Limiting Factors for Curved Sprinting Performance

by

Geng Luo

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The undersigned certify that they have read, and recommend to the Faculty of Graduate Studies for acceptance, a thesis entitled "Limiting Factors for Curved Sprinting Performance" submitted by Geng Luo in partial fulfilment of the requirements of the degree of Doctor of Philosophy.

__________________________
Supervisor, Dr. Darren Stefanyshyn, Faculty of Kinesiology

__________________________
Dr. John Bertram, Department of Medical Science

__________________________
Dr. Walter Herzog, Faculty of Kinesiology, Departments of Mechanical Engineering, Medical Science and Biomedical Engineering

__________________________
Dr. Campbell Rolian, Department of Veterinary Medicine

__________________________
Dr. Rodger Kram, Department of Integrative Physiology
University of Colorado

__________________________
Date
Abstract

Sprinting along a curved path is regularly performed in athletics. Yet, the locomotion mechanisms behind curved sprinting performance are relatively unexplored. The current dissertation aimed to explore the limiting factors for curved sprinting performance from a biomechanical perspective.

It was discovered that the available traction between footwear and ground can limit the maximum-effort curved sprinting performance, but only to a certain extent. As available traction was systematically increased from a traction coefficient of 0.26 to 0.82, the athletes leaned more into the ground, generated a greater impulse against the ground and achieved a higher performance. Further increases in the available traction could not be utilized by the athletes for additional performance benefits.

With an experimental perturbation of additional body mass, the idea that non-sagittal plane joint stabilizing moments may limit performance was examined. It was revealed that for the ankle and knee joints, non-sagittal plane moments higher than that experienced in maximum-effort curved sprints can indeed be endured. This observation challenged the stabilizing moment limit theory.

When sprinting with and without the additional mass, the total ankle moment generation remained constant, despite significant differences in the ground reaction force. It is possible that the ankle moment generation is at the limit. Through an induced acceleration analysis, it was identified that moment generated at the ankle contributed to the majority of the ground force generation. It is possible that the ankle moment generation was maximized for its importance in the ground force generation.

To examine the idea that ankle moment generation may limit curved sprinting performance, an experimental intervention of wedged footwear was implemented aiming to increase the ankle moment. By aligning the ankle joint closer to its neutral configuration in the frontal plane, ankle moment generation increased. That increase was associated with significant improvements in ground impulse generation, and overall curved sprinting performance.
Preface

Chapter 3, 4 and 6 are based, respectively, on the following manuscripts:


Luo, G. and Stefanyshyn, D. J. (accepted) Limb force and non-sagittal plane joint moments during maximum-effort curve sprint running in humans. *Journal of Experiment Biology*.

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To Mom and Dad
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<th>Description</th>
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<tbody>
<tr>
<td>µ</td>
<td>friction or traction coefficient</td>
</tr>
<tr>
<td>3D</td>
<td>three-dimensional</td>
</tr>
<tr>
<td>ANOVA</td>
<td>analysis of variance</td>
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<tr>
<td>BW</td>
<td>body weight</td>
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<tr>
<td>COM</td>
<td>centre of mass</td>
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<tr>
<td>COP</td>
<td>centre of pressure</td>
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<tr>
<td>GF</td>
<td>ground force</td>
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<tr>
<td>GI</td>
<td>ground impulse</td>
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<tr>
<td>GRF</td>
<td>ground reaction force</td>
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<tr>
<td>GRF_{a-p}</td>
<td>anterior-posterior ground reaction force</td>
</tr>
<tr>
<td>GRF_{cpt}</td>
<td>centripetal ground reaction force</td>
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<tr>
<td>GRF_{vert}</td>
<td>vertical ground reaction force</td>
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<tr>
<td>GRI</td>
<td>ground reaction impulse</td>
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<tr>
<td>HAT</td>
<td>head, arms and trunk</td>
</tr>
<tr>
<td>IAA</td>
<td>induced acceleration analysis</td>
</tr>
<tr>
<td>LCS</td>
<td>lab coordinate system</td>
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<tr>
<td>MT0.2</td>
<td>available traction condition of traction coefficient 0.26</td>
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<tr>
<td>MT0.5</td>
<td>available traction condition of traction coefficient 0.54</td>
</tr>
<tr>
<td>MT0.8</td>
<td>available traction condition of traction coefficient 0.82</td>
</tr>
<tr>
<td>MT1.1</td>
<td>available traction condition of traction coefficient 1.13</td>
</tr>
<tr>
<td>RMS</td>
<td>root mean square</td>
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<tr>
<td>s.d.</td>
<td>standard deviation</td>
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<td>SCS</td>
<td>segment coordinate system</td>
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The Journey is the Reward.

- a Chinese proverb
CHAPTER ONE: 
INTRODUCTION

Studying the limits to human performance is fascinating. Revealing the constraints for performing ‘faster, higher, and stronger’ not only serves to fulfill a profound curiosity about locomotion mechanisms but also provides tremendous practical implications. Ultimately, performance is determined in a complex manner by a variety of psychological, physiological and physical factors. The current dissertation focused on the identification of the performance constraints from a biomechanical perspective. More specifically, the investigation concentrated on the ‘faster’ aspect of performance through a series of studies involving maximum-effort curved sprints.

The main motivation for studying curved sprinting lies in its crucial role in sports demanding maneuverability. In professional soccer, for example, players perform more than 700 turns per game (Bloomfield et al., 2007). For basketball, it has been reported that athletes spend more than 40% of the game changing directions (Stacoff et al., 1993). Despite its importance, our understanding of locomotion along a curved path is rather limited (Alexander, 2002; Dickinson et al., 2000). The general goal of the current dissertation was, thus, to explore the potential limiting factors for maximum-effort curved sprinting performance.

The risk of skidding has been suggested as one key limiting factor for curved sprinting performance, especially in cases where the sprints were performed on curves of small radii (Alexander, 1982). Quantified as the maximum ratio of the horizontal over vertical GRF without skidding, the available traction at the shoe-ground interface confines the orientation of the GRF. Intuitively, for a given vertical GRF the more traction available, the more the athlete can utilize it to create the centripetal force without skidding. Thus, it would be expected that a higher available traction would result in a better performance. It is likely that once the available traction reaches a certain level, other factors become the predominant performance constraints. At and beyond such point, traction no longer limits performance. Research work investigating the scope to
which available traction limits performance is scarce in the literature. While the dramatic influence of footwear traction elements on cutting and accelerating performance has been demonstrated (Müller et al., 2009, 2010), the extent available traction limits performance remains to be systematically determined (Alexander, 2002).

The ability to generate limb force can limit the application of the centripetal GRF, and thus curved sprinting speed. Indeed, the peak limb force generation has been argued to be a key factor limiting top linear sprinting speed (Grabowski et al., 2010; Weyand et al., 2000). While earlier work with curved sprinting speed data tends to support predications based on a limb force limit model (Greene, 1985; McMahon, 1984), a direct examination of the GRF was not conducted until recently (Chang and Kram, 2007). It was observed by Chang and Kram (2007) that the limb force applied during sprints along small circles did not reach the maximum magnitude observed during straight sprints, which was used to define the peak limb force (also as in: Greene, 1985; Usherwood and Wilson, 2006), suggesting the limb may not operate at its force generation limit. While this finding challenged the idea of a constant peak limb force across radii, it may yet be insufficient in falsifying the hypothesis that the limb force was operating at the force generation limit - it is plausible that while the athletes were still exerting the limb force maximally, the peak magnitude reduced in small circles as a consequence of changes in body configuration. One potential way to further test the limb force limit theory could be by applying an external perturbation to the system – an additional mass for the limb to support. Based on the limb force limit theory, it would be hypothesized that as the requirement to generate GRI to support the additional mass increases, the peak limb force would remain operating at the threshold, and the stance time would extend. An experiment directly testing this hypothesis is currently lacking from the literature.

One novel theory was proposed by Chang and Kram (2007) in explaining their observations of the changes in the peak limb force across radii. It has been observed previously that during maximally exerted cutting movements (Wannop et al., 2010), the lower extremity joints experienced large non-sagittal plane joint moments – an indication of unwanted stress on joint soft tissues (Mizuno et al., 2009; Seering et al., 1980). Chang and Kram (2007) suggested that it is possible that the limb extensors reserved force
generation as the non-sagittal plane moments experienced at the lower extremity joints reached a safety threshold. Currently, experiments designed to directly examine this theory are not available in the literature. One possible approach to test hypotheses formulated based on this theory could be by prescribing experimental implementations that can potentially increase and/or decrease the non-sagittal plane joint moments. Based on the non-sagittal plane joint moment limit theory, it would be hypothesized that such implementations will induce changes in the limb force while the non-sagittal plane moments remain operating at the threshold level to perform maximally.

The lower limb is composed of multiple joints. As a chain is only as strong as its weakest link, it is possible that the moment generation at certain joints can limit the system’s performance. While investigation of this joint moment generation limit idea has been conducted for linear sprinting (Weyand et al., 2010), currently there is a lack of information regarding the function and operating states of the individual lower extremity joints during curved sprinting. Explorations of the moment generation at individual lower extremity joints may provide insight into the performance constraints. If the moment generation ability at certain joint(s) is at the limit, manipulating such ability can potentially lead to performance changes.

While direct kinematic measurement has not been reported, based on the GRF orientation (Chang and Kram, 2007), it can be expected that the ankle joint is possibly operating at an extreme angle during sprints along small circles. It is possible that the operation range of the ankle joint can embed a kinematic constraint limiting the body lean and the application of the centripetal GRF. Whether this kinematic constraint is a predominant factor limiting curved sprinting speed remains to be determined.

The current dissertation consisted of three experiments and one model to examine the potential limiting factors introduced above. The thesis is structured around these four individual analyses; however the information obtained from each analysis was used collectively in addressing the following six questions:

1. To what extent does available footwear traction limit body lean and overall curved sprinting performance?
2. When sprinting maximally along curves of small radii, is the supporting limb operating at its limit generating extension force?

3. During maximum-effort curved sprinting, is the non-sagittal plane joint loading experiencing its limit?

4. During curved sprinting, what are the contribution of joint torques, gravitational loading and Coriolis loading to the overall GF generation of the system? Furthermore, what are the individual contributions to the GF generation by the moment generated at each lower extremity joint?

5. During maximum-effort curved sprinting, are certain joints operating at the limit in generating torque to accelerate the body?

6. Is the ankle range of motion a predominant factor in limiting the body lean and overall curved sprinting performance?

In Chapter 2, background information and relevant literature is reviewed in detail. The influence of available traction on curved sprinting performance is addressed in Chapter 3. The investigation of whether limb force was operating at its limit was conducted in Chapter 4, where subjects performed maximum-effort curved sprints with and without an additional mass. The non-sagittal plane joint moment limit theory is first examined in Chapter 4, where the moments experienced at the ankle and knee joint were investigated for subjects sprinting with and without an additional mass. Such experimental implementation represents an attempt to increase the non-sagittal plane joint moments. In Chapter 6, the theory was further examined with an experiment where subjects performed the sprints in shoes with and without a frontal-plane wedge. This experimental implementation was introduced aiming to reduce the non-sagittal plane joint moments. The joint moment generation limit was collectively studied through Chapter 4,
5 and 6. In Chapter 4, the joint moment generation under normal and additional body mass conditions were examined. To gain an understanding of the individual joint moments’ contribution to the overall external force generation, a model was developed in Chapter 5. In Chapter 6, an attempt was made to manipulate the moment generation at a joint and its influence on the overall performance was investigated. The idea of ankle operation range as the limit to performance was examined in Chapter 6, where experimental footwear with and without a frontal plane wedge was implemented to change the ankle configuration. Chapter 7 summarizes the observations along with interpretations. It also addresses the limitations of the current work and makes suggestions for future research.
CHAPTER TWO:
REVIEW OF RELEVANT LITERATURE

2.1 Relevance of terrestrial locomotion along curves

Terrestrial locomotion is rarely linear. The ability to move along curves at high speed, often termed ‘maneuverability’, can have a direct influence on the survival of an animal (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 is an effective escape strategy for the prey, especially in the cases where the prey cannot outrun its predator along a straight path.

Turning at high speed is also common and crucial in athletic settings. An analogy can be drawn between the natural prey-predator scenario and a ball chase during a soccer match. Indeed, players in various sports spend a large portion of playtime changing directions. Through game video analysis, Bloomfield et al. (2007) observed that during professional soccer matches (England Premier League) players performed more than 700 turns per game. In Australian Football League, more than half of all the sprints during each game involve a change of direction (Dawson et al., 2004). For basketball, it has been reported that athletes spend more than 40% of the game changing directions (Stacoff et al., 1993).

Despite its importance, our current understanding of locomotion along curved paths is rather limited (Alexander, 2002; Dickinson et al., 2000; Higham, 2007). The current dissertation aimed to gain a better understanding from a biomechanics perspective through investigations on high speed curved sprints performed by human subjects. In this chapter, the general mechanics involved in curved sprinting are reviewed. Subsequently, potential limiting factors for maximum curved sprinting performance are discussed in detail.
2.2 Mechanics of curved sprinting

2.2.1 Ground reaction forces during curved locomotion

In order to travel along a curve, the body needs to create a centripetal acceleration (Figure 2.1). The magnitude of the centripetal acceleration is determined by the tangential travelling speed and the curve radius.

In order to achieve a tangential speed $v$, the runner needs to create a centripetal acceleration of magnitude $a$ directing toward the centre of the circle, where:

$$a = \frac{v^2}{r} \quad (2.1)$$

To accelerate the runner’s body centre of mass (COM), a centripetal ground reaction force (GRF) is required, and this is accomplished by generating a ground force (GF) at the foot-ground contact. Based on Newton’s second law of motion, the magnitude of the required centripetal GRF is a function of the runner’s body mass $m$ and the magnitude of the centripetal acceleration $a$, where:

$$GRF_{centripetal} = m \cdot a \quad (2.2)$$
With (2.1) and (2.2), it can be derived that:

\[
GRF_{\text{centripetal}} = m \cdot \frac{v^2}{r}
\]  

(2.3)

Equation 2.3 can be used to describe the relationship among tangential sprinting speed, curve radius and the centripetal GRF required (Figure 2.2; body mass = 70 kg).

\[
\begin{array}{ll}
\text{GRF}_{\text{centripetal}} & \text{[N]} \\
5000 & \hline \\
4000 & \\
3000 & \\
2000 & \\
1000 & \\
0 & \\
\end{array}
\]

\[
\begin{array}{ll}
r \text{[m]} & \\
5 & \hline \\
10 & \\
15 & \\
20 & \\
25 & \\
30 & \\
\end{array}
\]

\[
\begin{array}{ll}
v \text{[m s}^{-1}] & \\
0 & \hline \\
1 & \\
2 & \\
3 & \\
4 & \\
5 & \\
6 & \\
7 & \\
8 & \\
9 & \\
10 & \\
\end{array}
\]

Figure 2.2: The centripetal GRF required in order for a body (mass = 70 kg) to travel along curves of various radii \((r)\) at different tangential speed \((v)\).

Sprinting along a curve of a given radius, in order to increase the tangential speed, the required centripetal GRF will increase exponentially. Meanwhile, to maintain a
constant tangential speed, the required centripetal GRF is much bigger when sprinting along a circle of smaller radius.

In addition to the centripetal GRF, during level-ground locomotion, a vertical GRF is needed in order to counter the effect of gravity so that the body COM height can be maintained. Over time, the average magnitude of the vertical GRF equals the magnitude of the gravitational force acting on the body (2.4).

\[ GRF_{vertical} = mg \]  

where \( g \) is the magnitude of the gravitational acceleration. Unlike the centripetal component, the average magnitude of the vertical GRF over time is independent of either the travelling speed or the radius of the curve.

The anterior-posterior GRF affects the body’s acceleration/deceleration in this direction. At the initial stage of the run, as the body increases the forward traveling velocity, a net anteriorly directed GRF is needed. Once the runner achieves the steady state of the top tangential speed, the anterior and posterior GRF reach a balance so that the average GRF in this direction becomes zero. It shall be noted here that, different from straight running, at this steady-state stage the body COM still undergoes a net acceleration, but only toward the centre of the circle.

2.2.2 GRFs during stride cycles

The average GRF requirements outlined in the previous section apply as general physical rules to a locomotion system traveling along a curve. If a locomotion system is not in constant contact with the ground, in the case of a legged animal, greater GRFs are needed.

Locomotion of legged animals has been characterized by the cyclic movement of the limbs (Gray, 1968; Howell, 1944; Muybridge, 1899). A ‘stride’ cycle starts with the initial ground contact of a limb and ends with the next ground contact of the same limb. For each limb, the ground contact duration is finite. This ground contact period is referred to as stance phase. Accompanied with the stance phase is the swing phase where the limb
is lifted from the ground and moved through the air. It has been widely observed that the
stance duration reduces as the locomotion speed increases (McMahon, 1984; Weyand et
al., 2000). For human locomotion, the transition from walking to running is characterized
as the phenomenon that the stance phase of each limb is reduced to an extent that there is
no period where both limbs are in contact with the ground simultaneously. The period
where both limbs are off the ground is referred to as flight phase. Through flight phase,
the body COM experiences mainly the gravitation force and comparatively small air
resistance (Willems et al., 1995). The total duration (stance and flight) between the initial
ground contact of one limb and the initial ground contact of the opposite limb is referred
to as step time (Figure 2.3).

Figure 2.3: Illustration of a ‘step’ during running. A step is marked as the initial
touch-down of one limb to the initial touch-down of the opposite limb. The total
duration between these points are referred as step time and is the sum of the stance
and flight phase duration. Adapted from Muybridge, 1899.

The ratio of stance duration over the entire step duration ultimately determines the
average magnitude of the GRF that needs to be created during ground contact. Figure 2.4
helps illustrate this point using a simplified wave-shaped vertical GRF.
Figure 2.4: Vertical forces experienced at the COM over time. Gravity is constantly applying a downward force (body weight) pulling the body toward the ground. During ground contact, an upward GRF needs to be created in order to keep the COM from falling to the ground. The magnitude of this force is associated with the stance time – defined as the duration between touch-down \( (t = td_i) \) to toe-off \( (t = to_i) \) – and the step time – defined as the duration between touch-down of one limb \( (t = td_i) \) to the touch-down of the opposite limb \( (t = td_{i+1}) \)

In this example, over the entire step \( (t = td_i \text{ to } t = td_{i+1}) \), gravity is constantly applying a downward force (body weight) at the COM. The magnitude of the total downward impulse due to gravity over this step can be calculated as:

\[
I_{\text{gravity}} = \int_{td_i}^{td_{i+1}} m \cdot g \, dt
\]  

(2.5)

where \( m \) is the body mass and \( g \) is the magnitude of the gravitational acceleration.

To overcome this downward impulse, an upward ground reaction impulse (GRI) of equal magnitude needs to be created during the contact \( (t = td_i \text{ to } t = to_i) \):

\[
I_{\text{gravity}} = GRI = \int_{td_i}^{to_i} GRF \, dt
\]  

(2.6)

where GRF is the vertical force created during the ground contact. Thus:
\[ \int_{t_d}^{t_o} GRF \, dt = \int_{t_d}^{t_{d+1}} m \cdot g \, dt \] (2.7)

Since \( m \) and \( g \) remains unchanged over time, the average magnitude of the vertical GRF depends on the proportion of step time (\( t = t_d \) to \( t = t_{d+1} \)) that the limb is in contact with ground. It can be seen that the need to generate GRF increases as stance duration (\( t = t_d \) to \( t = t_o \)) reduces and/or flight duration (\( t = t_o \) to \( t = t_{d+1} \)) increases.

This relationship between required GRF and stance/flight duration also applies to the horizontal component of the GRF. In the medial-lateral direction, the need to support body weight over time is replaced by the need to create centripetal acceleration. For a given speed and curve radius, as the stance time decreases and/or flight time increases, the need to create centripetal GRF during ground contact increases.

