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The influence of swing leg motion on maximum running speed

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FAO: Professor Farshid Guilak: Editor

I enclose a manuscript entitled “The influence of swing leg motion on maximum running speed” for publication as a Full Length Article in the Journal of Biomechanics. Both authors have made substantial contributions to all of the following: (1) the conception and design of the study, or acquisition of data, or analysis and interpretation of data, (2) drafting the article or revising it critically for important intellectual content, (3) final approval of the version to be submitted. The authors received no writing assistance. The manuscript, including related data, figures and tables has not been previously published and is not under consideration elsewhere.

Yours sincerely

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The influence of swing leg motion on maximum running speed

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Abstract

The motion of the swing leg of elite sprinters at maximum speed is markedly different from that of slower sprinters, but the mechanisms by which this influences performance are unknown. The aim of this study was to use a computer simulation model to establish whether and, if so, how the motion of the swing leg influences maximum achievable running speed. A seven-segment planar computer model was constructed to simulate the stance phase of sprinting. Optimisation was used to maximise the running speed of the model using two different swing leg techniques, one representative of an elite sprint athlete, and the other of a college athlete. The maximum speed of the model increased when using the swing leg technique of the elite athlete compared with the technique of the college athlete (10.0 $m.s^{-1}$ vs 9.2 $m.s^{-1}$). This improvement in performance was due to greater horizontal displacement of the mass centre during stance (0.867 m vs 0.837 m), and an increase in average vertical ground force of 90 N (0.1 bodyweights). The increase in vertical force was due to a larger impact peak caused by more negative vertical momentum of the stance leg at touchdown, and subsequently greater torques in the joints of the stance leg which were placed in faster eccentric contractions and at angles closer to optimum during the first half of stance. It is likely that the swing leg technique contributes to the asymmetrical vertical ground reaction force traces observed in elite sprinters.

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1. Introduction

Maximal running speed in humans demonstrates great variability, Weyand et al. (2000) measured maximum speeds ranging from 6.2 to 11.1 $m.s^{-1}$ in a group of thirty three physically active men and women, and speeds of over 12 $m.s^{-1}$ have been recorded by elite male athletes in competition (Graubner and Nixdorf, 2011). This prompts the question of what allows some humans to run so much faster than others.

Running speed is the product of step length and step frequency and therefore faster speeds can be attained by increases in either or both. It has been observed that, at relatively low speeds, runners increase speed primarily by increasing step length (Weyand et al., 2000; Dorn et al., 2012). They achieve this via increases in vertical ground forces which lead to increased vertical impulses and consequently flight times (Weyand et al., 2000; Dorn et al., 2012). At relatively higher speeds, the capacity to apply vertical ground forces plateaus and briefer stance times result in reduced vertical impulses and flight times; increases in speed after this point are due primarily to increases in step frequency, which continue until maximum speed is reached (Weyand et al., 2000). Running speed is therefore theoretically limited by the ability to produce vertical ground force during brief stance times and the minimum swing time during which the limb can be repositioned for its subsequent stance phase. Weyand et al. (2000; 2010) showed that in humans it is the size of the vertical ground forces that differentiates between faster and slower runners, rather than the minimum swing time, which is similar for all runners at their top speeds. Furthermore it has subsequently been shown that it is not just the magnitude of the vertical ground force that differentiates between faster and slower runners, but also the shape of the vertical ground force-time graph which peaks earlier in stance for faster sprinters (Clark and Weyand, 2014).

Clear differences exist between the techniques of faster and slower sprinters, particularly in relation to the swing leg; the hip angle of the swing limb at touchdown of the contralateral leg has been shown to differentiate between elite and sub-elite sprinters and to be the largest single kinematic predictor of maximum running speed, despite exhibiting no differences in angular displacement and angular velocity throughout the stride (Sides, 2014). Trained sprinters exhibit a more flexed hip throughout the first half of swing, resulting in the thigh of the swing leg being closer to vertical at touchdown of the contralateral leg and closer to horizontal at contralateral toe-off (Kunz and Kauffman, 1981; Mann and Herman, 1985; Bushnell and Hunter, 2007; Sides, 2014). Furthermore, as running speed increases, the mechanical demands on the muscles of the swing leg, mainly at the hip, substantially increase, whereas those on the muscles of the stance leg plateau (Dorn et al., 2012; Schache et al., 2015) and it has been shown that muscle volumes of the hip flexors and extensors are extremely large in sprinters when compared with controls (Handsfield et al., 2017, Miller et al., 2020). These results indicate that, although the time taken to reposition the limb at maximum speed is similar in faster and slower runners, the kinematics and kinetics of the limb during this portion of the stride are different.

It has been shown that the motion of the swinging limbs can improve performance in standing (Harman et al., 1990; Ashby and Delp, 2006; Cheng et al., 2008; Domire and Challis, 2010) and running jumps (Dapena and Chung, 1988; Allen et al., 2010). In a vertical jump an arm swing can improve performance by increasing the height and velocity of the whole-body centre of mass (CoM) at takeoff (Cheng et al., 2008; Domire and Challis, 2010). The increase in takeoff velocity caused by the arm swing is partly due to energy imparted directly to the CoM and partly caused by a downwards reaction force which augments the force of the extensor muscles of the stance legs by putting them in faster eccentric or slower concentric conditions (Cheng et al., 2008; Domire and Challis, 2010). The performance requirements of maximum speed running are different to vertical jumping in that the athlete is not attempting to maximise vertical velocity but to achieve sufficient vertical velocity to allow adequate flight and hence swing time to reposition the limb. However, it is conceivable that runners could use their swing leg to increase vertical ground force in the same way as an arm swing in vertical jumping, and therefore achieve the requisite vertical impulses during briefer stance phases, facilitating higher running speeds.

Despite the observed differences in technique between faster and slower runners and this theoretical rationale, there has been limited consideration of the effect swing leg motion might have on maximum running speed. Therefore, the aim of this study was to use computer simulation to investigate the influence of swing leg motion on maximum achievable running speed.

2. Methods

2.1. Study design

A computer model was developed to simulate the stance phase of sprinting. Optimisation was used in two ways: to evaluate the capacity of the model to match performance data collected on a sprinter; then to maximise the running speed of the model using two different swing leg techniques, one representative of an elite athlete (ET), and the other of a college athlete (CT).

2.2. Experimental data

Following Loughborough University Ethics Approvals (Human Participants) Sub-Committee guidelines and having given informed consent, a male athlete (28 years, 91.1 kg, 1.86 m, 100 m personal best time: 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, and their positions were 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 head and trunk, upper and lower arm, thigh, shank, rearfoot, and toes, was written to extract segment and joint angles. Further analysis was performed in MATLAB (R2020a, Mathworks, MA, USA).

Kinetic and kinematic data were low-pass filtered above 50 Hz and 15 Hz respectively, using a fourth-order zero-lag Butterworth filter, 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 between legs to give representative data on one full stride at top speed. Contact was identified using an 80 N threshold in vertical force.

To determine the effect of different swing leg techniques, a publicly available video of an elite athlete and a college athlete running at their respective top speeds was used (LocomotorLabSMU, 2012), as it illustrated typical differences in technique, with the hip joint of the elite athlete more flexed throughout stance. 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). Segment and joint angle time histories were calculated, and quintic smoothing splines were fitted to these time histories.

