THE INFLUENCE OF SPRINT TECHNIQUE ON PERFORMANCE.

# Abstract

Despite clear differences in the techniques used by trained and non-trained sprinters, little research has focused on whether one technique has any benefit over the other and if so how that technique conveys an advantage. Therefore, the aim of this study was to develop a simulation model of sprinting and examine what influence technique had on performance.

A planar, seven segment, torque-driven model was developed to simulate the contact phase of a sprinting stride. The model was evaluated against data collected on a well-trained sprinter sprinting at 9.7 m.s-1. To examine the effects of different techniques a video of a team sport athlete and an elite sprinter running at their respective top speeds was digitised to specify swing leg joint angles in the model. The model was then optimised to maximise speed using these two different techniques.

The top speed of the model using the elite sprinter’s technique was 9% higher than when using the technique of the team sport athlete. The elite sprinter’s technique increased the average torque across the simulation at the knee and ankle joints, leading to an increase in average vertical force. The increase in torque was due to putting the torque generators in a more favourable position on their torque-angular velocity relationships.

These results show the importance that sprint technique has on performance.

# Introduction

The maximal running speeds attained by humans have been recorded across a broad range from 6.2–11.1 m.s-1 (Weyand et al., 2000) with elite sprinters in the Olympic 100 m final reaching speeds over 12 m.s-1 (Graubner and Nixdorf, 2011). This prompts the question of what allows some humans to attain much faster maximal speeds than others.

Running speed is the product of stride length and stride frequency. Therefore, faster speeds could be attained by either taking longer strides or by taking them more frequently. When running at a constant (maximal) speed on level ground, the runner must produce a net vertical force equal to bodyweight. As speeds increase, the portion of the stride in which force can be produced, the contact time, decreases, requiring greater vertical forces to be produced in these briefer contacts. All else being equal, greater vertical forces will result in longer aerial times and stride lengths. Alternatively, a runner could keep the vertical force the same but decrease the stride time so that this force is sufficient to support bodyweight over a shorter period, leading to increased stride frequencies. The maximal attainable speed could therefore be limited by either the ability to produce vertical force in shorter contacts or by the maximum speed in which legs could be cycled. Weyand et al. (2000; 2010) showed that in humans it is vertical force in relation to bodyweight that differentiates faster and slower runners, rather than the time it takes to cycle limb, the swing time, which remains roughly constant regardless of top speed. Further work has shown that it is not just the magnitude of vertical force that is different between faster and slower runners but the shape of the vertical force-time graph as well, with faster sprinters showing asymmetrical forces, peaking in the early portion of stance (Clark and Weyand, 2014).

As such, much previous research has focused on the stance limb during sprinting (Bezodis et al., 2008) and the physical characteristics of faster sprinters which allow them to produce greater forces, such as muscle size (Handsfield et al., 2017, Miller et al., 2020; Weyand and Davis, 2005), architecture (Abe et al., 2000; Kumugai et al., 2000) or musculoskeletal geometry (Lee and Piazza, 2009; Suga et al., 2020). In comparison, there has been a much smaller focus on the technical differences between faster and slower sprinters. This is surprising given clear differences exist between the two, particularly in relation to the swing leg. Trained sprinters show greater amounts flexion at the hip of the swing leg, resulting in the swing leg’s thigh being closer to vertical at contralateral touchdown and reaching a greater amount of hip flexion at contralateral toe-off (Bushnell and Hunter, 2007; Kunz and Kauffman, 1981; Mann and Herman, 1985; Sides, 2010). Further, when running at faster speeds the mechanical demands of the swing leg, mainly at the hip, rapidly increase (Dorn et al., 2012; Schache et al., 2015). These results indicate that although the time taken to reposition the limb does not vary with speed, how the limb moves during this portion of the stride does. It may therefore be expected that the technique of the swing leg has an important effect on sprinting performance.