During linear sprinting, stance duration decreases as speed increases (Weyand et al., 2000), while the flight duration remains largely constant (Grabowski et al., 2010; McMahon and Cheng, 1990; McMahon and Greene, 1979; Weyand et al., 2000). Consequently, at a sprint speed above 8 m/s, the average vertical GRF experienced by the supporting limb during ground contact has been observed to reach a magnitude of more than 2 times body weight (BW) (Grabowski et al., 2010; Weyand et al., 2010; Weyand et al., 2000).

### 2.2.3 Management of angular momentum

One additional task for a system traveling along a curve is to manage the overall body angular momentum (Alexander, 2002; Biewener, 2003). During walking and running, the orientation of the GRF varies over time. When the line of action of the GRF vector deviates away from the COM, it creates an external moment (Figure 2.5(a)). If not properly managed, the accumulation of angular impulse about the COM can result in falling. It has been observed that during walking (Herr and Popovic, 2008) the whole body angular momentum was highly regulated in all three dimensional spaces. Investigations on sprinting acceleration has shown that in the sagittal plane, the
orientation of the effective leg, defined as the vector from the centre of pressure (COP) to the COM, was closely coupled with the GRF direction (Kugler and Janshen, 2010).

Compared to running straight, during curved sprinting the resultant GRF needs to be oriented more horizontally in the frontal plane. Chang and Kram (2007) observed that during curved sprinting along circles with radii of 6 m or less, the resultant GRF was directed $35^\circ - 40^\circ$ away from the vertical. The effective leg orientation needs to be coupled with the orientation of the GRF in order to minimize the whole body rolling moment, and this was usually achieved by leaning into the ground (Figure 2.5 (b)).

Figure 2.5: As the line of action of the GRF deviates away from the COM, a rolling moment about the COM is generated (a); to avoid such rolling moment during curved sprinting, the athletes need to lean into the ground (b) so that the effective leg (a vector from the COP to COM) can be aligned with the GRF.

As shown in the previous section, as the curved sprinting speed increases and/or the radius of the curve becomes smaller, the need to create average centripetal GRF increases exponentially while the need to create the average vertical GRF remains constant. The unequal increases between the vertical and horizontal components of the GRF lead to an increasing need for the body to lean into the ground.

While in the frontal plane body rotation is likely undesirable for the stability of the system, in the transverse plane, during each contact the body needs to rotate into a
new traveling direction (Jindrich and Full, 1999). Control of the whole body yaw moment has been investigated for sidestep and crossover cuts during running (Jindrich et al., 2006), and compared between humans and ostriches – bipedal runners with a different yaw axial moment of inertia (Jindrich et al., 2007). It was demonstrated that for various turning conditions, runners use braking GRFs to compensate for potential over-rotations created by the centripetal GRF.

2.3 Potential limiting factors for curved sprinting performance

2.3.1 Definition of performance

The definition of athletic performance is task-specific. Generally speaking, performance can be assessed based on the final outcome of a specific sport activity (Nigg and Yeadon, 1987), which can range from jumping height to shooting accuracy. In this thesis, the ultimate movement goal was to achieve high sprinting speed along a curve. Performance in this case is defined as:

- The maximum body COM travelling speed that can be achieved sprinting along a curve of a prescribed radius.

2.3.2 Traction at the ground contact interface

The danger of skidding has been proposed as a factor limiting curved sprinting performance (Alexander, 1982; Tan and Wilson, 2010). The traction property of the ground contact interface constrains the maximum horizontal GRF that can be created without skidding, and as shown previously, the centripetal GRF is crucial for curved sprinting performance. In this section, the definition and measurement of traction is first outlined. Then, the relevant literature on the influence of traction on athletic performance is reviewed in detail.
2.3.2.1 Friction and traction

The sliding behaviours at the foot-ground interface are commonly described with dry sliding friction. Dry sliding friction indicates the ability to resist relative sliding motions for two surfaces in direct contact. It can be static or dynamic. Static friction describes the resistance that has to be overcome in order to initiate a relative motion between the two contacting objects. Dynamic friction, on the other hand, describes the resistance experienced at the interface during motion, and is usually of lower magnitude compared to the static friction.

Early investigations by da Vinci, Amontons, Coulomb and Euler contributed to the general understandings of the phenomenon of dry friction between uniform and rigid surfaces (Dowson, 1979). The empirical laws published in Amontons’s work (1699) stated that the static frictional force acting at the contact interface is proportional to the normal load applied (Figure 2.6) and is independent of the apparent contact surface area. The independence of friction to apparent contact area has been attributed to surface roughness (Tabor, 1981). While appearing to be in full contact, as the result of surface roughness the actual number of microscopic contact asperities providing frictional force may be limited. As normal force increases, the actual number of contacting asperities increases and so does the observed static frictional force. Another important empirical law of dry friction was referred to as Coulomb’s law. It states that the kinetic friction is independent of the sliding velocity (Coulomb, 1785).
Figure 2.6: Sliding frictional force is a dissipative force opposing the relative motion between two contacting bodies. In static equilibrium, the frictional force has an equal magnitude as the pulling force but is acting in the opposite direction. As the pulling force increases, the frictional force increases, but only to a point after which sliding will be initiated. For a given pair of objects, the magnitude of such sliding initiation threshold is largely dependent on the normal force applied to hold the two objects against each other.

Due to the linearity between the normal and static frictional force stated in the empirical laws, the variable ‘coefficient of friction’ has been widely used to describe the frictional properties of two surfaces in contact. An instantaneous coefficient of friction is calculated as:

\[ \mu_{\text{instantaneous}} = \frac{F_{\text{frictional}}}{F_{\text{normal}}} \] \hfill (2.8)

The maximum coefficient of friction that can be experienced between the two contacting surfaces without sliding is referred to as the static coefficient of friction. Thus:

\[ \mu_{\text{static}} \geq \mu_{\text{instantaneous}} \] \hfill (2.9)

In an ideal dry and solid contact scenario, the static coefficient of friction value is a material-specific constant representing a physical property of the two contacting surfaces. In realistic foot- or footwear-ground contacts, however, the static coefficient of
friction no longer behaves like a material constant (Shorten et al., 2003). Due to material behaviours such as asperity deformation, adhesion, wear and ploughing, the static coefficient of friction varies as the contact conditions change. It has been shown that the frictional properties of footwear and its contact surface change depending on the contact region (Bonstingl et al., 1975; van Gheluwe et al., 1983) and normal load (Schlaepfer et al., 1983; Valiant, 1987; Wannop and Stefanyszyn, 2012; Warren, 1996). In addition, the sliding speed has been shown to alter the kinetic coefficient of friction (Beschorner et al., 2007; Grönpvist, 2003). Since the classic friction laws are limited in describing the empirical sliding phenomenon between footwear and surface, the term ‘traction’ is commonly used when referring to these friction-like behaviours.

2.3.2.2 Quantifications of traction

The traction properties between footwear and surface vary as the contact conditions change. Thus, empirical mechanical assessments are needed to quantify traction under realistic loading conditions. The term ‘available’ traction (a representation of the ‘static coefficient of friction’) is used throughout this thesis for the material properties quantified through such mechanical tests. It is defined as the maximum ratio of the horizontal over normal force experienced at an interface without large-scale slippage. It sets the boundary at and under which the human subjects can apply force at this interface. The actual traction, or force ratio, used during movements is termed as ‘utilized’ traction (equivalent to ‘instantaneous coefficient of friction’).

A mechanical traction tester is generally composed of: 1) an adjustable shoe last where the footwear can be mounted, 2) a weight holder where mass can be placed to apply normal load over the footwear against the surface, 3) horizontal rails along which the shoe-weight holder can move freely, 4) a mechanism to apply horizontal force, and 5) a mechanism to measure the reaction forces at the footwear-ground interface (Figure 2.7). Modern traction testers vary in the implementations of these components to allow the experimenters to conveniently adjust the testing parameters to represent realistic contact conditions (Chang et al., 2001).
Figure 2.7: A typical footwear-surface traction tester. The tester allows the investigators to conveniently adjust the contact parameters to represent realistic loading conditions.

In a typical mechanical test, the shoe last is first adjusted to represent the foot contact orientation. A vertical load is applied to the shoe against the surface. Then a horizontal force is applied to the shoe-load unit. With the force measurement system, the ratio between the horizontal and vertical interface force (i.e. traction coefficient) can be quantified. The decision of quantifying static versus dynamic traction coefficient should be made based the nature of the research question. In ergonomic studies on falling, dynamic traction coefficient has been proposed to be of more importance in assessing the risk of slip accidents (Strandberg and Lanshammar, 1981). In studying human performance, however, where footwear-ground interface shall not display large sliding, evaluation of the static (or quasi-static) traction coefficient appears more appropriate (Valiant, 1990).
2.3.2.3 Traction and general locomotion performance

For a given vertical GRF, the traction available between the footwear and ground determines the maximum horizontal GRF that can be created without a large-scale slippage. In other words, it constrains the orientation of the GRF. The orientation of the GRF, on the other hand, has been shown to play a critical role in determining athletic performance, especially acceleration performance (Hunter et al., 2005; Kugler and Janshen, 2010; Morin et al., 2011).

Despite its intuitive importance, the number of studies investigating the influence traction has on general athletic performance is limited. In an earlier study on American football, Krahenbuhl (1974) showed that athletes wearing cleated boots could finish a running course faster on a synthetic compared to a natural grass surface. The difference in available traction was suggested to largely explain the observed performance difference but unfortunately the available traction properties of these two surfaces were not quantified. Factors such as surface compliance could have confounded the interpretation. In a recent study on soccer shoe cleat configuration, cleat length was systematically modified on otherwise identical shoes (Müller et al., 2009). The cleats of a regular soccer shoe were abraded to half of their original length in one condition, and totally removed in another. It was found that as cleat length decreased, cutting and accelerating performance reduced in a step-wise order. While the available and utilized traction were not quantified, these results clearly demonstrated the importance of traction related elements on performance. In a subsequent research study by the same research group (Müller et al., 2010) evaluating performance for four different types of soccer boots on synthetic turf, available traction, utilized traction and performance were assessed. Mechanical testing showed that one shoe had more than 30% higher available traction compared to the others. The subjects however achieved the lowest slalom run performance in this shoe. Interestingly, although the available traction of this shoe was the highest, the traction utilized by the subjects during the performance tests was lower compared to the other shoes. Thus, while it is universally accepted that footwear traction can influence athletic performance, a basic understanding between traction and performance is still missing.
One commonly used approach in determining footwear traction for maximum performance is to quantify the ‘required’ traction for a target movement (Valiant, 1990; Shorten et al., 2003). In this method, subjects perform a movement task in footwear that is assumed to not slide on the test surface. The utilized traction through stance is quantified with the GRF data from a force platform secured under the test surface. The peak utilized traction value observed is defined as the ‘required’ traction for the entire movement, and used as the standard for constructing shoes (Table 2.1).

<table>
<thead>
<tr>
<th></th>
<th>Peak utilized traction coefficient</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking</td>
<td>0.17 - 0.22</td>
<td>Redfern et al., 2001</td>
</tr>
<tr>
<td>Running</td>
<td>0.60 - 0.70</td>
<td>Valiant, 1993</td>
</tr>
<tr>
<td>90° – 180° turns</td>
<td>0.80 - 1.25</td>
<td>Shorten et al., 2003; Valiant, 1990</td>
</tr>
</tbody>
</table>

Table 2.1: Peak traction utilized during different types of human locomotion. These traction coefficients have often been used to define the ‘required’ available traction for performing the given tasks.

While the required traction provides important information regarding the GRF created during a movement, it needs to be interpreted with the following assumptions in mind.

First, it is assumed that the testing footwear-surface combination provides more than enough available traction for maximum performance. This assumption is often supported with the absence of sliding observed during the task. While the absence of sliding is necessary for the assumption, it is not sufficient. The main reason is that human subjects can sense changes in surface traction properties and adapt movement patterns accordingly (Cohen and Cohen, 1994). Adaptations such as changing step length, which in turn changes the attack angle (Cooper et al., 2008), will change the utilized traction values. The observation that there is no sliding can be a result of such adaptation. Since the adapted movement then may not represent the maximum execution of a task, the
required traction quantified can potentially underestimate the actual value for maximum performance.

Secondly, it is assumed that the peak utilized traction observed is relevant to maximum performance. The utilized traction value is quantified as the ratio of the horizontal over vertical GRF. As the magnitude of the denominator - the vertical GRF - approaches 0 N at the very early and late stance, high utilized traction values are observed (Figure 1 in Valiant, 1990). The significance of these high utilized traction values to performance may be limited due to the low magnitude of the GRF. Thus, using such peak utilized traction values may overestimate the actual traction needed for maximum performance.

Therefore, to systematically explore the influence of available traction on performance, experiments where subjects perform movement tasks in different available traction conditions are needed.

2.3.2.4 Traction and curved sprinting performance

Available traction has been suggested as a key limiting factor to curved sprinting performance for humans (Alexander, 1982, 2002) and racing horses (Tan and Wilson, 2010). Based on the relationship between GRF and curved sprinting speed outlined previously, a theoretical performance constraint by available traction may be formulated.

For simplicity, the following assumptions are made: 1) the force generation between limbs are symmetric, and 2) the effective limb length is constant across radius and curved sprinting speed.

If the available traction can be fully utilized, the utilized traction between footwear and ground is:

$$\mu = \frac{\text{GRF}_{\text{centripetal}}}{\text{GRF}_{\text{vertical}}}$$

(2.10)

Over time the magnitude of the vertical GRF should equal to BW:
The magnitude of the centripetal GRF is:

\[ GRF_{\text{centripetal}} = m \cdot \frac{v^2}{r} \]  \hspace{1cm} (2.3)

To minimize the rolling moment in the frontal plane, a body lean is needed. As the body leans into the ground the actual body COM travelling radius decreases. This reduction is a function of the body lean magnitude and effective leg length:

\[ r = R - \cos \left( \tan^{-1} \left( \frac{GRF_{\text{vertical}}}{GRF_{\text{centripetal}}} \right) \right) \cdot L \]  \hspace{1cm} (2.11)

where \( R \) is the distance between the centre of the curve and landing foot, and \( L \) is the effective leg length.

Based on these equations, the relationship between curved sprinting performance \( v \) and utilized traction \( \mu \) can be derived:

\[ v = \frac{2}{\sqrt{\mu r g}} \]  \hspace{1cm} (2.12)

where \( r = R - \cos \left( \tan^{-1} \left( \frac{GRF_{\text{vertical}}}{GRF_{\text{centripetal}}} \right) \right) \cdot L \)

This mathematical relationship is depicted in Figure 2.8 for a prescribed radius \( R = 2.5 \) m and constant effective leg length \( L = 1 \) m. It can be seen that if available traction is the sole limiting factor and can constantly be fully utilized, as it increases, the maximum curved sprinting speed would continuously increase.
Figure 2.8: Theoretical performance constraint imposed by available traction. The constraint was derived using Equation 2.12 for curved sprints performed along circles of $R = 2.5$ m.

The extent to which available traction is the primary performance constraint for curved sprinting performance has not been reported in the literature. However, based on assumed available traction values, it has been proposed that during curved sprinting along a small radius the danger of skidding could limit the overall performance.

Tan and Wilson (2010) investigated the traveling speed of polo and racing horses when turning along curves of various radii. When turning along curves of relatively smaller radii ($r < 30$ m), it was suggested that the travelling speed seemed to be limited by the available traction between the hoofs and ground (Figure 2.9). It can be seen that the theoretical performance constraint imposed by available traction of 0.6 correlated well with the top speeds measured. Unfortunately, neither the available traction nor the actual utilized traction was measured in the study. It remained unknown if the chosen available traction value represented the actual interface properties.
Figure 2.9: Adapted from Tan and Wilson (2010). The tangential traveling speed of racing horses (pink) and polo horses (blue) during turnings along curves of different radii. A theoretical performance constraint imposed by available traction between the hoof and ground was formulated and plotted as the solid line. The available traction value was assumed to be 0.6. The connected stars represent the 99 percentile of the speed achieved for different radii. The top speeds achieved by the horses when turning along small curves are generally under the performance constraint imposed by available traction. The dashed line depicts a theoretical performance constraint imposed by limb force (discussed in detail in the following section), and the dotted line represents the speed on a straight path.

In a review of locomotion stability and maneuverability, Alexander (2002) suggested that when humans sprint along curves of small radii, available traction could limit the maximum performance. The author demonstrated this point with empirical data of one subject from McMahon (1984) using an assumed available traction value of 0.6 (Figure 2.10). It can be seen that for radii smaller than 10 m, the experimental outcomes
agreed well with the theoretical predications made based on the traction constraint theory. However, it remains uncertain how representative the chosen available traction value is. The running trial was performed with a spiked shoe on grass, and available traction greater than 1.0 has been reported for cleated soccer shoes on grass (Müller et al., 2010; Stefanyshyn et al., 2010). Thus it is likely that the assumed available traction of 0.6 was an underestimation. The author - Alexander (2002) - concluded that “experiments on surfaces giving different coefficients of friction would be needed to test this hypothesis.”

![Figure 2.10](image)

**Figure 2.10:** Adapted from Alexander (2002). A theoretically formulated performance constraint (dashed) imposed by available traction seems to explain the maximum speed achieved during sprints along small circles (dots).

In conclusion, theoretically available traction can limit the maximum curved sprinting performance, especially along curves of small radius. However, the knowledge of the extent to which available traction is the main performance constraint is currently limited. Investigations with systematic traction modifications are needed to gain such understanding.
2.3.3 Limb extension force

The need to create GRF increases exponentially as curved sprinting speed increases (Section 2.2.1). While available traction constrains the orientation of the GRF, the ultimate ability of an athlete to generate limb force can potentially constrain the magnitude of the GRF, and thus, overall performance.

Treating the supporting limbs as extensible struts has allowed the investigation of several fundamental locomotion mechanisms (Dickinson et al., 2000; Lee et al., 2011). From a performance perspective, the maximum axial force generated by such ‘struts’ has been associated with human top sprinting speed (Weyand et al., 2000). As running speed increases, stance time decreases, and as a result the need to generate GF increases (Section 2.2.2). Weyand et al. (2000) found that faster sprinters were the ones associated with greater GRFs when sprinting at their top speed.

The peak GF generation by the limb has been proposed as a limiting factor for maximum curved sprinting performance (Greene, 1985, 1987; McMahon, 1984). In an early model by Greene (1985), GRF was described as an on/off square signal over variable contact durations. Assumed to be the limiting factor, the peak resultant GRF (representing the maximum GF generation) was set constant across sprints on various curvatures. Since the resultant GRF is constant, as the need to create centripetal GRF increases, the vertical GF generation is compromised. In order to create sufficient vertical GRI to support the BW, the stance duration needs to be extended. Given that the COM travel over stance remained nearly unchanged (Cavagna et al., 1976; McMahon and Greene, 1979; Weyand et al., 2000), the travelling speed - calculated as the COM travel divided by stance duration - would decrease. This model was later modified by Usherwood and Wilson (2006) in studying 200 m indoor sprinting speed. While in Greene (1985) the total stride duration was set as constant independent of the sprinting speed, Usherwood and Wilson (2006) set the swing time of each limb constant based on experimental data from Weyand et al. (2000). Nevertheless, the fundamental assumption of a constant peak limb force limit is consistent between the two models. When the theoretically predicted sprint speeds were compared with experimentally measured speed, good agreements were reached for sprints performed along curves of large radii. For
circles of smaller sizes on the other hand, the agreement was less satisfactory (Greene, 1985; Figure 2.11).  