2.3. Computer model

A seven-segment planar computer model was constructed to simulate the stance phase of sprinting (Figure 1). The seven segments comprised: the thigh, shank, rearfoot, and toe of the stance leg; the thigh, and shank of the swing leg; and a combined head, arms, and trunk (HAT) segment, containing the mass of the rest of the body. Segments were represented as one-dimensional rigid rods, except for the rearfoot which was two-dimensional, and were connected by frictionless pin joints. The foot-ground interface was modelled using horizontal and vertical spring-dampers situated at the MTP joint and the end of the toe (Allen et al., 2012).

*** insert Figure 1 here ***

The hip, knee, and ankle joints of the stance leg were actuated by mono-articular flexion and extension torque generators, representing the contractile components (CC) of the muscle-tendon units crossing the joint, in series with a rotational spring, representing the series elastic components (SEC). The MTP joint was actuated by a rotational spring-damper. The hip and knee joints of the swing leg were angle-driven using experimental data. To represent the arm swing, the position of the CoM of the HAT segment relative to the hip was also kinematically driven using experimental data.

The torque produced by each torque generator was a product of maximum isometric torque, activation level (Yeadon and Hiley, 2000), and torque-angle (Forrester et al. 2011) and torque-angular velocity (Yeadon et al. 2006) relationships of the contractile component.

System equations of motion were formulated using Autolev (Kane and Levinson, 1985) and output as Fortran code. Inputs to the model were the initial conditions, representing CoM position and velocity and segment angles and angular velocities, and parameters specifying the activation of each torque generator as a function of time. The equations of motion were integrated forwards using a fourth order Runge-Kutta algorithm with a variable timestep.

Segmental inertia parameters were obtained from a previous data collection on the same participant (McErlain-Naylor, 2017). Torque generator parameters were taken from a previous study on a triple jump athlete (Allen, 2009), and scaled linearly to body mass. Viscoelastic parameters were determined via optimisation (see Section 2.4).

2.4. Model Evaluation

The model was evaluated by minimising the difference between simulation output and experimental data collected on the participant. A simulated annealing algorithm, set to run in parallel (Higginson and Anderson, 2005), minimised a cost function by varying forty-two activation parameters, two MTP spring and eight contact spring parameters. The cost function comprised the sum of four components: the sum of the mean squared differences in hip, knee, ankle and MTP joint angle time histories; the mean squared difference in the global orientation angle time history; the absolute difference in horizontal CoM velocity; and the absolute difference in swing time. The differences in orientation angle, CoM velocity, and swing time were weighted one, two, and three orders of magnitude respectively higher than the differences in joint angles with the aim of prioritising the match to the performance outcomes of swing time and horizontal velocity.

Flight time was calculated from the CoM vertical velocity at take-off and the difference in height of the CoM between take-off and touchdown using equations of constant acceleration. Swing time was the sum of two flight times and one stance time. Horizontal CoM velocity was the product of step length and 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 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.

Viscoelastic parameters obtained from this evaluation process were used in all subsequent optimisations.

2.5. Optimisation

Optimisations were carried out to establish the maximum running speed achievable by the model whilst angle-driving the swing leg with each of the two techniques from video data (LocomotorLabSMU, 2012). To establish maximum running speed, initial horizontal velocity was incrementally increased by 0.1 $m.s^{-1}$ from the matched simulation whilst all other initial conditions remained the same. After each velocity increment, forty-two activation parameters were varied by the simulated annealing algorithm to minimise the loss of horizontal velocity during stance, whilst matching the swing time from experimental data, since this has been shown to be similar for all runners at their top speeds (Weyand et al., 2000). If the loss in horizontal velocity during stance was less than 0.01 $m.s^{-1}$ and the swing time was matched to within 0.001 s, bounds picked from the variation in the experimental data, then the model was considered capable of running at that speed. Maximum speed was the highest speed at which the model could meet these conditions.

3. Results

3.1. Evaluation

The matched simulation showed good agreement with the experimental data (Figure 2). Horizontal velocity (9.66 $m.s^{-1}$) was closely matched to the experimental data (9.67 $m.s^{-1}$). The swing time matched exactly (0.374 s) however, stance time was shorter (0.091 s vs 0.110 s), and flight time was longer (0.141 s vs 0.132 s) than the experimental data. Orientation and configuration angle time histories were all matched to within a root mean squared error (RMSE) of 5° or less (Figure 3). It was concluded that the model was sufficiently accurate to be used to explore the effects of technique on performance.

*** insert Figure 2 here ***

*** insert Figure 3 here ***

3.2. Optimisation

The maximum speed of the model when using CT was 9.2 $m.s^{-1}$ compared with 10.0 $m.s^{-1}$ when using ET (Figure 4). Stance time was slightly shorter and flight time and vertical impulse slightly larger using ET (Table 1). The vertical whole-body CoM displacement did not differ between the two techniques, but the horizontal displacement was greater in ET (Table 1).

*** insert Figure 4 here ***

*** insert Table 1 here ***

Using ET, the model produced an average of 90 N (0.1 bodyweights) more vertical ground force during stance. 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 5). When using ET, the initial vertical momentum of the stance limb was more negative (-32.1 $kg.m.s^{-1}$ vs -28.5 $kg.m.s^{-1}$) which led to a larger vertical impact force peak (Figure 5). Average extensor torques at the knee and ankle were higher using ET, whereas the average extensor torque at the hip was lower (Table 1).

*** insert Figure 5 here ***

Activations of the stance leg torque generators were virtually identical in both techniques, with the hip and ankle extensors fully active and the flexors inactive 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 followed by flexion torques (Figure 6). As well as the extensor torque being higher in ET, the average flexion torque was also lower, despite similar activations between techniques. This led to an even greater net extension torque during the first half of stance (Figure 6).

*** insert Figure 6 here ***

Angular velocities of the CC were more eccentric at the knee and ankle but more concentric at the hip in ET (Figure 7). The higher eccentric CC angular velocities put the CC angle closer to its optimal position at the knee and ankle for the majority of stance (Figure 7).

*** insert Figure 7 here ***

4. Discussion

The results showed that a 0.8 $m.s^{-1}$ or 9% improvement in running speed can be gained solely from changing how the swing leg moves during stance (Table 1). The increase in speed was a result of an increase in the average vertical ground force of 90 N, or 0.1 BW, allowing the necessary vertical impulse required to maintain swing time to be generated in a briefer stance phase (Table 1).

The vertical ground force impact peak was found to be larger when using ET. It has been shown that the size of the vertical impact force peak during running can be accurately predicted using momentum changes of the lower leg (Clark et al., 2014). Since the vertical momentum of the whole-body CoM at touchdown remained constant for all simulations, the more positive swing leg momentum in ET resulted in a more negative stance leg momentum and consequently a larger impact peak. This was a passive effect since stance leg joint torques did not differ during impact (Figures 5 and 6).

As activations of the torque generators were almost identical between the two techniques, the increases in torque following impact were due to CCs operating at angles closer to optimal and at faster eccentric or slower concentric angular velocities (Figure 7). Faster eccentric CC angular velocities at the ankle and knee which placed the CC further up the ascending limb of the torque-angle curve were seen in ET (Figure 7). The opposite was the case at the hip however as the higher concentric angular velocity of the extensor CC meant that torque was reduced (Table 1, Figure 7). As there was a net increase in vertical force with ET, this is consistent with previous findings that the hip torque contributes little to the vertical ground force (Dorn et al., 2012).