Swing leg technique may improve sprint performance through its effect on the stance leg. As the swinging leg is accelerated upwards it will cause an equal acceleration downwards on the rest of the body. This acceleration downwards would put the extensor muscles of the stance leg into a faster eccentric or slower concentric contraction, more favourable for force production. This would allow a runner to generate more vertical force - like what is observed with the arm swing in standing (Cheng et al., 2008; Harman et al., 1990) and running jumps (Allen et al., 2010). As different kinematics of the swing leg would cause different accelerations on the rest of the body, one technique may benefit from this whereas another may not, due to the non-linearity of the force-velocity relationship. For example, elite sprinters have the swing leg thigh close to vertical at touchdown so that the centre of mass (CoM) of the swing leg would mostly accelerate upwards during early phase of contact, causing the stance leg to be put in more favourable contractile conditions. If the swing leg is behind the body at touchdown, as is seen with untrained sprinters, active hip flexion will cause a downwards acceleration on the swing leg during the early phase of contact, causing a subsequent upwards acceleration on the rest of the body and putting the stance leg in less favourable contractile conditions.

Another way swing leg technique could affect performance is through changing the CoM height of the runner. Previous work has assumed that the difference in CoM height between touchdown and toe-off is zero (Weyand et al., 2000). However, this is not strictly true as the CoM rises slightly during stance causing it to be higher at toe-off than at touchdown. An increased difference in CoM height would lead to a longer aerial time (and therefore swing time) for a given vertical velocity at toe-off (Figure 1). Alternatively, this would mean that a runner could have the same swing time but with a smaller vertical velocity at toe-off requiring less vertical impulse during stance – allowing them to run at faster speeds. It is interesting to note that having the thigh closer to vertical at touchdown, resulting in a smaller angle between the thighs, and a more flexed hip at contralateral toe-off, characteristics of elite sprinters’ technique, would result in a greater difference in whole-body CoM height between touchdown and toe-off (assuming the rest of the body’s motion was unchanged). This may therefore be a mechanism by which technique aids sprinting performance, increasing the difference in CoM height between toe-off and touchdown and therefore requiring less vertical impulse to generate a sufficient swing time. However, a larger difference in CoM height between touchdown and toe-off would result in a greater CoM velocity at the next touchdown, requiring more vertical impulse to produce the same vertical velocity at toe-off. How these two competing factors interact is unknown.

Figure 1 Treating a runner as point mass, the effect of the difference in CoM height between touchdown and toe-off (ΔSy) can be determined using the equations of constant acceleration for a given vertical velocity at take-off. The two vertical lines show what the aerial time would be with a ΔSy of 0 cm and -3 cm.



Despite the strong theoretical rationale there have been limited investigations into what effect technique might have on performance, and those that have were limited to descriptive differences, not able to explain why a certain technique may be better than another. This is probably due to the difficulty in trying to manipulate technique experimentally (Yeadon and Challis, 1994). A computer simulation however does not suffer from these drawbacks, making it a viable approach to investigate this. Therefore, the aim of this study was to explore the ways in which technique affects sprinting performance.

# Methods

## Experimental data

After giving informed consent to participate, one male sprinter (28 years, 91.1 kg, 1.86 m, 100 m pb: 10.50 s) sprinted at 9.7 m.s-1 on an instrumented treadmill (3DI, Treadmetrix, Utah, USA) recording three-dimensional ground reaction forces (2000 Hz). Sixty-five retroreflective markers were placed on the participant, to allow identification of joint centres, and their positions recorded (250 Hz) using sixteen infrared cameras (Vicon Vantage, Oxford Metrics, Oxford, UK), synchronised with the force data. A custom Vicon BodyBuilder model, composed of the trunk, thigh, shank, rearfoot, and toes, was written to extract segment and joint angles and then exported into MATLAB (R2020a, Mathworks, MA, USA) along with the calibrated forces for further analysis.

Kinetic and kinematic data were both interpolated and resampled on the same time base at 1000 Hz, using cubic splines. Six full strides on each leg were time normalised then averaged across strides and legs to give data on one full stride at top speed. Contact was identified using an 80 N threshold in vertical force and the force was set to 0 N during aerial phases.

To determine the effect of differing techniques, a publicly available video of an elite sprinter and a team sport athlete running at their respective top speeds was used (https://www.youtube.com/watch?v=3exj1tlEjaQ&list=FLQ08DdH24EnUwvy9hYFpEaw&index=1&ab\_channel=LocomotorLabSMU), as it illustrated the major differences in technique. The location of the hip, knee, ankle and metatarsophalangeal (MTP) joint centres along with a point at the base of the neck representing the top of the trunk, were digitised using the open-source package DLTdv7 (Hedrick, 2008). From which, segment and then joint angles were calculated and quintic smoothing splines fitted to the angle data so that they could be used in the model.