![Figure 2.11: Adapted from Greene (1985). A theoretical performance constraint (solid line) for curved sprinting imposed by peak limb force. The dashed line represents the maximum sprinting speed along a linear path. The dots were experimentally observed curved sprinting speed along curves of different radii. The theory assumed that the peak force experienced by the limb remains constant across sprints on curves of different radii.](image)

While the speed data (Greene, 1985; Usherwood and Wilson, 2006) provided confidence for the aforementioned limb force limit theory, it was not until recently that a direct examination of the theory through GRF measurement was published (Chang and Kram, 2007). In this study, subjects performed maximum-effort sprints along a straight path and curves of 1, 2, 3, 4 and 6 m radius. The peak resultant GRF observed during straight sprint was defined as the limb force limit, and was compared with the peak GRFs observed during curved sprints. The authors hypothesized that, based on Greene (1985), if the physiological limb force generation is the limiting factor, the peak resultant GRF should remain constant across different curvatures. Statistically significant differences in peak GRF were found for the inside limb between straight sprint and curved sprints on radii of 1 and 2 m. Meanwhile, a systematic trend of decreases in the peak resultant GRF
as the curve radius decreases was apparent for both limbs (Figure 2.12). However, this observation was not statistically significant, which was likely associated with the limited sample size (n = 4 or 5 depending on the condition). These observations challenged the notion that the maximum physiological force would be generated along curves of different radii. The authors concluded that “each subject had ample leg extension force available...for a shorter ground contact time” and proposed that “the generation of the ground reaction forces was constrained instead by one or more other limiting factors”.

![Figure 2.12: Adapted from Chang and Kram (2007). The peak limb force generated by the inside and outside legs during curved sprinting on small curves. The peak limb forces during curved sprinting appear to be lower than the peak force observed during linear sprints (dashed), especially for the inside leg.](image)

One alternative interpretation of Chang and Kram’s (2007) findings may be that the peak limb force generation was still the limiting factor during the sprints on small circles, but the magnitude of this peak force was not constant with the peak force generated during straight sprint, different from what Greene (1985) assumed. This may
be due to, for example, the change of the lower extremity joint configurations. Unfortunately, this hypothesis cannot be tested with the data currently available in the literature. One plausible way to further examine the limb force limit theory could be by overloading the system, which may be achieved by manipulating the subject’s body mass. If the limb is indeed generating its maximum extension force, as the body mass increases, the average resultant GRF shall be expected to remain constant and the stance time would be extended for the additional GRI needed.

2.3.4 Non-sagittal plane lower extremity joint loading

To explain the large reduction in force generation by the inside limb during curved sprints on small radii, Chang and Kram (2007) proposed it might be that the non-sagittal plane lower extremity joint moments had reached their critical thresholds preventing further limb extension force generation.

Resultant joint moment calculated with an inverse dynamics approach represents the net moment required at a joint to satisfy the body equations-of-motion (Andrews, 1995). It indicates the combined effect of the tissues, such as the muscles, bones, tendons and ligaments, surrounding the joint, and has been suggested to provide a good indication for loading at a joint (Hurwitz et al., 1998; Scott and Winter, 1990; Stefanyshyn, 2003), given that direct measurements of tissue loading in vivo are impractical for human experiment.

The non-sagittal plane moment has been associated with unwanted stress on joint soft tissues, especially for the knee joint (Mizuno et al., 2009; Seering et al., 1980). Excessive non-sagittal plane joint moment and angular impulse have been associated with both running and lateral sports injuries (Hewett et al., 2005; Kristianslund et al., 2011; Stefanyshyn et al., 2006). It is likely that during maximum-effort curved sprints, lower extremity joints experience tremendous undesired non-sagittal plane moments. While baseline joint kinetic information for curved sprinting is currently unavailable in the literature, observations made on other movements involving full-effort direction changes can provide insight into the magnitude of such moment. For example, in an experiment by Wannop et al. (2010), subjects performed maximum-effort 45° cutting
manoeuvres. For the lateral limb, it was found that the peak knee frontal plane moment reached a magnitude as large as the peak sagittal plane moment.

Since the lower extremity joints are likely experiencing large non-sagittal plane moments, there may be an operating threshold at which the body would inhibit further generation of limb force in order to protect the joints (Chang and Kram, 2007). If there is such a safety threshold limiting force generation, it would be expected that by manipulating such non-sagittal plane moments experimentally, changes in limb extension force would be observed. Currently no experiment has been designed to directly examine this theory. However, two observations exist in the literature tend to indirectly support this notion. In an experiment by Greene (1987), subjects performed maximum-effort curved sprints along tracks built on various frontal plane bank angles (0° – 30°). By providing a banked surface, it is possible that the GRF was aligned closer to the lower extremity joint centres and reduced the non-sagittal plane moments. Such potential reduction in non-sagittal plane moment could in turn allow a greater exertion of limb force. Indeed, the authors observed a speed improvement up to 10% running on the banked surface compared to flat. Unfortunately, joint kinematic and kinetic data were not assessed in this experiment. The other indirect evidence to this theory is from another condition Chang and Kram (2007) implemented in their study. In this condition, subjects sprinted along the curves with a tether attached to their approximate body COM location supplying the centripetal force. Since the centripetal GRF was no longer needed, body lean could then be eliminated. The peak limb force generated during this tethered condition was compared to normal curved sprints. In the tethered condition, a clear increasing trend in the peak resultant GRF was observed for the inside limb across radii. It is possible that the tethered condition changed the body orientation and thus joint configurations, which in turn reduced the non-sagittal plane moment allowing further generation of limb forces. These speculations could not be further evaluated as joint kinematic and kinetic data were not available from this study.

In summary, it is likely that the lower extremity joints experience large non-sagittal plane moments during maximum-effort curved sprints. Such non-sagittal plane joint moments could have reached a safety threshold so that further limb force generation
is prohibited. If such a safety threshold is indeed the limiting factor, by applying experimental implementations that can increase and/or decrease joint moments, changes in limb extension force would be observed, however, the non-sagittal plane moments would be expected to continue operating at such a threshold level.

2.3.5 Joint actuation

While the ‘strut’ leg has been used to reveal some of the fundamental locomotion mechanisms, it alone may not be adequate in studying the overall ability of the limb to actively generate GF (Pandy, 2003). The human body is composed of segments connected through various joints. The resultant GRF observed essentially represents the collective acceleration of all the body segments. Thus, in studying the limit of the GF generation, it is important to investigate the biological factors associated with the creation of segment accelerations.

The equations of motion for a linked-segment system can be expressed in the generalized coordinate system in matrix form as Equation 2.12 (Erdemir et al., 2007; Otten, 2003; Pandy, 2001; Zajac et al., 2002).

\[ M(q) \ddot{q} = C(q, \dot{q}) + G(q) + T + E \]  

(2.12)

where \( q, \dot{q} \) and \( \ddot{q} \) are the vectors of generalized coordinates, velocities and accelerations; \( M(q) \) is the mass matrix of the system; \( C(q, \dot{q}) \) is a generalized force vector due to the centrifugal and Coriolis loading; \( G(q) \) is a generalized force vector due to gravitational loading; \( T \) is a vector of net joint torques; and \( E \) is a vector of the external forces applied to the body by the environment - in the case of legged human locomotion, \( E \) essentially represents the GRF.

Equation 2.12 reveals two main factors contributing to the overall GRF – passive resistance to the gravitational loading - \( G(q) \), and the net joint torque – \( T \) (Zajac et al., 2002). Depending on the body configuration, the generation of the GF can be achieved by mainly one of these factors or the combined effects of both. A physical example of this point is as follows. A body is in an upright (Figure 2.13(a)) versus a crouched (Figure
2.13(b)) posture. Assuming both systems are in static equilibrium, the resultant GRF in both cases will act vertically through the body COM with a magnitude equal to the BW. In the upright posture, the majority of the GF generation can be attributed to the passive resistance provided by the bones along the long axis of the segments. In this scenario, the GF generation is likely limited by the safety threshold of the bones to endure stress. In the crouched posture, however, the majority of the GF generation is likely contributed by the torques generated across joints by the muscle forces. The GF generation limit in this case may be the safety threshold of the cross-sectional stress of the muscle tendon units (MTUs) (Biewener, 1990).

![Diagram](image.jpg)

Figure 2.13: Limiting factor for GF generation can change as the body configuration changes. Although the magnitude of the GRF in (a) and (b) are the same, assuming static equilibrium, the GF generation was contributed by different biological factors.

Individual contribution of the factors in Equation 2.12 to the overall GF generation has been investigated with an induced acceleration analysis (IAA) for walking and running (Anderson and Pandy, 2003; Hamner et al., 2010; Kepple et al., 1997; Liu et al., 2006; Sasaki and Neptune, 2006), and it was found that for both movements the majority of the GF generation can be attributed to the net moment generated at the joints as opposed to the passive resistance to gravity. Even during walking, where the body has...
been widely modeled as an inverted pendulum, it has been found that the vertical GF
generation due to the passive skeletal resistance to gravity was less than 50% BW
through stance (Anderson and Pandy, 2003). In addition, centrifugal and Coriolis forces
contributed very little to the overall GF due to the low joint velocities. During fast
walking (Liu et al., 2008) and running (Hamner et al., 2010), as the lower extremity joints
adopt more flexed configurations, the contribution by the passive skeletal resistance to
GF generation becomes negligible. While IAA for lateral movements such as curved
sprinting is not available in the literature, it may be assumed that as the body is likely in a
crouched posture the joint torques would contribute to the majority of the GF generation.
If joint torque is the main contributor to the GF generation, limits to its generation could
ultimately constrain the overall performance.

Observations by Chang and Kram (2007) that the peak resultant GRF reduced as
the curvature became tighter may be partially explained with such joint torque limit
theory. Based on Equation 2.12, it can be seen that the effect of joint torque on overall
GF generation depends on both joint angles ($q$) and angular velocities ($\dot{q}$). As the joint
configurations change, the mass distribution of the system ($M(q)$) will change, which in
turn alters the principle inertia axis of the whole system. It is possible that the lower
extremity joints continuously operate at their limits during sprinting around different
curvatures, but the body posture has changed. The posture adaptations could have in turn
changed the overall force generation by the limb.

Currently, there has not been any reported experiment designed to specifically test
the joint torque generation limit theory for curved sprinting. In an experiment
investigating the limiting factors for top-speed linear sprinting performance, Weyand et
al. (2010) estimated the torque generation at the ankle, knee and hip joints, and compared
the values between forward running/sprinting and one-leg hopping over a similar speed
range. They found that during hopping, the torque generated across all the joints were
significantly greater compared to forward running. Based on this finding, it was
concluded that top sprinting performance is not limited by the maximum joint torque
generation. A closer look at the methodology revealed two major issues that may weaken
such interpretation. First, instead of using a ‘bottom-up’ inverse dynamics calculation,
joint torques were estimated by directly multiplying the GRF vector by the perpendicular distance between the GRF vector and the joint centres. Moment calculation errors based on this method are significant, especially for the proximal joints (Wells, 1981). Secondly, the GRF vectors, their lever arms and the locations of the COP were estimated based on data averaged over stance duration, then trials and then different speed conditions. Such data treatment would likely introduce significant errors in the estimation of peak torque generation at individual joints. One alternative approach to examine the joint torque generation limit theory could be to have the subjects perform maximally in both high and low external load conditions (such as the two-leg sprinting versus one-leg hopping in Weyand et al. (2010)), and determine and compare, instead, the peak joint moments generated. If the torque generated at certain joints are truly at the limit when sprinting maximally, it would be expected that across the external load conditions, the torque generation at such joints would remain unchanged.

2.3.6 Joint operation range

As discussed previously, body lean is needed for creating centripetal GRF without causing a rolling moment about the body COM. The need to lean in to the ground increases exponentially as curved sprinting speed increases (Section 2.2.3). Unlike digitigrade animals (dogs and horses for example), body lean for human runners is primarily associated with the articulation at the ankle joint. It is plausible that the maximum curved sprinting performance is constrained by the kinematics of the ankle joint.

Based on the GRF data, Chang and Kram (2007) estimated that a 35 – 40° of ankle inversion/eversion would be expected for the body lean. During side shuffle movements (Simpson et al., 1992), shoe-shank inversion angles of 39° have been reported. While such reported angle may contain errors – ranging from 10° to more than 20° – due to foot-shoe movements (Stacoff et al., 1996), it is still possible that the ankle complex is operating at the kinematic limit during maximum-effort lateral movements.

One approach to examine if the ankle joint kinematics are limiting the maximum performance could be to assess the ankle range of motion with a mechanical tester and then compare the values with the kinematic data collected during the experiment. This
method may be limited for 1) passively assessed range of motion tends to ‘creep’ as the number of trials increases (Nigg et al., 1995); 2) the in-shoe movement introduced during the experiment may make the comparison of the range limit values difficult. Alternatively, the range of motion theory may be tested with a banked surface such as the one implemented in the experiment by Greene (1987). If joint kinematics, ankle in/eversion in particular, are the limiting factor, when the subjects perform curved sprints maximally on a flat versus banked surface, the same peak joint angle would be expected across conditions.

2.4 Summary and Statement of Problem

Curved sprinting performance is crucial in both the natural environment and athletic settings. Our current understanding of the factors limiting curved sprinting performance is rather limited. This chapter outlined major mechanical and biomechanical factors that can have a direct influence on the generation of GF and overall curved sprinting performance. The influences of these factors are assessed through a variety of experiments in this thesis. The following questions are addressed:

1. To what extent does available footwear traction limit body lean and overall curved sprinting performance?

2. When sprinting maximally along curves of small radii, is the supporting limb operating at its limit generating extension force?

3. During maximum-effort curved sprinting, is the non-sagittal plane joint loading experiencing its limit?

4. During curved sprinting, what are the contribution of joint torques, gravitational loading and Coriolis loading to the overall GF generation of the system?
Furthermore, what are the individual contributions to the GF generation by the moment generated at each lower extremity joint?

5. During maximum-effort curved sprinting, are certain joints operating at the limit in generating torque to accelerate the body?

6. Is the ankle range of motion a predominant factor in limiting the body lean and overall curved sprinting performance?
3.1 Introduction

Sufficient frictional force between footwear and playing surface is extremely important for sport performance. It allows an athlete to accelerate/decelerate rapidly and turn sharply without skidding. Since the frictional behavior between compliant outsole and surface materials does not follow the traditional Amontons-Coulomb paradigm, the term ‘traction’ instead of ‘friction’ has been proposed to better represent the complex nature of such interactions (Shorten et al., 2003).

Traction is commonly quantified using a traction coefficient, which is defined as the ratio of the horizontal over vertical force. In discussion of traction coefficient values, the distinction between two concepts requires clarification: the available versus utilized traction. The available traction describes the material properties of two contacting surfaces in terms of their ability to resist relative sliding, and is usually quantified using a mechanical apparatus. Utilized traction on the other hand is the actual force ratio occurring between the footwear and play surface during movements. Skidding happens when the utilized traction exceeds the available traction.

While the available traction between different shoe/surface pairs has been widely examined (e.g. Bonstingl et al., 1975; Schlaepfer et al., 1983; Valiant, 1987; Valiant et al., 1985), the number of studies specifically investigating the relationship between available traction, actual utilized traction, and, ultimately, human performance is limited.

In an earlier study on American football, Krahenbuhl (1974) showed that athletes wearing cleated boots could finish a running course faster on a synthetic compared to a natural grass surface. The difference in available traction was suggested to largely explain the observed performance difference but unfortunately the available traction properties of the different surfaces were not quantified. In a recent study on soccer shoe
cleat configuration, cleat length was systematically modified on otherwise identical shoes (Müller et al., 2009). The cleats of a regular soccer shoe were abraded to half of their original length in one condition, and totally removed in another. It was found that as cleat length decreased, cutting and accelerating performance reduced in a step-wise order. While the available and utilized traction were not quantified, these results indicate the importance of traction related elements on performance. In a subsequent research study by the same group (Müller et al., 2010) evaluating four different types of soccer boots on synthetic turf, available traction, utilized traction and performance were assessed. In this study, one shoe was found to have higher available traction compared to the other three. That shoe was, however, associated with reduced slalom run performance and was perceived by the subjects as the shoe with the worst performance. Interestingly, although the available traction of this shoe was the highest, the traction utilized by the subjects during the performance tests was lower with this shoe. It was suggested by the authors that this might have been due to the existence of a confounding factor: instability due to a lack of full cleat-ground engagement of this shoe condition. In summary, while it is universally accepted that footwear traction can influence athletic performance, a systematic investigation of this relationship is currently missing.

Intuitively, the more traction available, the more an athlete can lean into the ground and direct the ground reaction force (GRF) toward the favoured direction, resulting in a greater acceleration. During curved sprinting, a centripetal acceleration needs to be created in the body frontal plane. The ability to orientate the GRF toward the curve centre thus can be crucial for curved sprinting performance. When sprinting at top speed \( v \) along a curve of radius \( r \), an available traction coefficient \( \mu \) is needed in order to generate sufficient centripetal force:

\[
\mu \geq \frac{F_c}{F_n} = \frac{m \cdot (v^2/r)}{m \cdot g} = \frac{v^2}{r \cdot g}
\]

where \( F_c \) is the magnitude of the centripetal force, \( F_n \) is the magnitude of the normal force, \( m \) is the body mass and \( g \) is the magnitude of the gravitational acceleration.
Assuming the available traction can be fully utilized, Equation 3.1 would suggest that performance \( (v) \) could be continuously improved by systematically increasing the traction available. However, it is clear that each athlete has some other limits on his or her ultimate performance. Theoretical predictions of human and animal curved sprinting speed have suggested that when sprinting maximally along small circles, the available traction could be the main performance constraint up to a point, beyond which proposed characteristics such as limb strength may become the predominant limiting factors (Alexander, 2002; Tan and Wilson, 2010).

In developing appropriate footwear, a simple solution from a performance perspective would be to provide an abundance of available traction. This would ensure that sufficient traction is always available and is, therefore, never the limiting factor of performance. However, the shortcoming of this approach is that high available traction has also been proposed to be associated with athlete injury (Lambson et al., 1996; Nigg and Segesser, 1988; Torg and Quedenfeld, 1971; Torg et al., 1974). While the injury mechanism still requires further investigation, it has been proposed that high available traction might lead to increased risk of foot fixation which might in turn increase the stress on biological tissues (Torg et al., 1974). Thus, if in fact a critical traction point exists above which traction is no longer the limiting factor for performance, it would be important to identify this point as it could serve as the design criteria for athletic footwear.

In studying performance constraints for curved sprinting, the knowledge of the critical traction point also has its significance. In previous investigations (Alexander, 2002; Chang and Kram, 2007; Tan and Wilson, 2010), assumptions of the available traction values have been made to help interpret the findings. With these assumptions, the unknown effects of available traction may confound the understanding gained. It has been suggested previously that a systematic investigation of the extent to which available traction limits curved sprinting performance is needed (Alexander, 2002).

The purpose of this study, thus, was to investigate the relationship between the available traction and maximum-effort curved sprinting performance. It was hypothesized that performance would increase as available traction increased but only to a point after
which performance would plateau and further increases in shoe-surface traction would not affect performance.

3.2 Materials and methods

3.2.1 Footwear and traction properties

Initially, four available traction conditions were targeted for this experiment. They were traction coefficients of: 0.2, 0.5, 0.8, and 1.2. This range of available traction was selected based on previously documented utilized traction during bipedal locomotion. During walking, the utilized traction coefficient ranges between 0.15 and 0.3 (Redfern et al., 2001; Strandberg and Lanshammar, 1981). Utilized traction coefficients of 0.6 – 0.7 were observed during running (Valiant, 1993). When athletes performed maximum-effort 45°, 90° and 180° cutting, the peak traction coefficient utilized by the highest 95th percentile of the sampled population was approximately 1.2 (Shorten et al., 2003).

