Kinematic modifications of the lower limb during the acceleration phase of bend sprinting.

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Kinematic modifications of the lower limb during the acceleration phase of bend sprinting

Running title: Joint kinematics in the acceleration phase of bend sprinting

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Key words: athletics, curve, motion capture, adduction

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Abstract

A decrease in speed when sprinting on the bend compared with the straight has been attributed to kinetic, kinematic and spatiotemporal modifications. Although maximal speed is dependent on an athlete's ability to accelerate, there is limited research investigating the acceleration phase of bend sprinting. This study used a lower limb and trunk marker set with 15 optoelectronic cameras to examine kinematic and spatiotemporal variables of the lower limb during sprinting on the bend and straight. Nine sprinters completed up to six 30 m maximal effort trials in bend (radius 36.5 m, lane one) and straight conditions. An increase in body lateral lean at touchdown resulted in a number of asymmetric kinematic modifications. Whilst the left limb demonstrated a greater peak hip adduction, peak hip internal rotation and peak ankle eversion on the bend compared with the straight, the right limb was characterised by an increase in peak hip abduction. These results demonstrate that kinematic modifications start early in the race and likely accumulate, resulting in greater modifications at maximal speed. It is recommended that strength and conditioning programmes target the hip, ankle and foot in the non-sagittal planes. In addition, sprint training should prioritise specificity by occurring on the bend.

Key words: athletics, curve, motion capture, adduction

In track and field sprint events longer than 100 m, more than half the total distance is ran on a curved portion of track (Meinel, 2008). Compared with the straight, anti-clockwise bend sprinting elicits a decrease in maximum speed at approximately 40 m (Churchill, Salo, & Trewartha, 2015). This reduction in speed has been attributed to kinetic, kinematic and spatiotemporal alterations (Churchill, et al., 2015; Churchill, Trewartha, Bezodis, & Salo, 2016). However, the current understanding of biomechanical modifications during anti-clockwise bend sprinting in the acceleration phase (0 - 30 m) is limited. Identification of these affected parameters could aid overall race performance through the development of targeted training programmes and increased specificity of athlete preparation.

During bend sprinting, mean body lateral lean angles of 14° and 11° at touchdown have been reported in the left and right step, respectively (Churchill et al., 2015). It is thought this lean is responsible for inducing a number of kinematic changes of the lower limb (Churchill et al., 2015) which occur predominantly in the frontal and transverse planes (Alt, Heinrich, Funken, & Potthast, 2015; Churchill et al., 2015). More specifically, the left limb is characterised by a mean increase in left peak hip adduction of approximately 6° during bend sprinting at 40 m compared with the straight (Alt et al., 2015; Churchill et al., 2015). Furthermore, a high peak left ankle eversion angle (e.g. 13° Alt et al., 2015; > 35° Luo & Stefanyshyn, 2012) has been reported, but in protocols that are not representative of a competitive elite environment, for example, at submaximal effort (Alt et al., 2015) and with a smaller radius (2.5 m, Luo & Stefanyshyn, 2012). It has been suggested that excessive and repetitive eversion may result in injury (Clarke, Frederick, and Hamill, 1984) thus highlighting the importance of further investigation with a protocol more closely replicating race conditions.

In the right limb, a mean 4° increase in internal knee rotation on the bend compared with the straight is thought to contribute to a rotational strategy which serves to control horizontal plane motion (Alt et al., 2015). This finding did not reach the alpha level $p < 0.05$, but due to the small sample size (n = 6) was reported as a tendency ($p < 0.1$, Alt et al., 2015), suggesting this should be interpreted cautiously until further evidence is available. However, bend sprinting did result in a 3° increase in peak right ankle external rotation compared with the straight ($p < 0.05$, (Alt et al., 2015). Despite Alt et al. (2015) providing some initial findings using a controlled submaximal velocity, gaining further evidence during representative performance conditions (such as during acceleration) would enhance the current evidence base.