When considered together, these results indicate that swing leg motion could contribute to the asymmetrical vertical ground force patterns observed in sprinters. The action of the swing leg increased ground force and stance leg torque mainly during impact and when the knee and ankle were operating eccentrically during the early part of stance (Figures 5 and 6), which is where the pattern of vertical force in sprinters differs substantially from the symmetrical pattern predicted by a spring-mass model (Clark and Weyand, 2014).

The vertical whole-body CoM displacement during stance was similar between techniques (Table 1). This was despite the CoM of the swing leg rising 7.3 cm with ET but lowering 3.4 cm with CT. The positive vertical momentum of the CoM of the swing leg in ET reduced the vertical momentum of the rest of the body, delaying takeoff (Figure 4) and leading to a larger horizontal displacement of the whole-body CoM (Table 1). Prolonging stance results in a reduction in the required flight time and hence vertical impulse to maintain a given swing time. It is also possible that the increased hip flexion angle and gravitational potential energy of the swing leg in ET will facilitate an increase in the negative vertical momentum of the limb and consequent impact force at the subsequent touchdown.

The mechanical demands on the swing limb required to achieve ET were beyond the scope of this study, but joint torques and work done by the swing leg have been shown to increase superlinearly with running speed (Dorn et al., 2012) and the muscles which flex and extend the hip, and flex the knee, have been shown to be substantially larger in sprinters compared with controls (Handsfield et al., 2017, Miller et al., 2020). These findings, in addition to the results of this study, would suggest that athletes wishing to increase their running speed should devote time to technical and strength training associated with the swing phase of running.

The main limitation of the model was that it was unable to match stance time closely (Table 1). This is likely to be due to a lack of compliance in the model, which has been shown previously to result in briefer stance times in simulations of triple jumping (Allen et al., 2012). This also meant that the vertical ground reaction force rose unrealistically quickly, leading to higher forces than are typically observed in sprinters (Weyand et al., 2000). Nevertheless, the stance times were within the range of those reported in the literature (Sides, 2014) and the model was able to match performance outcomes well; since the aim was to compare two techniques it should not affect the study conclusions. Another limitation of the briefer stance times was that the angle-driven joints did not complete the same movement amplitude as the performance, since they were functions of time, however since most of the differences occurred early in stance this is also unlikely to have affected the conclusions. The optimisations were not required to prepare the stance limb for its subsequent swing phase, and it has been shown in sprinters that the hip flexors activate, and the hip extensors relax during late stance (Bezodis et al., 2008) which was not observed in this study, but this constraint would be likely to affect the outcomes in ET less since the hip torque did decay to zero during late stance in this simulation (Figure 6). Lastly, the model inertia parameters and the performance data were taken at different times from the same athlete, and the strength parameters and swing leg techniques were taken from different athletes, which may seem inconsistent. In this regard the aim of this study was not to optimise the technique of one participant, but to compare the effects of contrasting swing leg techniques typically associated with faster and slower sprinters, and to achieve this aim we believed that the model parameters needed only to be representative of a sprint athlete.

5. Conclusion

The aim of this study was to investigate the influence swing leg motion has on maximum achievable running speed. The results showed that using the swing leg technique of an elite sprinter can augment the vertical ground force passive impact peak by increasing the negative vertical momentum of the stance leg at touchdown, put the muscles of the stance leg in faster eccentric contractions placing them closer to their optimum length and allowing more torque and vertical ground force to be produced in the early portion of stance, increase the horizontal displacement of the mass centre during stance, and ultimately increase the maximum running speed. We recommend that athletes aiming to increase their maximum running speed devote time to training associated with the swing phase of running. Future work could investigate the effect of swing leg technique on the subsequent stance phase, and the contribution of the arm swing to increases in ground force.

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List of figure captions

Figure 1. Seven-segment model with viscoelastic springs under the MTP joint and toes. Open circles represent torque-driven joints. Filled circles represent angle-driven joints. Grey circle represents a viscoelastic rotational spring. Half-filled circle indicates torque- and angle-driven hip joints of the stance and swing leg, respectively.

Figure 2. Visual comparison between participant (top) and matched simulation (bottom). Circle represents whole-body CoM.

Figure 3. Orientation and configuration angles of the participant (squares) and matched simulation (solid line) techniques.

Figure 4. Visual comparison between optimised simulations using ET (top) and CT (bottom). Circle represents whole-body CoM.

Figure 5. Vertical force in optimised simulations using ET (solid line) and CT (dotted line).

Figure 6. Stance leg net joint moments in optimised simulations using ET (solid line) and CT (dotted line). Extension torques are positive.

Figure 7. Angle and angular velocities of the CC in optimised simulations using ET (solid line) and CT (dotted line) superimposed on the normalised tetanic torque-angle-angular velocity surface for each extensor torque generator. Positive angular velocities represent eccentric contractions. Stars represent touchdown.

