

RESEARCH ARTICLE

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

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SUMMARY

Compared with 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 ability of the muscle-tendon units to generate plantarflexion moment may provide an experimental opportunity to further examine this speculation.

Key words: curve sprinting, curve running, human performance, performance constraint, additional mass.

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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; Domenici et al., 2011b). Theoretical prey-predator scenarios highlighting this point have been described in the literature for different species and environments (Howland, 1974; Weihs and Webb, 1984; Arnott et al., 1999; Domenici, 2001). It was demonstrated that moving rapidly laterally can be an effective escape strategy for the prey, especially in 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 (Dickinson et al., 2000; Alexander, 2002; 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 center of mass. During level-ground locomotion, the magnitude of the vertical ground reaction force, averaged over steps, is equal to mg , 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 mv^2/r is required to continuously accelerate the body center of mass towards the curve center. 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). These authors 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 (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 center to create a centripetal force. Because Greene 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 with the peak resultant ground force generated during top-speed linear sprinting (which was used to define the limb force limit). The authors hypothesized that 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 with linear sprints. Meanwhile, a trend of force reduction was observed for other curvatures [fig. 6 in Chang and Kram (Chang and Kram, 2007)]. This trend, however, was not of statistical significance, which was probably 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 with the outside leg. As a chain is only as strong as its weakest link, the authors reasoned that the overall curve sprinting performance is probably 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 (Chang and Kram, 2007) proposed that the constraints on the ability of the inside leg 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 maneuvers (Wannop et al., 2010). The non-sagittal plane moment has been associated with unwanted stress on joint soft tissues (Seering et al., 1980; Mizuno et al., 2009). Excessive non-sagittal plane joint moments and angular impulses have been associated with both running and lateral sports injuries (Hewett et al., 2005; Stefanyshyn et al., 2006; Kristianslund et al., 2011). It is possible that during maximum-effort curve sprinting, the non-sagittal plane joint stabilizing moments reach their operating safety threshold and prohibit further extension force generation by the muscle-tendon units surrounding the joint. Currently, studies directly testing hypotheses based on this idea are lacking from the literature.

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

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

Secondly, the additional mass implementation allows the investigation of the joint stabilization limit theory. Based on the joint stabilization limit theory, it would be expected that during

maximum-effort curve sprinting along small circles, the ankle and/or knee non-sagittal plane moments are at their operating limits. If such limits are the ultimate performance constraints, regardless of the changes in the external force, these joint moments should remain operating at such thresholds for maximizing performance. By changing the mechanical demand of the supporting limb, the additional mass was introduced to potentially alter the ground reaction force and test this hypothesis.

MATERIALS AND METHODS

Subjects

Thirteen male subjects were recruited for the current study (mean \pm 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. The study was approved by the university ethics committee, and written informed consent was obtained from the subjects prior to testing.

Set-up 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 previously (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 (model Z4852C, Kistler, Winterthur, Switzerland) 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 (model Eagle, Motion Analysis Corporation, Santa Rosa, CA, USA) 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 centers 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 (model Yushuai IV, Li Ning Company, Beijing, China) 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 and Stefanyshyn, 2011). The footwear outsole was cleaned between trials and the laboratory floor was cleaned between conditions.

A commercially available lifejacket (model Fulcrum PFD, Mountain Equipment Co-op, Vancouver, Canada; Fig. 1) was modified in order to securely place an additional mass on the subjects without causing drastic changes in the body moment of inertia. Six pieces of diving lead were glued in cut-outs made in the bottom region of the jacket near the body center of mass. The modified jacket contained a total mass of 12.4 kg, providing, on average, a 16.5% increase in the body mass for the sampled population. This magnitude was chosen to aim for an effective perturbation to the system without causing a drastic change in the locomotion pattern. This was confirmed

in a pilot study with one subject, where no changes in joint angle and angular velocity variables due to the additional mass were found. Throughout the study, no subjects reported any discomfort or movement hindrance caused by the apparatus.