Spatiotemporal parameters such as contact time, step frequency and step length are fundamental components of sprint performance with these parameters being affected during bend sprinting. For example, Alt et al. (2015) found an increase in left contact time on the bend compared with the straight at submaximal speed. This increase in contact time is consistent with others during the maximal speed phase (approximately 40 m, Churchill et al., 2015; Churchill et al., 2016; Ishimura & Sakurai, 2010; Ishimura, Tsukada, & Sakurai, 2013). However, the evidence base regarding spatiotemporal variables is sometimes contradictory. For example, a reduction in right step length on the bend compared with the straight at maximal speed has been reported by several authors (Churchill et al., 2015;

Churchill et al., 2016; Ishimura, et al., 2013). This reduction was considered responsible for a loss of speed on the bend compared with the straight (Churchill et al., 2015), highlighting the importance of spatio-temporal variables and their relationship with performance. However, step length was unaffected during sub-maximal effort bend sprinting (Alt et al., 2015). Moreover, the majority of available research has focussed on the maximal speed phase of bend sprinting (Alt et al., 2015; Churchill et al., 2015; Churchill et al., 2016; Ishimura et al., 2013). The limited research available in the acceleration phase showed both left and right step lengths were reduced on the bend compared with the straight (Stoner & Ben-Sira, 1979). Therefore, further research is required regarding the effect of the bend during the acceleration phase on spatiotemporal aspects of technique and performance.

Alt et al. (2015) suggested some modifications may be velocity dependent. Thus, during the acceleration phase, where athletes have not yet reached maximum speed, the kinematic demands of bend sprinting may be different. Whilst there is always an element of acceleration during bend sprinting due to constant change of direction, for the purpose of comparisons with straight-line sprinting, the acceleration phase during this paper is considered to occur at 0-30 m. It is possible that modifications such as increased hip adduction and ankle eversion are less prominent during acceleration. Moreover, the maximum speed a sprinter is able to attain is dependent on the sprinters' ability to accelerate. However, the acceleration phase has received little attention within the bend sprinting literature. Therefore, the aim of the present study was to investigate the effect of bend sprinting on the kinematic and spatiotemporal parameters of the lower limb during the acceleration phase. It was hypothesised bend sprinting would result in greater adaptations in the frontal and transverse planes than on the straight.

Method

Participants

Ethical approval was provided by the Sheffield Hallam Research Ethics Committee. Nine male sprinters (mean age 22 ± 4 years; body mass 71.48 ± 9.47 kg; stature 1.81 ± 0.06 m) with experience of bend sprinting (200 and /or 400 m) volunteered to participate in this study. The sample size was guided by previous bend sprinting literature (Alt et al., 2015; Churchill et al., 2015; Churchill et al., 2016; Judson et al., 2019) and the number of available skilled athletes meeting the inclusion criteria of the study. To standardise ability with previous research (Alt et al., 2015, 22.60 ± 0.33 s; Churchill et al., 2015, 22.15 ± 0.93 s), the inclusion criteria required a 200 m personal best of 23.5 s or faster (mean 22.70 ± 0.49 s, range 21.8 - 23.43 s). All athletes were active in training and injury free at the time of data collection. The study procedures were fully explained to participants who subsequently provided written informed consent.

Experimental set-up

Kinematic data were collected at 200 Hz, using a 15-camera (13 x Raptor model and 2 x Eagle model, Motion Analysis Corporation (MAC), Santa Rosa, CA, USA) optoelectronic motion capture system (calibration volume: 7 m long, 3 m wide and 1.5 m high). Data were recorded at approximately 10 - 17 m of the 30 m sprints. For the identification of gait events, a force plate (Kistler, Model 9287BA, 900 x 600 mm) was embedded into the track surface at approximately 12 m. For full details of the experimental set-up, please refer to Judson et al. (2019).

A modified Vicon Plug in Gait (PiG) marker set (lower limb and trunk; Judson, Churchill, Barnes, Stone, & Wheat (2017)) was used to model the torso, pelvis, thighs, shanks and foot segments (toebox, forefoot, rearfoot). For full details of marker locations please see (Judson et al., 2018). The marker set was applied by the same researcher for all participants.

Protocol

Data collection took place on a standard flat indoor track surface. A bend replicating lane 1 (radius 36.5 m) of a standard 400 m running track (IAAF, 2008) was reconstructed and a 30 m section of straight track was used for straight-line trials. The order of bend and straight trials were randomised between participants to minimise order effects. Participants completed a typical warm-up followed by up to six maximal effort trials for 30 m from starting blocks in both bend and straight conditions. Athletes were instructed to sprint at maximal effort for the full 30 m, and '*on your marks, set, go*' signal was used. To avoid the onset of fatigue, approximately eight minutes were allowed between trials (Churchill et al., 2015). Participants wore their own sprint spikes for the testing session.

