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KINETICS OF TURNING GAIT IN TYPICALLY DEVELOPING CHILDREN

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INTRODUCTION

Turning is an essential component of both in-home and community ambulation [1]; however, knowledge of the biomechanics of turning is rather limited, especially in children. Preliminary work by our group has revealed the existence of spatio-temporal differences [2] and lower-limb kinematic changes mainly in the coronal and frontal planes in this population, but the underlying neuromuscular control mechanisms required to complete turning tasks remain unknown. Analysis of joint kinetics may provide a better understanding of these processes and, taken in combination with previous results, will lead to a broader understanding of turning biomechanics. Moreover, in a clinical setting, turning may help expose underlying weaknesses and motor control deficits, making it a relevant task for planning the management of gait disorders including those arising from cerebral palsy (CP) [3].

Therefore, the aim of this study was to analyse the lower-limb joint reaction moments and power during 90° step and spin turns in comparison to straight-line walking in typically developing children. We hypothesized that modified coronal and transverse plane kinetics and decreased power generation, mainly at the ankle, compared to straight-line gait would be observed [4].

METHODS

The gait data from forty-three healthy children were extracted from our laboratory normative data set for the purposes of this study. All subjects were free of gait abnormalities and provided written, informed consent prior to the gait analysis sessions. Subjects were instructed to perform straight and 90° turning tasks, while fitted with both the Plug-in Gait (PiG) and Oxford Foot Model (OFM) marker sets, resulting in either step (outside leg) or spin (inside leg) turning trials based solely on subject preference. Marker data were collected at 100 Hz via a 12-camera motion capture system (Vicon, Oxford, UK) and simultaneously two force plates (Advanced Mechanical Technology, Inc, Watertown, US) collected ground reaction force data at 1000 Hz.

Knee and hip joint kinetics were computed by the Vicon Nexus software package (v.1.7 Vicon, Oxford, UK) while the ankle (hindfoot with respect to tibia (HF/TB)) kinetics were calculated via an inverse dynamics analysis approach applied to the OFM [5] using Matlab (v2011b, The Mathworks Inc, Natick, USA). Turning style (spin or step) was determined by identifying the single-limb support phase step associated with the largest transverse plane pelvic range of motion. Thus, if the ipsilateral limb was in stance during this period, the trial was identified as spin, otherwise the trial was considered a step turn. Data were partitioned over the stance phase (SP) from foot-strike to foot-off using the first and last frame of ground reaction force data and normalized to 100%. Kinetic data for the ankle are presented during forefoot only contact (heel rise (HR) to foot-off) since, without additional hardware (such as a pressure plate) or careful foot positioning, the inverse dynamics problem cannot be solved whilst both foot segments (hindfoot and forefoot) are in contact with the ground [5].

As not all subjects performed all conditions, a between-subject design with three groups of ten subjects (straight, spin, and turn) was implemented (Table 1). A representative trial for each subject within each group was obtained based on the minimum average root mean squared error compared to the subject mean. Variability over the entire stance phase was estimated using 95% confidence bands (CBs) computed via the Bootstrap approach as point-by-point Gaussian-based confidence intervals (CIs) may lead to an underestimation of variability [6]. All quantities were normalized to body weight. Net internal joint reaction moments are presented. Statistical analyses were conducted by investigating the mean difference 95% CBs. Statistical significance was achieved (at α = 0.05 level) when the paired differences did not contain zero [6].

Table 1: Subject Anthropometric Data by Group

Group	Age (years)	Height (m)	Weight (kg)	Sex
Straight	11.1 ± 3.2	1.4 ± 3.2	36.5 ± 14.5	5F
Spin	10.3 ± 2.6	1.4 ± 0.2	34.2 ± 9.8	6F
Step	12.0 ± 2.6	1.5 ± 0.1	45.4 ± 10.7	7F

Mean ± standard deviation

RESULTS AND DISCUSSION

Starting with the sagittal plane moments (Figure 1), a number of significant differences were found for the ankle, knee, and hip mainly between the step and straight groups: increased ankle plantarflexor moment following HR, increased knee flexor moment in late stance, and decreased hip flexor moment in late stance. In the coronal plane, the greatest differences occurred at the ankle with the spin group showing mainly an evertor moment after HR. At the knee and hip, decreased valgus and abductor moment, respectively, were found for the spin group near midstance.

Finally, in the transverse plane, the step group revealed decreased ankle external rotator moment in late stance.

For the joint power data (Figure 1), differences were mostly seen for the step group. The ankle showed a brief, but significant, decrease in power generation during the push off phase. The knee revealed decreased power generation in early stance and generation rather than absorption during late stance while at the hip the power profile showed an increase followed by a decrease in power generation.