Protocol

After a 20 min warm-up session, subjects performed maximum-effort curve sprints in the control and additional mass conditions. Subjects started by sprinting in the control condition for four trials, then for eight trials with the weighted lifejacket securely placed around their torso. After the additional mass condition, subjects performed an additional four trials in the control condition. At least three practice trials were required prior to the collection of each condition in order to minimize variations caused by adaptation. A minimum of a 3 min rest period was given between trials in order to minimize the effects of fatigue. Two-tailed paired *t*-tests ($\alpha=0.05$) were used to compare the control condition sprinting speed (quantified as the average pelvis marker speed over stance) before and after the additional mass condition in order to examine if there existed any learning and/or fatigue effects. No differences were detected for any of the 13 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 with 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 center 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 center 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 center 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 laboratory coordinate system as possible. A foot coordinate system parallel to the laboratory coordinate system was embedded at the segment origin, the ankle joint center – defined as the middle point between the markers placed at the medial and lateral malleoli. This coordinate system was then adjusted so that the plantarflexion/dorsiflexion axis 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



Fig. 1. Photograph of the modified lifejacket (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 center of mass.

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

Statistical analysis

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

RESULTS

Mostly due to the changes in the vertical and centripetal components (Fig. 2A,B), the peak resultant ground reaction force increased significantly in the additional mass condition compared with the control (Table 1).

In the frontal plane, a larger resultant ground reaction force was observed through the first half of stance in the additional mass condition compared with the control (Fig. 3A). When averaged over stance, the relative increase of the centripetal ground reaction force (9.9%) approximates the relative increase of the vertical ground reaction force (12.3%) – see Table 1. As a result, the orientation of

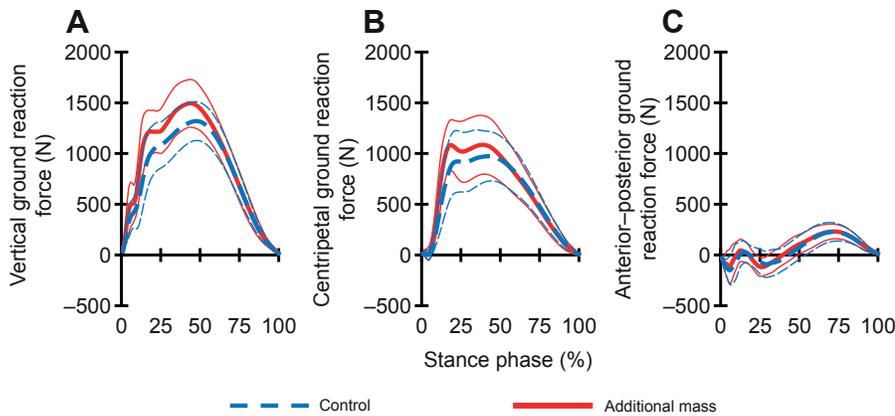


Fig. 2. Ground reaction forces in the (A) vertical, (B) centripetal and (C) anterior–posterior directions. Sample means (thick lines) and standard deviations (thin lines) were plotted against the normalized stance time. When the subjects sprinted with an additional mass, significant increases in the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

the average frontal plane ground reaction force changed by less than 1 deg (control: 53.5 ± 2.7 deg *versus* additional mass: 54.3 ± 3.2 deg with respect to the ground; $P=0.0201$; Fig. 3B).

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

Compared with curve sprints performed in the control condition, larger peak non-sagittal plane joint moments were observed in the additional mass condition (Fig. 4). At the ankle joint, a 19.0% difference was observed (control: 111.8 ± 30.6 Nm *versus* additional mass: 132.9 ± 28.5 Nm; $P=0.0012$; Fig. 4A,C). At the knee joint, the increase was 19.7% (control: 98.6 ± 33.4 Nm *versus* additional mass: 118.0 ± 44.2 Nm; $P=0.0158$; Fig. 4B,C).

In the sagittal plane (Fig. 5), peak ankle joint moment remained unchanged across conditions (control: 205.0 ± 41.0 Nm *versus* additional mass: 202.0 ± 37.6 Nm; $P=0.2615$; Fig. 5A,C). At the knee joint, a 15.2% increase in the peak extension moment was observed in the additional mass condition when compared with the control (control: 159.6 ± 27.5 Nm *versus* additional mass: 183.9 ± 30.1 Nm; $P=0.0104$; Fig. 5B,C).

DISCUSSION

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

Limb force generation

The maximum limb force generation has been proposed as a factor limiting curve sprinting performance (McMahon and Greene, 1979; Greene, 1985; 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. As 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 limit of the limb 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 with without the mass. While the stance duration was extended from 236.5 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 with 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 deg) more vertical in the additional mass condition. These observations supported Chang and Kram's statement (Chang and Kram, 2007) that during maximum-effort curve sprinting, the inside supporting limb seems to possess additional ability to generate force. It remains puzzling why the inside leg would reserve force generation when a critical part of the locomotion task is to maximize centripetal acceleration. Chang and Kram proposed that this reserved force generation may be associated with the need to stabilize the lower extremity joints.