Data processing

3D marker coordinate data were tracked using Cortex software (version 5.3, Motion Analysis Corporation, Santa Rosa, CA, USA) and automatic gap filling (cubic spline) performed on any gaps <10 frames. Raw marker positions were filtered at 18 Hz using a low-pass, fourth order recursive Butterworth filter. The cut-off frequency was determined with the use of residual analysis. Segments, local coordinate systems and joint centres were defined using Visual 3D software (version 6, C-Motion, Rockville, MD, USA) in accordance with ISB guidelines ((Wu et al., 2002; Wu et al., 2005). de Leva (1996) was used to estimate body segment inertial parameters. At the foot, segment inertia values were then adjusted by 150 to 189 g representing the mass of individual participants' spiked shoe according to manufacturer guidelines.

Vertical force data was used to identify touchdown and take-off events, where the mean plus two standard deviations of the vertical ground reaction force (with zero load on the force plate) was used as a threshold (Bezodis, Thomson, Gittoes, & Kerwin, 2007). All variables were measured individually for the left and right step. The foot that initiated the step defined whether the step was left or right. For touchdown of the second foot contact, or trials where the athlete did not make contact with the force plate and so force data was not available, methods described by Bezodis et al. (2007) were used where the mean plus two standard deviations of the fifth metatarsal head vertical coordinates in the static trial were used as a threshold to detect touchdown and take-off. Spatio-temporal variables were calculated following the methods of Churchill et al. (2015). The first central difference technique was calculated using the horizontal distance travelled in the anterior direction by the CoM to give *absolute speed. Race velocity*, which provides a measure of performance in terms of official race distance, was calculated by first using a four-quadrant inverse tangent to calculate the angle between the x and z CoM position at each time point. The difference between angles at two consecutive time points was used to calculate race displacement. Finally, first central difference technique was then used to calculate instantaneous velocity of the CoM relative to the race line. Similarly, *race step length* is a measure of the length of official race distance travelled with each step and was calculated using a the angle between the MTP at two consecutive ground contacts was calculated (θ), then multiplied by the radius of the race line (36.7 m). *Directional step length* was calculated relative to the direction of travel. A vector between the horizontal positions of the 2nd metatarsal head at consecutive ground contacts was created. A step progression vector was then created between the horizontal positions of the CoM at consecutive ground contacts and divided by its norm to create a unit vector. The dot product of the two vectors gave directional step length. *Step frequency* was calculated as absolute speed divided by directional step length. *Touchdown distance* was calculated as the horizontal displacement between the CoM and second MTP joint at touchdown. *Contact time* was the time from touchdown to take-off of the same leg and *flight time* the total step time (touchdown of one foot to touchdown of the contralateral foot) minus contact time. *Turn of centre of mass* (CoM; a measure of how much 'turning' occurred) was calculated using the angle between CoM progression vectors during the flight phase before and after the ground contact of interest.

Joint angles were defined as the distal segment relative to the proximal segment. Joint angles were calculated using the cardan sequence *zxy* (multi-segment foot angles: *zyx*) and cropped to the stance phase. To enable standardisation with previous bend sprinting research (e.g. Alt, et al., 2015), peak angle during stance was then calculated and averaged across three trials for each participant. For ease of interpretation, values of the left limb were multiplied by minus one. Body lateral lean at touchdown was calculated following methods of Yeadon (1990) in Matlab (v2017a, Mathworks, Natick, USA) and used as a measure of how much the athletes were 'leaning' into the bend (Churchill et al., 2015).

Statistical analysis

Shapiro-Wilk normality test ($p > 0.05$) was used to confirm normal distribution of data. Differences between the bend and straight conditions for the left and right limb were assessed using a two way repeated measures Analysis of Variance (ANOVAS), (condition: bend vs. straight, limb: left vs. right) for each dependent variable. Owing to the small sample size, effect size was also implemented for the interpretation of results. Cohen's *d* provides an estimate of effect with a population, and so can be biased for small samples (Lakens, 2013). Therefore, Hedges's *g* was used, which includes a correction for small sample size. Effect size (*g*) was interpreted based on Cohen (1988) guidelines where $g < 0.20$ represents a trivial difference, $0.20 \geq 0.50$ indicating a small difference, $0.50 \geq 0.80$ a moderate difference and ≥ 0.80 a large difference between means.

