Gait & Posture 108 (2024) 110–116

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Muscle coactivation during gait in children with and without cerebral palsy

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ARTICLE INFO

Keywords:
Cerebral Palsy
Gait
Muscle co-activation
Electromyography
Biomechanics

ABSTRACT

Background: Children with Cerebral Palsy (CP) walk with an uncoordinated gait compared to Typically Developing (TD) children. This behavior may reflect greater muscle co-activation in the lower limb; however, findings are inconsistent, and the determinants of this construct are unclear.

Research objectives: (i) Compare lower-limb muscle co-activation during gait in children with, and without CP, and (ii) determine the extent to which muscle co-activation is influenced by electromyography normalization procedures and Gross Motor Function Classification System (GMFCS) class.

Methods: An electromyography system measured muscle activity in the rectus femoris, semitendinosus, gastrocnemius, and tibialis anterior muscles during walking in 46 children (19 CP, 27 TD). Muscle co-activation was calculated for the tibialis anterior-gastrocnemius (TA-G), rectus femoris-gastrocnemius (RF-G), and rectus femoris-semitendinosus (RF-S) pairings, both using root mean squared (RMS)-averaged and dynamically normalized data, during stance and swing. Mann-Whitney U and independent t-tests examined differences in muscle co-activation by group (CP vs. TD) and GMFCS class (CP only), while mean difference 95% bootstrapped confidence intervals compared electromyography normalization procedures.

Results: Using dynamically normalized data, the CP group had greater muscle co-activation for the TA-G and RF-G pairs during stance ($p < 0.01$). Using RMS-averaged data, the CP group had greater muscle co-activation for TA-G (stance and swing, $p < 0.01$), RF-G (stance, $p < 0.05$), and RF-S (swing, $p < 0.01$) pairings. Muscle co-activation calculated with dynamically normalized, compared to RMS-averaged data, were larger in the RF-S and RF-G (stance) pairs, but smaller during swing (RF-G). Children with CP classified as GMFCS II had greater muscle co-activation during stance in the TA-G pair ($p < 0.05$).

Significance: Greater muscle co-activation observed in children with CP during stance may reflect a less robust gait strategy. Although data normalization procedures influence muscle co-activation ratios, this behavior was observed independent of normalization technique.

1. Introduction

Cerebral palsy (CP) is characterized by difficulties with posture, coordination, and motor function [1]. These challenges are evident during gait, wherein muscle spasticity, contractures, and altered motor control contribute to an uncoordinated walking pattern [1,2]. Compared to their typically developing (TD) peers, children with CP walk with a more synchronous and less variable lower-limb coordination strategy [3,4]. This behavior is likened to a rigid gait pattern and may reflect an impaired ability to selectively activate muscles during gait [5]. These impairments contribute to pathological gait in CP and can negatively impact quality of life [6].

Maintaining stability while walking requires synchronization across muscle groups [7]. Muscular co-activation (MCa) is a biomechanical metric which quantifies timing and activation levels of muscle pairs surrounding a joint [8]. Thus, changes in MCa can help describe neuromuscular control during movement (e.g., gait), in that, excessive or insufficient MCa may suggest a dysfunctional motor strategy [9,10].

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https://doi.org/10.1016/j.gaitpost.2023.11.012
Received 2 May 2023; Received in revised form 10 October 2023; Accepted 19 November 2023
Available online 21 November 2023

In children with CP, greater MCa of the rectus femoris/semitendinosus and tibiailis anterior/triceps surae pairings has been observed during gait [11]. This behavior likely restricts joint motion and necessitates greater energy expenditure [12], contributing to an uncoordinated and inefficient gait [7,13]; however, there is heterogeneity in the literature, particularly with respect to how methodology and clinical metrics influence MCa [14].

A scoping review highlighted (i) heterogeneity in procedures used to quantify MCa, and (ii) variability in CP samples across studies, as key items to help explain inconsistencies in the literature [11]. Gagnat et al., reported that group differences in MCa during gait (CP vs. TD) changed based on which electromyography (EMG) normalization technique was used prior to calculating MCa ratios [14]. Additionally, children with CP are diverse in terms of walking ability [1]. The Gross Motor Function Classification System (GMFCS) helps categorize this heterogeneity; unfortunately, GMFCS class is often underreported and leaves a gap in understanding how walking ability is related to MCa [11].

Therefore, we aim to (i) compare lower-limb MCa during gait in children with, and without CP; and (ii) determine the extent to which MCa is influenced by EMG normalization procedures and GMFCS class, respectively. We hypothesized that (i) children with CP will have greater MCa than their TD peers, (ii) EMG normalization procedures will influence MCa ratios and (iii) greater MCa will be observed in children with GMFCS class II, compared to GMFCS class I.