To identify appropriate materials for the targeted traction coefficients, different material samples were attached to the outsoles of otherwise identical mid-cut basketball shoes (Li Ning Yu Shuai IV) and tested on the laboratory floor. Thirty-six materials ranging from silk to latex were evaluated. Since footwear traction does not follow the classical Amontons-Coulomb laws of friction (Grönqvist et al., 2001), available traction properties of these materials were assessed under conditions simulating the loading experienced during the actual movements using an automated traction tester (Figure 3.1). The test shoe was mounted on a last and placed in 30° plantarflexion so the forefoot region was in contact with the ground. A vertical load of 600 N was applied directly above the metatarsal heads. These conditions were chosen based on a pilot study where kinematic and kinetic information of three subjects performing the testing movements were analyzed. A horizontal force was applied to the shoe by an electric actuator. A force platform (Kistler, Winterthur, Switzerland, model 9286A) located under the testing shoe was used to collect the GRF at 2400 Hz per channel (eight channels in total). High-speed video (240 Hz) was collected simultaneously in order to detect shoe motion. The traction
coefficient was calculated as the average ratio of the horizontal over the vertical GRF from the initiation of sliding motion to when the shoe reached its top speed of 30 mm s\(^{-1}\). This relatively low sliding speed was chosen to simulate the occurrence of small scale sliding and the calculated values represent the transition between static and dynamic traction coefficient when such sliding is initiated. The entire laboratory area, including the top of the force platform, shared a uniform floor covering. The consistency of the traction characteristics of this surface throughout the laboratory was confirmed at a multitude of locations with the on-board traction assessments provided by the traction tester (validated previously in Wannop et al., (2009)).

Figure 3.1: The traction tester used in the current study for assessing the available traction between different footwear outsole and the lab floor. The shoe orientation, contact region, and vertical load can be adjusted to represent realistic loading conditions.
Medical tape (Cantech® 988 Medsport Pro Adhesive Tape, Canadian Technical Tape Ltd., Montreal, Canada), Gorilla Tape™ (Gorilla Glue Company, Cincinnati, USA.), stiff rubber and soft rubber without outsole patterns were selected for the experiment based on their traction coefficient values, material uniformity and durability (designated as MT0.2, MT0.5, MT0.8 and MT1.1, respectively; Figure 3.2). The targeted 1.2 condition could be met with a regular herringbone sole pattern. However, although the traction of the flat rubber outsole was slightly less than the targeted value, it was selected to keep the outsole pattern consistent across conditions and to minimize the stick-slip phenomenon and its related traction non-linearity (Shorten and Xia, 2006). The available traction coefficients between the selected materials and the lab floor surface are summarized in Table 3.1.

Figure 3.2: Outsole materials used in the current study to provide different levels of available traction. From top left to bottom right are the MT0.2, MT0.5, MT0.8 and MT1.1 conditions.
<table>
<thead>
<tr>
<th>Name</th>
<th>MT0.2</th>
<th>MT0.5</th>
<th>MT0.8</th>
<th>MT1.1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Material</td>
<td>Medical tape</td>
<td>Gorilla Tape&lt;sup&gt;TM&lt;/sup&gt;</td>
<td>Stiff rubber</td>
<td>Flat soft rubber</td>
</tr>
<tr>
<td>Outcome from the mechanical assessments</td>
<td>0.26 ± 0.01</td>
<td>0.54 ± 0.01</td>
<td>0.82 ± 0.01</td>
<td>1.13 ± 0.01</td>
</tr>
</tbody>
</table>

Table 3.1: Description of the traction materials used in the current study and their available traction coefficients.

After each trial, the laboratory floor and footwear outsole were cleaned. Medical tape (MT0.2) and Gorilla Tape<sup>TM</sup> (MT0.5) were replaced after each testing session. In a pilot study with three subjects, the available traction properties of these four materials were evaluated before and after each testing session. In addition, the two rubber outsoles (MT0.8 and MT1.1) were assessed before and after the entire project. For both cases, no significant change in the traction coefficients across time was detected.

Differences in mass across shoe conditions (20 g maximum) were eliminated by gluing lead onto the heel counters of the lighter shoes. No influence of the attached traction materials on bending or torsional stiffness was observed.

3.2.2 Subjects

Thirty-two subjects (age 24.5 ± 5.4 years, body height 1.80 ± 0.05 m, body mass 77.4 ± 10.6 kg; mean ± 1 s.d.) volunteered for this study. The sample represented a heterogeneous population of athletes; the subjects were competing in local basketball, soccer, rugby, lacrosse, handball, baseball or track and field leagues. Written consent approved by the university ethics committee was obtained from the subjects prior to experimental testing.
3.2.3 Protocol

After a 20-min warm-up session, subjects performed maximum-effort curved sprints. The curved sprints were performed on a circle of 2.3 m radius (Figure 3.3) based on observations from a previous study. Larger traction is needed when the curve radius is small and/or the tangential speed is high (Chapter 2). In a study on curved sprint performance along circles of different radii (1 – 6 m), Chang and Kram (2007) commented that the largest utilized traction was observed when the subjects were sprinting along a curve of radius 2 m. Due to practical limitations on camera placement, a radius of 2.3 m was chosen. Subjects started the sprint at locations that allowed them to reach top speed when entering the motion analysis collection volume (width 1.5 m, length 3.5 m, height 2.3 m). Subjects remained sprinting at top speed until they passed another half loop after the collection volume.

Figure 3.3: Photograph of the experiment setup. The subjects performed maximum-effort curved sprints along a curve of 2.3 m radius (marked on the ground). Two pairs of timing lights were used to monitor fatigue.
Sufficient practice prior to testing was given to minimize learning effects. Seven trials were then performed in each traction condition. The condition order was randomized. Upon the completion of the fourth condition, three additional trials of the first condition were repeated in order to determine any learning or fatigue effects. Subjects were given 3-min rests between trials and 7-min rests between shoe conditions. Two pairs of timing gates (Figure 3.3) were placed in the collection volume in order to immediately detect any onset of fatigue. If a decrease in sprint speed larger than 10% was detected between two continuous trials, a longer rest period was prescribed.

3.2.4 Data acquisition

GRF data were sampled using an in-floor force platform (Kistler, Winterthur, Switzerland, model 9286A) at 2400 Hz per channel (eight channels in total). The starting location was adjusted so the subjects could naturally plant their outside foot on the platform. Trials with incomplete foot plant or unnatural execution were excluded.

Eight high-speed cameras (Motion Analysis Corporation, Santa Rosa CA, USA, model Eagle) were used for the kinematic assessment at 240 Hz. Spherical reflective markers (19 mm diameter) were placed at the sacrum, left and right anterior superior iliac spine (ASIS) and toe box of both the left and right shoe.

3.2.5 Data analysis

Data processing was conducted using customized MATLAB programs (The MathWorks, Inc., Natick, Massachusetts, USA). The raw kinetic and kinematic data were filtered with a 4th-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.

Location of the body centre of mass (COM) was determined using the sacrum and ASIS markers. It was calculated as the average 3D coordinates of the three markers and was used for the quantification of performance and body lean angle. Performance was quantified as the average speed for the COM to travel over a 1 m interval. This interval
distance was selected as it was large enough to measure meaningful performance differences but yet small enough to still represent a tangential displacement component on the curve.

Utilized traction during stance was quantified using the GRF data. The stance interval was determined using the vertical GRF at a 3% body weight threshold. Peak utilized traction and the weighted average utilized traction was determined. The weighted average utilized traction was calculated by using the magnitude of the centripetal GRF as the weighting factor, for its importance to curved sprinting speed.

Ground reaction impulse (GRI) and average GRF in the frontal plane were analyzed. Resultant GRF angle with respect to the ground was calculated. Body lean, defined as the orientation of the vector from centre of pressure to the COM, was calculated using force plate and kinematic data.

3.2.6 Statistical analysis

Initial Shapiro-Wilk tests indicated that the data were normally distributed. One-way repeated measures analysis of variance (ANOVA) tests were performed for comparisons across the traction conditions. When significant effects of the available traction were detected, a Scheffé's contrast post-hoc test was applied to further determine the differences between paired conditions. Statistical significance level was set a priori at $\alpha = 0.05$. 
3.3 Results

Four subjects were not able to perform consistently (statistically significant reduction in performance during repeated conditions) and were thus excluded from the analysis.

Available traction had a significant effect on curved sprint performance ($F_{(3, 81)} = 448.9, p<0.001$). The post-hoc tests indicated that performance improved as the available traction increased, up to the level of MT0.8; no significant performance benefits were detected in MT1.1 compared to MT0.8 (Figure 3.4).

Figure 3.4: Curved sprinting performance (average speed over a 1 m interval) across available traction conditions. Inequality symbols are used to express statistically significant differences detected using Scheffé's contrast post-hoc tests.
Average centripetal GRF (GRF\textsubscript{cpt}) increased significantly from MT0.2 to MT0.5 to MT0.8 then remained constant between MT0.8 and MT1.1 (Table 3.2, Figure 3.5(a)). Centripetal GRI (GRI\textsubscript{cpt}) showed increases in the same manner, while stance time decreased from MT0.2 to MT0.5 to MT0.8 (Table 3.2). Peak utilized traction (µ\textsubscript{utilized}) differed across conditions (Table 3.2, Figure 3.5 (b)). The weighted average utilized traction increased from MT0.2 to MT0.5 to MT0.8 then remained unchanged in MT1.1 (Table 3.2).

<table>
<thead>
<tr>
<th></th>
<th>MT0.2</th>
<th>MT0.5</th>
<th>MT0.8</th>
<th>MT1.1</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Peak µ\textsubscript{utilized}</strong></td>
<td>0.27 ± 0.03 &lt; 0.61 ± 0.06 &lt; 0.90 ± 0.13 &lt; 1.21 ± 0.23</td>
<td></td>
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</tr>
<tr>
<td><strong>Average µ\textsubscript{utilized}</strong></td>
<td>0.23 ± 0.02 &lt; 0.48 ± 0.07 &lt; 0.61 ± 0.09 = 0.64 ± 0.10</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Average GRF\textsubscript{cpt} [BW]</strong></td>
<td>0.25 ± 0.04 &lt; 0.60 ± 0.11 &lt; 0.76 ± 0.13 = 0.80 ± 0.15</td>
<td></td>
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<tr>
<td><strong>Average GRF\textsubscript{vert} [BW]</strong></td>
<td>1.24 ± 0.12 &lt; 1.35 ± 0.14 = 1.39 ± 0.14 = 1.39 ± 0.15</td>
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<tr>
<td><strong>GRI\textsubscript{cpt} [BW s]</strong></td>
<td>0.07 ± 0.01 &lt; 0.15 ± 0.02 &lt; 0.18 ± 0.03 = 0.18 ± 0.03</td>
<td></td>
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<tr>
<td><strong>Stance time [s]</strong></td>
<td>0.30 ± 0.03 &gt; 0.25 ± 0.02 &gt; 0.23 ± 0.02 = 0.23 ± 0.02</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

| µ\textsubscript{utilized}: utilized traction | GRF\textsubscript{cpt}: centripetal GRF | GRF\textsubscript{vert}: vertical GRF |
| GRI\textsubscript{cpt}: centripetal GRI |

Table 3.2: Summary of the kinetic data and stance time (mean ± 1 s.d.). GRF and GRI were normalized to body weight. Inequality symbols are used to indicate statistically significant differences detected using Scheffé’s contrast post-hoc tests.
Figure 3.5: (a) Centripetal GRF (sample mean) versus normalized stance time; (b) utilized traction (sample mean) over normalized stance time.
The subjects leaned more into the ground as the available traction increased from MT0.2 (75.5 ± 1.8°) to MT0.5 (62.4 ± 1.0°) to MT0.8 (57.6 ± 2.2°), then remained unchanged for MT1.1 (56.4 ± 2.2°) (Figure 3.6). Average GRF angle decreased from 77.2 ± 1.4° to 63.9 ± 1.8° to 58.2 ± 2.0° as the available traction increased from MT0.2 to MT0.5 to MT0.8, then remained constant for MT1.1 (56.1 ± 4.6°).

Figure 3.6: As the available traction increased, subjects leaned closer to the ground (lines) from 75.5° to 56.4°. The body orientation was closely coupled with the average force orientation (arrows).
3.4 Discussion

The purpose of this study was to investigate the influence of available traction on the maximum-effort curved sprinting performance. Increases in available traction from MT0.2 to MT0.5 to MT0.8 provided performance advantages. However, no further performance enhancements were detected when the available traction increased beyond MT0.8.

3.4.1 Available traction and curved sprint performance

When sprinting along a curve, the locomotion system needs to produce force to support the body COM and create centripetal acceleration. More traction needs to be utilized as the curve radius decreases and/or the running speed increases (Equation 3.1). In the current work, the running radius was prescribed and the effects of the available traction on top curve sprinting speed were investigated.

As the available traction increased from MT0.2 to MT0.5 to MT0.8, performance improved. Greater amounts of traction (both peak and average) were utilized as the available traction increased. Leaning more toward the centre of the circle, the subjects showed significantly larger average centripetal GRFs with small (MT0.2 vs. MT0.5) or no (MT0.5 vs. MT0.8) increases in the average vertical GRF. Increases in the average centripetal GRF were accompanied with decreases in stance time. However the centripetal GRI still increased significantly as the available traction increased from MT0.2 to MT0.5 to MT0.8. Increased GRI and shortened stance time are both beneficial from a performance perspective (Mero et al., 1992).

Further increases in the available traction from MT0.8 to MT1.1 no longer enhanced performance. While the peak utilized traction differed across these conditions, the average utilized traction, body lean angle and centripetal GRI remained constant. Based on the data from the MT0.8 and MT1.1 conditions, at average utilized traction values of approximately 0.6, it seemed that factors other than traction began to limit the subjects’ ability to lean into the ground and create larger centripetal GRF, and thus further performance improvement. Interestingly, this value coincides with the previously
assumed available traction for proposing traction as the limiting factor for curved sprinting speed along small circles (Alexander, 2002).

3.4.2 Peak utilized traction – implications on determining required traction for performance footwear

Available traction provides an upper boundary within which utilized traction can be generated without large scale skidding. Peak utilized traction generally occurred at very early and late stance. Results from this study indicated that the subjects were able to utilize the full range of available traction values available at these phases. On the other hand, the current findings showed that utilization of these peak traction values was not necessarily related to final performance. To illustrate this point, Figure 3.7 shows the average time history of the accumulation of centripetal GRI along with the utilized traction coefficient during curved sprinting in MT1.1 for all subjects. It can be seen that the relative contribution to the total impulse generation was minimal during the first and last 5% of stance when peak utilized traction occurred.

![Graph showing utilized traction and cumulative centripetral GRI](image)

**Figure 3.7:** Utilized traction and cumulative centripetal GRI as a function of normalized stance time during curved sprinting in MT1.1. Sample population means were used for the calculation.
This observation raises a concern regarding the current method of evaluating ‘required’ traction. In both ergonomic and athletic footwear research fields, the required available traction is generally assessed by having subjects perform target movements in a non-slipping shoe/surface condition (Valiant, 1993). The average (or in some cases the highest 95th percentile of the sample population (Shorten et al., 2003)) peak utilized traction was quantified and used as the target traction value. The current results indicated that findings based on such an approach may overestimate the peak traction value necessary for uncompromised performance. Without a complete understanding of injury mechanisms associated with high traction conditions, the danger of such overestimation remains unclear. However, based on the current understanding of the relationship between traction and joint loading (Wannop et al., 2010), a lower yet sufficient traction may be preferred.

3.4.3 Study limitations

This experiment was designed with the intent to examine the effects of available traction on performance in non-contact situations. The conclusions on performance should be interpreted within this context. When designing footwear for sports such as American football where body contact is frequent and important in determining the final performance outcome, it is expected that higher available traction could lead to further performance improvement. Future investigations are needed for such scenarios.

Four conditions of available traction were tested in the current study. This was the maximum number of conditions that the subjects could perform without observable effects of fatigue. This on the other hand limited the resolution of the independent variable. Results from the current work showed that MT0.8 is sufficient for optimal performance for curved sprinting. Due to the lack of additional intervals between MT0.5 and MT0.8, this result may be biased toward the higher end of what is truly required. To try to address this concern, the total impulse generated at and below different traction thresholds from 0.1 to 1.3 was calculated using data from the MT1.1 condition. In Figure 3.8, it can be seen that 97% of the impulse was generated when the utilized traction was
0.7 while 99.7% was generated when the utilized traction was 0.8. Only 0.3% of total impulse was generated beyond a utilized traction coefficient of 0.8.

Figure 3.8: Total centripetal GRI generated at and below different utilized traction thresholds. Sample population means from the MT1.1 conditions were used for the calculation.

Finally, the available traction values identified are based on the specific testing methodology. A critical available traction value of 0.82 is appropriate provided similar available traction testing methodology is used. However, critical absolute traction values could change with different mechanical test conditions (Beschorner et al., 2007; Schlaepfer et al., 1983; Valiant, 1987). While the mechanical test conditions were chosen to represent the majority of the loading conditions experienced during stance for the tested movements, the available traction could have changed during extreme conditions such as the very early and very late stance. This explains the observation that the peak utilized traction that occurred during these periods exceeded the available traction measured.
3.4.4 Beyond MT0.8, what might limit curved sprint performance?

Available traction has been suggested as a key performance constraint for curved sprinting along small radii (Alexander, 2002). Results from the current study showed that available traction below certain thresholds can indeed limit performance. To illustrate this point, experimentally measured curved sprinting speeds across different available traction conditions were graphed against a theoretical traction-imposed performance constraint derived from Equation 2.11 in Chapter 2 (Figure 3.9). For the calculation of the theoretical constraint, the actual circle radius (2.3 m) was used to define $R$, and the average limb length (hip to ground distance during quiet standing) of the sampled population (0.9 m) was used to define $L$. It can be seen that at low available traction, the theoretical performance constraint agreed well with the actual performance measures.

![Figure 3.9: The Maximum curved sprinting speeds at different available traction conditions. The diamonds represents the speed measured experimentally in this study and the associated bars indicate ± 1 s.d.. The dotted line represents a theoretical performance constraint imposed by available traction. This constraint was calculated based on Equation 2.11 with experimentally assessed input parameters.](image)
As available traction increases, the experimentally measured speed gradually diverged from the theoretically derived performance constraint imposed by traction. This divergence indicated that available traction was no longer the predominant limiting factor. An additional observation worth noting is the increasing size of the standard deviation of the measured speed as available traction increases. The size of the standard deviation represents the variance of the measured speed for the sampled population. At low available traction, the tight range of measured speed indicated that, despite the heterogeneous nature of the sampled population, the subjects’ performance was likely constrained by the same factor – traction. In the higher available traction conditions, the performance may become constrained by other more individualized factors, such as strength. The increased variance in performance is likely a reflection of the individual differences in such factors.

The maximum limb force has been suggested as a limiting factor for curved sprinting performance (Greene, 1985; Usherwood and Wilson, 2006). As the curved sprinting speed increases, the need to generate ground force (GF) increases exponentially (Chapter 2). Results from the current study showed that, up to MT0.8 higher available traction allows a greater exertion of such limb force. Beyond such traction level, it is possible that the limb force generation reaches its limit, and can no longer supply the exponential increase in the centripetal GRF needed for a higher speed. An alternative theory suggested that the non-sagittal plane joint stabilizing moment may constrain curved sprinting performance (Chang and Kram, 2007) along curves of small radii. During maximum-effort cutting, the ankle and knee joint experience large non-sagittal plane joint moments (Wannop et al., 2010), and excessive non-sagittal plane joint moment and angular impulse have been associated with both running and lateral sports injuries (Hewett et al., 2005; Kristianslund et al., 2011; Stefanyshyn et al., 2006). Chang and Kram (2007) proposed that during top speed curve sprinting, the moment experienced by the lower extremity joints in the frontal and transverse plane may have reached critical operation limits. Such critical loading conditions could be the factor limiting further force generation. These theories will be examined in the subsequent Chapters.
3.5 Conclusion

Available traction has significant effects on maximum-effort curved sprinting performance, but only to an extent, after which factors other than traction seemed to limit performance. Further increases in the available traction beyond such point were not reflected in the average utilized traction or overall impulse generation, and caused no further performance improvement.
4.1 Introduction

The ability to negotiate curves at high speed is crucial for the performance of a locomotion system. During a prey-and-predator encounter in a natural environment, such ability can dictate survival (Howland, 1974). Executions of quick and sharp turns are also of high frequency and importance in numerous athletic settings. In a basketball game, for example, players spent more than 40% of the game changing directions (Stacoff et al., 1993). Despite its importance, the number of studies on the limiting factors for curved sprinting performance is rather limited.