Results

Joint kinematics

For joint kinematics, there was a condition x limb interaction for peak hip abduction joint angle ($F_{(1,\,8)} = 6.075$, $p = 0.039$), with the right limb being more abducted on the bend compared with the straight (Table 1). There was a condition x limb interaction for peak hip adduction angle, $F_{(1,\,8)} = 12.093$, $p = 0.008$. Peak left step hip adduction was greater on the bend (8°) compared with the left step on the straight (4°) and the right step on the bend (6°, Table 1). In addition, a large effect size suggesting higher peak left hip external rotation (*g* = 0.89) on the bend compared with the straight, F $_{(1,\,8)}$ = 3.859, *p* = 0.085. There was a condition x limb interaction for body lateral lean at touchdown, *F* $_{(1,\,8)}$ = 26.697, *p* = 0.001 which was greater in both the left and right step on the bend (left step -5°; right step -12°) compared with the straight (left step 6°; right step -5°). Left step peak ankle internal rotation was greater on the bend compared with the straight and the right step on the bend resulting in a condition x limb interaction (*F* $_{(1,\,8)}$ = 17.091, *p* = 0.003). Although no main effect was reported for peak ankle eversion (*F* $_{(1,\,8)}$ = 1.247, *p* = 0.297), left step peak ankle eversion was 55% greater on the bend compared with the straight (*g* = 0.88). No significant interactions were reported for any variables at the knee.

***** Table 1 near here *****

Spatiotemporal variables

There was no main effect for condition on absolute speed, *F* $_{(1,\,8)}$ = 0.574, *p* = 0.47. For race velocity, which takes into consideration the progression of the athlete with respect to the actual race distance, there was no main effect for condition (*F* $_{(1,\,8)}$ = 2.673, *p* = 0.141, Table 2). However, there was a significant condition x limb interactions (*F* $_{(1,\,8)}$ = 19.467, *p* = 0.002) due to shorter race step lengths on the bend compared with the straight (Table 2).

For step frequency, there was a significant condition x limb interaction, *F* $_{(1,\,8)}$ = 12.144, *p* = 0.008, due to the left step on the bend being lower compared with the left step on the straight (*g* = 0.66) and the right step on the bend (*g* = 0.61). There was a condition x limb interaction for touchdown distance, *F* $_{(1,\,8)}$ = 5.477, *p* = 0.04, where left step touchdown distance was longer on the bend (0.30 ± 0.05 m) compared with the straight (0.25 ± 0.05 m).

***** Table 2 near here *****

Discussion

The aim of this study was to investigate the effect of bend versus straight-line sprinting during the acceleration phase of a race on the kinematic and spatiotemporal parameters of the lower limb. During bend sprinting, a non-significant increase in peak left hip adduction (8°, *g* = 1.09), combined with a non-significant increase in peak left hip external rotation (-14°, *g* = 0.89) was reported compared with straight-line sprinting. This supports research during sub-maximal effort bend sprinting at approximately 9.26 m/s (Alt et al., 2015) which also reported peak left hip adduction (14°) and external rotation (22°). The excessive hip adduction observed might have implications for injuries, particularly at the knee (Li et al., 2015). For example, it is expected that iliotibial band tension would increase with hip adduction, potentially resulting in iliotibial band syndrome (Chuter & Janse de Jonge, 2012; Powers, 2010). Therefore, the high peak hip adduction observed during bend sprinting may be a precursor for injury. Strength and conditioning programmes should aim to ensure the hip joint is capable of withstanding high loads and prevent long-term implications for athletes.

The present study observed a large, but non-significant, increase in peak left step ankle eversion on the bend compared with the straight (*g* = 0.88). These findings support the theory from Alt et al. (2015) that the left limb is associated with a stabilising role achieved through the combination of greater hip adduction and ankle eversion. Whilst increased eversion enables the attenuation of impact forces (Hreljac, 2004), it is also linked with medial tibial stress syndrome and patellofemoral pain syndrome (Chuter & Janse de Jonge, 2012) both of which were amongst the most frequently reported injuries in indoor bend sprinters over a season (Beukeboom, Birmingham, Forwell, & Ohrling, 2000). Therefore, strengthening evertor muscles at the foot and ankle should be prioritised to reduce the risk of injury in bend sprinters. It is apparent that the left limb is in a complex segmental arrangement which might compromise force production and therefore be responsible for the loss of speed observed on the bend. As Chang and Kram (2007) suggested, it is possible that modifications in the transverse and frontal planes restrict the capacity of muscles to operate and produce force in the sagittal plane. Therefore, future analysis of joint moments during bend sprinting is warranted.