## Computer model

### Model construction

The model was constructed through an iterative process, beginning simple and progressively adding complexity until it was sufficient for the current purposes (Alexander, 2003; Yeadon and King, 2018). The model was two-dimensional, focusing on the sagittal plane, and composed of seven rigid segments connected by frictionless pin joints. These segments represented the thigh, shank, rearfoot, and toe segment of the stance leg, the thigh and shank for the swing leg, and a combined head, arms and, trunk (HAT) segment, comprising the mass of the rest of the body (Figure 2). Given that the trunk is not completely rigid, and to include the effects of the arm swing, the CoM of the HAT segment was allowed to vary. Its position from the hip joint was specified in the model by another quintic spline, determined from the experimental data.



Figure 2. The seven-segment model developed with viscoelastic springs under the MTP joint and toes. Open circles represent torque-driven joints, with torque generators at the hip, knee, and ankle and a viscoelastic spring at the MTP joint. Swing hip and knee joints are angle-driven by quintic splines.

The foot-ground interface was modelled using viscoelastic springs at the end of the MTP joint and the end of the toes. The force at each contact point was calculated as:

Where x,y are the vertical and horizontal spring compressions and ki, bi are the spring stiffness and damping coefficients. The damping component of the vertical force was multiplied by the absolute spring depression to avoid negative force values while the spring was still stretched.

The hip, knee, and ankle joint of the stance leg was actuated by one flexion and one extension mono-articular torque generator, representing the contractile component (CC) of the muscle-tendon unit, in series with a rotational, linear spring, representing the series elastic component (SEC). The MTP joint was actuated by a viscoelastic spring, the equation for which is the same as the vertical force at each contact spring. The hip and knee joint of the swing leg were angle-driven, with the angle specified by either the data from the participant or from the digitised video of the two techniques

The torque produced at each torque generator was calculated as:

Where is the activation of the torque generator, T0 the maximum isometric torque, Tv and Ta are the normalised torque-angular velocity and angle relationships. The torque-angle relationship was taken from Forrester et al. (2011) and the torque-angular velocity relationship was taken from Yeadon et al. (2006), with the exception of differential activation. Differential activation is usually included to represent the lower muscle activity commonly seen in eccentric versus concentric movements (Westing et al., 1991) and was not included for two reasons. Firstly, it was likely much smaller during a movement well accustomed to the participant (e.g., sprinting for a well-trained sprinter) than in an isokinetic dynamometer where the measurements were taken, as increases in muscle activity during eccentric contractions are seen with training (Hortobagyi et al., 1996). Secondly, by removing the differential activation it allowed a closed-form solution to calculate the angular velocity from the CC torque, which helped when including series elasticity.

At each timestep the CC angle was calculated via numerical integration from the CC angular velocity. This then allowed the SEC angle to be calculated, from which the torque could be found by multiplying the SEC angle by the angular stiffness. By assuming the torque in the SEC was equal to the torque in the CC, the CC angular velocity could then be found, and the process repeated. This left just the initial CC angle to be determined which was done by assuming, initially, that the CC angular velocity equalled the joint angular velocity and numerically solving for the CC angle using the binary split method.

Activations of the torque generators were specified using the quintic function from Yeadon and Hiley (2000) with a slight modification. Each quintic function specified a ramp from one activation level to another, and so given an initial activation level, three parameters were required to define the function to the new activation level, the ramp onset time, the ramp time and the new activation level. The functions were pieced together such that the next ramp began at a specified time from the end of the previous ramp. In preliminary tests it was found that two ramps allowed the torque generators sufficient freedom, while keeping computing time low.

Input to the model was the initial segment angles and angular velocities, the initial position and velocity of the end of the toe segment and the parameters defining the activations for each torque generator. As CoM velocity was a more useful input this was specified and then the position and velocity of the end of the toe calculated. Given the participant was running on a treadmill and so would have small horizontal velocities in the laboratory frame, input was the measured CoM velocity plus the belt speed. Simulations began at touchdown of the stance leg and finished at take-off when the vertical force dropped to zero.

The equations of motion for the model were generated in Autolev (Kane and Levinson, 1985) and the Fortran code generated was then modified to include additional features of the model e.g., torque generator profiles. The equations of motion were integrated forwards using the fourth order Runge-Kutta method with a variable timestep.