When sprinting along a curve, a ground reaction force (GRF) is needed in order to withstand gravity and create a centripetal acceleration for the body centre of mass (COM). During level-ground locomotion, the magnitude of the vertical GRF, 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 centripetal GRF of magnitude \( m \cdot \frac{v^2}{r} \) is required to continuously accelerate the body COM toward the centre of the circle. It can be seen that as speed increases, the need to supply GRF increases exponentially.

The maximum amount of ground force (GF) that the supporting limb can generate has been suggested as a key factor in limiting human sprinting performance (Weyand et al., 2000). By treating the maximum limb force generation as a constant limit, Greene (1985) formulated a mathematical model to examine the maximum attainable curved sprinting performance, 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 ground to create a centripetal GRF. As the magnitude of the resultant GRF was treated as a constant (the limb force limit), the redirection of the GRF vector associated with body
lean would compromise the generation of vertical GF. To maintain sufficient vertical
ground reaction impulse (GRI) 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 reached 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 10m), however,
performance predictions based on this model became less satisfactory (Alexander, 2002).

An examination of this limb force limit theory by directly assessing the GRF was
not conducted until recently (Chang and Kram, 2007). In that investigation, GRFs during
maximum-effort sprints along circles of various radii (1, 2, 3, 4, and 6 m) were compared
to the peak GRF generated during top-speed linear sprinting (which was used to define
the limb force limit). The authors hypothesized that, based on Greene (1985), if the
maximum GF generation was the limiting factor, the peak resultant GRF should remain
constant across different curvatures. The results revealed statistically significant
differences in peak GRF for the inside limb between straight sprint and curved sprints on
radii of 1 and 2 m. Meanwhile, a systematic trend of decreases in the peak resultant GRF
as the curve radius decreases was apparent for both limbs (Chang and Kram, 2007;
Figure 2.12). 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 limits curved sprinting
speed along small circles. The authors suggested that despite the ample limb force
available, the generation of this force may be inhibited as other factors reached the
operating limits.

Chang and Kram (2007) proposed that the constraints to the inside leg’s ability to
generate GF might be related to the need of joint stabilization. Inverse dynamic analyses
have shown that the lower extremity 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,
especially for the knee joint (Mizuno et al., 2009; Seering et al., 1980). Excessive non-
sagittal plane joint moment and angular impulse have been associated with both running and lateral sports injuries (Hewett et al., 2005; Kristianslund et al., 2011; Stefanyshyn et al., 2006). It is possible that during maximum-effort curved sprinting, the non-sagittal plane joint stabilizing moments have reached their operating safety threshold and prohibit further limb force generation by the muscle tendon units (MTUs) surrounding the joint. Currently, studies directly testing hypotheses based on this theory 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 at the subjects’ COM. The purpose of this perturbation was two-fold.

First, it allows a further examination of the limb force limit theory. While peak forces generated during curved sprints along small circles were observed to be smaller than the peak GRF during top-speed linear sprinting (Chang and Kram, 2007), the observation itself may not be sufficient in concluding the limb is not generating its maximum force for the specific configuration experienced. It is possible that the limb force generation is at its limit during the movements but the magnitude of this limit is reduced compared to 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 perturbation, the peak limb force will remain constant.

Second, the perturbation allows the investigation of the joint stabilization limit theory. Based on the joint stabilization limit theory, during maximum-effort curved 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 the changes in the GRF, these joint moments should remain operating at such limits in order to perform maximally. The perturbation was introduced to potentially alter the GRF and test this hypothesis.
4.2 Materials and methods

4.2.1 Setup and equipments

Sprinting trials were performed along a circle of 2.5 m radius in a laboratory (Figure 4.1). This radius was chosen based on the observations reported by Chang and Kram (2007) that statistically significant reductions in the force generation by the inside leg began to occur when circle radius decreased below 3 m. Subjects started the sprint at a location that allowed them to reach top speed when entering the 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. GRF data were sampled using an in-ground force platform (Kistler, Winterthur, Switzerland, model Z4852C) at 2400 Hz per channel (eight channels in total). Only trials where the subjects could plant 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 reflective markers at 240 Hz. Twelve markers were placed on each subject to represent the pelvis, left thigh, shank, and foot segment. Hip, knee and ankle joint centres were determined using additional markers during neutral standing trials prior to the movement trials.

Figure 4.1: Photograph of the experimental setup. Subjects performed maximum-effort curved sprints along a curve of 2.5 m radius.
To prevent any confounding effects caused by differences in shoe-ground available traction, all subjects used the same pair of athletic shoes (Li Ning Company Limited, Beijing, China, model Yushuai IV; Figure 4.2) providing sufficiently high available traction on the laboratory floor. Mechanical traction tests of the current shoe-floor interface were conducted using a previously validated protocol (Chapter 3), and the available traction quantified under this protocol was a traction coefficient of 1.13. It has been shown that available traction beyond a traction coefficient of 0.82 provided no performance benefits for maximum-effort sprinting along a curve of 2.3 m radius (Chapter 3). The footwear outsole was cleaned between trials and the laboratory floor was cleaned between conditions.

Figure 4.2: Photograph of the experimental footwear. The experimental footwear consists of a specially constructed outsole to provide uniform traction properties across the contact area, and a mid-foot lateral support to minimize the in-shoe movements of the foot.
A commercially available life jacket (Mountain Equipment Co-op, Vancouver, Canada, model Fulcrum PFD; Figure 4.1 and 4.3) was modified in order to securely place additional mass on the subjects without causing drastic changes in the body moment of inertia. Six pieces of diving lead were glued in the cut-outs made surrounding the bottom region of the jacket close to the body COM. 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.

Figure 4.3: Photograph of the modified life jacket (total mass = 12.4 kg) worn by a subject. Six pieces of diving lead were placed surrounding the bottom region of the jacket.
4.2.2 Subjects

Thirteen male subjects were recruited for the current study (age 22 ± 2 years, mass 75.4 ± 5.5 kg, height 177.5 ± 5.5 cm; mean ± 1 s.d.). 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.

4.2.3 Protocol

After a 20-min warm-up session, subjects performed maximum-effort curved sprints in the control and additional mass conditions. Subjects started by sprinting in the control condition for four trials, then, eight trials with the weighted life jacket 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 (α = 0.05) were used to compare the control sprinting speed (quantified over stance using the pelvis markers) 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.

4.2.4 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 (Chapter 3).

Peak and average GRFs over stance were determined. The stance interval was defined using the vertical GRF at a 3% body weight threshold. In addition, the average frontal plane GRF angle was calculated. To ensure available traction was not a limiting factor in the current study, the maximum traction utilized by the subjects, calculated as
the resultant horizontal GRF divided by the vertical GRF, were quantified and compared with the available traction provided by the shoe-ground interface.

Stance time and curved sprinting speed were quantified. The curved sprinting speed was defined as the average COM speed over stance. The location of the COM was estimated by averaging the coordinates of the three markers placed on the pelvis segment.

Joint moments at the ankle and knee of the inside leg were calculated with a conventional inverse dynamics approach (Andrews, 1995). The segment inertial properties for individual subjects were estimated based on Winter (2005). The centre of pressure (COP) location, GRFs, and vertical free moment were calculated using force plate data. The GRF and vertical free moment were applied to the foot segment at the COP location for the “bottom-up” joint moment calculation. The ankle joint moments were expressed in the foot segment coordinate system (SCS). The foot SCS is defined as follows. During a neutral quiet standing trial, the subject’s foot was placed so that its long axis was as closely aligned with the anterior-posterior axis of the lab coordinate system (LCS) as possible. In the process of constructing the SCS, a coordinate system parallel to the LCS was first embedded at the ankle joint origin - defined as the middle point between the markers placed at the medial and lateral malleoli. The plantar-/dorsiflexion axis of the SCS was then aligned with the vector connecting the markers placed at the medial and lateral malleoli. Knee joint moments were expressed in the shank SCS. The process of the shank SCS construction is similar to that of the foot SCS. The knee joint centre was defined as the middle point between the markers placed at the medial and lateral epicondyles of the femur. The shank SCS was aligned so that the long (internal/external rotation) axis was aligned to the vector connecting the ankle and knee joint centres. The vector sums of the frontal and transverse plane moments were calculated to represent the demand for joint stabilization, and they were denoted as the non-sagittal plane moments. The vector sums of the moments in all three planes were calculated to allow a further investigation of the overall moment generation ability at each joint.
4.2.5 Statistical analysis

One-tailed paired t-tests were used to compare results across testing conditions. One-tailed analyses were used due to the nature of the intervention, where increases in the GRF 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 \textit{a priori} at $\alpha = 0.05$.

4.3 Results

Mostly due to the changes in the vertical and centripetal GRF (Figure 4.4), the peak resultant GRF increased significantly in the additional mass condition compared to the control condition (Table 4.1).

![Figure 4.4: GRF in the vertical (GRF$_{\text{vert}}$), centripetal (GRF$_{\text{cpt}}$) and anterior-posterior (GRF$_{\text{a-p}}$) directions. Sample means (thick lines) and standard deviations (thin lines) plotted against the normalized stance time.](image-url)
<table>
<thead>
<tr>
<th></th>
<th>Control</th>
<th>Additional Mass</th>
<th>One-tailed paired t-test</th>
</tr>
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<tbody>
<tr>
<td><strong>Peak GRF&lt;sub&gt;resultant&lt;/sub&gt; [N]</strong></td>
<td>1730.1 ± 273.1</td>
<td>1919.1 ± 292.0</td>
<td>p = 0.0002</td>
</tr>
<tr>
<td><strong>Peak GRF&lt;sub&gt; cpt&lt;/sub&gt; [N]</strong></td>
<td>1091.0 ± 240.4</td>
<td>1215.6 ± 218.9</td>
<td>p = 0.0007</td>
</tr>
<tr>
<td><strong>Peak GRF&lt;sub&gt; vert&lt;/sub&gt; [N]</strong></td>
<td>1369.6 ± 169.4</td>
<td>1528.0 ± 207.5</td>
<td>p = 0.0003</td>
</tr>
<tr>
<td><strong>Average GRF&lt;sub&gt; cpt&lt;/sub&gt; [N]</strong></td>
<td>583.0 ± 120.3</td>
<td>640.4 ± 129.6</td>
<td>p = 0.0003</td>
</tr>
<tr>
<td><strong>Average GRF&lt;sub&gt; vert&lt;/sub&gt; [N]</strong></td>
<td>797.0 ± 96.5</td>
<td>894.8 ± 104.3</td>
<td>p &lt; 0.0001</td>
</tr>
<tr>
<td><strong>Stance time [ms]</strong></td>
<td>236.5 ± 11.3</td>
<td>254.5 ± 14.7</td>
<td>p &lt; 0.0001</td>
</tr>
<tr>
<td><strong>COM speed [m s&lt;sup&gt;-1&lt;/sup&gt;]</strong></td>
<td>3.6 ± 0.2</td>
<td>3.5 ± 0.2</td>
<td>p = 0.0302</td>
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**GRF<sub>resultant</sub>:** 3D resultant GRF  
**GRF<sub> cpt</sub>:** centripetal GRF  
**GRF<sub> vert</sub>:** vertical GRF

Table 4.1: Peak and average GRF generated by the inside limb, stance time and average COM speed over stance (mean ± 1 s.d.).
In the frontal plane, a larger resultant GRF was observed through the first half of stance in the additional mass condition compared to the control (Figure 4.5 (a)). When averaged over stance, the relative increase of the centripetal GRF (9.9%) approximates the relative increase of the vertical GRF (12.3%) – Table 4.1. As a result, the average orientation of the frontal plane GRF with respect to the ground changed by less than 1 degree (control: $53.5^\circ \pm 2.7^\circ$ versus additional mass: $54.3^\circ \pm 3.2^\circ$; $p = 0.0201$; Figure 4.5 (b)). The peak utilized traction in both conditions (control: 1.02 and additional mass: 0.97) were below the available traction provided (1.13), confirming that available traction was not hindering the maximal execution of the movement.

Figure 4.5: (a) Resultant frontal-plane ground reaction force generated during stance. Thick lines represent the average values across all the subjects and thin lines indicate $\pm 1$ s.d.; (b) average frontal-plane GRF generated over stance.
Compared to the control condition, when sprinting in the additional mass condition, the peak ankle non-sagittal plane moment increased significantly (p = 0.0012; Figure 4.6 (a) and (b)), while the peak plantarflexion moment remained unchanged (p = 0.0584; Figure 4.6 (b) and Figure 4.8). The differences observed in the non-sagittal plane ankle moment were largely associated with the changes in the ankle inversion moment (Figure 4.8).

**Figure 4.6:** Non-sagittal plane ankle moment experienced at the inside limb (a). Thick lines represent the average values across all the subjects and thin lines indicate ± 1 s.d.. Peak sagittal and non-sagittal plane ankle moments are plotted in (b) with the s.d. bars. *Statistically significant difference.
For the knee joint, both peak non-sagittal (p = 0.0158) and sagittal (p = 0.0104) plane moments increased significantly as the mass of the body increased from the control to the additional mass condition (Figure 4.7 and Figure 4.8). The differences in the peak non-sagittal plane moment can largely be attributed to the changes in the knee abduction moment (Figure 4.8).

Figure 4.7: Non-sagittal plane knee moment experienced by the inside limb (a). Thick lines represent the average values across all the subjects and thin lines indicate ± 1 s.d.. Peak sagittal and non-sagittal plane knee moments are plotted in (b) with the s.d. bars. *Statistically significant difference.
Figure 4.8: Joint moments at the ankle and knee joints. Thick lines represent the average values across all the subjects and thin lines indicate ± 1 s.d..
The peak total moment at the ankle joint did not differ across conditions (control: 227.4 ± 45.0 Nm versus additional mass: 233.0 ± 44.0 Nm; p = 0.124). Indeed, the total joint moment generation at the ankle remained unchanged between conditions through the entire stance (Figure 4.9 (a)). Peak total knee joint moment increased from 170.3 ± 28.6 Nm in the control condition to 193.4 ± 32.1 Nm in the additional mass condition (p = 0.022). The difference in total knee joint moment occurred mostly during the first half of the stance (Figure 4.9 (b)).

![Figure 4.9: Total moment generated at the ankle (a) and knee (b) joint of the inside limb. The total moment was calculated as the vector sum of joint moment across all three anatomical planes. Thick lines represent the average values across all the subjects and thin lines indicate ± 1 s.d.](image)
4.4 Discussion

By implementing an external perturbation to the body – an additional body mass of 12.4 kg, the study aimed to examine, during maximum-effort curved sprinting, whether:

1) the inside leg reached its limit in generating limb force;
2) the non-sagittal plane joint stabilizing moments experienced at the ankle and knee were at the operating limits.

4.4.1 Limb force generation

The peak limb force generation has been proposed as a factor limiting curved sprinting performance (Greene, 1985; McMahon and Greene, 1979; Usherwood and Wilson, 2006). A recent study (Chang and Kram, 2007) investigating the GRFs during curved and straight sprinting found that, during sprints along curves of small radii the peak GRF was smaller than during top-speed straight sprints. Since this study, and the studies before (Greene, 1985; Usherwood and Wilson, 2006), treated the peak GRF during straight sprints as the limb force limit, the discrepancy observed between the peak GRFs during curved and straight sprints tends to suggest that the supporting limb was not at its limit in generating force. An alternative interpretation of the results can be that the reduced peak GF generation during curved sprinting still represents the limb’s limit in generating force; however, the magnitude of this limit is reduced due to factors such as changes in body configuration. By implementing an additional mass, the current study first investigated whether the limb can generate additional GF. The additional mass condition essentially represents an additional need for GRI 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 impulse (GI) increases in the additional mass condition, it would be expected that the peak resultant GRF remains unchanged however the frontal plane GRF would be directed more vertically, and/or, the stance time will be extended.

It was found that when the subjects performed the curved sprints with the additional mass, the peak resultant GRF increased significantly (10.9%). Despite an
extended stance time, the average frontal plane GRF over stance was 11.4% larger with the additional mass compared to in the control condition. The increase in the frontal plane GRF was a result of increases in both the vertical and centripetal GRF; over stance, the resultant force orientation only became slightly (less than 1°) more vertical. These observations supported Chang and Kram’s (2007) statement that during maximum-effort curved sprinting, the inside supporting limb seems to possess additional ability for force generation.

4.4.2 Non-sagittal plane joint stabilizing moment

The second purpose of the study was to examine the theory proposed by Chang and Kram (2007) that during maximum-effort curved sprinting, the non-sagittal plane stabilizing moment at the lower extremity joints of the inside leg might have reached their physiological operating limits, and thus constrained the overall performance of the locomotion system. If these non-sagittal joint stabilizers were indeed operating at their limits, with the increase in the external force acting upon the system it would be expected that the non-sagittal plane joint moments at the inside leg ankle and/or knee joints would remain under such thresholds. In the presence of a larger GRF, this can potentially be achieved by adapting the segmental kinematics to realign the GRF vector closer to joint centres to reduce the lever arms (Biewener, 1989).

In the current study, as the GRF increased from the control to 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). These results contradicted the predictions made based on the theory that the joint stabilizing moments experienced at the ankle and/or knee joint were operating at their limits during maximum-effort curved sprinting. It indicated that the joint stabilizers were indeed able to endure external loading larger than the amount experienced under normal control conditions. While increases in peak non-sagittal plane moments were detected across joints, extension moments responded to the GRF change in a rather different manner. As GRF rose, the peak knee extension moment increased by 56.8%. Meanwhile, no changes in the ankle plantarflexion moment were observed. Since plantarflexors are the primary
ankle moment generators, this observation may indicate that the overall ability of MTUs surrounding the ankle joint to generate moment have reached a limit.

### 4.4.3 Overall moment at the ankle joint

During dynamic analyses of human movements, it is conventional to separate the moment experienced at a joint into three orthogonal planes. Such anatomical definitions facilitate the description of joint kinematics and kinetics. They, however, may limit the interpretation of the overall functioning mechanism of the system. This is especially the case when the joint is in a complex configuration. During curved sprinting, the ankle joint is placed in such a situation and likely the function of the MTUs’ differs from when the joint is in a neutral position. As the subtalar joint becomes everted or inverted, for example, the frontal-plane lever arm of the Achilles tendon changes, and as a result the function of the triceps surae changes. In vitro experiments have shown that, in addition to generating sagittal plane moment, the triceps surae indeed function as inverters during eversion and evertors during ankle inversion (Klein et al., 1996; Zifchock and Piazza, 2004).

By treating the moment generated at a joint as an entity, one could potentially gain insight in the overall functional properties of the synergistic MTUs. In the current study, the vector sums of the joint moments at the ankle and knee were calculated. These variables provided an indication of the MTUs’ overall load tolerance under the specific operating scenario experienced. Despite the 11.4% increases in the external GRF over stance, the total moment generated at the ankle joint remained unchanged. Based on this finding, it is possible that the ankle joint MTUs’ overall ability to withstand external load had reached the limit. While information on the moment experienced at the ankle joint during maximum-effort curved sprinting is scarce in the literature, results from linear sprint running seem to support this interpretation. In a study by Kuitunen et al. (2002), 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 GRF experienced at higher speeds, peak plantarflexion moment remained constant, indicating the plantarflexors may have reached their operating limit. While it is difficult to make a
direct numerical comparison between the two studies, due to the differences in sampled population, movement task and joint definition, the peak ankle moment (an approximate average of 250 Nm plantarflexion moment estimated based on the graph) reported by Kuitunen et al. (2002) is in line with the present results (an average of 230 Nm of total moment across conditions).