The modifications reported in the present study during the acceleration phase on the bend compared with the straight are not as great as those reported during the maximal speed phase. For example, Churchill et al. (2015) and Alt et al. (2015) reported peak left hip adduction values during bend sprinting of 11° and 14° respectively, compared with the 8° in the present study. This suggests kinematic modifications between the bend and straight become more prominent as velocity increases. However, greater modifications have also been found at smaller radii when running at slower speeds (Chang & Kram, 2007; Luo & Stefanyshyn, 2012). Therefore, it is likely that a combination of radius and velocity are responsible for inducing kinematic modifications. However, the effect of lane allocation has not yet been investigated during the acceleration phase despite performance differences being observed across lanes during the maximal speed phase of bend sprinting (Churchill, Trewartha, & Salo, 2018).

Based on the results of the present study, modifications of the right limb during the acceleration phase of bend sprinting can be characterised by an increase in hip abduction. In addition, the present study did not observe a change in peak right ankle external rotation or internal knee rotation. These two factors were suggested by Alt et al. (2015) to contribute towards a rotational strategy of the right limb. However, findings of this study cannot support this during the acceleration phase. This further highlights the left and right limb have differing functions on the bend, whilst also advancing the notion that the key to understanding the limits of bend sprinting performance lie within the left limb. Hence, limb specific training, with an appreciation of these between-limb differences, should be considered when developing training programmes.

There was a large increase in body lateral lean at touchdown on the bend compared with the straight. These findings, combined with the aforementioned kinematic modifications support the suggestion from Churchill et al. (2015) that the increase in lateral lean angle found on the bend might be responsible for inducing kinematic modifications in the lower limb. However, this was lower than the -10° (left step) and -15° (right step) lean angles reported during maximum speed (Churchill et al., 2015), suggesting these smaller changes accumulate during the acceleration phase, resulting in greater changes at faster speeds. To ensure the transfer of strength training to sports performance, the principle of training specificity is of paramount importance (Young, 2006). Coaches tend to prefer the specificity of training is addressed by adding resistance to sporting movements rather than attempting to make gym exercises more sports specific (Burnie et al., 2017). Therefore, as suggested by Churchill, Trewartha, Bezodis, and Salo (2016) the use of ropes or harnesses in training to provide resistance in a leaning position might be beneficial to performance. In addition, undertaking representative sprint training on the bend to further promote specificity is essential.

Small effect sizes were observed when comparing absolute speed on the bend and the straight for the left (g = 0.52, 2%) step. Whilst the reported effect sizes suggest these reductions are small, a 2% reduction may be meaningful in terms of competitive race performance. For the left step, the 2% reduction found in the present study is the same as reported in previous research into the acceleration phase of bend sprinting (Stoner and Ben-Sira, 1979). Similarly to Churchill et al. (2015) at maximal speed, the reduction in left step velocity on the bend can be attributed to a reduction in left step frequency. Moreover, an increase in left step touchdown distance was apparent on the bend compared with the straight. An increase in touchdown distance has previously been shown to be related to a decrease in ratio of force (Bezodis, Trewartha, & Salo, 2015). A decrease in ratio of force during bend sprinting was also demonstrated by Judson et al. (2019); suggesting reducing touchdown distance may be a key consideration for improving performance on the bend.

Churchill et al. (2015) suggest the reduction in right step absolute speed is due to a shorter right directional step length on the bend compared with the straight. However, no reduction in right step absolute speed was observed in the present study. A small (*g* = 0.35) decrease in right directional step length on the bend compared with the straight was reported, although the difference (0.03 m) does not meet the minimum detectable difference (MDD) of 0.08 m identified by Judson, et al. (2018). Therefore, from the results suggest maintaining a similar right directional step length on the bend and straight aided in avoiding a reduction in right step absolute speed.

A main effect for condition was reported with a reduction in race step length on the bend compared with the straight in both the left (*g* = 2.89) and right steps (*g* = 6.16). The reductions reported here are up to twice as great as those found by Churchill et al. (2015) and Churchill et al. (2016) during the maximal speed phase. The radius in the present study was 36.5 m (lane one), whilst Churchill et al. (2015) and Churchill et al. (2016) examined a 37.72 m radius (lane two), which might have some impact on the results. Furthermore, athletes tend to try and maintain a straight path for as long as possible during the acceleration phase. Qualitative analysis of video data shows athletes were not running straight at the point of data collection. However, doing so in the earlier phases of the race may result in athletes not closely following the race line and consequently, a 3% and 2% decrease in race velocity for the left and right steps, respectively. Therefore, sprinting with the aim of maintaining a straight path during the acceleration phase may not be an effective strategy.

