

**Limb force and non-sagittal plane joint moments during maximum-effort curve sprint
running in humans**

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Summary

Compared to running straight, when human runners sprint along a curve, the ability of the inside leg to generate force is compromised. This decreased force generation has been suggested to limit the overall performance of the runner. One theory for this force loss is that the large non-sagittal plane joint moments of the inside leg reach their operating limits, thus prohibiting further generation of the performance-related sagittal plane joint moments. We investigated the inside leg force generation and the ankle and knee joint moments when 13 subjects sprinted with and without an additional mass of 12.4 kg along a curve of 2.5 m radius. The increase in the subjects' mass evoked a significant increase in the resultant ground reaction force. The peak non-sagittal plane moments increased significantly for both the ankle and knee joints. This observation suggests that when sprinting normally with maximum effort, the non-sagittal plane joint moments are not operating at their limits. The large increases in ground reaction force were associated with greater extension moments generated at the knee joint. In contrast, the peak ankle plantarflexion moment remained unchanged across conditions. It is possible that for the specific joint configuration experienced, the overall ability to generate plantarflexion moment reaches the limit. Future studies with interventions increasing the muscle tendon units' ability to generate plantarflexion moment may provide an experimental opportunity to further examine this speculation.

Introduction

Animal locomotion is commonly non-linear. The ability to move along curves at high speed, often termed ‘manoeuvrability’, can have a direct influence on survival (Domenici et al., 2011a, 2011b). Theoretical prey-predator scenarios highlighting this point were framed in the literature for different species and environments (Arnott et al., 1999; Domenici, 2001; Howland, 1974; Weihs and Webb, 1984). It was demonstrated that moving rapidly laterally can be an effective escape strategy for the prey, especially in the cases where the prey cannot outrun its predator along a straight path. Despite their importance, our current understanding of the performance constraints for locomotion along curved paths is rather limited (Alexander, 2002; Dickinson et al., 2000; Higham, 2007). Studying human subjects performing maximum-effort curve sprinting in a controlled laboratory environment provides an opportunity for gaining such understanding for legged terrestrial locomotors.

When sprinting along a curve, a ground reaction force is necessary to withstand gravity and create a centripetal acceleration for the body centre of mass. During level-ground locomotion, the magnitude of the vertical ground reaction force, averaged over steps, is equal to $m \cdot g$, where m is the body mass and g is the magnitude of the gravitational acceleration. In addition, for a given tangential traveling speed v along a curve of radius r , a horizontal ground reaction force of magnitude $m \cdot v^2/r$ is required to continuously accelerate the body centre of mass toward the curve centre. It can be seen that as speed increases, the need to supply ground reaction force increases exponentially.

The maximum ground force generated by the supporting limb has been suggested as a key factor in limiting human sprinting performance (Weyand et al., 2000). Weyand et al. (2000) observed that as running speed increases, the ground contact period reduces; to sustain a sufficient amount of vertical ground reaction impulse, the magnitude of the limb supporting force increases. By treating the maximum limb force generation as a constant limit, Greene (1985) formulated a mathematical model to examine the maximum attainable curve sprinting speed, using the maximum linear sprinting speed as input. Based on this theoretical model, in order to sprint along a curve, subjects need to lean towards the curve centre to create a centripetal force. Because Greene (1985) treated the magnitude of the resultant limb force as a constant (the limb force limit), the redirection of the force vector associated with body lean would compromise the generation of the vertical force. To maintain sufficient vertical ground reaction impulse to

support the body weight, subjects would need to extend the ground contact time to compensate for the loss in vertical force generation. This extended ground contact time would then hinder the overall performance. Theoretical predictions based on this limb force limit model are in good agreement with empirical speed data for sprints performed along curves of large radii (Greene, 1985; Usherwood and Wilson, 2006). For curves of smaller size (radius smaller than 10 m), however, performance predictions based on this model are less satisfactory (Alexander, 2002).

An examination of this limb force limit theory by directly assessing the ground reaction force was only conducted recently (Chang and Kram, 2007). In this investigation, ground reaction forces during maximum-effort sprints along circles of various radii (1, 2, 3, 4, and 6 m) were compared to the peak resultant ground force generated during top-speed linear sprinting (which was used to define the limb force limit). The authors hypothesized, if the maximum limb force generation was the limiting factor (Greene, 1985), the peak resultant ground reaction force should remain constant across different curvatures. The results revealed statistically significant reductions in the peak resultant ground reaction force for the inside limb during curve sprints on radii of 1 and 2 m when compared to linear sprints. Meanwhile, a trend of force reduction was observed for other curvatures (Fig. 6 in Chang and Kram, 2007). This trend however was not of statistical significance, which was likely associated with the limited sample size ($n = 4$ or 5 depending on the condition). These findings directly challenged the notion that the maximum limb force limits curve sprinting speed along small circles. Furthermore, the force reduction appeared to be larger for the inside compared to the outside leg. As a chain is only as strong as its weakest link, the authors reasoned, the overall curve sprinting performance is likely limited by the hindered force generation of the inside limb. The authors suggested that despite ample limb force, the generation of this force may be inhibited as other factors reached the operating limits.

Chang and Kram (2007) proposed that the constraints on the inside leg's ability to generate force might be related to the need of joint stabilization. Inverse dynamic analyses have shown that the ankle and knee joints experience large non-sagittal plane joint moments during maximum-effort cutting manoeuvres (Wannop et al., 2010). The non-sagittal plane moment has been associated with unwanted stress on joint soft tissues (Mizuno et al., 2009; Seering et al., 1980). Excessive non-sagittal plane joint moments and angular impulses have been associated with both running and lateral sports injuries (Hewett et al., 2005; Kristianslund et al., 2011;

1 Stefanyshyn et al., 2006). It is possible that during maximum-effort curve sprinting, the non-
2 sagittal plane joint stabilizing moments reach their operating safety threshold and prohibit further
3 extension force generation by the muscle tendon units surrounding the joint. Currently, studies
4 directly testing hypotheses based on this idea are lacking from the literature.