2. Methods

2.1. Participants

Data from 46 children with, and without, CP were extracted from two partnering databases located at the Oxford Gait Laboratory (12 CP and 27 TD participants) and the New Jersey Institute of Technology (7 CP participants). Both groups were similar in age, height, and body mass ($p > 0.05$). Participant demographics are provided in Table 1. Inclusion criteria for participants with CP were a diagnosis of spastic diplegia, a GMFCS class of level I or II, the ability to follow verbal instructions, and age < 18 years. TD children (age < 18 years) were included if they had no history of gait, neurological, or musculoskeletal abnormalities. All parents/guardians provided written consent, and study procedures were approved by the local ethics boards.

2.2. Data collection

Kinematic and EMG data were collected while participants walked barefoot along a 10 m walkway at their comfortable gait speed. Participants were fit with retro-reflective markers based on a modified Plug-in Gait model [15] or the conventional gait model (CGM) 2.5 template [16]. A motion-capture system collected marker data at 100 Hz with 12–17 optoelectronic cameras (Vicon Motion Systems Ltd., Oxford, UK). Surface EMG data were collected using multi-channel systems (Delsys Inc., Natick, USA), at 2000 Hz. EMG sensors were placed bilaterally over the rectus femoris, semitendinosus, lateral gastrocnemius, and tibialis anterior muscles. Sensor locations were identified according to SENIAM guidelines [17].

Table 1
Demographics for children with cerebral palsy and their typically developing peers.

	CP		TD
Participants (n)	19		27
Age (years)	11.3 (3.2)		11.1 (3.0)
Height (cm)	145.8 (15.9)		150.8 (17.4)
Weight (kg)	40.4 (14.4)		43.0 (15.2)
Sex (males)	12		14
GMFCS	I (n = 11)	II (n = 8)	

Mean (standard deviation) for all demographics variables in children with cerebral palsy (CP) and their typically developing (TD) peers. GMFCS: Gross motor functional classification system.

2.3. Data processing and analyses

Initial data processing (e.g., marker labelling, trajectory filling) and computation of hip, knee, and ankle kinematics were conducted using Vicon Nexus (Vicon Motion Systems Ltd., Oxford, UK). Next, data were imported into Matlab (v2020a, The Mathworks Inc., Natick, USA) for further processing with the biomechZoo toolbox (v1.9.7) [18] and custom code. Kinematic data from the heel, toe, and sacrum markers were used to identify gait events (i.e., foot strike and foot-off) as described by Zeni et al. [19]. Hip, knee, and ankle sagittal plane joint angles were retained. Next, raw EMG data were full-wave rectified; linear enveloped using a low-pass 4th order Butterworth filter (frequency cut offs of 20 and 450 Hz); and smoothed using a root mean squared (RMS) moving window (50 ms) [20]. Processed data were then duplicated and (i) kept as is (RMS-averaged) or (ii) dynamically normalized to muscle specific maximum activations during gait cycles (dynamically normalized). Last, data were partitioned into stance and swing phases, using foot strike and foot-off gait events [19]. EMG waveforms were inspected for signal distortion and removed when judged to interfere with our analyses (n = 21/182 observations). Due to the symmetrical nature of motor limitation in our CP group, the lower limb with the most available trials (i.e., left lower limb) was retained for analyses, in both groups.

Muscle coactivation ratios were calculated for the rectus femoris and semitendinosus (RF-S), rectus femoris and gastrocnemius (RF-G), and tibialis anterior and lateral gastrocnemius (TA-G) muscle pairs [21]. This approach quantifies the simultaneously contraction of a pair of agonist-antagonist muscles during a movement cycle as a percentage of overall muscle activity. MCa was calculated based on the ratio of the common area under the curves of a muscle pair, to the sum of the areas under the curve for the respective muscles in that pair (Eq. 1). [22].

$$\text{Percent } coactivation = 2 \text{ x } \frac{\text{Common area of } A\&B}{\text{Area of } A + \text{B}} \text{ x } 100 \quad (1)$$

Participant MCa ratios during stance and swing phases for the RF-S, RF-G, and TA-G muscle pairs were calculated, for each available trial, using both RMS-averaged and dynamically normalized EMG data. For all participants and conditions, the RMS error (RMSE) between each MCa curve and the mean MCa curve was computed. The trial with the average overall minimum RMSE was selected as the representative trial and retained [23].