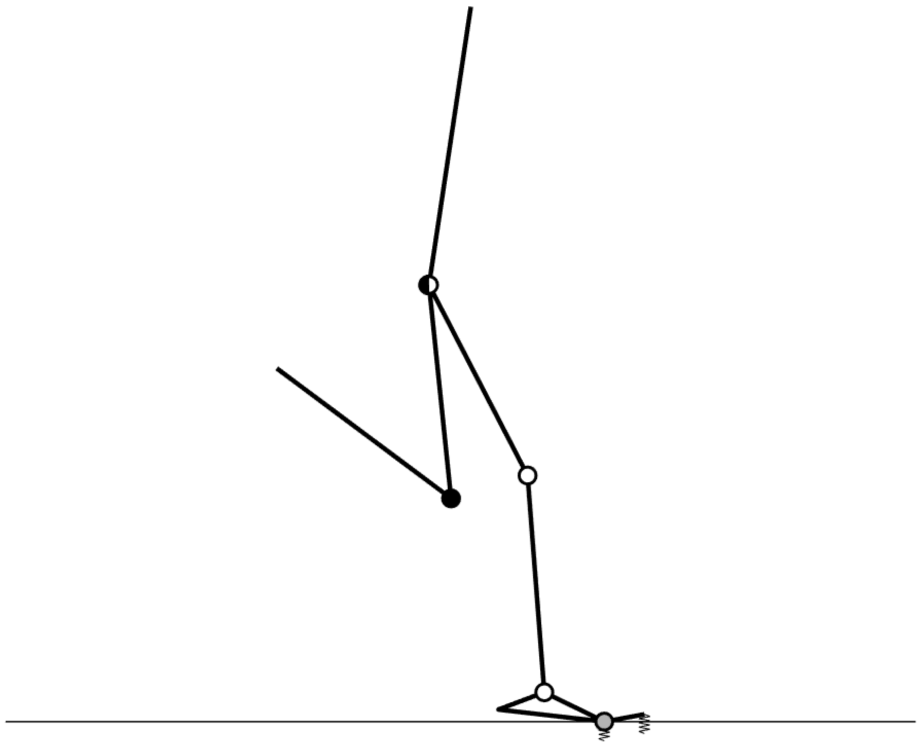


Figure 1

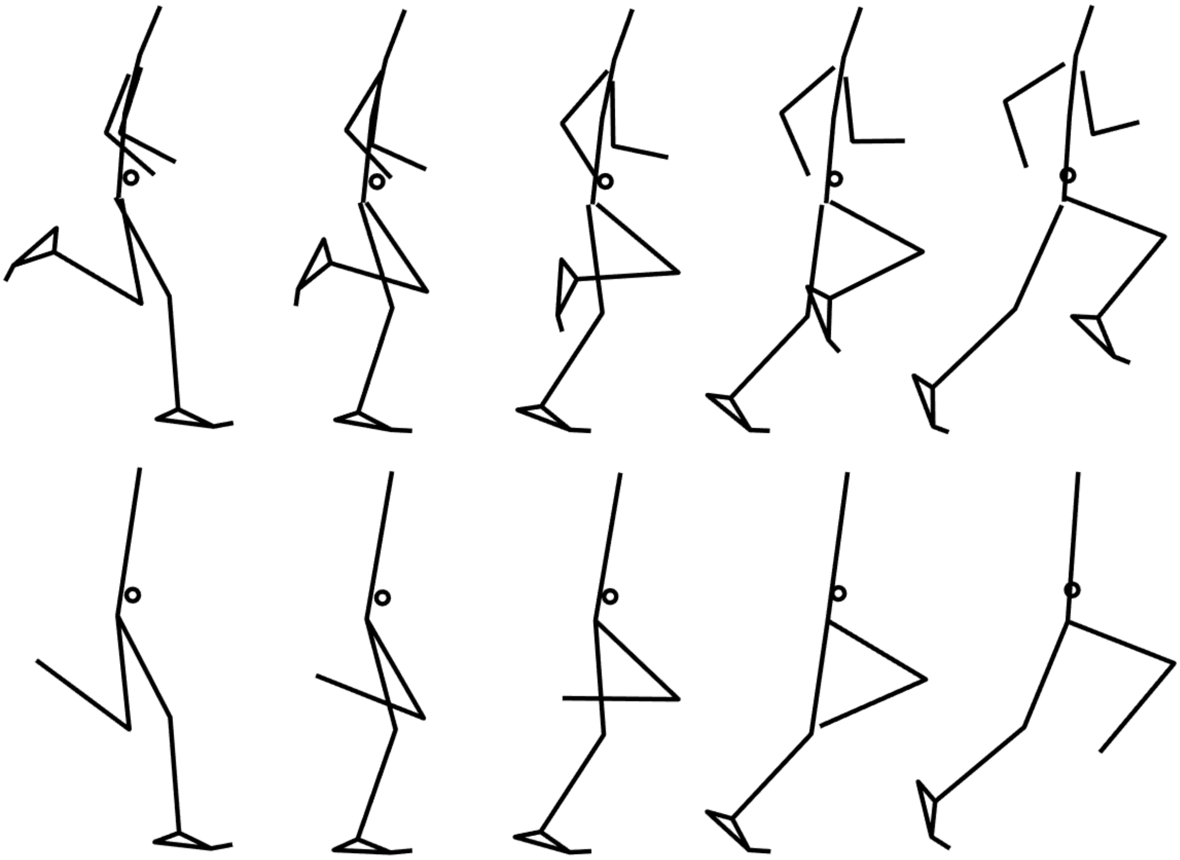


Figure 2

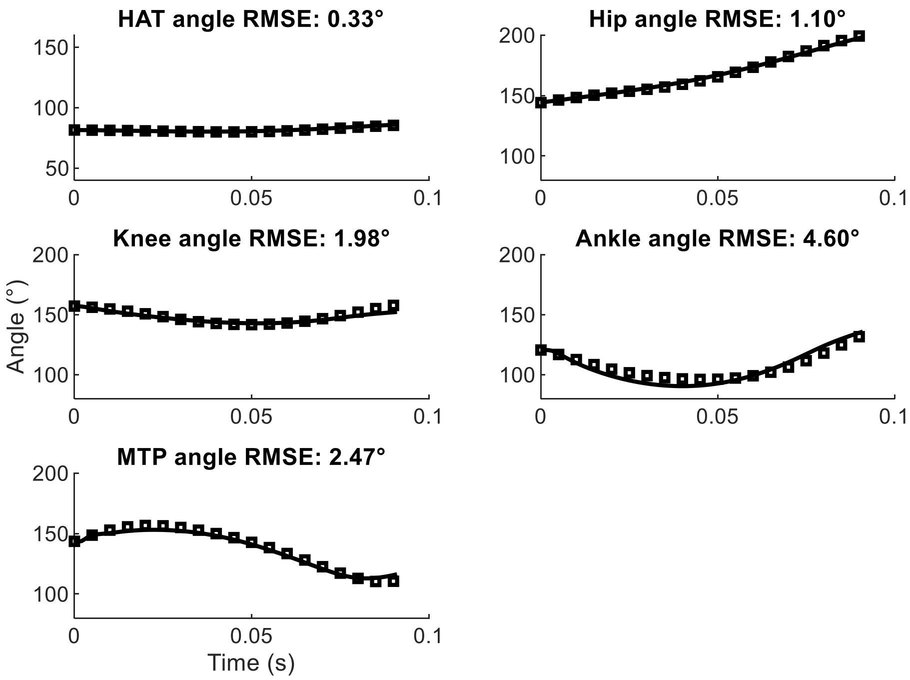

HAT angle RMSE: 0.33°
Hip angle RMSE: 1.10°
Knee angle RMSE: 1.98°
Ankle angle RMSE: 4.60°
MTP angle RMSE: 2.47°
Angle (°)
Time (s)
150
100
50
200
0
0.05
0.1

