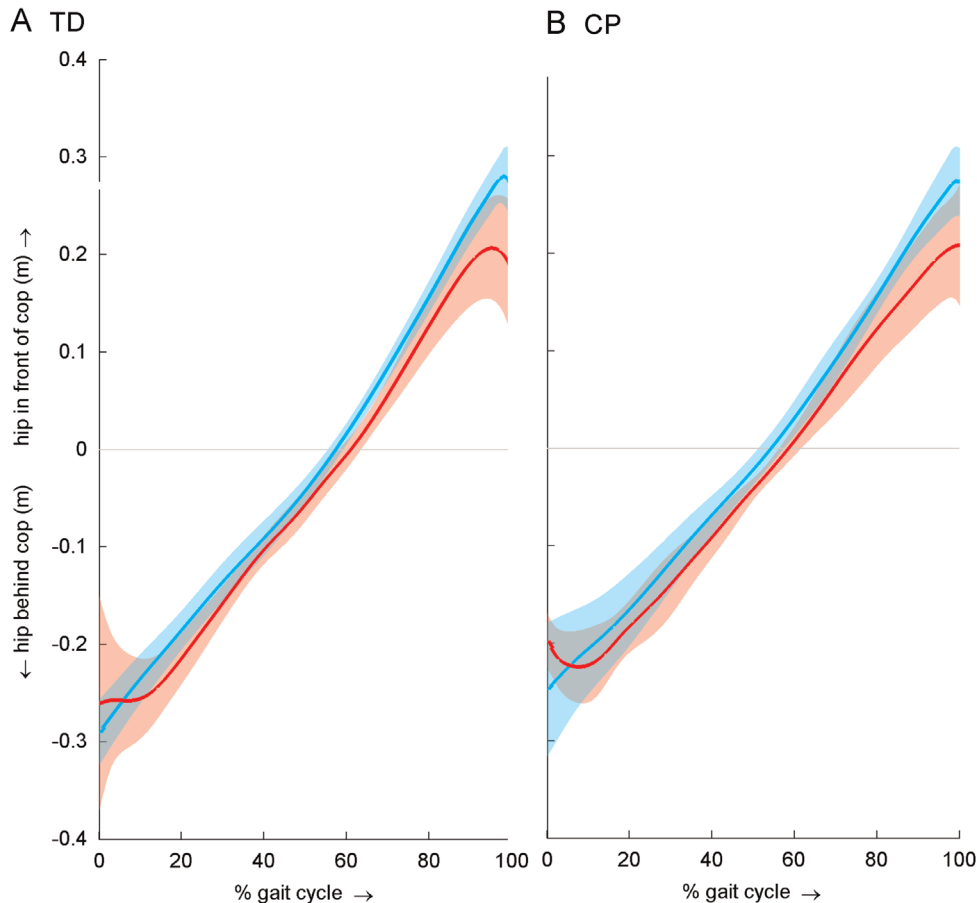
width and some increased deviation in knee and ankle angles in CP, considerable differences in kinetics were found.

Most importantly, in TM compared with OG a shift was found from an ankle to a hip strategy, with higher hip extension moments and a trend toward more net hip work, and less power generation at the ankle. The difference between labs was considerable, with 20% more net hip work and a shift in ankle work from a net neutral to a net dissipation of 0.07 J/kg, averaged over both groups. The increase in hip moments was in line with Rozumalski et al. (2014,2015) for both walking and running in TD children. This effect could not be explained by a difference in the GRF values, as these were similar between conditions in both vertical and anteroposterior direction. Anteroposterior CoP values were affected in opposite direction for TD and CP, and hence could also not explain the shift towards a hip strategy in both groups. Furthermore, kinematics were very similar between TM and OG for these subjects (Van der Krogt et al., 2014). However, when looking in more detail at the subjects' body postures, we found that the hip joint center was more posterior relative to the CoP in TM (Fig. 3). Furthermore, we found significantly more forward trunk lean in both TD and CP subjects (2.8° on average,  $p = 0.037$ , Table 1). Such a shift from ankle to hip work may lead to increased energy cost of walking (Donelan et al., 2002). These findings suggest that it may be important to keep an eye on the upright posture of subjects when walking on a treadmill, for instance by instructing them to look forward to the VR screen, rather than to the front of the treadmill. Such instructions were not given explicitly during the current experiments, but may help to reduce the shift from ankle to hip work.