2.4. Statistical analyses

To test our hypothesis that lower-limb MCa ratios during gait differ in children with and without CP, independent sample t-tests were run. Separate tests were conducted for the RF-S, RF-G, and TA-G MCa ratios, during stance and swing phases, using both RMS-averaged and dynamically normalized EMG data. These data were checked for normality using QQ plots and the Shapiro-Wilk test, while Levene's test examined homogeneity of variance. To test our secondary hypothesis that lower-limb MCa ratios were greater in children with higher GMFCS class, independent sample t-tests were run on the same set of 8 dependent variables, comparing children with CP separated by GMFCS class (i.e., GMFCS I vs. GMFCS II). For all comparisons, if assumptions of normality or homogeneity of variance were not met, the Mann-Whitney U test or Greenhouse-Geisser correction were used, respectively (see results for test implemented). Mean difference and 95% confidence intervals (CI) were calculated. Cohen's d (d) and Glass's delta (Δ) determined effect sizes for the results of the t-tests and Mann-Whitney U tests, respectively. Level of significance for all tests was set at $\alpha = 0.05$. Statistical analyses were performed using SPSS (Version 23, IBM, Chicago, USA).

To test the hypothesis that EMG normalization techniques influence MCa ratios, mean difference and Cohen's d 95% bootstrapped confidence intervals were calculated for MCa pairings (TA-G, RF-G, and RF-S) [24,25]. This was done because RMS-averaged and dynamically normalized MCa ratios were judged to be distinct outcome variables and not appropriate for comparison using t-tests. Bootstrapping procedures were performed by group (CP, TD, full sample), for stance and swing phases. Participant data were resampled 1000 times, with replacement. For each iteration, differences between MCa (dynamically normalized) and MCa (RMS-averaged) were calculated. Based on these 1000 iterations, a 95% CI was determined for the averaged mean difference and Cohen's d. If the 95% CI did not include 0, we concluded that the approaches were significantly different [25]. Analyses were performed in Matlab (v2020a, The Mathworks Inc., Natick, USA).

3. Results

3.1. Dynamically normalized MCa ratios

Using dynamically normalized data, the CP group had greater MCa for the TA-G muscle pair during stance ($\bar{x}$ difference=18.4; CI=[11.4, 25.5]; *d*=1.64), but not swing phase ($\bar{x}$ difference=5.7; CI=[−4.7, 16.0]; *d*=0.34; Table 2, Fig. 1). Similarly, for the RF-G pair, the CP group had greater MCa during stance ($\bar{x}$ difference=11.5; CI=[3.3, 19.6]; *d*=0.86), but not swing phase ($\bar{x}$ difference= −4.2; CI=[−15.4, 7.0]; *d*=−0.23). Group differences in MCa were not observed for the RF-S muscle pair during stance ($\bar{x}$ difference=3.7; CI=[−4.6, 12.0]; $\mathbf{\Delta}$ = 0.27) nor swing ($\bar{x}$ difference=5.3; CI=[−2.3, 19.2]; *d*=0.50).

3.2. RMS-averaged MCa ratios

Using RMS averaged data, the CP group had greater MCa for TA-G during stance ($\bar{x}$ difference=17.4; CI=[9.7, 25.1]; *d*=1.45) and swing ($\bar{x}$ difference=13.1; CI=[4.6, 21.7]; $\mathbf{\Delta}$ = 0.92; Table 2, Fig. 1). The CP group also had greater MCa for RF-G during stance ($\bar{x}$ difference=9.2; CI=[1.4, 17.1]; *d*=0.73), but not swing ($\bar{x}$ difference=−6.8; CI=[−16.3, 2.7]; d=−0.43). For the RF-S pair, we observed greater MCa during swing ($\bar{x}$ difference=15.3; CI=[4.8, 25.8]; $\mathbf{\Delta}$ = 0.91), but not stance ($\bar{x}$ difference=0.9; CI=[−10.4, 12.1]; *d*=0.05).

3.3. Gross Motor Function Classification System (GMFCS) level

Using RMS-averaged EMG data, children with CP classified as GMFCS II had greater muscle co-activation during stance in the TA-G pair, compared to those classified as GMFCS I ($\bar{x}$ difference=10.1; CI=[2.0, 18.1]; *d*=0.35, Table 3). No other differences were observed.

3.4. Comparing normalization techniques

Considering all participants, MCa ratios determined using dynamically normalized EMG data showed larger MCa ratios in the RF-S (stance/swing) and the RF-G (stance) parings, but not TA-G (Table 4). When considered by group, MCa ratios using normalized EMG data were greater in both the CP (RF-S and RF-G pairings during stance) and the TD group (TA-G and RF-S pairings during swing, RF-G during stance). RF-G MCa ratios using normalized EMG data were smaller during swing, for all groups.