It is acknowledged that the sample size is small, however, the number of participants (*n* = 9) is within the range of those previously reported in bend sprinting research (Alt et al., 2015; Churchill et al., 2015; Churchill et al., 2016; Churchill et al., 2018). In addition, the statistical analysis was appropriate to account for the small sample size, with the use of Hedge's *g*, which includes a correction for smaller samples (Lakens, 2013). In some instances, although not statistically significant, small effect sizes were observed when comparing the left and right limb on the straight. For example, small effect sizes of *g* = 0.26, 0.34 and 0.41 were reported for directional step length, ankle inversion and hip internal rotation, respectively. However, existing sprint literature offers some plausible explanations for these differences. For example, Exell, Irwin, Gittoes, and Kerwin (2017) reported asymmetry in all kinetic and kinematic variables analysed during maximal velocity straight-line sprinting. Furthermore, asymmetrical differences in the strength of invertor and evertor muscle groups of indoor bend sprinters have also been observed (Beukeboom et al., 2000). It is possible the differences observed between the left and right limbs on the straight may be a result of muscular changes due to expertise in the discipline of bend sprinting. Importantly, Excell et al., (2017) concluded that asymmetry was athlete-specific and not necessarily detrimental to performance. Finally, the evaluation of one lane is a limitation, since Churchill et al. (2018) demonstrated differences in kinematic modifications across lanes. However, Judson et al. (2018) demonstrated that the reliability of kinematic variables decreases when sessions take place across two days. Therefore, the collection of data in one session was prioritised, although this consequently constrained the number of trials available due to the risk of fatigue induced injury.

Conclusion

The results of the present study demonstrate that the bend impacts upon kinematic and spatiotemporal parameters of technique and performance during the acceleration phase. These results show that altered kinematics start early in the race and likely accumulate, resulting in greater modifications during the maximal speed phase and thus a greater reduction in speed. Furthermore, the reported kinematic modifications are more prominent in the left limb. A recent study by Ohnuma, Tachi, Kumano, and Hirano (2018) compared technique on the bend and straight in 'good' and 'poor' bend sprinters - where athletes were categorised by their ability to maintain their maximum straight-line speed on the bend (i.e. those with a higher percentage difference in running speed were categorised as poor, and vice versa). It was concluded that better bend sprinters are those who are able to more closely maintain the same sagittal plane kinematics and kinetics as on the straight path. Therefore, coaches should prioritise strategies to address the reported modifications of the left limb, such as reducing touchdown distance and increasing step frequency on the bend. Moreover, an investigation of joint moments may be warranted to understand the mechanisms responsible for these modifications and identify their role in restricting or aiding performance.

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Table 1 Joint kinematics. Group mean ± standard deviation. Significant main effects are marked with *. Significant interactions are marked with $^{\#}$.