5 One way to examine whether certain variables are operating at the limits is by
6 introducing a perturbation to the system. In the current study, the perturbation to the system was
7 implemented by placing an additional mass near the subjects' centre of mass to alter the
8 mechanical demand for the supporting limb. The purpose of this perturbation was two-fold.

9 First, the implementation of an additional mass allows a further examination of the inside
10 limb force limit theory. While peak inside leg force during curve sprints along small curves were
11 observed to be smaller than the peak force during top-speed linear sprinting (Chang and Kram,
12 2007), that observation itself may not be sufficient in concluding the limb is not generating its
13 maximum force for the specific joint configuration experienced. It is possible that the force
14 generation of the inside limb is at its limit during the movements but the magnitude of this limit
15 is reduced compared to linear sprinting. Such reduction may be due to the differences in lower
16 extremity joint configurations among the movements. Based on the limb force limit theory, a
17 hypothesis can be formed: as the need to support body weight increases with the additional mass,
18 the peak inside limb resultant force will remain constant during curve sprint running.

19 Second, the additional mass implementation allows the investigation of the joint
20 stabilization limit theory. Based on the joint stabilization limit theory, it would be expected that
21 during maximum-effort curve sprinting along small circles, the ankle and/or knee non-sagittal
22 plane moments are at their operating limits. If such limits are the ultimate performance
23 constraints, regardless of the changes in the external force, these joint moments should remain
24 operating at such thresholds for maximizing performance. By changing the mechanical demand
25 of the supporting limb, the additional mass was introduced to potentially alter the ground
26 reaction force and test this hypothesis.

Materials and methods

Subjects

Thirteen male subjects were recruited for the current study (mean \pm 1 s.d.: age 22 ± 2 years, mass 75.4 ± 5.5 kg, height 177.5 ± 5.5 cm). All subjects participated in recreational sports on a regular basis and had no lower extremity injuries in the past year prior to the experiment. Written consent approved by the university ethics committee was obtained from the subjects prior to testing.

Setup and equipments

Counterclockwise sprints were performed maximally along a circle of 2.5 m radius in a laboratory. This radius was chosen based on the observations reported by Chang and Kram (2007) that statistically significant reductions in the limb force generation by the inside leg occurred for circle radii below 3 m. Subjects started the sprint at a location that allowed them to reach top speed when entering the motion capture data collection volume (width 1.5 m, length 3.5 m, height 2.3 m). Kinetic and kinematic data were sampled simultaneously while the subjects entered the collection volume. Ground reaction forces were sampled using an in-ground force platform (Kistler, Winterthur, Switzerland, model Z4852C) at 2400 Hz per channel. Only trials where the subjects planted their inside foot naturally on the force platform were treated as successful. Eight high-speed cameras (Motion Analysis Corporation, Santa Rosa CA, USA, model Eagle) were used to capture the trajectories of the reflective markers at 240 Hz. Twelve markers were placed on each subject to represent the pelvis, left thigh, shank, and foot segment. Knee and ankle joint centres were determined using additional markers during neutral standing trials (anatomical position) prior to the movement trials.

To prevent any confounding effects caused by differences in shoe-ground traction, all subjects used the same pair of athletic shoes (Li Ning Company Limited, Beijing, China, model Yushuai IV) that provided sufficiently high available traction on the laboratory floor. Mechanical traction tests of the current shoe-floor interface were conducted using a previously validated protocol (Luo and Stefanyshyn, 2011). The available traction quantified under this protocol was a traction coefficient of 1.13. It has been shown that beyond a traction coefficient of 0.82, available traction no longer constrains sprinting performance along a curve of 2.3 m radius (Luo

1 and Stefanyshyn, 2011). The footwear outsole was cleaned between trials and the laboratory
2 floor was cleaned between conditions.

3 A commercially available lifejacket (Mountain Equipment Co-op, Vancouver, Canada,
4 model Fulcrum PFD; Fig. 1) was modified in order to securely place an additional mass on the
5 subjects without causing drastic changes in the body moment of inertia. Six pieces of diving lead
6 were glued in the cut-outs made surrounding the bottom region of the jacket near the body centre
7 of mass. The modified jacket contained a total mass of 12.4 kg, providing on average a 16.5%
8 increase in the body mass for the sampled population. This magnitude was chosen to aim for an
9 effective perturbation to the system without causing a drastic change in the locomotion pattern.
10 This was confirmed in a pilot study with one subject, where no changes in joint angle and
11 angular velocity variables due to the additional mass were found. Throughout the study, no
12 subjects reported any discomfort or movement hindrance caused by the apparatus.

13
14 **(Fig. 1)**

15
16 *Protocol*

17 After a 20-min warm-up session, subjects performed maximum-effort curve sprints in the
18 control and additional mass conditions. Subjects started by sprinting in the control condition for
19 four trials, then, for eight trials with the weighted lifejacket securely placed around their torso.
20 After the additional mass condition, subjects performed an additional four trials in the control
21 condition. At least three practice trials were required prior to the collection of each condition in
22 order to minimize variations caused by adaptation. A minimum of a 3-min rest period was given
23 between trials in order to minimize the effects of fatigue. Two-tailed paired t-tests ($\alpha = 0.05$)
24 were used to compare the control condition sprinting speed (quantified as the average pelvis
25 marker speed over stance) before and after the additional mass condition in order to examine if
26 there existed any learning and/or fatigue effects. No differences were detected for any of the 13
27 subjects.

Data analysis

Prior to any analyses, the raw kinetic and kinematic data were filtered with a fourth-order recursive Butterworth low-pass filter. The cut-off frequency was chosen at 60 Hz for the kinetic data and 20 Hz for the kinematic data. The filtered data contained more than 99% of the integrated power content of the original signal.

Peak and average ground reaction forces over stance were determined. The stance interval was determined using the vertical force at a 3% body weight threshold. In addition, the average frontal plane ground reaction force angle was calculated. To ensure that available traction was not a limiting factor in the current study, the maximum traction utilized by the subjects, calculated as the peak ratio of the horizontal over the vertical ground reaction force, was compared to the available traction provided by the shoe-ground interface.