4. Discussion

Children with CP had greater MCa during stance (TA-G and RF-G pairs), independent of the normalization technique used. Using RMS-averaged data, greater MCa ratios were also seen during swing (RF-S and TA-G). Relative to RMS-averaging, dynamically normalization of EMG data increased MCa ratios mostly in the RF-S (stance/swing) and RF-G (stance) pairs; although, this observation varied depending on group and gait phase.

4.1. Differences between CP and TD (Using dynamically normalized data)

A trend of greater MCa in children with CP was observed for both muscle pairs and gait phases, although statistical differences were only observed for TA-G and RF-G during stance. Past work also reported greater MCa in TA-G during weight acceptance; however, their observation of greater MCa in RF-S during various sub-phases of gait were inconsistent with our results [14]. Differences may relate to sample characteristics, methodology to quantify MCa, or how gait was partitioned. To our knowledge MCa in RF-G hasn't been studied in the context of CP; however, increased RF-G MCa has been reported in healthy adults during distinct sub-phases of stance and swing [26]. Regardless, collectively, findings point to greater lower-limb MCa in children with CP during stance.

4.2. Differences between CP and TD (Using RMS-averaged data)

We observed greater MCa ratios for TA-G (stance, swing), RF-G (stance), and RF-S (swing) muscle pairs in children with CP. Others used comparable procedures and reported both similar (i.e., greater TA-G MCa during sub-phases of stance) and conflicting (i.e., greater RF-S during weight acceptance and no differences during swing) findings [14]. Otherwise, our results are somewhat aligned with work reporting greater MCa in TA-G and RF-S pairings in gait; however, these studies described MCa qualitatively and statistical comparisons with TD children were not performed [27–29].

4.3. Differences between GMFCS levels

Children with CP classified as GMFCS II had greater MCa in the TA-G

Table 2
Differences in muscle coactivation ratios in children with, and without cerebral palsy during gait.

EMG normalisation technique	Muscle Pair	Gait Phase	CP MCa Ratios	TD MCa Ratios	P-value
Dynamically normalized	TA-G	Stance	70.1 (9.7)	51.0 (13.1)	**< 0.01**
		Swing	43.6 (15.3)	36.3 (17.1)	0.27
	RF-S	Stance	64.9 [10.6]	59.1 [18.9]	0.37
		Swing	53.3 (18.3)	44.6 (17.3)	0.12
	RF-G	Stance	59.1 (12.4)	46.7 (14.1)	**< 0.05**
		Swing	48.9 (15.7)	51.9 (20.5)	0.46
RMS-avg	TA-G	Stance	68.2 (9.7)	50.5 (14.5)	**< 0.01**
		Swing	43.1 [14.6]	29.5 [10.4]	**< 0.01**
	RF-S	Stance	55.9 (16.9)	53.5 (19.1)	0.88
		Swing	55.5 [30.7]	34.0 [24.3]	**< 0.01**
	RF-G	Stance	49.3 (12.3)	39.5 (13.2)	**< 0.05**
		Swing	57.6 (17.7)	64.1 (14.3)	0.16

Mean (standard deviation) or median [interquartile range] muscle coactivation (MCa) ratios for the tibialis anterior-gastrocnemius (TA-G), rectus femoris-semitendinosus (RF-S), and rectus femoris-gastrocnemius (RF-G) muscle pairs during gait in children with, and without, cerebral palsy, using dynamically normalized and RMS averaged (RMS-avg) data. CP: cerebral palsy; TD: typically developing.