Figure 3

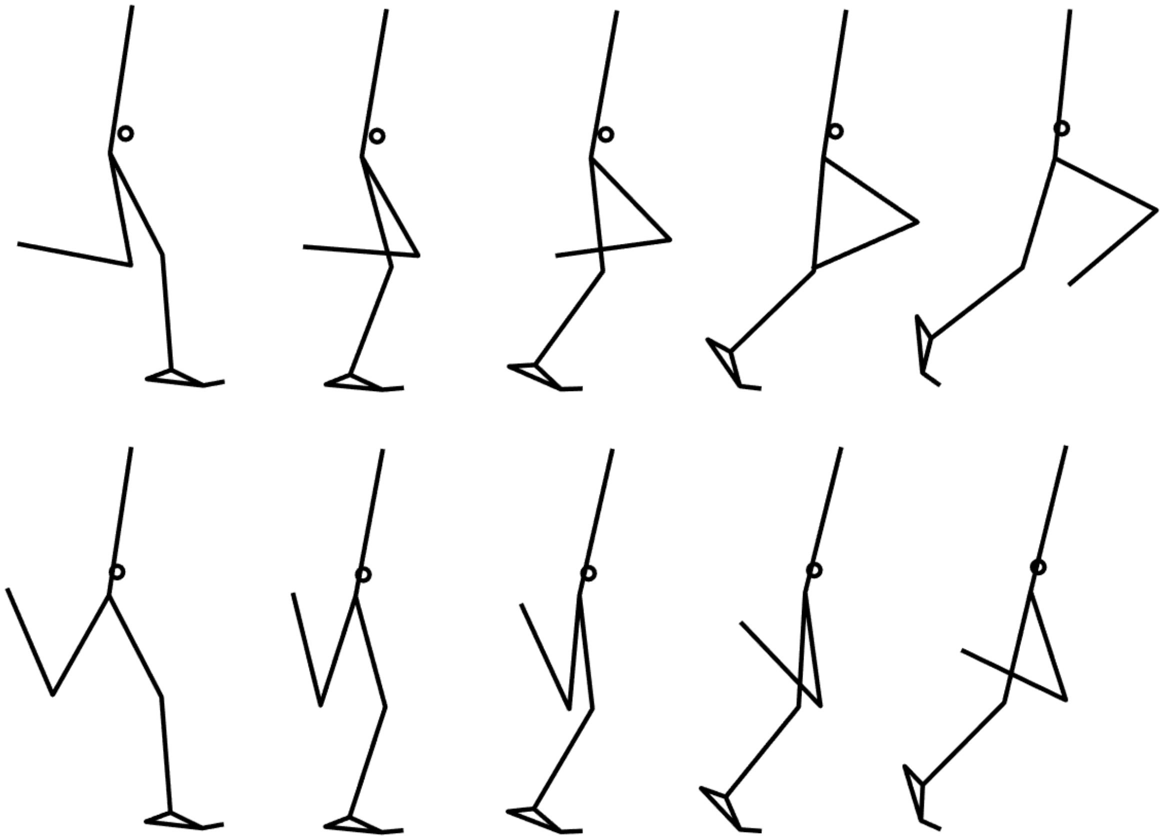


Figure 4

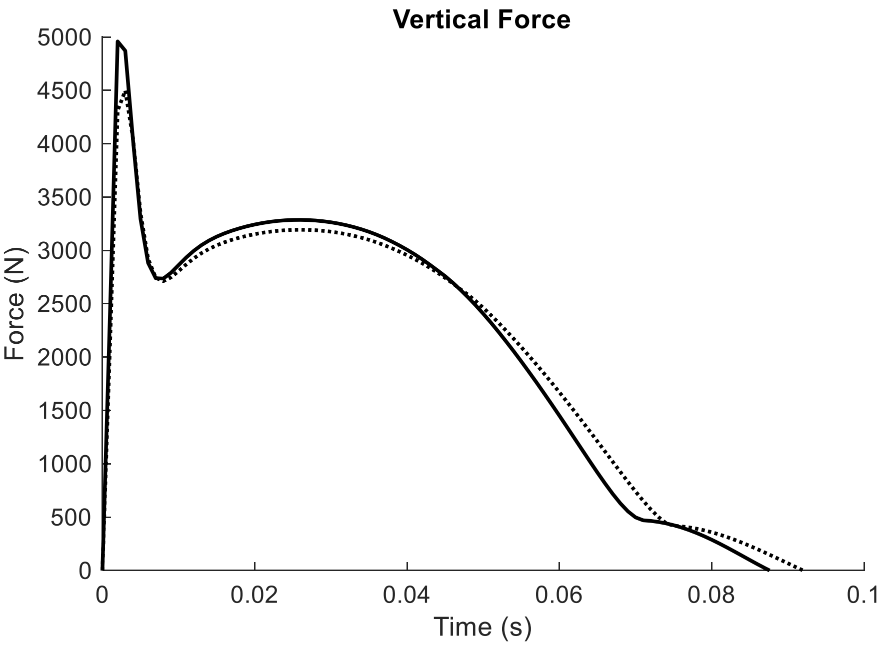
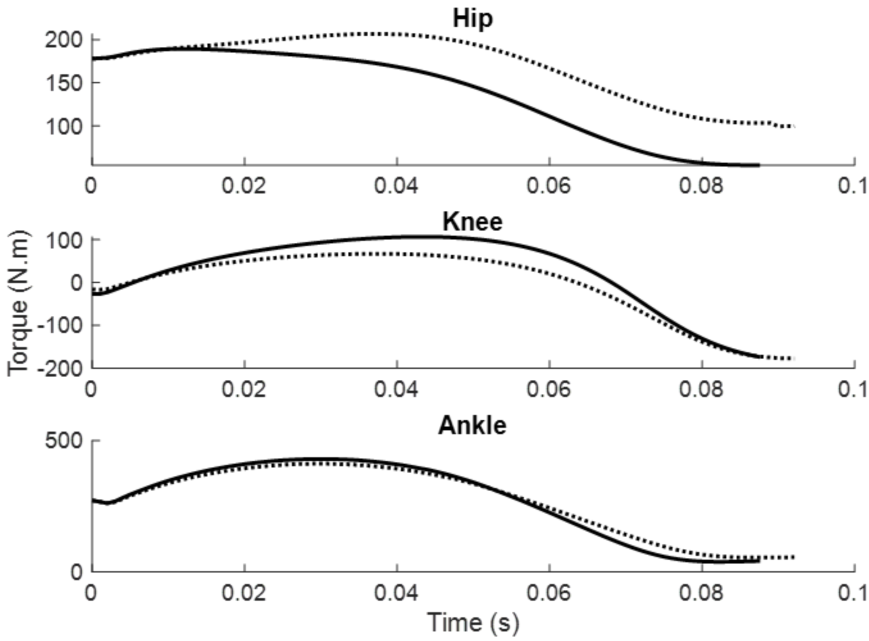

Vertical Force
Force (N)
5000
4500
4000
3500
3000
2500
2000
1500
1000
500
0
0
0.02
0.04
0.06
0.08
0.1
Time (s)

Figure 5



Hip
200
150
100
0
0.02
0.04
0.06
0.08
0.1
Knee
Torque (N.m)
100
0
-100
-200
0
0.02
0.04
0.06
0.08
0.1
Ankle
500
0
0
0.02
0.04
0.06
0.08
0.1
Time (s)