One reason why the ankle joint MTUs would operate at their limits may lie in their critical role in generating GF. The contribution of net moments generated at a specific joint to the resultant GF has been investigated previously with an induced acceleration analysis (Kaya et al., 2006; Kepple et al., 1997; Pandy, 2003). By representing the linked segments in a generalized coordinate system, the induced acceleration analysis allows the investigator to input joint moments individually and examine the resultant acceleration of all the segments. While such analyses on curved sprinting have not been reported in the literature, investigations on fast walking (Liu et al., 2008) and running (Hamner et al., 2010) suggested that during mid- and late stance (when peak GRF occurs) soleus and gastrocnemius were the primary contributors to both propelling and supporting the body COM. If the ankle joint MTUs were also the main contributors to the GF generation in curved 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 GF generation during curved sprinting is, however, needed to further elaborate on this speculation.

If the ability to generate ankle moment is among the predominant factors limiting the top curved sprinting performance, by changing such moment generation, performance changes should be observed. Future studies where experimental implementations altering the ankle moment generation may help further reveal its importance in curved sprinting performance.

4.4.4 The reserved moment generation capacity at knee

While the total moment generated around the ankle joint remained constant, the total knee joint moment increased significantly from the control to the additional mass condition. In both sagittal and non-sagittal planes, the MTUs surrounding the knee joint
were able to generate larger external moments as the GRF increased. This finding is rather puzzling - why would the subjects not fully utilize the moment generation potential at the knee joint when sprinting without the additional mass?

In order to address this question, it is important to first examine the salient determinants of the overall performance of a legged locomotion system. Sprinting performance, defined as maximum COM travelling speed, is the product of step length and frequency. Step length is the sum of the COM travel during stance and flight, and step frequency is determined by the stance and flight time. When running in a linear direction, high speed was achieved by increasing both step length and frequency (Weyand et al., 2010). During curved sprinting, however, maximizing step length may become less favourable, if not counterproductive, for the overall performance. Firstly, since the task requires the subjects to finish 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 finish a circle (increasing step length), the redirection of the COM travel becomes more acute, and a more acute COM redirection has been associated with a greater loss of momentum (Bertram and Gutmann, 2009). For these two reasons it is plausible that during curved sprinting, maximizing step length could become rather costly. For a circle of given radius, there may exist an optimal step length beyond which further increases will lead to no performance improvements. Based on this speculation, it would be expected that, when sprinting in circles, subjects would achieve maximum overall performance by constantly maximizing step frequency but adjusting step length based on the circle radius. Step length and frequency data reported by (Chang and Kram, 2007) support this prediction. In their study, as the circle radius decreased from infinity (straight sprint) to 6, 4, 3, 2, and 1 m, sprinting with full effort subjects decreased step length while maintaining a constant step frequency. While the performance optimization strategy warrants future research, the association between knee joint moments and step length and frequency may help gain insight into why moment generation ability was reserved in the control condition.
The knee joint is generally viewed as a hinge joint, and in the frontal plane its function can be seen as a linear actuator along the long axis of the leg. In the case of curved sprinting, where a substantial body lean is present, increases in knee extension moment were associated with increases in both the centripetal and vertical components of GRF (as observed in the current study). While an increase in the centripetal GRF is desired to redirect the body COM at higher travelling speed, a linear increase in the vertical component of the GRF may yield negative consequences for maximizing speed. Over a constant stance time, increases in vertical GRF will lead to a larger vertical GRI. For a given body mass, the increase in vertical GRI will increase the net vertical impulse (calculated as the GRI – gravitational impulse) experienced by the body COM, which will in turn result in a longer flight time. The overall step time will thus be extended, in other words, step frequency will be decreased. In addition, the extended flight time can result in a longer flight travel which can in turn increase step length. Based on the potential performance optimization strategy discussed in the previous paragraph, these changes may yield a negative impact on the overall performance.

If, for a given traveling speed, a particular net vertical impulse existed enabling the system to operate within the proposed optimal step length and frequency range, the net vertical impulse would remain constant as the mass of the body changed. In the current experiment, as the mass increased, the subjects generated a larger GF. A post-hoc analysis of the force data revealed that the net vertical impulse experienced by the body COM over stance was not different across conditions (control: 13.25 ± 14.04 Ns vs. additional mass: 8.06 ± 15.56 Ns, p = 0.79). Unfortunately step length and frequency information are not available for further elaboration. Future research is needed to investigate these speculations.
4.5 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 theories. The current observations did not support the hypotheses formed based on either theory. During maximum-effort curved sprinting: the limb reserves additional force generation ability; the ankle and knee joint non-sagittal plane stabilizing moments were not at their operating limits.

In addition, it was found the total moment generated at the ankle joint remained constant despite significant increases in the GRF indicating that during curved sprinting, the ankle joint MTUs may operate at their limits. Future attempts to increase the ankle joint’s ability to generate moment may induce performance benefits and facilitate the examination of this ankle moment limit theory.
CHAPTER FIVE:  
DECOMPOSITION OF THE GROUND FORCE GENERATION DURING CURVED SPRINTING

5.1 Introduction

The ability to generate ground force (GF) can have a direct influence on curved sprinting performance (Chang and Kram, 2007; Usherwood and Wilson, 2005). In order for the body centre of mass (COM) to travel along a curved path, a centripetal acceleration is needed, and this is achieved by creating a centripetal ground reaction force (GRF) toward the centre of the curve. For a given radius, as curved sprinting speed increases, the need for the centripetal GRF increases exponentially. Thus factors associated with such force generation can potentially limit the ultimate curved sprinting speed.

In a previous experiment (Chapter 4) exploring the limiting factors for the GF generation, it was found that the ankle joint’s ability to generate moment remained unchanged despite an 11.4% perturbation in the GRF. Based on this observation it was proposed that for the specific joint configuration experienced, the muscle tendon units (MTUs) surrounding the ankle joint may be operating at their limit in generating joint moment. One potential reason for the body to maximize out the ankle moment generation may be associated with its contribution to the generation of overall GF.

Due to dynamic coupling, the moment generated at one joint can accelerate all the body segments in the linked-segment system (Zajac et al., 2002; Zajac and Gordon, 1989). According to Newton’s Second Law of Motion, the accelerations of the body segments are collectively reflected by the GRF. For a given joint moment, its effects in accelerating the body segments can change as the body configuration changes. While a ‘bottom-up’ inverse dynamics approach allows the calculation of individual joint moments, it is usually limited in studying the induced body acceleration (or GF generation) by a given joint moment. This is due to the often lack of information of the
head, arms and trunk (HAT). In other words, the equations-of-motion of the whole body is incomplete. A dynamic model containing the entire linked-segment body is needed in investigating the contribution of a specific joint moment to the overall GF generation.

Induced acceleration analysis (IAA) has been used to study the contribution of individual joint moments (Kepple et al., 1997) and measured (Kaya et al., 2006) or estimated muscle forces (Anderson and Pandy, 2003; Hamner et al., 2010; Liu et al., 2008; Sasaki and Neptune, 2006) on the GF generation during legged locomotion. In an IAA, a full-body dynamic model along with a ground contact model is constructed. To study the instantaneous effect of a joint moment in accelerating the body, the moment of interest is applied in isolation into the equations-of-motion of the linked body, assuming a rigid ground contact (Anderson and Pandy, 2003; Hamner et al., 2010; Kaya et al., 2006; Kepple et al., 1997; Lin et al., 2011; Pandy et al., 2010). An alternative approach is by perturbing the factor of interest (Liu et al., 2006; Neptune et al., 2001). In this approach, the joint moment or muscle force of interest is altered by a small amount while all other factors remain unchanged. The equations of motion of the system are then integrated forward over a brief time. The perturbed and unperturbed conditions over time are then used to calculate the segmental accelerations or foot sprint-damper forces due to the perturbed factor. The inputs for an IAA can either be derived from a simulation (Anderson and Pandy, 2003; Neptune et al., 2001), or directly from information acquired experimentally (Kaya et al., 2006; Kepple et al., 1997; Hamner et al., 2010). The outcomes of an IAA represent a series of ‘snapshots’ describing the contribution of different factors to the overall GF generation at discrete time points.

IAAs on linear sprinting (Dorn et al., 2012) have identified soleus and medial and lateral gastrocnemius as the main contributors to both the vertical and propulsive force generation. While these findings provide an insight to the role of ankle plantarflexors in generating GF during sprinting, they cannot be generalized to movements of other body configurations, such as the body lean during curved sprinting.

The purpose of the current work was to investigate the contribution of lower extremity joint moments to the overall GF generation during curved sprinting with an IAA approach.
5.2 Materials and methods

The current IAA was conducted using experimental data. The general analysis workflow was as follows (Figure 5.1). Kinematic and kinetic data of subjects performing maximum-effort curved sprinting were first collected in a laboratory. Anthropometric information from individual subjects was used to scale the rigid body dynamic model and define segment inertial properties. A conventional “bottom-up” inverse dynamics approach was then used to calculate joint moments. During the IAA, the analysis for each time frame was conducted independently. For each frame, experimentally collected segment orientation was used to configure the model, and the GRF data and foot kinematic data were used to construct the ground contact. A particular factor of interest, ankle moment for example, was applied to the model in isolation (all other factors, such as gravity of the environment, were set to zero). GRF predicted for this specific time instant was then calculated as the reaction force at the ground contact ‘joint’. This procedure iterates through different factors for each time instant, after which the same analysis is conducted for the next instant.
Figure 5.1: Outline of the general workflow for the current IAA. Data from experimental measurement were used as parameters and inputs for the dynamic model.
5.2.1 Experimental data

Seven male subjects participated in the experiment (age 24 ± 2 years, mass 77.1 ± 4.9 kg, height 176.0 ± 7.5 cm; mean ± 1 s.d.). All subjects were free from lower extremity injuries during the year prior to the experiment. Written consent approved by the university ethics committee was obtained before each testing session.

Eight trials of maximum-effort curved sprints were performed along a circle (radius = 2.5 m). As the subjects reached their maximum speed, kinematic data (240 Hz) and kinetic data (2400 Hz per channel; eight channels) were collected simultaneously during an inside-leg stance. Kinematic data included the three-dimensional trajectories of 18 reflective makers securely attached to the skin and footwear upper representing the upper trunk, pelvis, and left thigh, shank, rearfoot and forefoot segments (Figure 5.2). Eight high-speed cameras (Motion Analysis Corporation, Santa Rosa CA, USA, model Eagle) were used to ensure sufficient field of view. GRF was recorded with an in-floor force platform (Kistler, Winterthur, Switzerland, model Z4852C). Prior to the movement trials, a neutral standing trial was collected with additional markers to identify joint centre locations. The hip joint was estimated to be located at the marker on the greater trochanter in the sagittal plane and directly below the marker placed on the anterior superior iliac spine (ASIS) in the frontal plane. The knee joint centre was defined as the middle point between the markers placed at the medial and lateral epicondyles of the femur, and the ankle joint centre was defined as the middle point between the markers placed at the medial and lateral malleoli. Joint rotations were performed prior to the placement of the joint markers to help identify the joint axes locations and orientations.

A fourth-order recursive Butterworth low-pass filter was used to remove the high frequency noise for both the raw kinematic (cut-off frequency: 10 Hz) and kinetic (cut-off frequency: 100 Hz) data. GRF, COP location, and orientation, angular velocity and angular accelerations of the segments and joints were calculated. A conventional ‘bottom-up’ inverse dynamics analysis was performed to quantify joint moments generated at the ankle, knee and hip joints. These experimentally quantified values were later used as parameters and inputs to the rigid-body dynamic model constructed for the IAA.
5.2.2 Rigid body dynamic model

The three-dimensional (3D) linked-segment model was constructed in the MATLAB SimMechanics® environment (The MathWorks, Inc., Natick, Massachusetts, USA). The segments were modeled as rigid bodies and the joints were assumed frictionless.

A total of four segments were used to represent the body: left foot, left shank, left thigh and one for head, arms, torso and the non-contact leg (HAT). The decision of including the non-contact leg as part of the HAT segment was made based on findings from Hettinga (2009), where the kinetic influences of including a swing leg were examined in a forward dynamics speed skating model. It was found that the swing leg did not have a significant effect on the development of the GRF. The inertial properties of the
body segments were scaled for each subject with the anthropometric information obtained (body mass and segment and joint locations during the neutral standing trial). Segment mass was defined as a proportion of the body mass based on Clauser et al. (1969). COM locations for the segments were located at a proportion of the segment lengths based on Winter (2005). For the foot length, since all subjects performed the experiment in the same pair of shoes, the shoe length (275 mm) was used. For each thigh and shank, the segment length was defined as the distance between joint centres, hip-to-knee and knee-to-ankle respectively. For the HAT segment, the segment length was defined as the distance between the average location of the three upper trunk markers and the average location of the three pelvis markers. The location of the HAT COM was defined as the middle point along the HAT segment. Finally, moment of inertia of the segments was estimated based on Forwood et al. (1985). A visual representation of the segment inertial properties can be seen in Figure 5.1 (Dynamic model), where the shape and size of the segments depict the mass distribution (uniform density was assumed).

Three joints were used to connect the body segments. The HAT and thigh segments were connected through a ball-and-socket hip. The knee was modeled as a hinge joint connecting shank and thigh allowing flexion/extension. The ankle was modeled as a universal joint between shank and foot allowing in-/eversion and plantar-/dorsiflexion. The location and orientation of the joint axes were configured for each subject so that they are consistent with the axes definitions implemented in the inverse dynamics analysis.

One challenging aspect of modelling a legged locomotion system is to properly represent the shoe-ground interactions. The nature of the shoe-ground interface has a direct influence on the equations-of-motion of the whole system and thus can significantly affect the model predictions. Multiple factors, such as friction and surface stiffness, dictate such shoe-ground interactions. In order to encapsulate the effects of these different material properties, intensive modeling effort is needed to construct the interfaces and foot-shoe structures (Cheung and Zhang, 2005). During an IAA, to simplify the shoe-ground interaction without losing generality, the contact has been widely modelled with hard kinematic constraints (Anderson and Pandy, 2003; Hamner et
The hard kinematic constraint approach assumes rigid contact point(s) – zero acceleration - between certain area of the shoe and ground, and allows a computationally-efficient (integration over time is not needed) estimation of the effects of individual joint moments to the GF generation for each time step (Lin et al., 2011). In contrast to a spring-and-damper approach where the contact force explicitly depends on the location and velocity of the contact points relative to the ground, the hard kinematic constraint approach includes the GRF implicitly into the equations-of-motion – as the reaction force enforcing the kinematic constraints of the ground contact ‘joint’ (Kaya et al., 2006). Instead of modeling the dynamics itself of the interface, as in a spring-and-damper approach (Neptune et al., 2000), hard kinematic constraints are used to model the effects of the dynamics. Thus, it is critical to constrain shoe motion in a way that it is consistent with the manner the shoe actually accelerates in the experiment. During walking for example (Anderson and Pandy, 2003), the shoe-ground has been modeled as a weld joint during foot-flat and a hinge joint after the heel leaves the ground.

In order to determine a proper kinematic constraint, the motions of the shoe relative to ground for all the subjects were qualitatively examined with a finite helical (screw) axis calculation. The relative motion between two rigid bodies can by described with a rotation about and a translation along an axis. With the finite helical axis calculation, the instantaneous rotation axis of the shoe between two continuous frames can be expressed. Figure 5.3 shows a representative time history of the rotation axis orientation and magnitude (represented by the axis length). It was found that during touch-down, or the impact phase, the shoes underwent a large rotation about the lateral edge. Meanwhile, large translation along the axis was observed. After the impact phase, the location of the floating rotation axis progresses along the long axis of the foot with a relative consistent orientation (rolling around a hinge). Furthermore, it was observed that the location of the helical axes progressed largely with the progression of the COP (Figure 5.3). Based on these observations a hinge axis passing through the COP approximating the post-impact helical axes orientation was chosen as the hard kinematic constraint for the current model. For each time frame, the axis was anchored at the
corresponding COP location (Hamner et al., 2010). The orientation of the axis was calculated as the average helical axis orientation of the current frame and its neighbouring 10 frames. The rotation axis is not allowed to penetrate through or translate along the floor surface.

Figure 5.3: The COP location and between-frame (interval duration = 0.00417 s) shoe rotation axis expressed in the lab coordinate system (LCS). The photographs illustrate the general orientation of the shoe in the LCS across stance, starting with blue at initial touch-down and finishing with red at toe-off. The length of the rotation axis indicates the magnitude of the rotation (ranging from 0.3° to 3.7°).

Since the hard kinematic constraints do not properly represent the shoe-ground interaction during the initial contact (approximately the first 25% of stance, (Lin et al., 2011)), analyses of this phase were excluded from the current study. The maximum GRF and ankle and knee moments occur at the mid-to-late stance phase (Chapter 4), when the realistic ground contact approaches practically a rigid constraint (high stiffness at the contact area). It is believed that the exclusion of the initial contact phase should not have a significant influence on revealing performance-related information.
5.2.3 Data analysis

With the IAA, the contributions of ankle, knee and hip moment, gravitational loading and Corolis loading to the overall GRF development were evaluated for each time step \((t = 0.00417 \text{ s})\). The overall effects of all the factors were first calculated and compared with the experimentally measured GRF. Then the individual effects of each factor were evaluated by setting other factors to zero.

A necessary condition for the model’s validity is that the modelled overall GRF reproduces the actual GRF measured experimentally with the force platform. To quantify the model-experiment agreement, the root-mean-square (RMS) errors were calculated for each GRF component. Violation of the rigid contact assumption was qualitatively examined by calculating the acceleration of the contact point (COP) through stance.

To examine the contribution of individual factors to the overall GF generation, the factor of interest was applied in isolation in the model. For example, to examine the contribution of the knee extension moment to the body acceleration for a given time instant, all other joint moments are set to zero, the gravity in the modelling environment is turned off, and the angular velocity of body segments are set to zero. At this very instant, the knee joint moment calculated through inverse dynamics was applied and the reaction force at the ground contact ‘joint’ was calculated. This analysis was repeated for each time instant over the time period of interest. The total contribution to the GRF development by a particular factor was calculated as the percentage of total ground reaction impulse (GRI) contributed by this factor over the time period of interest. The IAA was conducted for each of the seven subjects. In the end, the average values of the sampled population were calculated.
5.3 Results

Joint moments calculated with a bottom-up inverse dynamics approach for the inside leg ankle, knee and hip joints are presented in Figure 5.4, and they were used as the inputs for the IAA. Comparable peak moments were observed across the joints.

![Graphs showing ankle, knee, and hip joint moments](image)

Figure 5.4: Average ankle, knee and hip joint moments calculated with a conventional inverse dynamics approach.
The overall modelled GRF agreed well with the experimental measurement after the initial 25% of ground contact (Table 5.1; Figure 5.5). The RMS errors for each GRF component reduced after the initial ground contact (Table 5.1). As described in the method session, during this initial contact phase the shoe was rolling about and translating along its long axis; the contact point (COP) experienced large acceleration on the ground (Figure 5.5). The rigid contact assumptions during this phase were likely violated.

<table>
<thead>
<tr>
<th></th>
<th>0 – 25% stance</th>
<th>26 – 100% stance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vertical [N]</td>
<td>281.5</td>
<td>55.2</td>
</tr>
<tr>
<td>Centripetal [N]</td>
<td>227.7</td>
<td>33.2</td>
</tr>
<tr>
<td>Anterior-posterior [N]</td>
<td>66.3</td>
<td>14.5</td>
</tr>
</tbody>
</table>

Table 5.1: RMS errors of the model GRF prediction during and after the initial ground contact (0 -25%) of stance.
Figure 5.5: Sample averages of the modelled versus experimental GRF over stance. During early stance, the rigid contact assumption was violated and this was reflected in the large contact point (COP) acceleration.
After the initial contact phase (first 25% of stance) joint moments contributed to the majority of the GRF development, and the influence of gravitational and Coriolis loading was found to be minimal (Table 5.2). Analysis of the post-impact stance showed that ankle and knee moments contributed to the majority of the GRF in supporting the body mass (Figure 5.6). The centripetal GRF was largely contributed by the knee and ankle joint moments prior to mid-stance, and mostly by the ankle moment after mid-stance. In the anterior-posterior direction, the knee extension moment was associated with the braking GRF and the ankle moment was the main contributor for the propulsive GRF. Hip moment contributed minimally to the GRF development.