The influence of the additional mass on stance time and curve sprinting speed was examined. The speed measure was defined as the average centre of mass (calculated as the average coordinate of the three pelvis markers) speed over stance.

Joint moments at the inside leg ankle and knee were calculated with a conventional inverse dynamics approach (Andrews, 1995). The centre of pressure location, ground reaction force, and vertical free moment were calculated using force plate data and parameters. The ground reaction force and vertical free moment were applied to the foot segment at the centre of pressure location for the “bottom-up” joint moment calculation. The ankle joint moments were expressed in the foot coordinate system. The foot coordinate system is defined as follows. During a neutral quiet standing trial, where the subjects were in the anatomical position, the subjects’ foot was placed so that its long axis was as closely aligned with the anterior-posterior axis of the lab coordinate system as possible. A foot coordinate system parallel to the lab coordinate system was embedded at the segment origin, the ankle joint centre - defined as the middle point between the markers placed at the medial and lateral malleoli. This coordinate system was then adjusted so that the plantar-/dorsiflexion axis of was aligned with the vector connecting the markers placed at the medial and lateral malleoli. Knee joint moments were expressed in the shank coordinate system. The process of the shank coordinate system construction is similar to that of the foot. The shank segment origin was located at the knee joint

1 centre - defined as the middle point between the markers placed at the medial and lateral
2 epicondyles of the femur. The shank coordinate system was adjusted so that the long (in-
3 /external rotation) axis was aligned to the vector connecting the ankle and knee joint centres. The
4 vector sums of the frontal and transverse plane moments were calculated to represent the demand
5 for joint stabilization; they were denoted as the non-sagittal plane moments. Furthermore, the
6 joint extension moments were quantified and compared between conditions.

7 8 *Statistical analysis*

9 One-tailed paired t-tests were used to compare results between testing conditions. One-
10 tailed analyses were used due to the nature of the intervention, where increases in the ground
11 reaction force and joint moment variables were expected. In addition, findings of no differences
12 are critical for determining the limiting factors; one-tailed tests provide additional power in such
13 cases. Statistical significance level was set *a priori* at $\alpha = 0.05$.

14 15 16 **Results**

17 Mostly due to the changes in the vertical and centripetal components (Fig. 2a & 2b), the
18 peak resultant ground reaction force increased significantly in the additional mass condition
19 compared to the control (Table 1).

20
21 **(Fig. 2)**

22 **(Table 1)**

23
24 In the frontal plane, a larger resultant ground reaction force was observed through the
25 first half of stance in the additional mass condition compared to the control (Fig. 3a). When
26 averaged over stance, the relative increase of the centripetal ground reaction force (9.9%)
27 approximates the relative increase of the vertical ground reaction force (12.3%) – Table 1. As a
28 result, the orientation of the average frontal plane ground reaction force changed by less than 1
29 degree (control: $53.5^{\circ} \pm 2.7^{\circ}$ versus additional mass: $54.3^{\circ} \pm 3.2^{\circ}$ with respect to the ground; $p =$
30 0.0201; Fig. 3b).

1
2 **(Fig. 3)**
3

4 The stance time was 7.6 % longer in the additional mass condition compared to control;
5 associated with the elongated stance was a 2.8 % decrease in speed (Table 1). The peak utilized
6 traction coefficient in both conditions (control: 1.02 and additional mass 0.97) were below the
7 available traction provided (1.13, measured with a mechanical apparatus), confirming that
8 available traction was not hindering the maximal execution of the movement.

9 Compared to curve sprints performed in the control condition, larger peak non-sagittal
10 plane joint moments were observed in the additional mass condition (Fig. 4). At the ankle joint, a
11 19.0 % difference was observed (control: 111.8 ± 30.6 Nm versus additional mass: 132.9 ± 28.5
12 Nm; $p = 0.0012$; Fig. 4a & 4c). At the knee joint, the increase was 19.7 % (control: 98.6 ± 33.4
13 Nm versus additional mass: 118.0 ± 44.2 Nm; $p = 0.0158$; Fig. 4b & 4c).
14

15 **(Fig. 4)**
16

17 In the sagittal plane (Fig. 5), peak ankle joint moment remained unchanged across
18 conditions (control: 205.0 ± 41.0 Nm versus additional mass: 202.0 ± 37.6 Nm; $p = 0.2615$; Fig
19 5a & 5c). At the knee joint, a 15.2 % increase in the peak extension moment was observed in the
20 additional mass condition when compared to the control (control: 159.6 ± 27.5 Nm versus
21 additional mass: 183.9 ± 30.1 Nm; $p = 0.0104$; Fig. 5b & 5c).
22

23 **(Fig. 5)**
24

25 **Discussion**

26 By implementing an external perturbation to the body – an additional mass of 12.4 kg,
27 the study aimed to examine whether: 1) the inside leg reached its limit in generating limb force;
28 2) the non-sagittal plane joint stabilizing moments experienced at the ankle and knee were at
29 their operating limits, during maximum-effort curve sprinting
30

Limb force generation

The maximum limb force generation has been proposed as a factor limiting curve sprinting performance (Greene, 1985; McMahon and Greene, 1979; Usherwood and Wilson, 2006). A recent study (Chang and Kram, 2007) investigating the limb force generation during curve and straight sprinting found that, during sprints along curves of small radii, the peak ground force generated by the inside leg was significantly smaller than during top-speed straight sprints. Since that study, and the studies before (Greene, 1985; Usherwood and Wilson, 2006), treated the peak ground force during straight sprints as the constant limb force limit, the discrepancy observed between the force generation during curve and straight sprints tends to suggest that the inside supporting limb was not at its limit in generating force. An alternative interpretation of this finding, however, can be that the reduced peak ground reaction force during curve sprinting still represents the limb's limit in generating resultant force, but the magnitude of this limit is altered due to factors such as changes in joint configuration among movement tasks. By implementing an additional mass, the current study further investigated the force generation ability of the inside leg while the subjects executed the same task between conditions of different supporting limb mechanical demands. The additional mass condition represents a situation requiring the inside limb to generate greater ground impulses for weight support. If the peak limb force generation in the control condition is truly at the limit, as the need to generate vertical ground reaction impulse increases in the additional mass condition, it would be expected that the peak resultant ground reaction force remains unchanged, the frontal plane ground reaction force would be directed more vertically, and/or, the stance time would be extended.