Figure 6

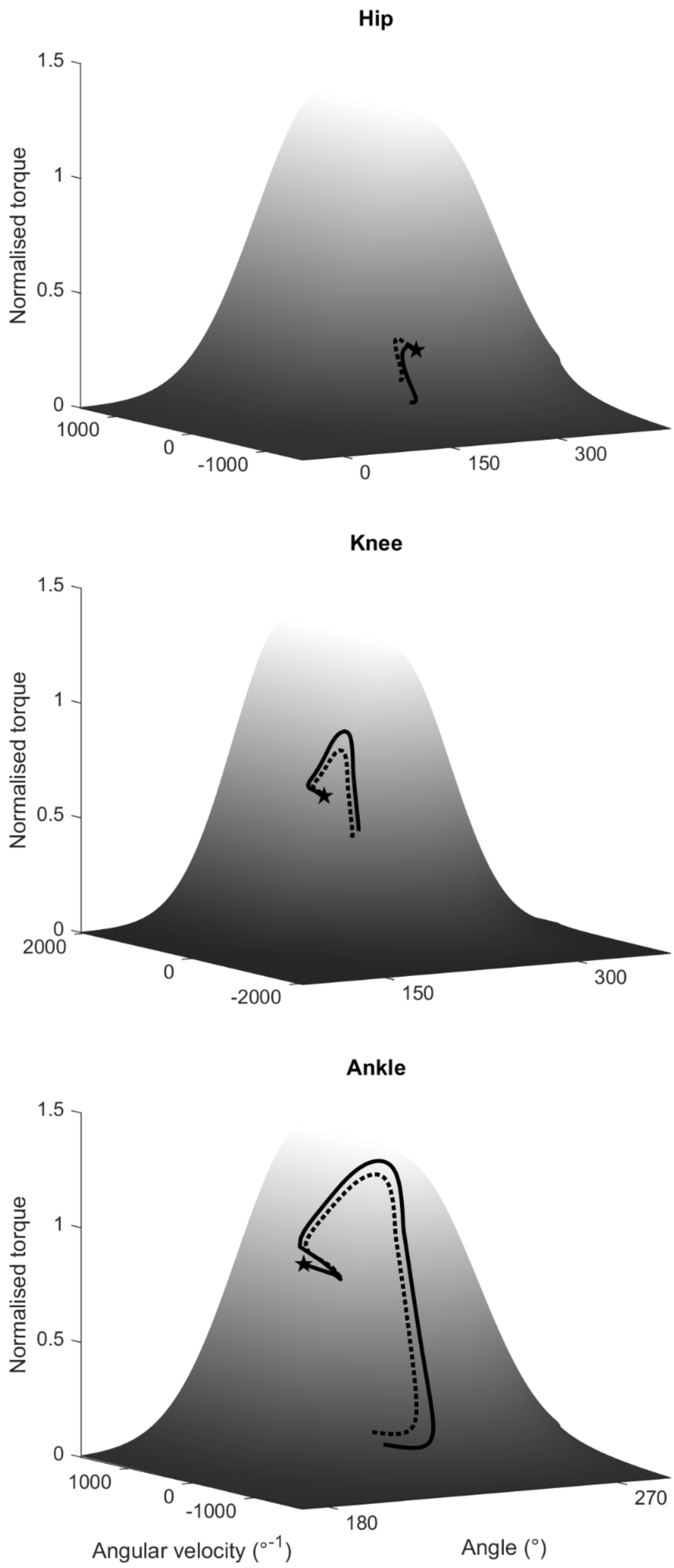

Hip
Normalised torque
1.5
1
0.5
0
1000
0
-1000
0
150
300
Knee
Normalised torque
1.5
1
0.5
0
2000
0
-2000
150
300
Ankle
Normalised torque
1.5
1
0.5
0
1000
0
-1000
180
270
Angular velocity (°$^{-1}$)
Angle (°)

Figure 7

Table 1. Comparison between the two swing leg techniques for selected variables.

	Elite athlete	**College athlete**
Horizontal velocity ($m.s^{-1}$)	10.0	9.2
Swing time (s)	0.374	0.374
Stance time (s)	0.088	0.092
Flight time (s)	0.143	0.141
Horizontal CoM displacement (m)	0.866	0.837
Vertical CoM displacement (m)	0.021	0.022
Vertical impulse (N.s)	111.7	109.8
Mean vertical force (N)	2123	2033
Mean hip extensor torque (N.m)	138.2	168.6
Mean knee extensor torque (N.m)	208.8	186.3
Mean ankle extensor torque (N.m)	277.2	267.8