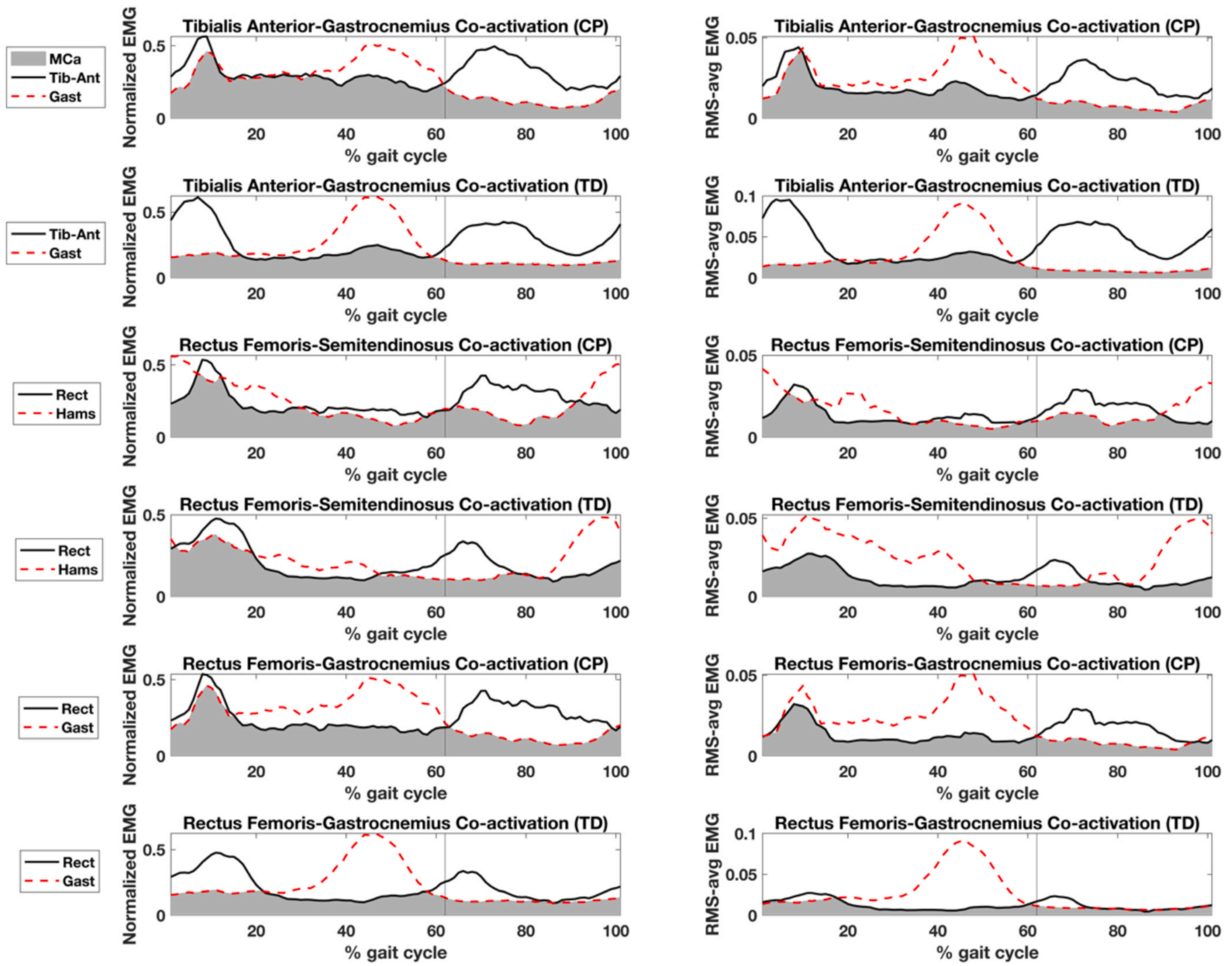

Fig. 1. Mean ensemble muscle co-activation ratios (MCa) and lower-extremity muscle activations during gait using both dynamically normalized (Normalized EMG) and RMS-averaged (RMS-avg) EMG data, for children with Cerebral Palsy (CP) and their typically developing (TD) peers. Shaded area represents the MCa for the respective muscle pairings. Vertical lines denote start of swing phase. Tibialis anterior (Tib-Ant), Gastrocnemius (Gast), Rectus femoris (Rect), Semitendinosus (Hams).

pair during stance, only when using RMS-averaged data. Since a correlation between GMFCS level and restricted motor development has been previously demonstrated [30], the lack of additional differences in our work was unexpected. To our knowledge, no comparable studies exist [11]. Regardless, our results suggest that MCa ratios may not differ greatly based on GMFCS class, possibly due to heterogeneity of impairments reflected in respective GMFCS levels [31]. Alternatively, a different motor classification (e.g., by gait pattern) may have led to different findings.

4.4. Clinical implications

Greater MCa in children with CP was observed in the TA-G and RF-G pairings during stance, a behavior that likely limits motor flexibility/adaptability and contributes to less robust gait [32]. Excessive MCa is also aligned with an in-phase and invariable lower-limb coordination strategy observed in CP [3], and may reflect difficulties with segment decoupling during movement (E.g., knee and ankle in early stance, Fig. 2) or decreased selective motor control [5]. Thus, findings of excessive MCa during challenging phases of gait (stance) are logical. Otherwise, greater MCa at the TA-G muscle pair may relate to the adoption of an ankle-dominant strategy to help maintain balance during gait [33], counteract deficits in plantar flexor muscle power [34], or serve as a "vaulting" mechanism to assist with contralateral foot clearance during swing [34]. Similarly, greater MCa at the RF-G may underlie dysfunctional motor control; although, this behavior likely restricts knee motion (Fig. 2) due to the flexor/extensor actions of the bi-articular muscles, and may help stabilize the knee [26]. The role of increased TA-G MCa during swing, however, is less clear. Increased MCa might suggest a compensatory strategy to control foot drop [35], or this co-activation may serve no functional role and result from decreased selective motor control or weakness. Last, greater MCa in the CP group may be due to decreased muscle activity of a select muscle (e.g., tibialis anterior). This could attenuate peaks in the signal of the CP, relative to TD group, resulting in higher MCa due to greater common area in the signals. However, lower muscle activation would be underscored by greater relative noise in the signal.