Results from the current study contradict the notion that the maximum limb force is always generated during maximum-effort curve sprinting. We found that when the subjects performed curve sprints maximally with the additional mass, the peak force generated by the inside leg increased significantly (10.9%) compared to without the mass. While the stance duration was extended from 236.5 ms to 254.5 ms, the average frontal plane ground reaction force over stance was found to be 11.4% larger in the additional mass condition compared to the control. The increase in the frontal plane ground reaction force was a result of the increases found in both the vertical and centripetal components; averaged over stance, the resultant force orientation became only slightly (less than 1°) more vertical in the additional mass condition. These observations supported Chang and Kram's (2007) statement that during maximum-effort

1 curve sprinting, the inside supporting limb seems to possess additional ability to generate force.
2 It remains puzzling why the inside leg would reserve force generation when a critical part of the
3 locomotion task is to maximize centripetal acceleration. Chang and Kram (2007) proposed that
4 this reserved force generation may be associated with the need to stabilize the lower extremity
5 joints.

6 7 *Non-sagittal plane joint stabilizing moment*

8 The second purpose of the study was to examine the idea proposed by Chang and Kram
9 (2007) that during maximum-effort curve sprinting, the non-sagittal plane stabilizing moments at
10 the ankle and knee joints of the inside leg might reach their physiological operating limits, and
11 thus constrain the limb force generation. The observed differences in ground reaction forces
12 between conditions in the current study permitted such examination. If the non-sagittal joint
13 stabilizers were indeed operating at their limits when the subjects sprint normally, with the
14 increase in the external force in the additional mass condition it would be expected that the non-
15 sagittal plane joint moments at the inside leg ankle and/or knee joints would remain at or even
16 under such thresholds. In the presence of a larger ground reaction force, this can potentially be
17 achieved by adapting the segmental kinematics to align the resultant force vector closer to joint
18 centres to reduce the lever arms (Biewener, 1989).

19 Observations made in the current study do not support the notion that during maximum-
20 effort curve sprints, the non-sagittal plane ankle or knee joint moments were at their
21 physiological limits. As the ground reaction force increased from the control to the additional
22 mass condition, the peak non-sagittal plane moment at both the ankle and knee joint increased
23 significantly (19.0% for the ankle joint and 19.7% for the knee). The greater non-sagittal plane
24 moments in the additional mass condition indicated that the joint stabilizers were indeed able to
25 endure external loading larger than the amount experienced when sprinting normally without the
26 mass. If the external force is to be further increased, it is likely that the magnitude of these non-
27 sagittal plane moments would eventually reach a safety threshold and constrain the overall limb
28 force generation. Yet, observations from the current study indicated that during maximum-effort
29 sprints along small curves, the non-sagittal plane stabilizing moments at neither the ankle nor
30 knee joint of the inside leg was the ultimate limiting factor for limb force generation and overall
31 performance. To further probe into the performance constraints for curve sprinting, we took a

1 closer examination of the extension moments generated by the muscle tendon units surrounding
2 the ankle and knee joints, who likely contribute significantly to the overall limb force generation.

4 *Ankle and knee extension moments*

5 While increases in peak non-sagittal plane moments were detected for both the inside
6 ankle and knee joints in the additional mass condition compared to the control, joint extension
7 moments responded in a rather different manner. In the additional mass condition, the peak knee
8 extension moment was 56.8% greater than in the control condition. In contrast, no changes in the
9 peak ankle plantarflexion moment were observed, and this was despite the significant differences
10 in peak and average ground reaction forces. Could it be possible that the plantarflexion moment
11 generation reaches a limit during normal curve sprinting and becomes one of the performance
12 constraints?

13 The idea of joint extension moment generation as a performance constraint has not been
14 directly examined for curve sprinting in the literature. A recent study, however, investigated this
15 idea for straight sprinting (Weyand et al., 2010). Similar to the current study, experimental
16 conditions in that study were implemented to alter the mechanical demand of the supporting
17 limb. Instead of altering the body mass, Weyand et al. (2010) compared the joint moments
18 during straight sprinting and one-legged hopping; the one-legged hopping aimed to increase the
19 limb mechanical demand, by increasing the aerial time for swinging the limb forward and thus
20 the need for vertical ground reaction impulse. The authors reported that the ankle, knee and hip
21 extension moments were all significantly larger during one-legged hopping when compared to
22 sprinting normally. Based on this finding, the authors concluded that the maximum extension
23 moments were not the performance constraints for linear sprinting.

24 A closer examination of the methodology of that study, however, revealed some critical
25 issues that likely weaken the confidence in this conclusion. Mainly, the joint moments were
26 calculated using average values, more specifically average lever arms and ground reaction forces
27 over stance phase, and over steps. These moments were then averaged over a broad range of
28 speed to estimate the extensor muscle force needed for the two movement tasks. Such data
29 treatment likely introduced significant errors to the estimation of the maximum joint moment
30 generation. Furthermore, the force generation of a muscle depends on its length and contracting
31 velocity. In that study, the joint kinematic differences between the two movement tasks may

1 have lead to different muscle operation lengths. In addition, the comparison of average muscle
2 force generation between forward running and one-leg hopping was made over a different range
3 of speed. The dependence of muscle force generation on contracting velocity may thus confound
4 the interpretations.

5 In the current study, instantaneous joint moments were calculated, and the peak values
6 observed during the maximum-effort executions of the same movement task were compared. In
7 order to explore whether the extensor muscles were operating at similar length and velocity, a
8 post-hoc analysis of the ankle and knee kinematics was conducted. While direct measurements of
9 muscle length and contracting velocity are not available, joint kinematics may help provide
10 insight into the joint extensors' operating conditions. In the analysis, joint angle was used as an
11 estimate for muscle length, and joint velocity was used to approximate the contracting velocity.
12 One-tailed paired t-tests were used and the significance level was set *a priori* at $\alpha = 0.05$.