Figure 1

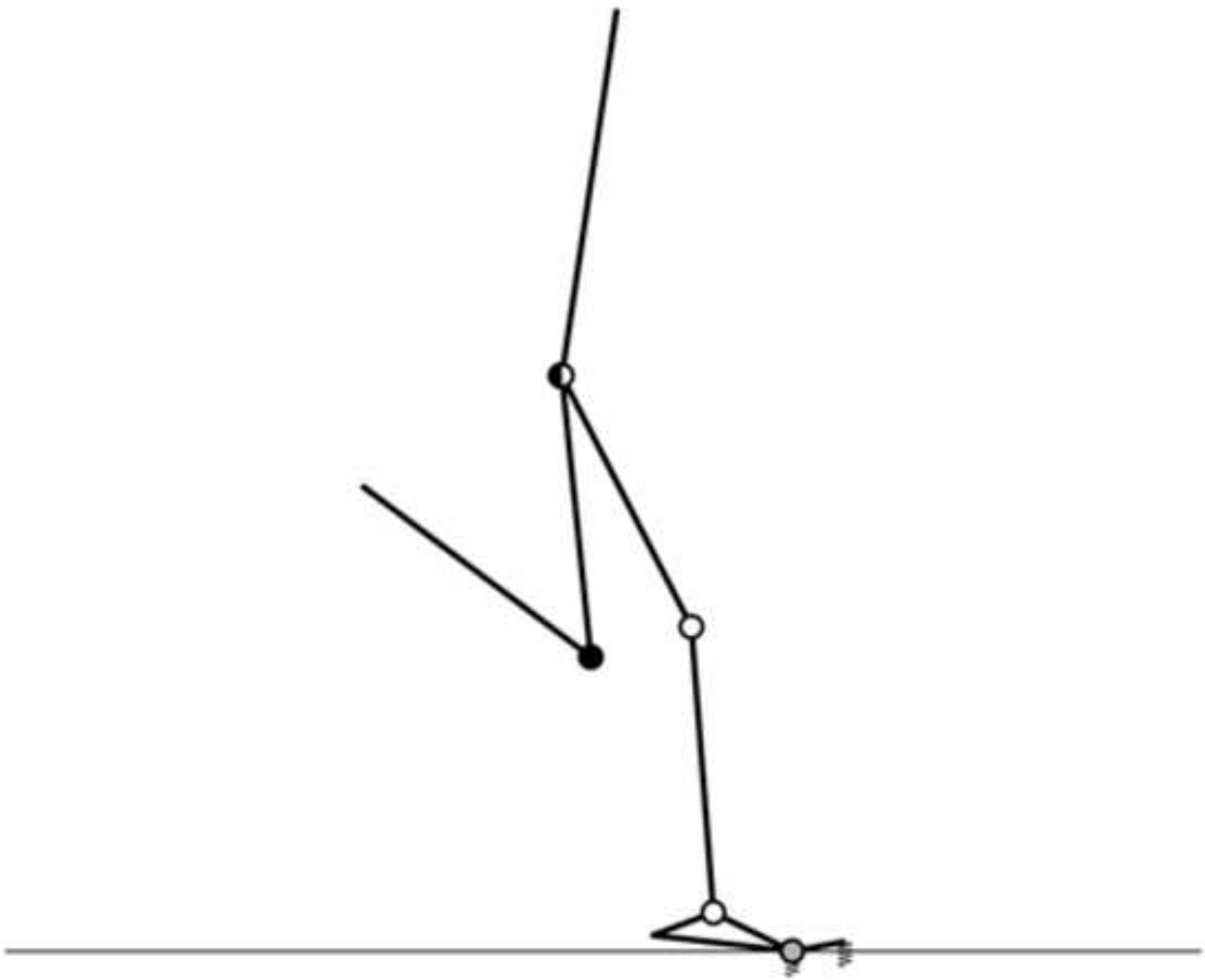


Figure 2

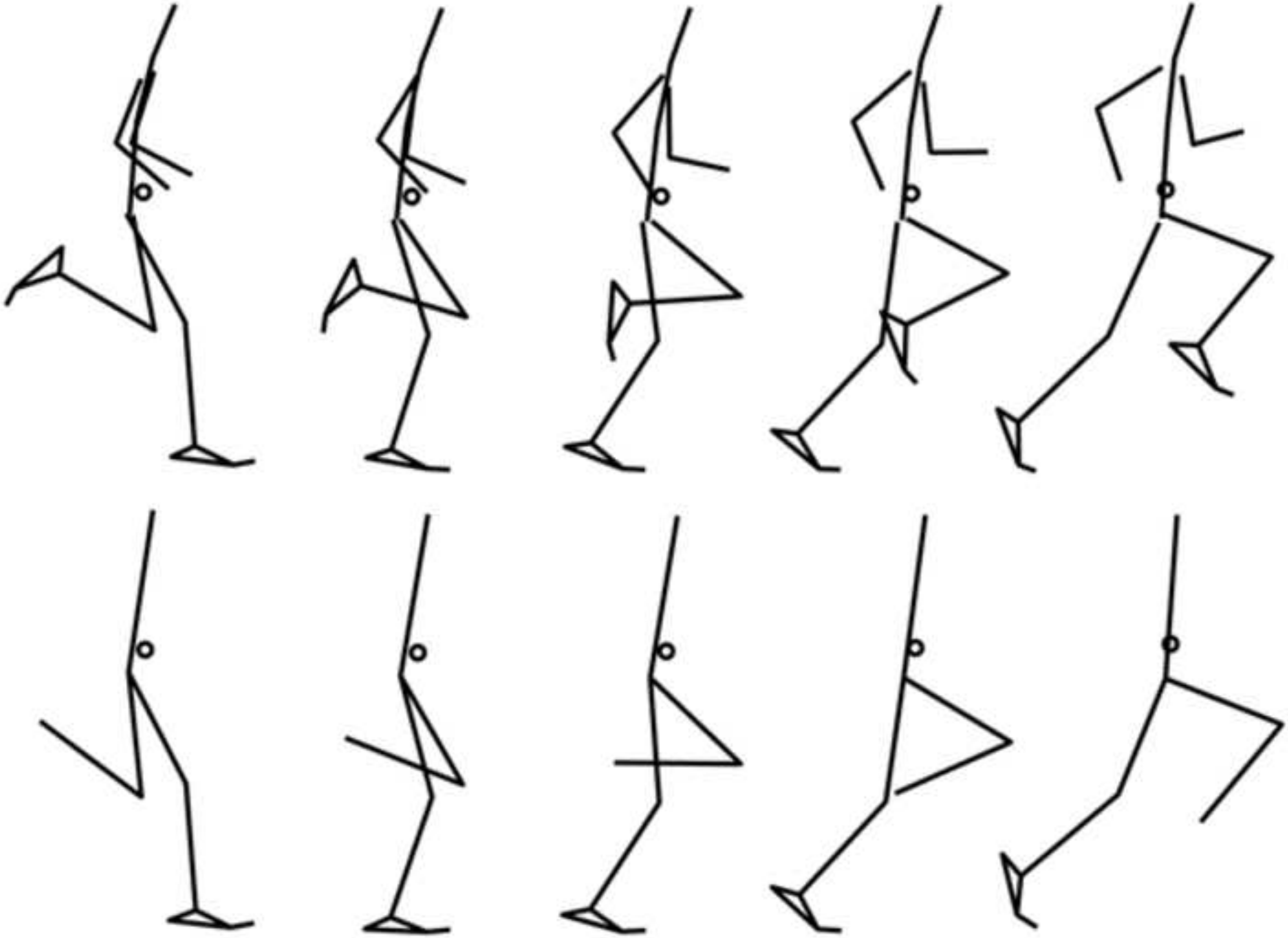
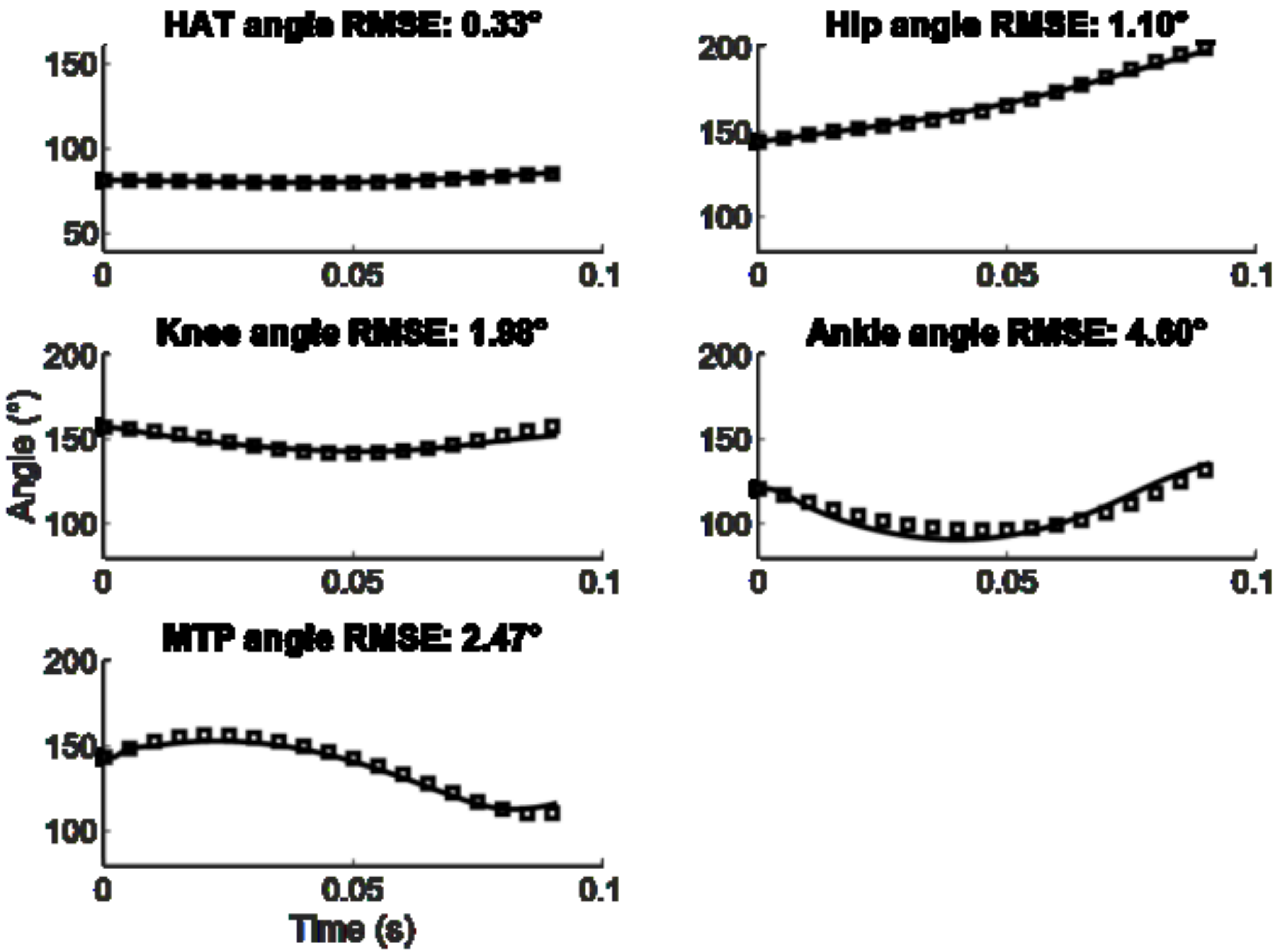


Figure 3


HAT angle RMSE: 0.33°
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MTP angle RMSE: 2.47°
Angle (°)
Time (s)
150
100
50
200
0
0.05
0.1