Greater MCa in the RF-S pairing was observed during swing in the CP group when using RMS-averaged data. During swing phase in TD children, knee extension is largely passive and followed by an eccentric lengthening of the of the hamstrings to slow extension [36]. In children with CP, however, this quick lengthening of the hamstrings would trigger a spastic response and may necessitate greater rectus-femoris contraction to actively extend the knee, resulting in greater MCa, as

Table 3
Comparison of muscle co-activation ratios during gait in children with cerebral palsy, based on gross motor function classification system level.

EMG normalisation technique	Muscle Pair	Gait Phase	GMFCS I n = 11	GMFCS II n = 8	P-value
Dynamically normalized	TA-G	Stance	67.5 (8.6)	72.7 (10.6)	0.25
		Swing	40.6 (12.6)	47.8 (17.6)	0.33
	RF-S	Stance	59.0 (17.2)	65.7 (6.6)	0.32
		Swing	47.1 (19.6)	61.0 (14.0)	0.11
	RF-G	Stance	58.4 (10.1)	59.9 (15.1)	0.80
		Swing	45.9 (17.0)	50.4 (15.3)	0.56
RMS-Avg	TA-G	Stance	63.7 (7.4)	73.8 (9.2)	**< 0.05**
		Swing	42.9 (11.4)	50.6 (16.4)	0.24
	RF-S	Stance	51.4 (18.2)	58.4 (17.4)	0.42
		Swing	48.7 (19.9)	55.3 (15.4)	0.44
	RF-G	Stance	47.2 (12.6)	50.9 (12.3)	0.53
		Swing	55.2 (21.1)	60.1 (10.8)	0.51

Mean (standard deviation) coactivation ratios for the tibialis anterior-gastrocnemius (TA-G), rectus femoris-semitendinosus (RF-S), and rectus femoris-gastrocnemius (RF-G) muscle pairs during gait in children with cerebral palsy, using dynamically normalized and RMS averaged (RMS-Avg) data. Data are segregated based on gross motor function classification system (GMFCS) levels.

observed herein.

4.5. Effect of EMG normalization technique

When comparing normalization techniques, dynamic normalization generally led to larger MCa ratios relative to RMS-averaging, except in the RF-G pair during swing. Larger MCa ratios may be partially explained by the presence of lower-limb weakness in CP, wherein normalizing to a % max value of a muscle exhibiting low activity may exaggerate MCa values [37]. Why this occurred predominantly in the RF-S and RF-G pairs, and also in healthy participants is less clear. One explanation may be linked to the similar waveform dynamics and amplitude of the tibialis anterior and gastrocnemius RMS-averaged EMG signals (Fig. 1). These similarities could attenuate the effects of scaling muscle activity to a common value during dynamic normalization procedures, resulting in similar MCa ratios, regardless of whether RMS-averaged or dynamically normalized data were used. In contrast, the other pairings have more distinct relative maximums (i.e., rectus femoris vs. semitendinosus and rectus femoris vs. gastrocnemius RMS-averaged curves), which can distort the waveforms across the dynamic normalization approach due to the impact of scaling data to a common value (Fig. 1). This may misrepresent relationships between waveforms and incorrectly distort MCa values, in the positive or negative direction; thus, using RMS-averaged data to calculate MCa may be more appropriate. Qualitatively, Table 2 shows that we observed greater MCa in children with CP in 4/6 group comparisons when using RMS-averaged data, compared to only 2/6 comparisons when using dynamically normalized MCa ratios. This may reflect increased sensitivity in detecting group differences when using less processed data to calculate MCa ratios (i.e., RMS-averaged instead of dynamically normalized EMG signals), possibly to due to improved data fidelity. This conflicts, however, with similar work reporting more group differences in MCa during gait in children with CP when using amplitude normalized, compared to RMS-averaged data [14]. Pending standardized procedures, researchers might consider this issue with respect to their research question and population of study. A discussion will highlight some considerations.