<table>
<thead>
<tr>
<th></th>
<th>Ankle moment</th>
<th>Knee moment</th>
<th>Hip moment</th>
<th>Gravity</th>
<th>Coriolis</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Vertical GRI [%]</strong></td>
<td>77.8</td>
<td>22.5</td>
<td>-2.1</td>
<td>1.9</td>
<td>-0.1</td>
</tr>
<tr>
<td><strong>Centripetal GRI [%]</strong></td>
<td>72.4</td>
<td>23.9</td>
<td>5.1</td>
<td>0.0</td>
<td>-1.4</td>
</tr>
<tr>
<td><strong>Anterior GRI [%]</strong></td>
<td>271.9</td>
<td>-176.2</td>
<td>0.08</td>
<td>-4.0</td>
<td>0.0</td>
</tr>
</tbody>
</table>

Table 5.2: Contribution of the individual factors to the overall GRF development after the initial contact. The overall contribution of each factor over the post-impact phase was assessed by calculating the percentage of total GRI contributed by this factor.
Figure 5.6: GRF induced by individual joint moments during the post-impact stance. The plots represent the average values of the sampled population.
5.4 Discussion

The purpose of this study was to explore the contribution of moments generated at the ankle, knee and hip joints to the overall GRF development during maximum-effort curved sprinting. Through a 3D IAA, it was found that the moment generated at the ankle joint contributed most to the overall GF generation.

Due to dynamic coupling, the moment generated at a joint can accelerate segments that are not immediately connected to this joint (Zajac and Gordon, 1989). The effect of a given joint moment in accelerating the entire linked-segment system is dependent not only on the magnitude and direction of the moment but also the configuration of the body. Through the IAA approach, studies have been conducted to investigate such acceleration effects induced by moments generated by individual muscles during walking (Anderson and Pandy, 2003), running (Hamner et al., 2010), and sprinting (Dorn et al., 2012).

During curved sprinting, the majority of the GRF development was contributed by joint moments, in comparison to the effects of gravitational and Coriolis loading. The reason for the minimal effects of gravitational loading is likely associated with the flexed body posture associated with the movement. The contribution of gravitational loading to the vertical GRF is high when the body is in an upright posture with the lower limb extended (e.g. mid-stance of walking in Anderson and Pandy (2003)). The reason is that the weight of the segments is then transmitted along the bones to the ground. In contrast, when the limb is in a more flexed posture, instead of transmitting along the segments, the gravitational force tends to accelerate the segments angularly about the joints and collapse the linked system. In such a scenario, if no counter-acting joint moments are present, the body will experience near free-falling and the vertical GRF will be small. In such body postures, joint moments become the main contributors to supporting the body weight and generating GF. Previously, the threshold for bones to endure long-axial compression has been assumed as a limiting factor in curved sprinting performance (Greene, 1987). Findings from the current analysis tend to challenge this assumption.

The ankle and knee joint moments contributed to the majority of vertical support for the body weight throughout stance. During linear running/sprinting (Dorn et al., 2012)
at various speeds (3.5 – 9.0 m s\(^{-1}\)), it was found that both the plantarflexor and knee extensor muscles contributed the weight support throughout the majority of stance. Meanwhile, the contribution by the ankle plantarflexor muscles was found to largely exceed the contribution of the knee extension muscles. The current findings for curved sprinting are consistent with these observations for linear sprinting.

The majority of the centripetal GRF was contributed by the ankle moment throughout post-impact stance, while contribution from the knee joint moment was observed prior to mid-stance. Little information is available in the literature to allow a comparison on this finding across studies. During linear sprinting, the ankle plantarflexion and knee extension muscles were found to be the main contributors to the medial-lateral GF generation, but their contributions were in opposite directions (Dorn et al., 2012). In contrast, during walking (Pandy et al., 2010), knee extensors and ankle plantarflexors have been found to accelerate the body laterally, while the hip abductors accelerate the body medially. Overall, the net GRF was oriented medially throughout the walking stance. While the function of the knee extensors and ankle plantarflexors in accelerating the body laterally (i.e. toward the circle centre) is observed in the current work, the contribution of the hip abductors in generating GF was found to be minimal. This observation may explain the more laterally oriented GRF found in the current work.

In the anterior-posterior direction, the ankle moment was found to contribute to the propulsive force and the knee joint moment contributed to the braking force. This finding was consistent with previous observations for linear sprinting (Dorn et al., 2012) and running (Hamner et al., 2010).

In the previous chapter, it was proposed that during maximum-effort curved sprints the MTUs surrounding the ankle joint might be operating at their moment generation limit. One plausible reason for the ankle MTUs to operate at their maximum could be for their large contribution to the generation of the GF to both supporting the body weight and creating centripetal acceleration, as identified in the current work. While the current IAA with experimental data provides a descriptive GRF decomposition, it is limited in that it cannot be used to predict performance changes over the entire stance where individual moments are altered. The current IAA is not equivalent to a forward
dynamics optimization control model since each time instant is analyzed independently. For each time instant, the experimentally measured body configuration and contact condition were used as inputs to allow the evaluation of the contribution by individual joint moments. The effects of altering certain joint moments are not integrated over time. In order to gain further understanding on the influence of ankle moment generation on the overall curved sprinting performance, an experimental approach needs to be taken. If during maximum-effort curved sprinting, the ankle moment is indeed operating at its limit and thus constrains the overall performance, by experimentally increasing the ankle moment generation, improvements in performance would be expected.

A major limitation of the current work is that the foot-ground interaction model is a large simplification of the real-life situation. It has been shown that the definition of the kinematic constraints can have a direct influence on the IAA outcomes (Dorn et al., 2011). More specifically, during walking and running, by reducing the degrees of freedom at the contact, from a hinge to a weld joint for example, the GRF contribution of the proximal joint moments would be increased. In a pilot study where the IAA was performed with a weld joint, the modeled GRF was found to differ substantially from the experimental measurement. The RMS errors in the weld joint model nearly doubled compared to the current model. In reality, the kinematic constraint should lie in between a hinge and a weld joint as the foot rolls forward. The much larger RMS errors associated with the weld joint model indicate that the hinge model is likely a more appropriate choice. Future work with a more realistic ground contact model, such as a cam-like foot shape, is needed to increase the IAA validity.

5.5 Conclusion

Through an IAA with experimental data, it was identified that joint moments contributed to nearly all the GRF development during curved sprinting. Among all, the moment generated at the ankle joint contributes to the majority of the vertical and centripetal GRF development.
CHAPTER SIX:
ANKLE MOMENT GENERATION AND
MAXIMUM-EFFORT CURVED SPRINTING PERFORMANCE

6.1 Introduction

The ability to negotiate turns at high speed is critical for athletic performance. In sports such as basketball, players spend more than 40% of the game changing directions (Stacoff et al., 1993). One crucial biomechanical constraint for turning performance is the ground force (GF) that can be generated by the supporting limbs (McMahon, 1984). In addition to the need to overcome gravity with the vertical ground reaction force (GRF), during curved sprinting a horizontal GRF is needed to create a centripetal acceleration to redirect the body centre of mass (COM). As curved sprinting speed increases, the need for GRF would be expected to increase exponentially (Chapter 2.2).

One of the factors limiting the GF generation may be the strength of the muscle tendon units (MTUs) surrounding certain lower extremity joint(s). In a recent investigation on ankle and knee moments during curved sprinting, it was observed that when subjects performed maximum-effort curved sprints with and without an additional mass of 12.4 kg, the total moment generated at the ankle joint over stance remained unchanged (Chapter 4). This was despite an 11.4% difference in the average frontal-plane GRF over stance. This observation led to the speculation that for the specific operating conditions (joint angles and angular velocities), the MTUs surrounding the ankle joint may have reached the limit in terms of generating joint moments.

The ideal means to examine this speculation would be to directly measure the MTUs’ maximum moment generation ability and compare the assessed values with the results from the values derived from an inverse dynamic analysis of the real movement. In order to achieve a valid assessment, the measurement conditions need to closely represent the actual operating conditions. The prevalent method for joint torque assessments in vivo is with the use of a dynamometer (e.g. Bobbert and van Ingen
Schenau, 1990; Herzog, 1988), where maximum-effort isokinetic and/or isometric torques generated around an isolated joint are measured. A survey of the literature, however, revealed that the maximum ankle plantarflexion during such isolated measurement tends to underestimate the ankle’s ability to generate torque during a leg extension involving multiple joints (Hahn et al., 2011). In addition, the majority of dynamometers are limited in assessing joint torques surrounding one fixed rotation axis and assuming the moment about the two other axes are negligible. As shown in Chapter 4 and 5, the magnitudes of the non-sagittal plane joint moments during curved sprinting are substantial. Furthermore, misalignment of the prescribed measurement axis with the actual joint axis can lead to errors (Herzog, 1988). During curved sprinting, the ankle joint is likely rotating about an axis that changes orientation over time, making a realistic representation of the push-off on a dynamometer challenging.

An alternative approach to investigate the speculation that the ankle moment generation limits curved sprinting performance may be by implementing interventions that can potentially alter the overall ankle moment generation. If the ankle moment generation was indeed at the limit and constrains the overall performance, with interventions raising such moment generation ability, improvement in the GF generation and overall performance would be expected. One experimental approach to alter the ankle MTUs’ ability to generate moment may be by changing their operating configuration.

To avoid a rolling moment in the body frontal plane, curved sprinting is performed with a body lean, which positions the ankle joint at an everted/inverted configuration. The deviation from its frontal plane neutral position may place the ankle joint in a less optimal condition in generating joint moments. Wedged footwear can potentially be used as an intervention to align the ankle joint without compromising the overall body lean. Such joint alignment may consequently increase the MTUs’ overall ability to generate moment. Early work by Greene (1987) showed that when sprinting along banked curves, peak running speed can be improved on the order of 10%. Unfortunately, no kinetic data were available to reveal the underlying mechanisms for such changes.
The main purpose of the current study was to gain a better understanding of the influence of ankle moment generation on curved sprinting performance. The null hypotheses were: with the implementation of wedged footwear, no increase will occur in the 1) total ankle moment generation, 2) GF generation, and 3) curved sprinting speed.

6.2 Materials and methods

6.2.1 Setup and equipments

Maximum-effort sprinting trials were performed along a circle of 2.5 m radius in a laboratory. Subjects started the sprint at a location that allowed them to achieve top speed when entering the motion analysis collection volume (width 1.5 m, length 3.5 m, height 2.3 m), where kinetic and kinematic data were sampled simultaneously. GRF data were collected with an in-ground force platform (Kistler, Winterthur, Switzerland, model Z4852C) at 2400 Hz. Only trials where the subjects could plant their inside foot naturally on the force platform were used in later analyses. Eight high-speed cameras (Motion Analysis Corporation, Santa Rosa CA, USA, model Eagle) were used to capture the reflective marker trajectories at 240 Hz. Twelve markers were placed on each subject to represent the pelvis, left thigh, shank, and foot segment. Ankle and knee joint centres were determined using additional markers during neutral standing trials prior to the movement trials. The ankle joint centre was defined as the middle point between markers placed on the medial and lateral malleoli. The knee joint centre was defined as the middle point between markers placed on the medial and lateral epicondyles of the femur.

Testing footwear (size: US10) with and without modified midsole structures were used in the current study (Figure 6.1). The modifications were made in the frontal-plane aiming to reduce ankle eversion/inversion. The modified shoe condition was termed ‘wedged’, and the unmodified condition was referred to as ‘control’. Since the subjects were required to perform counter-clockwise turns, the medial side of the left shoe (inside foot) was raised, and the lateral side of the right shoe (outside foot) was raised. The modifications were made so that the height raise was uniform longitudinally (Figure 6.2).
Other aspects of the testing shoes - upper structure and outsole material - were identical between conditions. Parameters of the modifications are summarized in Table 6.1.

Figure 6.1: Modification of the midsole structure to align the ankle joint during counter-clockwise curved sprinting.

Figure 6.2: The midsole modification was uniform along the longitudinal direction of the testing shoes.
### Table 6.1: The modification parameters of the experimental footwear.

<table>
<thead>
<tr>
<th></th>
<th>Lower edge</th>
<th>Raised edge</th>
<th>Resting angle [°]</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>First metatarsal head</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Control</td>
<td>2.5</td>
<td>2.5</td>
<td>0</td>
</tr>
<tr>
<td>Wedged</td>
<td>2.5</td>
<td>3.5</td>
<td>6.3</td>
</tr>
<tr>
<td><strong>Calcaneal tuberosity</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Control</td>
<td>3.5</td>
<td>3.5</td>
<td>0</td>
</tr>
<tr>
<td>Wedged</td>
<td>3.5</td>
<td>4.5</td>
<td>7.5</td>
</tr>
</tbody>
</table>

6.2.2 Subjects

Seventeen male subjects were recruited for the current study (age 23 ± 3 years, mass 75.5 ± 6.2 kg, height 178.5 ± 6.5 cm; mean ± 1 s.d.). All subjects regularly participated in recreational sports 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.

6.2.3 Protocol

A 20-min warm-up session was provided before the data collection. Three maximum-effort practice runs were required prior to each testing condition to familiarize the subjects with performing in the testing footwear. Four maximum-effort trials were first collected in the control condition. The subjects then performed eight trials in the wedged condition, after which four additional trials were collected for the control condition. The number of trials was chosen in order to minimize the potential influence of fatigue. A minimum of a 3-min resting period was given between trials, while the floor surface was cleaned. In order to examine if there existed any learning and/or fatigue effects, two-tailed paired t-tests ($\alpha = 0.05$) were used to compare the control condition performance (defined in next section) before and after the wedged footwear condition. Two subjects were not able to sustain a consistent performance level throughout testing, and were thus excluded from further analysis.
6.2.4 Data analysis

Prior to any analysis, 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 (Chapter 3).

To evaluate the effects of the midsole modification, ankle in-/eversion angle during stance was calculated with a joint coordinate system convention (Wu et al., 2002). Based on this convention the in-/eversion axis was defined as the cross-product of the hinge (flexion/extension) axis of the shank segment and the long (ad-/abduction) axis of the foot segment (details on the construction of segment coordinate system can be seen in Chapter 4). The neutral standing trials were performed in the control condition, while the on-shoe locations of the markers were kept consistent between conditions. Joint moments at the ankle and knee joint of the inside leg were calculated using an inverse dynamics approach (Andrews, 1995). The inside leg was chosen since its ability to generate force was hindered to a greater extent compared to the outside leg when sprinting along small circles (Chang and Kram, 2007). The segment inertial properties used for the inverse dynamics calculation were estimated using values summarized in Winter (2005). The vector sum of the moments in all planes was calculated to allow investigations of the overall ability to generate moment at each joint. Furthermore, the vector sum of the frontal and transverse plane moments was calculated to represent the demand for joint stabilization (Chapter 4); they were denoted as the non-sagittal plane moments.

Average GRFs over stance and GRIs were calculated. The stance interval was defined using the vertical GRF at a 3% body weight threshold. Performance was quantified using the average speed of the body COM over stance. The location of the body COM was estimated by averaging the 3D coordinates of the three markers placed on the pelvis segment.

6.2.5 Statistical analysis

Paired t-tests were used to compare results between conditions. Based on the nature of the theory, one-tailed analyses were chosen. The statistical significance level was set a priori at $\alpha = 0.05$. 
6.3 Results

Averaged over stance, the wedged footwear intervention reduced the inside ankle eversion by 4.2° (p<0.0001). While the magnitude of the effect varies across time, it was apparent throughout the majority of the stance phase (Figure 6.3).

Figure 6.3: Eversion angle of the inside leg ankle was reduced with the wedged shoes. Thick lines represent sample averages and thin lines represent ± 1 s.d..
Compared to the control condition, a 13.6% increase (p<0.0001) in the peak total ankle moment generation was observed when the subjects sprinted in the wedged footwear (control: 200.3 ± 44.2 Nm versus wedged: 227.5 ± 50.7 Nm; Figure 6.4). When averaged over stance, the total average ankle moment increased 12.4% (p<0.0001) from the control (120.4 ± 28.2 Nm) to the wedged (135.3 ± 32.0 Nm) condition.

Figure 6.4: Total ankle moment generation increased in the wedged shoe condition compared to control. Thick lines represent sample averages and thin lines represent ± 1 s.d..
The increased total ankle moment generation in the wedged shoe condition was due to the increase in the plantarflexion moment; compared to the control condition a lower non-sagittal plane ankle moment was observed (Figure 6.5).

![Sagittal and non-sagittal plane ankle moment generations.](image)

**Figure 6.5:** Sagittal and non-sagittal plane ankle moment generations. Thick lines represent sample averages and thin lines represent ± 1 s.d.. In addition, peak moments were plotted with s.d. bars. *Statistically significant differences.*

No differences in the total knee moment were detected between conditions (peak: p = 0.0769; stance average: p = 0.1008; Figure 6.6). While no changes (p = 0.2589) were found for the maximum knee extension moment between conditions, a larger peak non-sagittal plane moment (p = 0.0025) was observed in the wedged condition compared to the control (Figure 6.7).
Figure 6.6: Total knee moment generation remained unchanged between conditions. Thick lines represent sample averages and thin lines represent ± 1 s.d..

Figure 6.7: Sagittal and non-sagittal plane knee moment generation. Thick lines represent sample averages and thin lines represent ± 1 s.d.. In addition, peak moments were plotted with s.d. bars. *Statistically significant differences.
With the wedged footwear, a greater centripetal GRF over stance was observed, while the vertical GRF remained unchanged (Figure 6.8(a); Table 6.2). The increase in the centripetal GRF was apparent through the post-impact stance phase (Figure 6.8 (b)). Despite shortened stance duration, the centripetal GRI was significantly larger in the wedged condition compared to control (Table 6.2). The curved sprinting speed increased by 4.3% ($p = 0.0001$) from the control to the wedged footwear condition (Figure 6.9).

![Figure 6.8: (a) The average frontal-plane GRF, where GRF$_{vert}$ is the vertical GRF and GRF$_{cpt}$ is the centripetal GRF. (b) The magnitude of the centripetal GRF over normalized stance. Thick lines represent sample averages and thin lines represent ± 1 s.d..](image)
### Control vs. Wedged

<table>
<thead>
<tr>
<th></th>
<th>Control</th>
<th>Wedged</th>
<th>One-tailed paired t-test</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Average GRF(\text{cpt}) [N]</strong></td>
<td>569.8 ± 129.3</td>
<td>632.1 ± 148.4</td>
<td>(p &lt; 0.0001)</td>
</tr>
<tr>
<td><strong>Average GRF(\text{vert}) [N]</strong></td>
<td>848.3 ± 122.2</td>
<td>855.8 ± 136.9</td>
<td>(p = 0.2193)</td>
</tr>
<tr>
<td><strong>Average GRF(\text{frontal}) [N]</strong></td>
<td>1029.1 ± 166.8</td>
<td>1070.9 ± 192.4</td>
<td>(p = 0.0012)</td>
</tr>
<tr>
<td><strong>Stance time [ms]</strong></td>
<td>236.1 ± 18.5</td>
<td>231.2 ± 16.5</td>
<td>(p = 0.0074)</td>
</tr>
<tr>
<td><strong>GRI(\text{cpt}) [Ns]</strong></td>
<td>134.3 ± 29.4</td>
<td>145.9 ± 33.5</td>
<td>(p &lt; 0.0001)</td>
</tr>
<tr>
<td><strong>GRI(\text{vert}) [Ns]</strong></td>
<td>200.3 ± 32.6</td>
<td>198.0 ± 34.5</td>
<td>(p = 0.1519)</td>
</tr>
</tbody>
</table>

**Table 6.2: GRF and GRI results from the inside leg (mean ± 1 s.d.).**

**Figure 6.9: Maximum curved sprinting speed (calculated as the average COM speed over stance).** The diamonds represent group averages and the grey lines are data from individual subjects. *Statistically significant difference.
6.4 Discussion

The purpose of this study was to determine the influence of ankle moment generation on curved sprinting performance. By implementing wedged footwear, the ankle joint configuration was altered throughout stance, and increases were observed in the peak overall ankle moment generation, GRFs and top curved sprinting speed.