Non-sagittal plane joint stabilizing moment

The second purpose of the study was to examine the idea that during maximum-effort curve sprinting, the non-sagittal plane stabilizing moments at the ankle and knee joints of the inside leg might reach their physiological operating limits, and thus constrain the limb force generation (Chang and Kram, 2007). The observed differences in ground reaction forces between conditions in the current study

Table 1. Peak and stance-average ground reaction forces, stance time and curve sprinting speed

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

GRF_{resultant}, 3-D resultant ground reaction force; GRF_{cpt}, centripetal ground reaction force; GRF_{vert}, vertical ground reaction force.

The additional mass increased the ground reaction forces, elongated the stance time and reduced the maximum curve sprinting speed.

permitted such examination. If the non-sagittal joint stabilizers were indeed operating at their limits when the subjects sprinted normally, with the increase in the external force in the additional mass condition it would be expected that the non-sagittal plane joint moments at the inside leg ankle and/or knee joints would remain at or even under such thresholds. In the presence of a larger ground reaction force, this can potentially be achieved by adapting the segmental kinematics to align the resultant force vector closer to joint centers to reduce the lever arms (Biewener, 1989).

Observations made in the present study do not support the notion that during maximum-effort curve sprints, the non-sagittal plane ankle or knee joint moments were at their physiological limits. As the ground reaction force increased from the control to the additional mass condition, the peak non-sagittal plane moment at both the ankle and knee joint increased significantly (19.0% for the ankle joint and 19.7% for the knee). The greater non-sagittal plane moments in the additional mass condition indicated that the joint stabilizers were indeed able to endure external loading larger than the amount experienced when sprinting normally without the mass. If the external force is to be further increased, it is likely that the magnitude of these non-sagittal plane moments would eventually reach a safety threshold and constrain the overall limb force generation. Yet, observations from the current study indicated that during maximum-effort sprints along small curves, the non-sagittal plane stabilizing moments at neither the ankle nor knee joint of the inside leg was the ultimate limiting factor for limb force generation and overall performance. To further investigate the performance constraints for curve sprinting, we made a closer examination of the extension moments generated by the muscle–tendon units surrounding the ankle and knee joints, which probably contribute significantly to the overall limb force generation.

Ankle and knee extension moments

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

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

A closer examination of the methodology of that study, however, revealed some critical issues that probably weakened the confidence in this conclusion. Mainly, the joint moments were calculated using

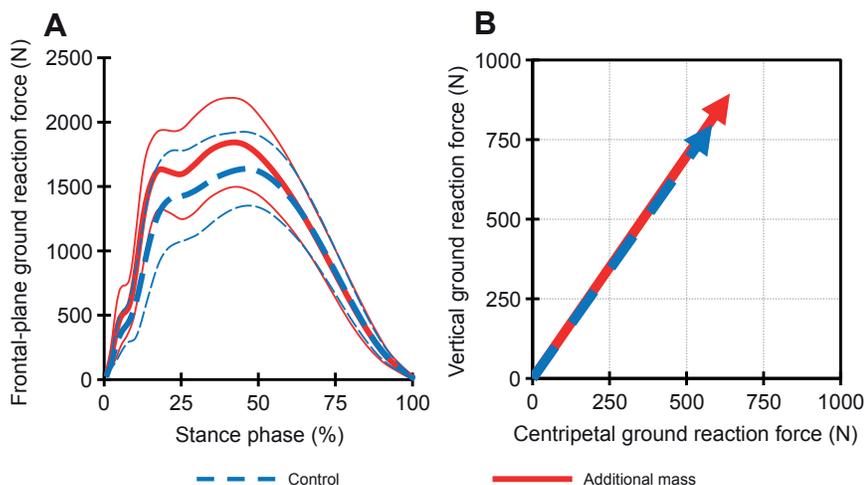


Fig. 3. (A) Resultant frontal-plane ground reaction force generated during stance. It was calculated as the vector sum of the centripetal and vertical ground reaction force. The thick lines represent the average values across all the subjects and the thin lines indicate ± 1 s.d. (B) Average frontal-plane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