Peak angle (°)	**Straight**		**Bend**		**Effect size (*g*) (% difference)**			
	Left	**Right**	**Left**	**Right**	**Left vs right straight**	**Left vs right bend**	**Straight vs bend left**	**Straight vs bend right**
Hip abduction	-6 ± 4	-6 ± 3	- 6 ± 3	-8 ± 3	0.08 (27%)$^{\#}$	0.66 (21%)$^{\#}$	0.22 (20%)$^{\#}$	0.50 (28%)$^{\#}$
Hip adduction	4 ± 4	7 ± 3	8 ± 4	6 ± 4	0.49 (72%)$^{\#}$	0.63 (30%)$^{\#}$	1.09 (106%)$^{\#}$	0.24 (31%)$^{\#}$
Hip internal rotation	2 ± 8	5 ± 4	1 ± 9	7 ± 5	0.41 (104%)	0.75 (126%)	0.15(86%)	0.41 (38%)
Hip external rotation	-9 ± 8	-10 ± 9	-16 ± 7	-8 ± 4	0.13 (11%)	1.22 (89%)	0.89(51%)	0.28 (14%)
Knee abduction	-2 ± 4	-1 ± 7	-2 ± 4	0 ± 4	0.03(75%)	0.39 (23%)	0.03 (0%)	0.21(93%)
Knee adduction	5 ± 5	5 ± 4	5 ± 4	6 ± 6	0.10(26%)	0.15(17%)	0.05(5%)	0.05 (21%)
Knee internal rotation	-1 ± 8	-5 ± 8	-1 ± 6	-5 ± 9	0.37 (70%)	0.52 (80%)	0.02 (18%)	0.07 (11%)
Knee external rotation	-15 ± 7	-13 ± 7	-14 ± 6	-15 ± 8	0.18 (6%)	0.18 (4%)	0.14(5%)	0.22 (6%)
Ankle inversion	14 ± 9	10 ± 9	11 ± 9	12 ± 9	0.34 (48%)$^{\#}$	0.06 (6%)$^{\#}$	0.22 (33%)$^{\#}$	0.19 (16%)$^{\#}$
Ankle eversion	-2 ± 9	-4 ± 10	-5 ± 9	-3 ± 10	0.15 (36%)	0.12 (23%)	0.88 (55%)	0.02 (8%)
Ankle internal rotation	2 ± 4	3 ± 5	12 ± 7	1 ± 7	0.25 (44%)*$^{\#}$	1.95 (562%)*$^{\#}$	1.70 (346%)*$^{\#}$	0.46(108%)*$^{\#}$
Ankle external rotation	-10 ± 5	-10 ± 3	-5 ± 5	-9 ± 5	0.13 (9%)	0.85 (42%)	0.95(50%)*	0.20 (6%)
Body lateral lean at touchdown	6 ± 3	-5 ± 1	- 5 ± 2	-12 ± 2	3.24 (221%)*$^{\#}$	2.98 (62%)*$^{\#}$	4.01 (168%)*$^{\#}$	3.39(78%)*$^{\#}$

Table 2: Spatiotemporal variables. Group mean ± standard deviation . Significant main effects are marked with *. Significant interactions are marked with $^{\#}$.

	Straight		**Bend**		**Effect size (*g*) (% difference)**			
	Left	**Right**	**Left**	**Right**	**Left vs right straight**	**Left vs right bend**	**Straight vs bend left**	**Straight vs bend right**
Absolute speed (m/s)	7.98 ± 0.34	8.00 ± 0.20	7.81 ± 0.30	7.98 ± 0.34	0.12(0%)	0.48(2%)	0.52 (2%)	0.05 (0%)
Race velocity (m/s)	7.98 ± 0.34	8.00 ± 0.20	7.76 ± 0.32	7.86 ± 0.27	0.12 (0%)	0.32 (1%)	0.64(3%)	0.56 (2%)
Contact time (s)	0.107 ± 0.007	0.111 ± .012	0.119 ± 0.007	0.114 ± 0.008	0.34 (4%)$^{\#}$	0.57 (4%)$^{\#}$	1.50 (11%)*$^{\#}$	0.27 (24%)*$^{\#}$
Flight time (s)	0.135 ± 0.019	0.133 ± 0.016	0.135 ± 0.021	0.124 ± 0.016	0.10(2%)	0.53 (9%)	0.01 (0%)	0.51 (7%)
Step Frequency (Hz)	4.33 ± 0.25	4.29 ± 0.22	4.11 ± 0.37	4.30 ± 0.22	0.18 (1%)$^{\#}$	0.61 (5%)$^{\#}$	0.66 (5%)$^{\#}$	0.09 (1%)$^{\#}$
Directional step length (m)	1.84 ± 0.11	1.87 ± 0.08	1.90 ± 0.12	1.84 ± 0.07	0.26 (2%)$^{\#}$	0.55 (3%)$^{\#}$	0.48 (3%)$^{\#}$	0.35 (2%)$^{\#}$
Race step length (m)	1.84 ± 0.11	1.87 ± 0.08	1.53 ± 0.10	1.37 ± 0.08	0.26 (2%)$^{\#}$	1.70 (12%)*$^{\#}$	2.89 (17%)* $^{\#}$	6.16 (37%)* $^{\#}$
Touchdown distance (m)	0.25 ± 0.06	0.26 ± 0.07	0.30 ± 0.05	0.27 ± 0.08	0.12 (4%)$^{\#}$	0.49 (10%)$^{\#}$	0.95 (29%) $^{\#}$	0.11 (4%)$^{\#}$
Turn of CoM (°)			2.48 ± 0.91	2.61 ± 0.86		0.13 (5%)		