Normalization procedures are typically performed to reduce non-physiological variability in EMG data and improve comparability between muscles and study participants [38]. Procedures can involve referencing data to a standard value, such as maximal voluntary isometric contraction [39]; however, it is unclear if such procedures accurately reflect maximal activation capacity in people with CP [40]. Others dynamically normalize EMG signals to a local maximum obtained during the gait cycle [2]; however, this assumes that a local maximum would occur at the same gait phase in children with CP and their TD peers. Thus, some question the use of normalization procedures altogether when calculating MCa in a neurological population and argue that absolute EMG data be used instead [37]. Further, the construct of MCa is typically represented as a ratio and serves as a form of normalization, possibly limiting the need for further data transformation [41]. Overall, further research is required; however, our data suggest that

Table 4
The influence of normalization techniques on muscle coactivation ratios during gait.

Group	Muscle Pair	Gait Phase	Mean difference (95% CI)	Cohen's d (95% CI)
All participants	TA-G	Stance	1.03 [− 1.07, 2.92]	0.07 [− 0.07, 0.20]
		Swing	1.61 [− 1.62, 4.79]	0.10 [− 0.10, 0.32]
	RF-S	Stance	**5.96 [1.70, 10.87]**	**0.38 [0.11, 0.68]**
		Swing	**5.50 [2.21, 9.35]**	**0.31 [0.12, 0.54]**
	RF-G	Stance	**8.84 [6.53, 11.39]**	**0.65 [0.44, 0.88]**
		Swing	**-11.02 [− 15.33, − 6.52]**	**-0.65 [− 0.95, − 0.37]**
CP only	TA-G	Stance	1.67 [− 2.29, 5.28]	0.19 [− 0.23, 0.65]
		Swing	-2.60 [− 7.54, 1.99]	-0.19 [− 0.60, 0.15]
	RF-S	Stance	**7.54 [0.59, 15.3]**	**0.51 [0.03, 1.10]**
		Swing	1.89 [− 1.36, 5.78]	0.11 [− 0.08, 0.34]
	RF-G	Stance	**10.20 [6.13, 14.49]**	**0.87 [0.53, 1.27]**
		Swing	**-9.38 [− 15.1, − 2.70]**	**-0.59 [− 1.07, − 0.15]**
TD only	TA-G	Stance	0.71[− 1.15, 2.75]	0.05 [− 0.09, 0.22]
		Swing	**4.91 [0.39, 9.81]**	**0.32 [0.03, 0.69]**
	RF-S	Stance	4.91 [− 0.90, 10.71]	0.31 [− 0.05, 0.71]
		Swing	**8.65 [3.66, 14.15]**	**0.55 [0.21, 1.02]**
	RF-G	Stance	**8.00 [5.12, 11.18]**	**0.62 [0.34, 1.04]**
		Swing	**-12.08 [− 18.10, − 5.91]**	**-0.72 [− 1.15, − 0.33]**

Mean difference and 95% confidence intervals comparing the influence of normalization techniques (RMS-averaged vs. dynamically normalized) on muscle coactivation ratios in children with and without cerebral palsy during gait. Positive mean differences indicate larger MCa values when using dynamically normalized, compared to RMS-averaged, EMG data. Bolded denotes statistical significance. MCa: Muscle Coactivation; CP: Cerebral palsy; TD: Typically developing; TA-G: Tibialis anterior-gastrocnemius; RF-S: Rectus femoris-semitendinosus; RF-G: Rectus femoris-gastrocnemius.