### Parameter determination

Segments’ inertial properties were calculated using the geometric model of Yeadon (1990) for a previous study on the same participant and were left unchanged as there was only a small difference in the participant’s body mass on the two occasions. As torque generator parameters and estimates of SEC angular stiffness were unavailable, they were taken from a triple jumper with a similar performance level to the participant used here (Allen, 2009), and scaled by body mass. SEC stiffness were scaled so that they maintained the same stretch at T0. Parameters for the viscoelastic springs were determined during the model evaluations (see Section 2.3).

## Model Evaluation

The model was evaluated against the experimental data collected on the participant to assess how well it could match the contact phase. A simulated annealing algorithm, set to run in parallel (Higginson and Anderson, 2005), minimised a cost function by varying 42 activation parameters, 2 MTP spring and 8 contact spring parameters. The cost function to be minimised is shown below:

Where θtrunk is the mean squared error in global orientation of the model, θjoint the sum of the mean squared errors in hip, knee, ankle and MTP joint angles, tswing the absolute difference in swing time and, vcmx the absolute difference in horizontal CoM velocity. Aerial time was calculated, using the equations of constant acceleration, from the CoM vertical velocity at the end of a simulation and the difference in height of the CoM between take-off and touchdown. Horizontal CoM velocity was calculated from the step length multiplied by the step frequency.

The initial temperature for Simulated Annealing was chosen to give an appropriate step length so that roughly 50% of all function evaluations were accepted (Goffe et al., 1994) and NT, the number of cycles before a temperature reduction, was determined through preliminary tests aiming to minimise the computing time while still finding the optimum. All other parameters were left at their default values. Upper bounds on the activation level were 1.0. Ramp times for the activation profiles were given a lower bound of 100 ms (Tillin et al., 2012). The upper bound for the stiffness of spring at the MTP joint was 300 N.m and the contact spring parameters were not constrained in anyway. While this may lead to unrealistic spring compression during contact, it would account for the lack of compliance elsewhere in the model (Allen et al., 2012).

## Optimisation

Optimisations were carried out to assess how fast the model could run while still producing the necessary swing time. With the spring parameters kept the same as in the evaluation, the 42 activation parameters were again varied. The cost function to be minimised is shown below:

Where tsw is the absolute difference in swing time, between the performance and the simulation and vcmx the absolute difference in horizontal CoM velocity. If the model could match the swing time to within 0.001 s and there was not a drop in horizontal velocity of more than 0.01 m.s-1 between the start and end of the simulation, bounds picked from the variation in the experimental data, then the model was said to be able to run at that speed. The speed was then increased, and the optimisation repeated until the model could not meet those conditions. Bounds on the activation parameters were kept the same as in the evaluation. An optimisation was first performed using the spline specifying the swinging limb angles from the matching data. Optimisations were then performed to compare the two techniques; everything was kept the same between optimisations except the spline coefficients used to drive the swing leg.

# Results

## Evaluation

The matching simulation showed good agreement with the experimental data (Figure 3). Overall orientation and configuration angles were all matched to within a RMSE of 5° or less (Figure 4). The swing time was matched exactly; however, the contact time was shorter than what was measured, and the aerial time was longer. Despite not being explicitly included in the cost function the vertical impulses and the raise in CoM height during stance showed good agreement with the data (Table 1). Overall, it was concluded that the model was sufficient to model the stance phase of sprinting and could be used to explore the effect of technique on performance.

Table 1. Comparison between the performance and the matching simulation. Δscmy is the difference in CoM height between take-off and touchdown.

|  |  |  |
| --- | --- | --- |
|  | Performance | Matching |
| Horizontal velocity (m.s-1) | 9.67 | 9.66 |
| Swing time (s) | 0.374 | 0.374 |
| Contact time (s) | 0.110 | 0.091 |
| Aerial time (s) | 0.132 | 0.142 |
| Δscmy (m) | 0.019 | 0.020 |
| Vertical impulse (N.s) | 108.3 | 112.3 |



Figure 3 Comparison between participant (top) and matching simulation (bottom).



Figure 4 Orientation and configuration angles of the participant (squares) and matching simulation (solid line).

## Optimisations

Optimising the activation parameters improved the top speed of the model using the participant’s technique by 4% to 10.0 m.s-1. Optimising the activation parameters of the model using the technique of the team sport athlete led to a top speed of 9.2 m.s‑1 compared to 10.0 m.s‑1 when using the sprinter’s technique, a 9% difference. Swing time remained the same for both techniques, with only slight differences in contact and aerial times and vertical impulses. The sprinter’s technique resulted in the CoM of the model being lower at touchdown and higher at toe-off, but the change in CoM height was not different with the two techniques (Table 2).