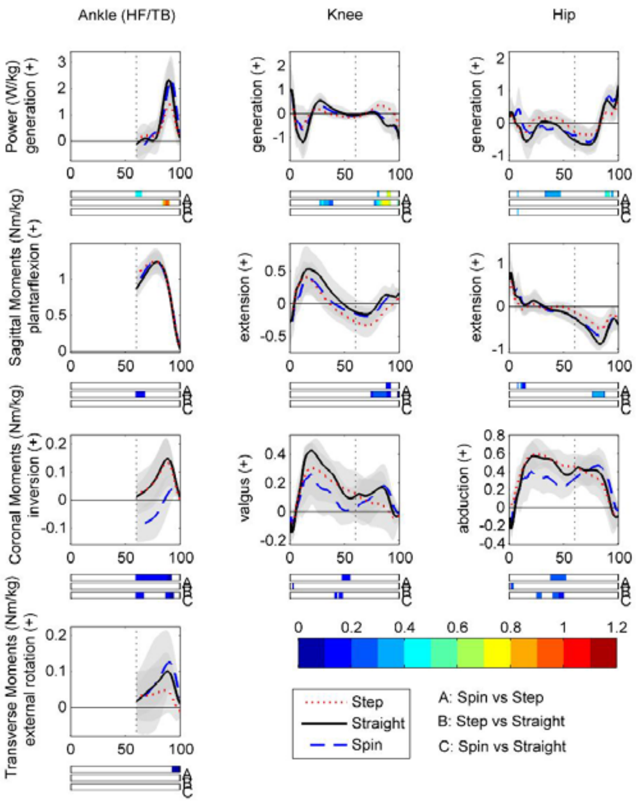

Figure 1: Ankle (HF/TB), knee, and hip net internal joint moments and power normalized to 100% of SP. Dashed vertical line represents estimated timing of HR. All data presented with corresponding bootstrap CBs. Colorbars show areas where mean difference CBs revealed significant differences ($p < 0.05$). Intensity of colorbar varies from dark blue (0.1) to dark red (1.2).

This study investigated the lower-limb kinetics of 90° turning in typically developing children and found that both step and spin turns require a number of specific coordinated biomechanical adaptations compared to straight walking. In general, the step turn seemed to affect sagittal plane moments and power while the spin turn was best characterized by coronal plane changes. These differences may reveal separate motor control strategies used to control balance and body reorientation between turn types.

Our results for straight walking are in agreement with previous age-matched normative data, while, as hypothesized, our turning data generally agrees with previous results from a study in adults by Taylor et al. [4]. For the ankle, most notable differences between studies were found in the sagittal plane and power generation data perhaps due to the differences in biomechanical models used (single rigid foot PiG vs. multi-segment foot OFM). The subdivision of the foot into hindfoot and forefoot segments allows for the isolation of the ankle kinetic behavior and therefore does not include the activity of the midfoot musculature and supportive structures present in the PiG estimations [5]. Nonetheless, greater changes in power generation profiles during the push-off phase were expected for the turn conditions. Investigation of midfoot kinetics [5] and increasing sample size may help provide more definitive data regarding the role of the foot and ankle during turning. For the knee and hip, step turn results are similar between studies; however, it is difficult to compare our current results for the spin turn group as Taylor et al. identified two spin sub-strategies [4]. The sub-strategies are likely to emerge at the higher gait velocities used by the subjects in the adult study [4]. More generally, adaptations, particularly in the sagittal plane, identified in both studies are consistent with decreased stride-length observed in our previous work [2].

Transverse plane kinetics at the knee and hip were not presented due to the inherent inaccuracies in this plane using the PiG model. More robust methods need to be explored in order to accurately describe motion in this plane during turning gait.

The statistical analysis procedure implemented in the current study represents a robust approach allowing the identification of differences between groups or conditions over the entire stance phase; however, the method is blind to differences arising due to covariates, such as gait velocity, that has been shown to decrease during turning [2]. Further work may need to be undertaken to verify the effect of gait speed on turning kinetics.

CONCLUSIONS

Turning is a crucial component of gait and may be relevant to investigate in a clinical setting [3]. The biomechanical adaptations that occur during turning may be difficult for children with gait disabilities, such as those arising from CP, to accomplish as this clinical population often present with lower-limb joint instability, weakness, and contractures. Turning gait may expose underlying problems that are not obvious in straight gait and may help classify severity of involvement and lead to improvements in surgical interventions and rehabilitative outcomes.

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