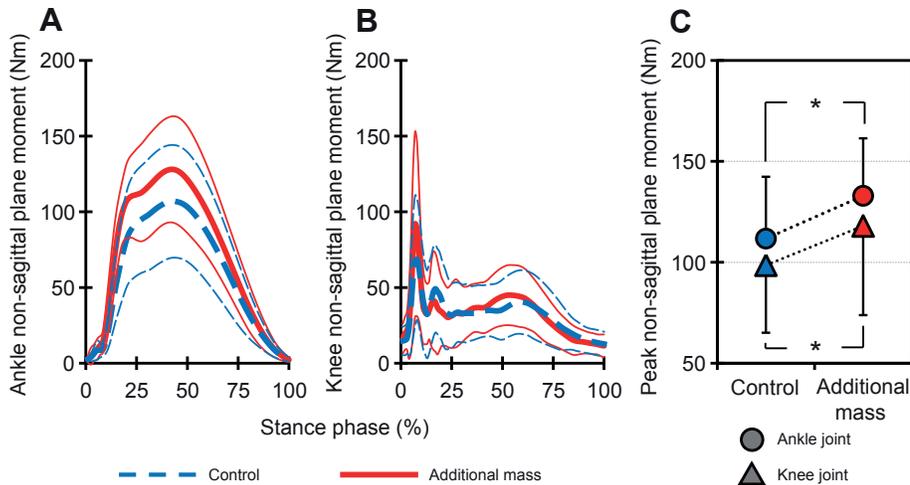


Fig. 4. Non-sagittal plane joint moments experienced at the inside limb (A) ankle and (B) knee joints normalized over stance. The non-sagittal plane moment was calculated as the vector sum of the frontal and transverse plane joint moments; it was used to represent the demand for joint stabilization. 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.

average values, more specifically average lever arms and ground reaction forces over stance phase, and over steps. These moments were then averaged over a broad range of speeds to estimate the extensor muscle force needed for the two movement tasks. Such data treatment probably introduced significant errors to the estimation of the maximum joint moment generation. Furthermore, the force generation of a muscle depends on its length and contracting velocity. In that study, the joint kinematic differences between the two movement tasks may have lead to different muscle operation lengths. In addition, the comparison of average muscle force generation between forward running and one-legged hopping was made over a different range of speeds. The dependence of muscle force generation on contracting velocity may thus confound the interpretations.

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

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

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

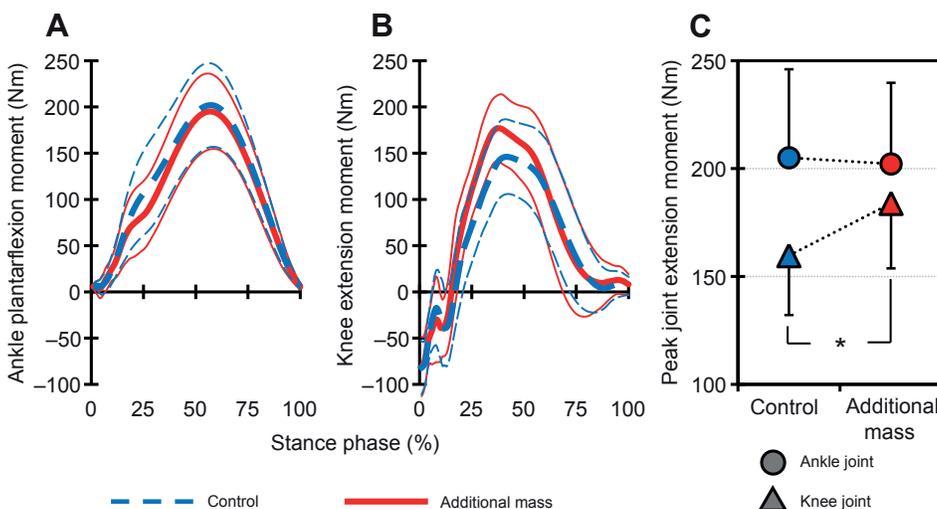


Fig. 5. Joint extension moments generated at the inside limb (A) ankle and (B) knee joints normalized over stance. The thick lines represent the average value across all the subjects and the thin lines indicate ± 1 s.d. (C) Peak extension moment increased in the additional mass condition for the knee joint. For the ankle joint, the peak plantarflexion moment remained constant between conditions. *Statistically significant difference.