Using the technique of the sprinter, the model produced 90 N (0.1 BW) more vertical force. This increase was due to higher forces in the first half of stance, offsetting lower forces during the second half, giving a net positive effect (Figure 6). Average extensor torques at the knee and ankle were higher using the sprinter’s technique, whereas the average extensor torque at the hip was lower (Table 2).

Activations of the torque generators were virtually identical with the two techniques, with the hip and ankle extensors being fully activated and the flexors inactivated for the duration of stance, resulting in net extension torques at these joints. Both the extensor and flexor torque generators were activated at the knee, leading to a period of net extension then flexion torques (Figure 7). As well as the extensor torque being higher with the sprinter’s technique, the average flexion torque was also lower, despite similar activations across techniques. This led to an even greater net extension torque during the first half of stance (Figure 7).

CC angular velocities were more eccentric at the knee and ankle but was more concentric at the hip with the sprinter’s technique (Figure 9). There was also some evidence that the higher eccentric CC angular velocities put the CC angle closer to its optimal position.

Net joint torques at the swing hip and knee also differed with technique (Figure 8). The technique of the sprinter showed a greater peak in hip extension towards the end of the stance, compared with the technique of the team sport athlete who had a greater flexion torque at the start of the stance. The team sport athlete’s technique also showed a much larger knee extension torque.

Table 2. Comparison between the optimised simulations using the technique of the elite sprinter and the team sport athlete. Vertical force and torques represent the average value over the simulation.

|  |  |  |
| --- | --- | --- |
|  | Sprinter | Team sport |
| Horizontal velocity(m.s-1) | 10.0 | 9.2 |
| Swing time (s) | 0.374 | 0.374 |
| Δscmy (m) | 0.021 | 0.22 |
| Contact time (s) | 0.088 | 0.092 |
| Aerial time (s) | 0.143 | 0.141 |
| Vertical impulse (N.s) | 111.7 | 109.8 |
| Vertical force (N) | 2123 | 2033 |
| Hip extensor torque (N.m) | 138.2 | 168.6 |
| Knee extensor torque (N.m) | 208.8 | 186.3 |
| Ankle extensor torque (N.m) | 277.2 | 267.8 |



Figure 5 Comparison between optimised simulations using the sprinter’s (top) and team sport athlete’s (bottom) technique.



Figure 6 Vertical force in the optimised simulations using the sprinter’s (dashed) and team sport athlete’s (dotted) technique.



Figure 7 Net joint moments in the optimised simulations using the sprinter’s (dashed) and team sport athlete’s (dotted) technique. Extension torques are positive



Figure 8 Swing leg torques in the optimised simulations using the sprinter’s (dashed) and team sport athlete’s (dotted) technique. Extension torques are positive.



Figure 9 CC angular velocities in the optimised simulations using the sprinter’s (dashed) and team sport athlete’s (dotted) technique. Positive angular velocities represent eccentric contractions.

# Discussion

The main outcome of the study showed that a 9% improvement in speed could be gained just from changing how the swing leg moves during stance (Table 2). The improvement in speed came from an increase in the vertical ground force of 90 N, or 0.1 BW, allowing the necessary vertical impulse to be generated in a shorter stance phase (Table 2). This increase in force produced was despite the strength of the model remaining constant across all simulations, indicating the torque generators were operating more optimally.

As activations of the torque generators were almost identical when using the different techniques, the increase in torque came from the torque generators operating at more favourable conditions for torque production, i.e., angles closer to optimal and faster eccentric or slower concentric angular velocities. There were clear differences showing that the sprinter’s technique put the ankle and knee CC in more eccentric angular velocities (Figure 9), but also that the higher eccentric angular velocities put the CC in more optimal angles too. The opposite was the case at the hip however as the faster concentric angular velocity of the extensor CC meant that torque was reduced (Table 1, Figure 9). As there was still a next increase in vertical force with the sprinter’s technique, this could mean that the hip torque contributes little to the vertical force.