The increased ankle moment in CP especially during the first peak in early stance is consistent with the increased deviation in ankle kinematics seen in these patients during treadmill walking (Van der Krogt et al., 2014). The CoP shifted forward more quickly in TM in CP subjects, indicative of an increased toe-walking pattern. This could be due to increased fatigue as a result of the longer walking time in TM compared with OG. In contrast, the ankle moment in TD subjects was decreased, with a more backward shift of the CoP. This is not in line with results of Rozumalski et al. (2014,2015) who also found a more forward shift of the CoP and increased ankle moments in TD subjects. This difference could possibly be explained by the shorter belt used by Rozumalski et al. (2014,2015) compared with ours. Such a short belt could make subjects more cautious of their position on the belt, and hence look down more, in order to not get too far forward or backward. Furthermore, no VR screen or SP speed were used in their study, but we found these effects to be small both in healthy adults (Sloot et al., 2014a,2014b) and children with CP (Sloot et al., 2015a).

The increase in hip and knee abduction moments could be fully explained by a 3–4 cm wider step width seen in TM (Van der Krogt et al., 2014). This wider step width has been found also in previous studies of treadmill walking (Altman et al., 2012; Gates et al., 2012) and is likely due to the split belt (Altman et al., 2012). TD subjects partly compensated for this wider step width, by redistributing their weight more medially on the foot, as illustrated by a more medial CoP position relative to the foot. CP subjects did not show this displacement of the CoP, possibly due to a more limited flexibility in their gait pattern.

The individual differences found between OG and TM kinetic curves were on average 2–5 times larger than the normal stride-to-stride variation (Table 2), with several extremes up to 19 times for individual cases. This is in contrast to an earlier study performed with healthy adults, where the average differences were smaller than stride-to-stride repeatability (Riley et al., 2007). The larger differences in our study between conditions may partly be due to the fact that children, especially those with CP, may need more time to get used to the treadmill, SP walking, and virtual



**Fig. 3.** Horizontal distance between the hip joint center and the center of pressure (CoP) averaged over A. all typically developing (TD) children and B. all children with cerebral palsy (CP), for overground (cyan) and treadmill walking (red), with standard deviation. Note that the hip is further behind the CoP in TM than in OG. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

reality environment compared with adults. Indeed, the magnitude of differences was more in line with other comparisons made in children (Rozumalski et al., 2014, 2015). This stresses the need for sufficient habituation time on the treadmill especially in children.

An obvious limitation of our study is the limited sample size, with a small and heterogeneous group of CP patients. However, most of the differences found between OG and TM were considerable and highly significant despite the small group, except for some of the hip and ankle power parameters, with p-values around 0.10, which may have become significant with a larger group of subjects. This would even further strengthen the idea of the power shift from ankle to hip.

The considerable differences in kinetics found between OG and TM indicate that kinetic data collected on a TM cannot be readily compared with OG data, and that treadmill-specific normative data sets should be used when performing kinetic gait analysis on a treadmill. For clinical applications, it must be considered that both OG and TM conditions constitute highly controlled experimental settings. Therefore the key clinical question – which controlled situation is most relevant for clinical treatment decision and evaluation to improve walking performance – remains open.

#### Conflict of interest statement

This study was partly supported by Motekforce Link BV. We have in place an approved plan for managing any potential conflicts arising from this arrangement. The authors had full access to

all data in this study and take complete responsibility for the integrity of the data and the accuracy of the data analysis. Motekforce Link BV had no role in the study design; collection, analysis, and interpretation of data; writing the report; or the decision to submit the report for publication.

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