13 Joint angle and velocity at the instances when peak moments occurred were compared
14 between conditions. No difference was detected for either the ankle ($p = 0.1967$) or knee ($p =$
15 0.3166) joint angles (Fig. 6). Furthermore, no difference in joint velocity was observed for either
16 joint (ankle: $p = 0.0656$; knee: $p = 0.1226$; Fig. 7). The consistent operating kinematics provide
17 us with confidence in interpreting the operating state of the joint extensors.

18
19 (Fig. 6)

20 (Fig. 7)

21
22 In presence of the large change in external force, if the moment generation at a certain
23 joint is at the limit, it would be expected to remain constant across conditions. Our finding that
24 the ankle plantarflexion moment remained unchanged across conditions suggested that it is
25 possible that during maximum-effort curve sprinting, the ability to generate ankle plantarflexion
26 moment of the inside leg reaches its limit for the given operating states. This speculation seems
27 to be supported by another observation made for linear sprinting. In a study by Kuitunen et al.
28 (2002), the authors investigated the ankle and knee joint stiffness while the subjects sprinted at
29 70%, 80%, 90% and 100% of their top speeds. They found that, despite the larger ground
30 reaction force experienced at higher speeds, peak plantarflexion moment remained constant.

One reason why the ankle plantarflexors would operate at their limits may lie in their critical role in generating ground force. The contribution of net moments generated at a specific joint to the resultant ground force generated during linear sprinting has been investigated previously with an induced acceleration analysis (Dorn et al., 2012). It was found that during mid- and late stance (when peak ground reaction force took place) soleus and gastrocnemius were the primary contributors to the generation of ground force. If the ankle plantarflexors were also the main contributors to the ground force generation in curve sprinting, it would be logical to fully utilize their performance potential when the movement task is to sprint at maximum-effort. Future research to identify the ankle joint moment contribution to overall ground force generation during curve sprinting is, however, needed to further elaborate on this speculation.

If the ability to generate plantarflexion moment is among the predominant factors limiting the top curve sprinting performance, by changing such moment generation, performance changes should be observed. One plausible method to increase such moment generation may be by aligning the ankle joint toward a more optimal configuration for pushing off the ground (e.g., less ev-/inverted). In an early investigation of curved sprinting performance (Greene, 1987), subjects sprinted on surfaces with various frontal-plane bank angles (10 – 30°). Improvements in sprinting speed, as large as 10%, were observed in the banked track condition compared to flat. Unfortunately joint kinematics and kinetics were not assessed in that study. It is possible that the banked surface realigned the ankle joint and allowed a greater plantarflexion moment generation. Future investigations of joint kinematics and kinetics during sprints performed on flat versus banked surfaces may help further reveal the relationship between plantarflexion moment generation and curved sprinting performance.

It remains puzzling why the knee joint extension moment was not at the maximum level when the sprints were performed without the additional mass. One potential explanation may be that increases in the knee extension moment can result in an increase in both the centripetal and vertical ground force and impulse (as observed in the current study); while the increase in the centripetal ground force is favorable from a performance perspective, excessive vertical ground impulse may be counter-productive. Unlike their ankle counterpart, in the body frontal plane, the knee extensors' function can be seen as a linear actuator along the long axis of the leg which has both a horizontal and vertical component. While the increase in the centripetal ground force is needed for a higher curve sprinting speed, additional vertical ground impulse can result in an

extended flight and thus a longer step length, which may hinder performance. Firstly, since the task requires the subjects to complete a circle without passing inside it, by increasing the step length, the total distance traveled would increase as the step length increases. As the step length decreases, approximating zero, the total travel approaches the true perimeter of the circle. Secondly, by reducing the number of steps used to complete a circle (increasing step length), the redirection of the centre of mass travel becomes more acute, and a more acute centre of mass redirection has been associated with a greater loss of momentum (Bertram and Gutmann, 2009). Future investigations of the optimal step length/frequency for maximizing curve sprinting speed are needed to further evaluate the aforementioned speculation.

Limitations and future works

While the current study followed up closely on the previous work by Chang and Kram (2007) to examine the force generation and joint moments of the inside limb, it is limited in concluding the performance constraints for the whole system as the outside limb was not investigated. In Chang and Kram (2007), an asymmetric force reduction was observed between the limbs with the inside limb suffering a greater and significant force loss when sprinting along curves compared to straight lines. This observation made a strong case for the authors to propose that the inside limb may be the weakest link of the chain and thus constrains the overall performance. While the current study directly examined the hypotheses formulated based on the idea from Chang and Kram (2007), it is limited in that the outside limb's response to the experimental intervention remains unknown. Future work is needed to examine the outside limb force generation and joint moments as the supporting limb mechanical demand is altered, to provide additional information for the further exploration of the predominant limiting factors for curve sprinting performance.

The observation that the ankle plantarflexion moment remained unchanged between conditions lead to our speculation that such moment generation may be at its limit during maximum-effort curve sprinting. This speculation is limited in that the joint moment calculated through an inverse dynamics approach only represents the net effect of all the tissues surrounding a joint that satisfies the equations of motion. Based on the current methodology, we cannot conclude the moment generation capacity surrounding a given joint. In the overdetermined musculoskeletal system, the interpretations of the ankle plantarflexors' operating

1 state can be confounded by, for example, the co-contraction of the agonist and antagonist muscle
2 pairs and force sharing between uniarticular and biarticular muscles. Future work is needed to
3 directly assess the subjects' maximum ability to generate plantarflexion moment with a protocol
4 replicating the actual movement (e.g. Hahn et al., 2011 for leg extension). Understanding gained
5 from such experiments will provide strong evidence for or against the idea that the ability to
6 generate plantarflexion moment may be a constraint for curve sprinting performance.