These results can explain the asymmetrical vertical force patterns observed in sprinters. The action of the swing leg increases torque mainly when the knee and ankle are operating eccentrically, during the early part of stance, which is where the pattern of vertical force mainly differs (Clark and Weyand, 2014). The greater torques could result in a greater deceleration of the lower limb of the stance leg, which has been suggested to be the cause of the asymmetrical force pattern (Clark et al., 2014; 2017). In addition, Clark and Weyand (2014) found that non-sprinters showed symmetrical force patterns, even when running at maximal speeds. As the non-sprinters were likely to not have received much technical training, this points to the technique used being responsible for the asymmetry. Together these results indicate that the technique used by sprinters puts their muscles of the stance leg in more favourable conditions for force production, allowing them to produce more force early in stance and giving rise to the observed asymmetrical force-time pattern.

The torques of the swing leg also showed differences with the two techniques (Figure 8). The technique of the team sport athlete resulted in a greater hip flexion torque during early stance as the swing thigh was behind the body and needed to be brought through. This greater hip flexion torque could result in the less favourable conditions for the stance leg torque generators as it would accelerate the swing leg down. As only the stance phase was simulated, whether greater hip flexion torques were required to get the swing leg into its initial conditions for the sprinter’s technique is not known but is likely to be the case. The sprinter’s technique did require much greater hip extension torques to decelerate the swinging limb, which may explain why the hip extensors are found to be much larger in elite sprinters than other athletes (Handsfield et al., 2017; Miller et al., 2020). The hip extension torque towards the end of the stance caused the model’s global orientation to become more vertical, with the lack of it when using the team sport’s athlete resulting in the more horizontal trunk (Figure 5). The large knee extension torque when using the technique of the team sport athlete may not be solely from the muscles as the greater degree of knee flexion used would probably introduce large contributions from passive structures such as ligaments or soft tissue.

There was not an effect of technique on the difference in CoM height (Table 2). Surprisingly, this was despite the CoM of the swing leg rising ~9 cm with the sprinter’s technique but lowering ~5 cm with the team sport athlete’s technique. This might again be due to the swing leg’s effect on the rest of the model. The greater upwards acceleration of the sprinter’s technique, accelerating the rest of the body down, would prolong contact with the ground. Conversely, the team sport athlete’s technique would therefore result in a shorter contact meaning that the stance leg was more upright by toe-off, offsetting the lower swing leg CoM. Whether this was due to the model being too simplistic (e.g., specifying the HAT’s segment CoM relative to the hip) or is what is seen with different sprint technique is unknown as experimental data is lacking. The less upright stance leg when using the sprinter’s technique meant the CoM travelled further during stance which may also benefit performance (Weyand et al., 2000).

The main limitation of the model was that it was unable to match contact times well (Table 1). This may be expected given the rigidity of the model, such as the lack of soft-tissue movement (Pain and Challis, 2006) or using pin joints. This also meant that the force rose unrealistically fast, leading to higher average vertical forces than may be expected (Weyand et al., 2000). The overall effect was the same though, as impulses matched well (Table 1). Further, as this was a comparison of the two techniques it should not affect the results. Another issue with the shorter contacts was that it meant the angle-driven joints did not complete the same movement as they did from the digitised video, as they were functions of time. The contact times were long enough though, so that the majority of the movement was complete and any effect on the rest of the body was small at this point.

The top speeds reached in the optimisations are likely a slight overestimation of the true top speeds. This is because the optimisations were not constrained in any way to match joint kinematics or torque generator activations at take-off, whereas during an actual contact phase the stance leg will also need to prepare for swing. It is likely that the hip flexors will ramp up during the second half of stance and the extensors ramp down, allowing the leg to be swung forwards (Bezodis et al., 2008; Sides, 2014). However, the magnitude of this is likely to be similar for both the elite sprinter’s and the team sport athlete’s technique so the relative difference between the two techniques would remain the same. The high CC angular velocities of the extensor torque generators mean that the torque produced would be small anyway.

Although the model was based on one participant, the inertia, strength, and sprinting data were all taken at different times, possibly leading to inconsistencies in the model. However, the aim of this study was not to optimise the technique of the participant, which would require accurate model parameters, but to compare the effects of two different techniques associated with faster and slower sprinters and therefore parameters just needed to be typical of an athlete. Optimisation of the model using the technique of the participant resulted in a top speed of 10.0 m.s-1, which was not unrealistic and close to the top speed they achieved on the day of data collection (9.9 m.s-1).

In conclusion, the technique of a sprinter can have a large effect on their top speed. The benefit of using the technique of an elite sprinter is that it puts the muscles of the stance leg in faster eccentric contractions, allowing more force to be produced in the early portion of stance.

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