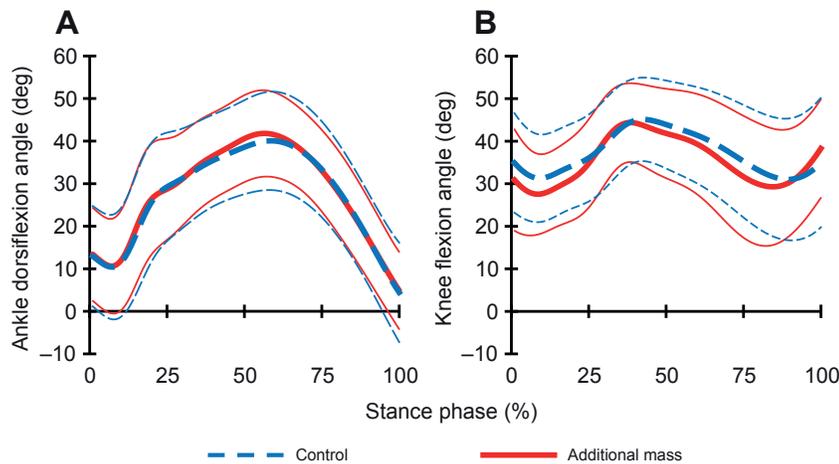


Fig. 6. Inside limb (A) ankle and (B) knee angles during stance. For the instants where the maximum joint moments were generated, no difference was observed between conditions for either the ankle or knee joint angle. The thick lines represent the average value across all the subjects and the thin lines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

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 towards a more optimal configuration for pushing off the ground (e.g. less everted/inverted). In an early investigation of curved sprinting performance (Greene, 1987), subjects sprinted on surfaces with various frontal-plane bank angles (10–30 deg). Improvements in sprinting speed, as large as 10%, were observed in the banked track condition compared with 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 counterproductive. Unlike their ankle counterpart, in the body frontal plane, the function of the knee extensors can be seen as a linear actuator along the long axis of the leg that 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 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 center of mass travel becomes more acute, and a more acute center of mass redirection has been associated with a greater loss of momentum (Bertram and Gutmann, 2009). Future investigations

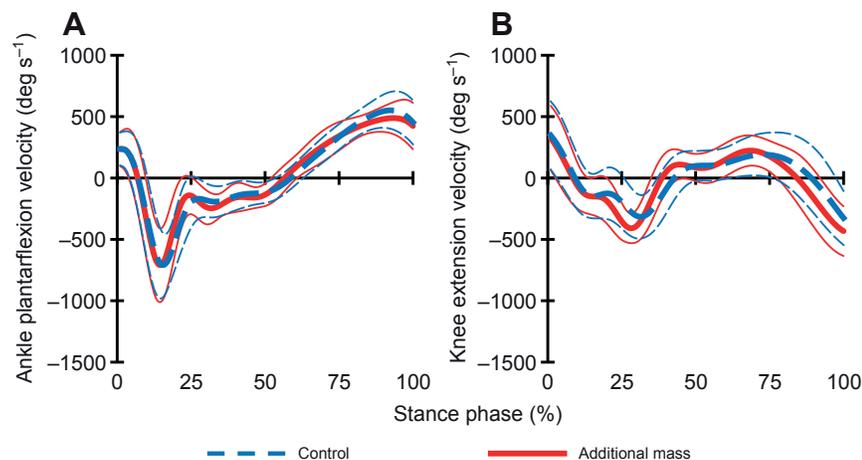


Fig. 7. Inside limb (A) ankle and (B) knee angular velocities during stance. For the instants where the maximum joint moments were generated, no difference was observed between conditions for either the ankle or knee joint velocity. The thick lines represent the average value across all the subjects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

of the optimal step length/frequency for maximizing curve sprinting speed are needed to further evaluate these speculations.

Limitations and future work

While the current study followed up closely on the previous work by Chang and Kram (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 the study of Chang and Kram (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 with 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 ideas formulated based on this logic, it is limited in that the response of the outside limb 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 leads 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 operating state of the ankle plantarflexors can be confounded by, for example, the co-contraction of the agonist and antagonist muscle pairs and force sharing between uniaxial and biaxial muscles. Future work is needed to directly assess the subjects' maximum ability to generate plantarflexion moment with a protocol replicating the actual movement [e.g. for leg extension, see Hahn et al. (Hahn et al., 2011)]. Understanding gained from such experiments will provide strong evidence for or against the idea that the ability to generate plantarflexion moment may be a constraint for curve sprinting performance.

Conclusions

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

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