8 **Conclusions**

9 The original design of this study aimed to examine the 1) limb force limit and 2) non-
10 sagittal plane ankle and knee joint moment limit as performance constraints for curve sprinting.
11 The current observations did not support the hypotheses formed based on either theory. During
12 maximum-effort curve sprinting: the inside limb reserves additional force generation ability; the
13 inside ankle and knee joint non-sagittal plane stabilizing moments were not at their operating
14 limits. It was found that the plantarflexion moment generated at the ankle joint remained
15 constant despite significant increases in the ground reaction force. We proposed that during
16 maximum-effort curve sprinting, the plantarflexors may operate at their limits.

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21 Futures and Li Ning Company Limited.

23 **References**

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Figure Legends

Fig. 1. Photograph of the modified life jacket (total mass = 12.4 kg; 16.5% body mass for sampled population) worn by a subject. Six pieces of diving lead were placed surrounding the bottom region of the jacket near the body centre of mass.

Fig. 2. Ground reaction forces in the (a) vertical, (b) centripetal and (c) anterior-posterior directions. Sample means (thick lines) and standard deviations (thin lines) were plotted against the normalized stance time. When the subjects sprinted with an additional mass, significant increases in the vertical and centripetal components of the ground reaction force were observed.

Fig. 3. (a) Resultant frontal-plane ground reaction force generated during stance. It was calculated as the vector sum of the centripetal and vertical ground reaction force. The thick lines represent the average values across all the subjects and the thin lines indicate ± 1 s.d.. (b) Average frontal-plane ground reaction forces over stance. Both the vertical and centripetal stance-average ground reaction forces increased in the additional mass condition compared to control. Consequently, only a slight change (less than 1°) in the ground reaction force orientation was observed.

Fig. 4. The non-sagittal plane joint moments experienced at the inside limb (a) ankle and (b) knee joints normalized over stance. The non-sagittal plane moment was calculated as the vector sum of the frontal and transverse plane joint moments; it was used to represent the demand for joint stabilization. The thick lines represent the average value across all the subjects and the thin lines indicate ± 1 s.d.. (c) Peak non-sagittal plane moments were great in the additional mass condition for both the ankle and knee joints. *Statistically significant difference.

Fig. 5. The joint extension moments generated at the inside limb (a) ankle and (b) knee joints normalized over stance. The thick lines represent the average value across all the subjects and the thin lines indicate ± 1 s.d.. (c) Peak extension moment increased in the additional mass condition

for the knee joint. For the ankle joint, the peak plantarflexion moment remained constant between conditions. *Statistically significant difference.

Fig. 6. Inside limb (a) ankle and (b) knee angles during stance. For the instants where the maximum joint moments were generated, no difference was observed between conditions for either the ankle or knee joint angle. The thick lines represent the average value across all the subjects and the thin lines indicate ± 1 s.d..

Fig. 7. Inside limb (a) ankle and (b) knee angular velocities during stance. For the instants where the maximum joint moments were generated, no difference was observed between conditions for either the ankle or knee joint velocity. The thick lines represent the average value across all the subjects and the thin lines indicate ± 1 s.d..

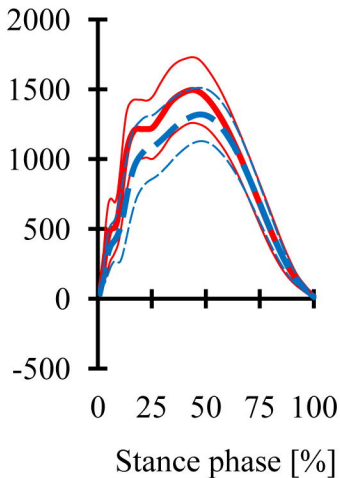
Table Legends

Table 1. Peak and stance-average ground reaction forces, stance time, and curve sprinting speed. The additional mass increased the ground reaction forces, elongated the stance time and reduced the maximum curve sprinting speed.

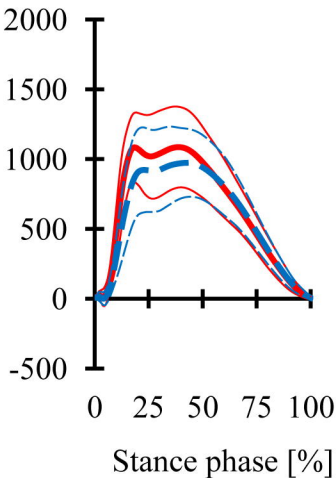


(a)

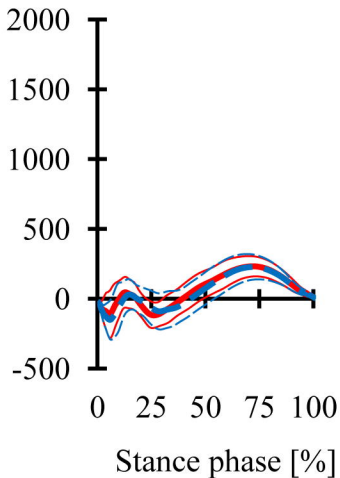
Vertical
ground reaction force [N]

**(b)**

Centripetal
ground reaction force [N]