6.4.1 Ankle moment and curved sprinting performance

In Chapter 4, by overloading the system with an additional mass of 12.4 kg, the frontal plane GF generation of the inside leg increased by 11.4% over stance. The total ankle moment generation, however, remained unchanged. This observation led to the speculation that during maximum-effort curved sprinting, the MTUs surrounding the ankle joint might have reached the moment generation limit for the specific joint configuration. If this was the case, ankle moment generation may in turn impose a limit for the overall performance of the locomotion system. In the current experiment, an attempt was made to alter such moment generation ability by inducing changes in joint configuration. With the wedged footwear the ankle joint was aligned 4.2° closer to its frontal plane neutral position. With the changes of ankle eversion angle, significant increases in the overall ankle moment generation were observed.

The increase in ankle moment was associated with a significant increase in the GRF, the centripetal component in particular, while no changes were observed in the total moment generated at the knee joint. Despite a shortened ground contact period, the 10.9% increase in the average centripetal GRF leads to a greater centripetal GRI in the wedged footwear condition. The top curved sprinting speed for the current sample increased by 4.3%. While to the authors’ knowledge little information is available from previous research that would allow comparisons across studies, this non-linear relationship between changes in force and overall performance is in line with what would be expected from a mathematical prediction (Chapter 2). In addition, it shall be noted that the overall performance improvement was likely the result of changes occurring in both limbs. In pilot work with 11 subjects of the current sample, the GF generated by the
outside leg was compared between conditions. With the wedged footwear, the average centripetal GF generation of the outside leg increased by a similar extent, 11.7%.

6.4.2 Further exploration of the non-sagittal plane stabilizing moment theory

Previously, the non-sagittal plane stabilizing moment experienced at the lower extremity joints has been suggested as a potential limiting factor for curved sprinting performance (Chang and Kram, 2007). The experiment conducted in Chapter 4 provided a direct test of the hypotheses formed based on this theory. The results showed that during curved sprinting with an additional mass (12.4 kg), the subjects could endure higher non-sagittal plane moments at both the ankle and knee joint of the inside leg. This finding indicates that the joint stabilizing moments were not at their limits during maximum-effort sprints in the control condition (without the mass). Observations made in the current experiment further challenge the notion that the non-sagittal plane moment at the lower extremity joints is limiting the overall performance. If the non-sagittal plane moment experienced at certain joints is the predominant performance constraint, in response to experimental interventions, this moment would be expected to remain operating at such limit for the subjects to perform maximally. At the knee joint, it was found that the peak non-sagittal plane moment increased significantly when the subjects sprinted with the wedged footwear. In contrast, at the ankle joint the wedged footwear reduced the non-sagittal plane moments. These findings further challenge the theory of non-sagittal plane joint stabilizing moments as the limit for curved sprinting performance.

6.4.3 Kinematic constraint imposed by the ankle joint range of motion

Ankle joint range of motion can be a plausible factor limiting body lean angle and overall curved sprinting performance. When sprinting along a curve of 2.5 m radius, the subjects experienced a peak ankle eversion greater than 35° (Figure 6.3) for the inside leg. This value approximates the maximum ankle eversion angle assessed in passive range of motion tests (Robinson et al., 1986). It is possible that during curved sprinting in the control footwear a kinematic constraint for further ankle eversion was imposed limiting the curved sprinting performance. While results from the current study cannot
directly address whether the ankle eversion is at the limit in the control condition, they tend to suggest that such a kinematic constraint may not be the predominant limit for performance. If this kinematic constraint was the ultimate performance limit, it would be expected that in both control and wedged footwear conditions, the subjects would operate at the threshold to perform maximally. With the wedged footwear, it would be expected that the subjects would increase the body lean to the extent that the ankle eversion angle reaches the limit again. Instead, it was observed that the ankle eversion angle was reduced throughout the stance in the wedged footwear condition, and the subjects did not further increase the body lean by operating at the ankle eversion threshold.

One limitation of the current joint angle evaluation is that the foot was represented using markers placed on the shoe, and the shoe position captured likely contains error in representing the actual foot (Stacoff et al., 1996). Effort to reduce such error was made in the current study by using footwear with identical upper construction between conditions, yet, future studies are needed to further investigate the ankle range of motion limit theory.

6.4.4 Control of the vertical GRI

While results from the current study supported the speculation that ankle moment generation may limit curved sprinting performance, it is likely that there exist other performance constraints.

When sprinting along a curve of given radius, an optimal step length may constrain the generation of GF, the vertical component in particular. In Chapter 4, with the additional mass attached, the subjects exerted a larger knee moment and a larger GRF compared to the control condition. In discussion of the potential reason for why the knee extensors reserved the moment generation ability during sprints without the mass, it was speculated that an excessive vertical GRI associated with a larger knee moment might elongate the flight travel and thus step length, which could be counterproductive for sprinting along curves of small radii. For a given body weight, it was proposed that there might exist a vertical GRI level for maintaining an optimal step length for performance. Results from the current study tend to support this notion. In contrast to Chapter 4, in the
current experiment the need to support body weight was unchanged between experimental conditions. With the wedged footwear, the subjects increased the centripetal GRI significantly while maintaining the vertical GRI when compared to the control footwear. Such directional increase in GRI seemed to be accomplished mostly by altering the ankle moment generation as the knee moment remained constant. Future work is needed to explore the relationship among GRI, step length and step frequency during curved sprinting from a performance optimization perspective.

6.5 Conclusion

During curved sprinting, as the ankle joint was aligned closer to its neutral position in the frontal plane, a larger total ankle joint moment was generated. The increase in the ankle moment generation was associated with a larger centripetal GRI despite a shortened stance time; overall, curved sprinting speed increased significantly. These observations supported the notion that during curved sprinting, the ankle joint moment generation may be one of the factors limiting the overall performance of the system.
CHAPTER SEVEN:
SUMMARY

7.1 Observations and interpretations

The current dissertation focused on the locomotion task of sprinting along a curved path, and aimed to gain knowledge on the mechanical and biomechanical constraints for achieving higher curved sprinting speeds. Six specific questions were addressed with experimental and mathematical approaches. The observations and interpretations are summarized as follows.

1. To what extent does available footwear traction limit body lean and overall curved sprinting performance?

In Chapter 3, the influence of available footwear traction on curved sprinting performance was systematically examined. By increasing the available traction from 0.26 to 0.54 then to 0.82, the subjects were able to utilize the incremental traction increases to further body lean, and generate greater ground impulse (GI) within a shortened stance period. Significant performance improvements were observed. Further increase of the available traction from 0.82 to 1.13 was not associated with detectable performance changes. Despite a greater amount of traction available, the subjects were no longer able to utilize it to achieve a higher speed.

At the traction coefficient of 0.82, it appeared that available traction at the shoe-ground interface no longer constrained performance. A closer examination of the utilized traction at the high available traction conditions revealed that an average traction coefficient of 0.6 was utilized, corresponding to a 59° ground reaction force (GRF) orientation with respect to the ground.
2. When sprinting maximally along curves of small radii, is the supporting limb operating at its limit generating extension force?

This question was addressed in Chapter 4 with an experimental perturbation – added body mass. If the overall limb force generation is at its limit during maximum-effort curved sprints, with the additional mass, an elongated ground contact is expected for the generation of the additional GI needed, but the peak limb force is expected to remain constant. As the body mass increased, a longer stance time was observed. Meanwhile, a significantly higher vertical and centripetal GRF was found. This observation indicated that during curved sprinting without the additional mass, the limb likely has further potential to generate ground force (GF).

3. During maximum-effort curved sprinting, is the non-sagittal plane joint loading experiencing its limit?

Chapter 4 tested the hypothesis that during maximum-effort curved sprinting, non-sagittal plane lower extremity joint moments may be operating at their limits thus constraining the overall performance. By introducing an experimental perturbation – an additional mass placed around the body COM, GRF experienced was altered. As the external force increased in the added mass condition, the non-sagittal plane ankle and knee joint moments increased significantly compared to sprinting maximally without the mass.

Although a tremendous amount of loading was observed at the knee and ankle joints in the non-sagittal planes, the current observations indicated that such joint loading was not at the limits during maximum-effort sprints. These findings challenged the theory that non-sagittal plane joint moments limit the maximum curved sprinting performance (Chang and Kram, 2007).

Furthermore, in Chapter 6, subjects performed curved sprints in flat shoes versus shoes with a wedge implemented in the midsole. By aligning the ankle joint in the frontal
plane closer to the neutral position, during curved sprints the wedge implementation was expected to reduce the lever arm of the GRF with respect to the ankle joint. This would reduce the non-sagittal plane ankle moment if the GRF remained the same. If the non-sagittal plane moment was the limiting factor, when the subjects sprinted in the wedged compared to the flat shoe, the non-sagittal plane moments would be expected to remain operating at the limit, and thus a greater GRF would be expected. In the experiment, when the subjects were sprinting in the wedged compared to the flat shoes, the resultant GRF increased significantly, however the non-sagittal plane ankle moment reduced. Furthermore, the peak knee non-sagittal plane moment increased significantly in the wedged condition, indicating when sprinting in the control condition, the knee non-sagittal plane moment was not at the limit. These observations together provided further indication that the non-sagittal plane joint moments may not be the predominant factor limiting curved sprinting performance.

4. During curved sprinting, what are the contribution of joint torques, gravitational loading and Coriolis loading to the overall GF generation of the system? Furthermore, what are the individual contributions to the GF generation by the moment generated at each lower extremity joint?

In Chapter 5, an induced acceleration analysis was performed to decompose the GRF. In this analysis, a rigid body linked-segment model was constructed based on information of individual human subjects. For each time instant, the body configuration quantified experimentally was used to configure the model, and then the effects of gravity, angular velocity, and joint torques on the GF generation were examined by applying them to the model individually. All the input values were calculated from experimental data.

It was observed that joint torques contributed to the majority of the force generation. Among the lower extremity joints, torques generated at the ankle joint contributed the largest to both supporting the body weight and creating the centripetal acceleration.
5. **During maximum-effort curved sprinting, are certain joints operating at the limit in generating torque to accelerate the body?**

In Chapter 4, as the subjects sprinted with an additional mass, greater GRFs were observed. A close examination of the total moments generated at the ankle and knee joint revealed that the increases in GF generation were associated with an increase in knee moment generation, meanwhile the ankle moment generation remained unchanged. Upon a perturbation - in this case an 11% increase in GRF over stance, the unchanged variables tend to provide clues to the factors operating at their limits. Since ankle moment contributes significantly to the GF generation (Chapter 5), it is possible that during curved sprints the moment generation at this joint is operating at the limit.

In Chapter 6, an experimental implementation was attempted to increase the moment generation ability of the ankle joint by aligning it to a configuration closer to its neutral position. This was achieved with experimental footwear with a wedge in the midsole for both feet. In the experiment, it was found that the wedged footwear increased the subjects’ ability to generate ankle moment significantly. Force generation for both limbs increased. The overall curved sprinting performance showed significant improvement. These observations indicated that the ability to generate moment at the ankle joint is likely one of the factors limiting curved sprinting performance.

6. **Is the ankle range of motion a predominant factor in limiting the body lean and overall curved sprinting performance?**

In Chapter 6, it was confirmed that during curved sprints, the ankle joint underwent large ev-/inversion. To address the question of performance constraint imposed by the ankle range of motion, the inside leg ankle eversion angle during curved sprints in a flat versus a wedged shoe was compared.

If ankle range of motion is the ultimate performance constraint, when sprinting in both shoe conditions it would be expected that the subjects would operate at this kinematic limit to perform maximally. For a given body lean, the wedged shoe would be
expected to reduce the ankle eversion. If the kinematic limit imposed by the ankle range of motion is the predominant factor limiting performance, it would then be expected that in the wedged shoe the subjects would increase their body lean to the extent that the ankle joint reaches the kinematic limit that it operates at in the flat shoe condition.

It was found that when the subjects performed in the wedged compared to the flat shoes, the ankle eversion of the inside leg was reduced throughout stance. This observation indicated that during maximum-effort curved sprints, the subjects did not necessarily achieve the maximum performance by operating at the kinematic limit.

7.2 Limitations, and future

One of the main limitations of the current study is that the muscle tendon units’ (MTU) ability to generate moment was not directly measured. Joint moment calculated with inverse dynamics was used instead as an indirect measure. The joint moment calculated indicates only the net effects of all the tissues surrounding a joint, e.g. muscle, tendon and ligaments. In the overdetermined musculoskeletal system, effects such as the co-contraction of the agonist and antagonist muscle pairs and force sharing between uniarticular and biarticular muscles can confound the interpretations. The proposal of ankle moment generation as a limiting factor was initiated based on the observation that moment generation at this joint remained constant despite the large difference in the GRF across the control versus added mass conditions (Chapter 4). It is possible that the unchanged net ankle moment was the consequence of simultaneous increases in the force generation by the tibialis anterior and the soleus and gastrocnemius. Maybe the net moment generation can indeed be increased (by, for example, an unbalanced force increase of the agonist and antagonist muscle pairs), but is kept constant to fulfill the need of certain unknown control mechanisms. Future developments of a direct measure of the joint moment generation ability in vivo would be ideal. While results from the added mass (Chapter 4) and wedged shoe (Chapter 6) experiments increased confidence in the notion that the ankle joint moment generation may be at the limit during regular maximum-effort curved sprints, a direct and valid strength measure can provide strong
evidence for or against this theory. The challenge in this direct approach would lie in the construction of an apparatus and protocol with which the joint configuration and operating conditions can closely replicate reality.

During the investigation of the contribution of individual joint moments to the overall GRF development (Chapter 5), a kinematic hard constraint was used to simplify the ground contact. Inaccuracy of such a simplified model can lead to errors in estimating the distribution of the contributions (Dorn et al., 2011). Future work with more realistic contact models are needed to improve the validity of the analysis. More importantly, experimental works are needed to validate the model predictions. This may be achieved with the use of robotic exoskeletons, or controlled electrical stimulations of targeted muscles. With these methodologies, the investigators may selectively change the moment output at a joint at a given time instant during the movement and quantify the induced segmental accelerations. Comparison between outcomes from such an experiment and the model prediction can be used for the validation of the model.

In studying the kinematic constraints at the ankle joint, the foot segment was represented using shoe markers. It has been shown substantial foot-shoe movements can occur during cutting movements (Stacoff et al., 1996), and these movements will lead to errors in calculating joint angles. Despite the effort in choosing footwear with stiff upper construction, deformation is unavoidable. Future work where in-shoe movement of the foot can be quantified without compromising the upper construction is needed for more accurate representation of the joint angles.

One fascinating question arose from the findings in Chapter 4 where the subject performed curved sprinting maximally with and without an additional mass. With the additional mass, the subjects were able to generate a greater knee moment and total limb force. It remains to be answered why this knee moment generation was reserved when sprinting maximally without the mass. In the discussion of Chapter 4, it was proposed that this reserved knee moment generation may be associated with the control of vertical GRI, and step length and frequency for performance optimization. The trade-off functions between the increases in step length versus frequency during curved sprinting along a small circle require future investigation.
REFERENCES


APPENDIX A:

JOINT KINEMATICS DURING CURVED SPRINTING PERFORMED WITH VERSUS WITHOUT AN ADDITIONAL MASS

In Chapter 4, subjects performed maximum-effort curved sprinting with and without an additional mass. Despite a significant difference in the magnitude of the ground reaction force, the peak ankle total moment, provided mainly by the plantarflexion moment, remained unchanged. Meanwhile, a significant increase in the peak knee total moment, provided mainly by the extension moment, was observed when the sprints were performed with the mass compared to without. Based on these observations, it was speculated that the ankle moment generation might reach its limit; for sprints performed without the mass, some knee extension moment might be reserved.

Muscle force generation depends on the operating conditions – specifically the muscle length and contracting velocity. Information of such operating conditions can serve to strengthen or weaken the aforementioned speculations. For the ankle joint, if the contracting length and/or velocity changed between conditions, the speculation that the moment generation may be at its limit across conditions will be weakened. For the knee joint, if the contracting length and/or velocity differ between conditions, the observed change in the moment may no longer be interpreted as a reserved ability.

While direct measurements of muscle length and contracting velocity are not available in the current study, joint kinematics may help provide insight into the joint extensors’ operating conditions. In the analyses below, joint angle was used as an indicator for muscle length, and joint velocity was used to represent muscle contracting velocity. One-tailed paired t-tests were used for their higher statistical power, and the significance level was set a priori at $\alpha = 0.05$.

Joint angle and velocity at the instances where peak moments occurred were compared between conditions. No difference was detected for the ankle ($p = 0.1967$) or knee ($p = 0.3166$) joint angles (Figure A.1). Furthermore, no difference in joint velocity was observed for either joint (ankle: $p = 0.0656$; knee: $p = 0.1226$; Figure A.2). These results increase the confidence of the aforementioned speculations.
Figure A.1: Inside leg ankle and knee joint angles during ground contact. Thick lines represent sample averages and thin lines represent ± 1 s.d..

Figure A.2: Inside leg ankle and knee joint angular velocities during ground contact. Thick lines represent sample averages and thin lines represent ± 1 s.d.
APPENDIX B:  
ANKLE JOINT POWER AS A PERFORMANCE CONSTRAINT

In Chapter 4, the observation that the ankle total moment generation remained unchanged between conditions (control versus additional mass) led to the speculation that such moment generation ability may be at its limit.

An alternative performance-related variable to examine is the joint power generation. It is the rate at which the muscle-tendon-units surrounding the joint do work. During late stance, the foot undergoes push-off while generating positive ankle power. Would it be possible that during maximum-effort curved sprinting, the ability to generate ankle joint power reaches its limit and in turn constrains the overall performance?

To address this speculation, peak positive ankle power was calculated and compared between the control and additional mass conditions. If the ankle peak power is the performance constraint, it would be expected that this variable remains constant between the control and additional mass conditions. For the comparison, a one-tailed paired t-test was used, for its higher statistical power, at the significance level of $\alpha = 0.05$.

It was observed that the peak positive ankle power was significantly reduced in the additional mass condition compared to control ($p = 0.0147$; Figure B.1). This result does not support the speculation that the ankle power generation operates at its limit during maximum-effort curved sprints performed in various conditions.

![Peak ankle power](image)

Figure B.1: Peak ankle power (mean and s.d.). *Statistically significant difference.
APPENDIX C:
ANKLE MOMENT GENERATION AT VARIOUS EV-/INVERSION ANGLES

In Chapter 5, an attempt was made to alter the moment generation at the ankle joint - wedged footwear was used to change the joint configuration in the frontal plane. To explore the potential effects of ankle configuration on moment generation, a pilot study with dynamometer measurements was conducted. Three subjects performed maximum voluntary plantarflexion at various ev-/inversion angles. The subjects performed isometric contractions in a seated posture. Based on the kinematics of the supporting leg (Appendix A), the ankle joint was placed at 30° dorsiflexion and the knee at 40° flexion. Three trials of maximum-effort plantarflexion were performed for each of the 10° ankle eversion, neutral and 10° ankle inversion setups. The testing order was randomized. Statistical analysis was not performed due to the limited sample size.

Shown in Figure C.1 were the average values of the sample group. It can be seen that ankle ev-/inversion may reduce the maximum plantarflexion moment generation, with eversion having a relatively larger influence.

![Figure C.1: Maximum isometric plantarflexion moments generated at various ankle joint configurations.](image)