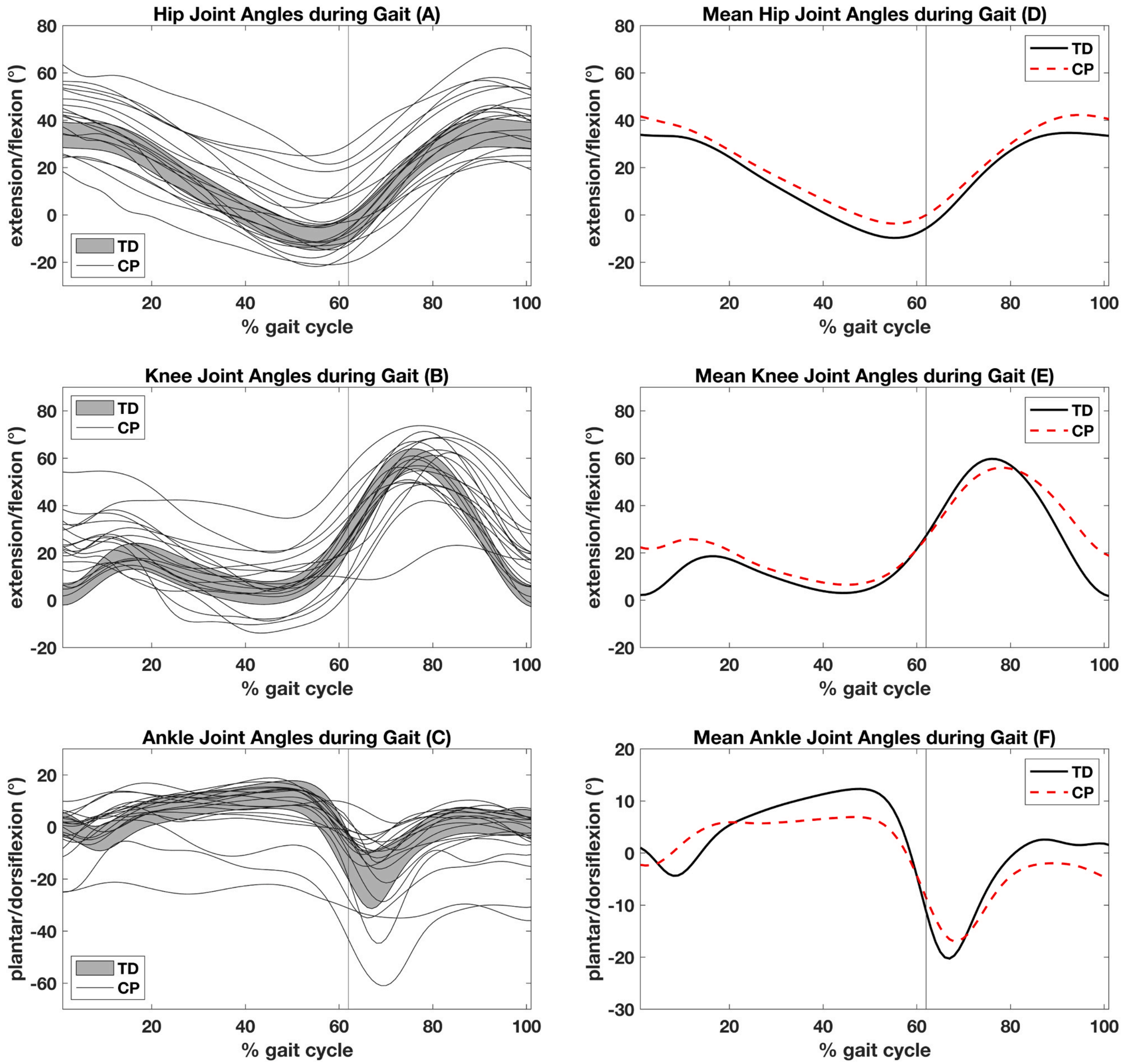

Fig. 2. Lower extremity sagittal plane joint angles during gait in children with Cerebral Palsy (CP) and their typically developing (TD) peers. A-C: Representative gait trials for all CP participants (light grey lines) and group mean + /- standard deviation for TD children (shaded grey area). D-F: Ensemble-averaged waveforms for children with CP (dashed red line) and their TD peers (solid black line). Vertical line denotes start of swing phase.

calculating MCa ratios using RMS-averaged data may be more appropriate.

4.6. Limitations

We only considered single muscle pairs in MCa calculations. Considering more muscles when comparing MCa in dynamic activities such as walking might provide a more complete picture. Next, the gait cycle was partitioned into 2 phases (i.e., stance and swing). Including more sub-phases (e.g., 6 phases as Gagnat et al. [14]) may have generated some additional findings; however, this process results in fewer data points in certain phases (e.g., pre-swing phase [36]), which may make the data more sensitive to aberrations. Last, our MCa index focuses on magnitude of co-activation, while temporal information is lost. Future work may consider comparing traditional measures of MCa with emerging methods [42].

5. Conclusion

Children with CP have greater MCa than their TD peers, at the ankle joint (tibialis anterior-gastrocnemius pairing) and knee (rectus femoris-gastrocnemius) during stance. This muscle activity pattern is aligned with a rigid and invariable gait strategy, possibly due to decreased selective motor control. The EMG normalization technique used impacts the resulting MCa ratios, which emphasizes that clinical decisions should not be taken based solely on these ratios. Rather, clinicians may consider incorporating MCa ratios alongside traditional gait metrics common in gait reports.

Declaration of Competing Interest

None.

Acknowledgements

All authors acknowledge the contribution of the Oxford Gait Laboratory and New Jersey Institute of Technology gait laboratory staff for data collection and pre-processing. P. Ippersiel is supported by a scholarship from the TransMedTech Institute and its main financial partner the Apogee Canada first Research Excellence Fund, and a Fonds de Recherche du Québec Santé(FRQS) post-doctoral fellowship. P.C. Dixon acknowledges the Fonds de Recherche du Québec Santé(FRQS) Junior 1 Research Scholar grant program.

Code and sample data associated with investigation are available at

https://github.com/PhilD001/cp_muscle_cocontraction.

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