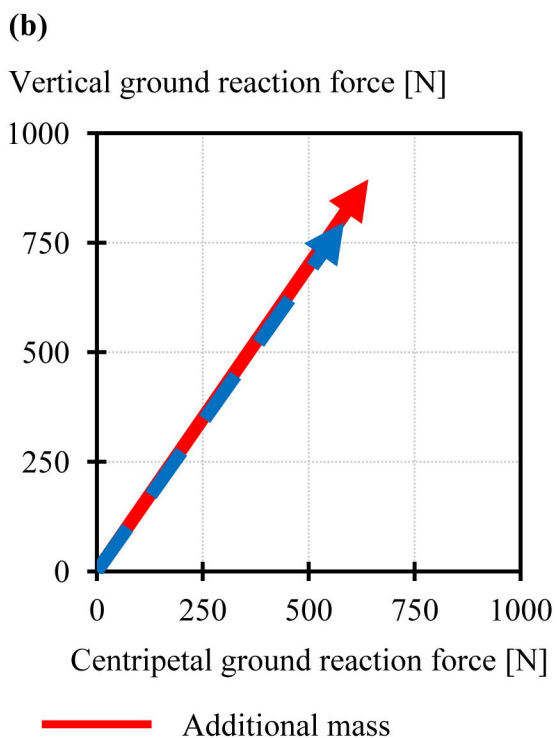
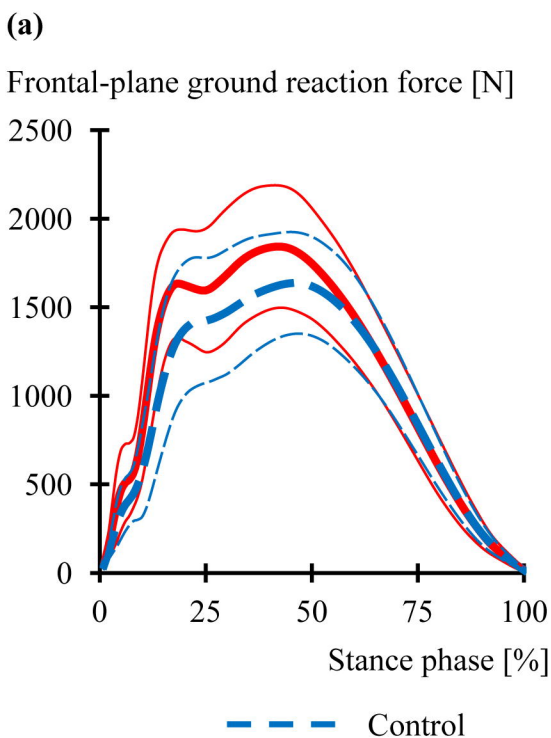
**(c)**

Anterior-posterior
ground reaction force [N]



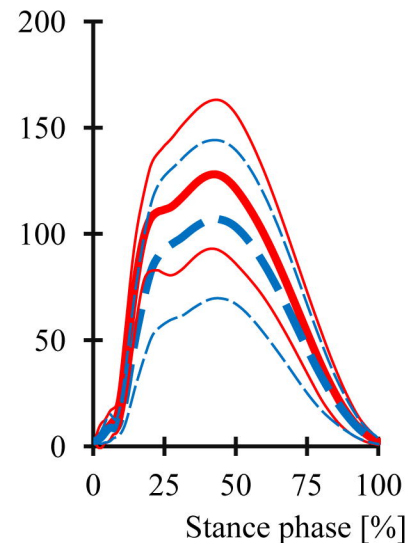
— — — Control

— Additional mass



(a)

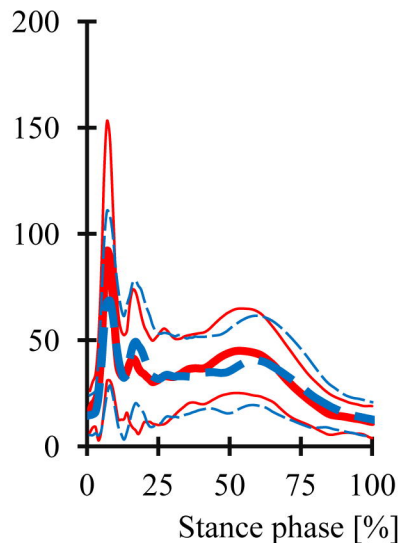
Ankle non-sagittal plane
moment [Nm]



— — Control

(b)

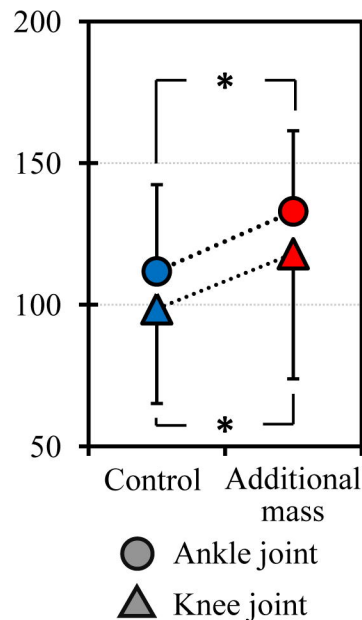
Knee non-sagittal plane
moment [Nm]

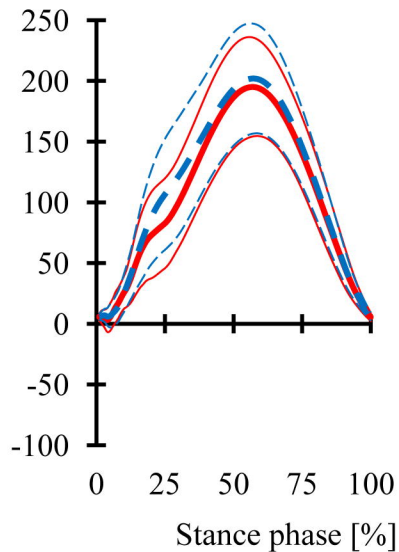


— Additional mass

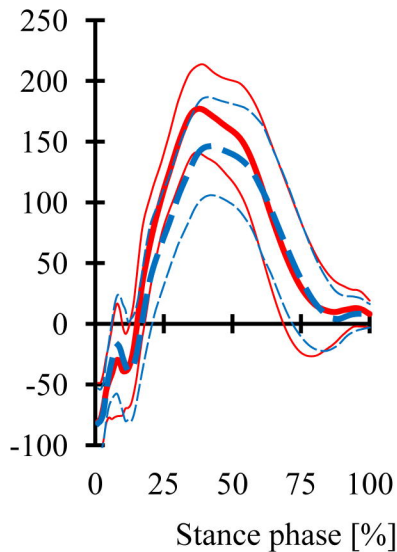
(c)

Peak non-sagittal plane
moment [Nm]

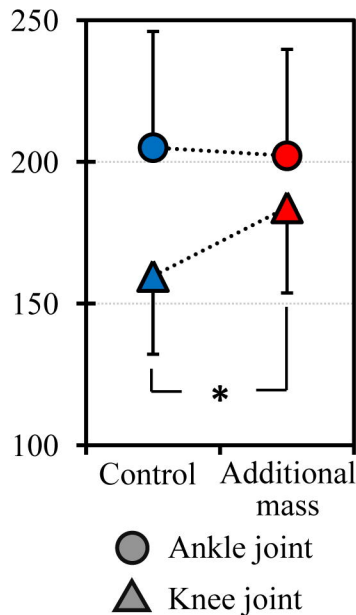


(a)Ankle plantarflexion
moment [Nm]

--- Control

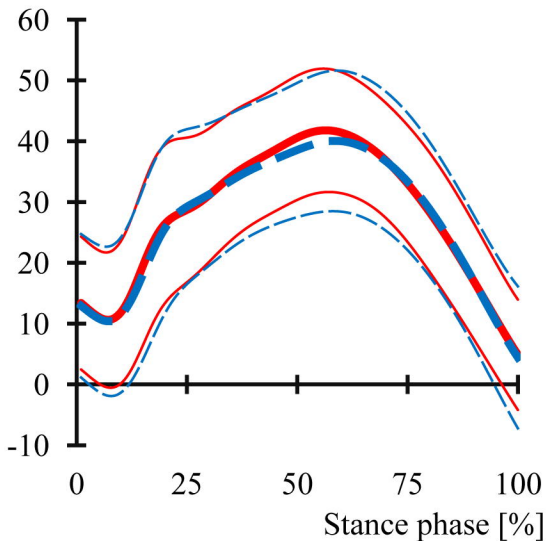
(b)Knee extension
moment [Nm]

— Additional mass

(c)Peak joint extension
moment [Nm]

(a)

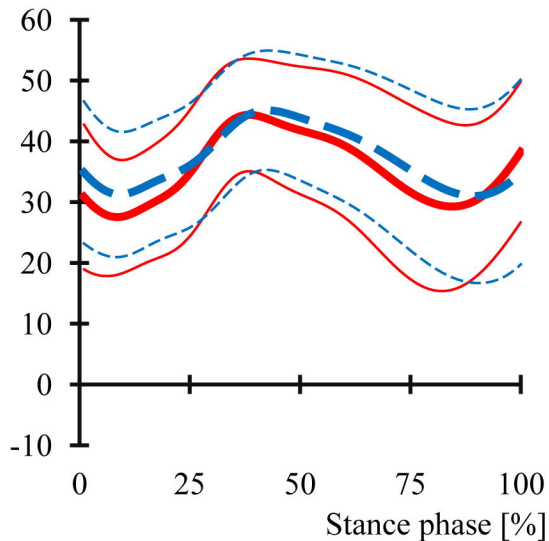
Ankle dorsiflexion
angle [$^{\circ}$]



Control

(b)

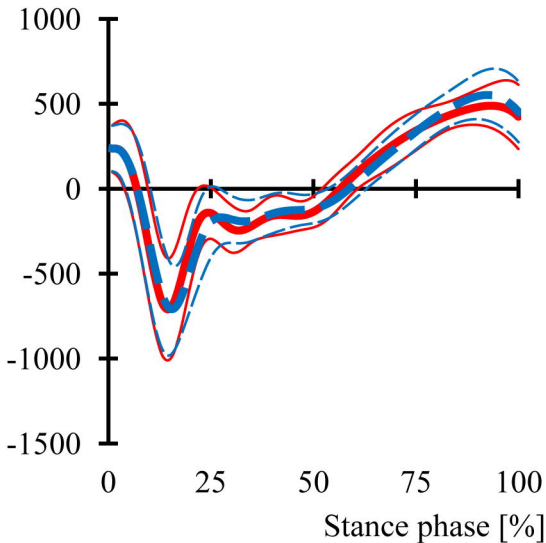
Knee flexion
angle [$^{\circ}$]



Additional mass

(a)

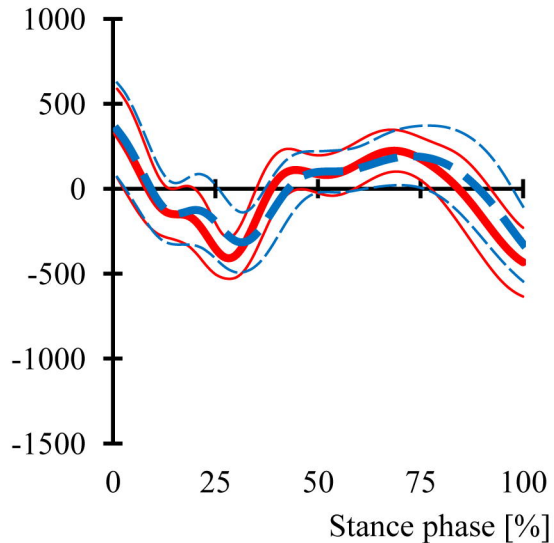
Ankle plantarflexion
velocity [$^{\circ} \text{s}^{-1}$]



— — — Control

(b)

Knee extension
velocity [$^{\circ} \text{s}^{-1}$]



— Additional mass

	Control	Additional Mass	One-tailed paired t-test
Peak GRF_{resultant} [N]	1730.1 ± 273.1	1919.1 ± 292.0	p = 0.0002
Peak GRF_{cpt} [N]	1091.0 ± 240.4	1215.6 ± 218.9	p = 0.0007
Peak GRF_{vert} [N]	1369.6 ± 169.4	1528.0 ± 207.5	p = 0.0003
Stance-average GRF_{cpt} [N]	583.0 ± 120.3	640.4 ± 129.6	p = 0.0003
Stance-average GRF_{vert} [N]	797.0 ± 96.5	894.8 ± 104.3	p < 0.0001
Stance time [ms]	236.5 ± 11.3	254.5 ± 14.7	p < 0.0001
Speed [m s⁻¹]	3.6 ± 0.2	3.5 ± 0.2	p = 0.0302

GRF_{resultant}: 3D resultant ground reaction force **GRF_{cpt}**: centripetal ground reaction force
GRF_{vert}: vertical ground reaction force