Figure 4

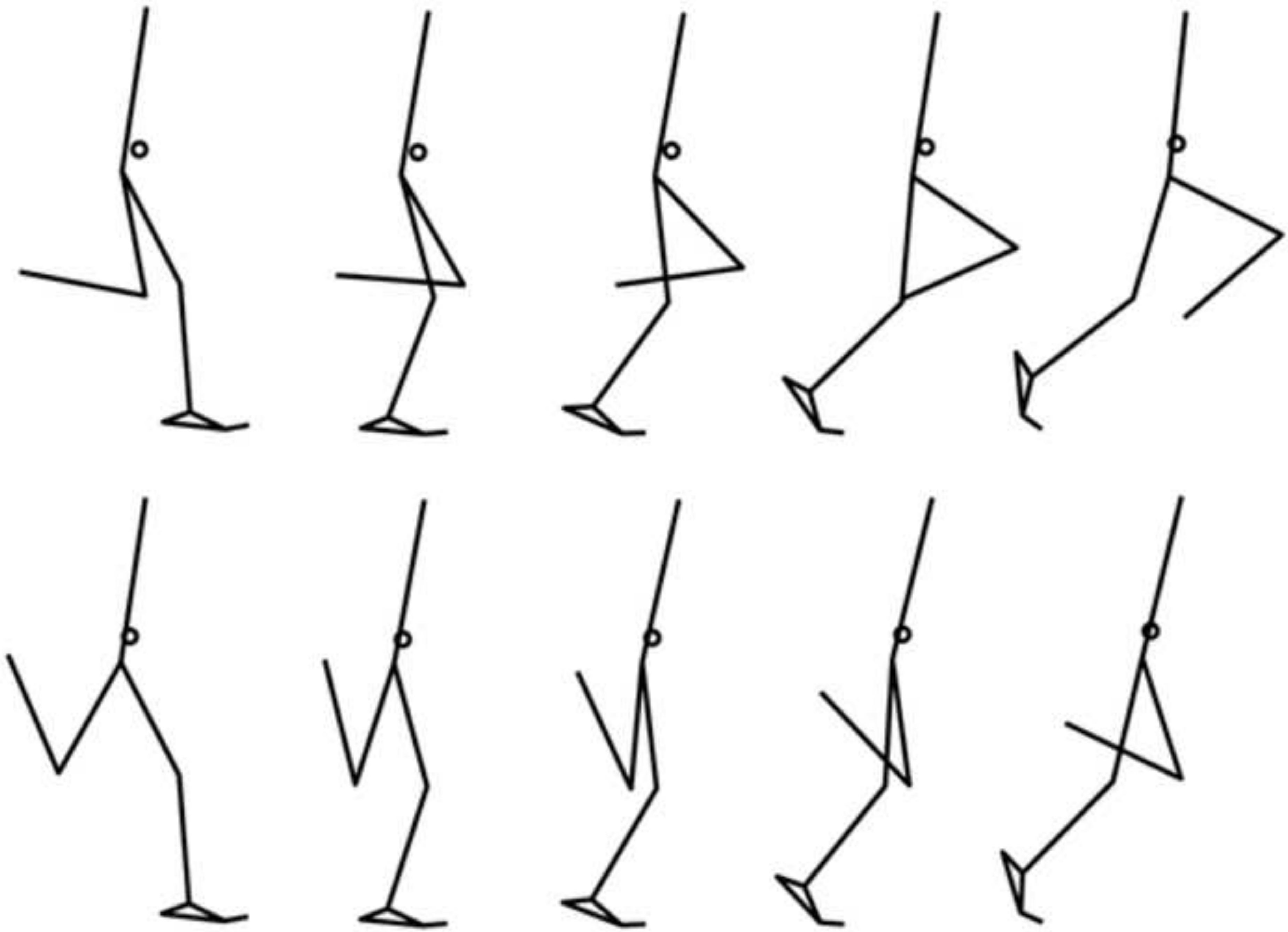

Figure 5

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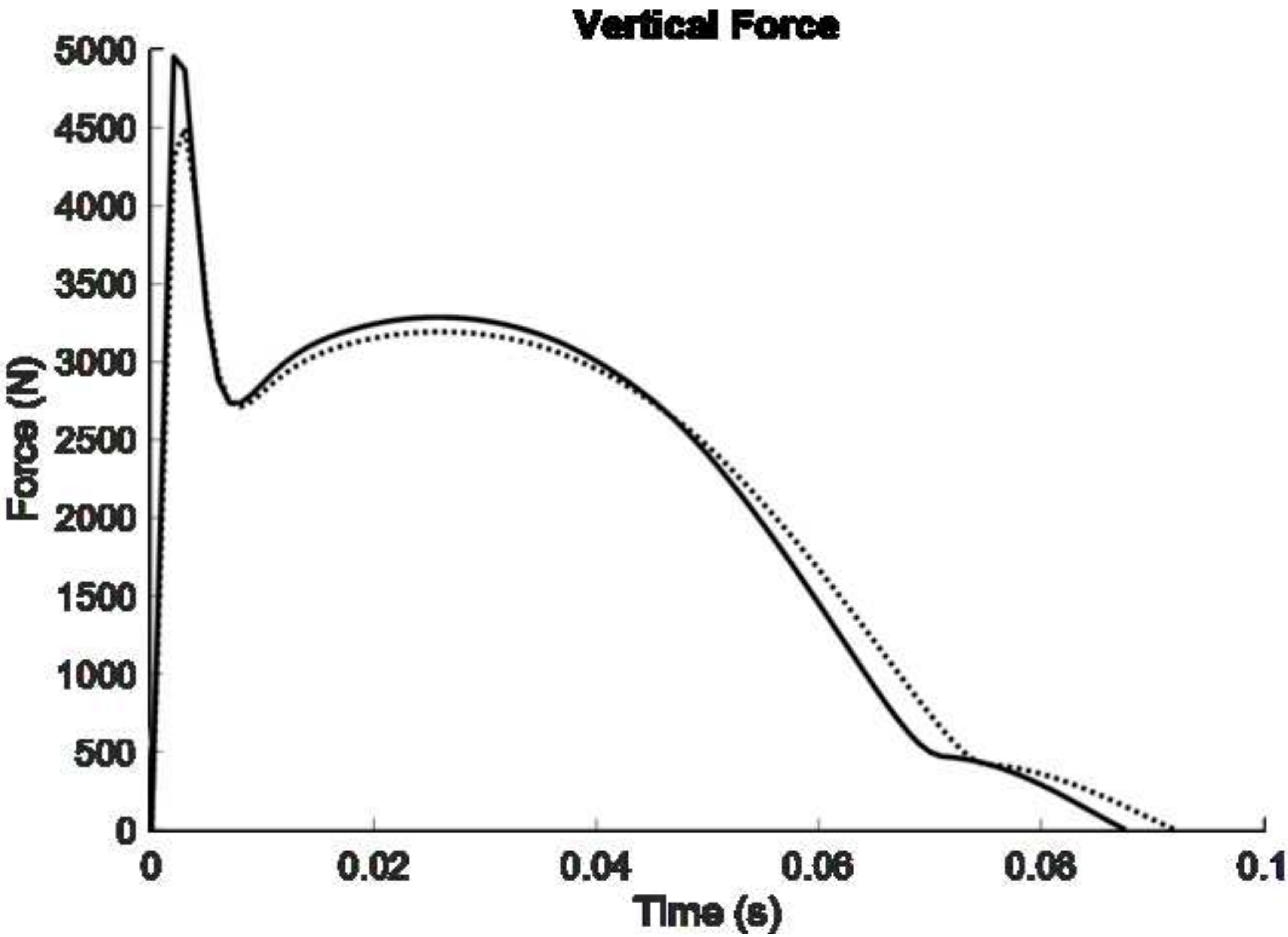


Figure 6

Click here to access/download;Figure;figure_6.tif

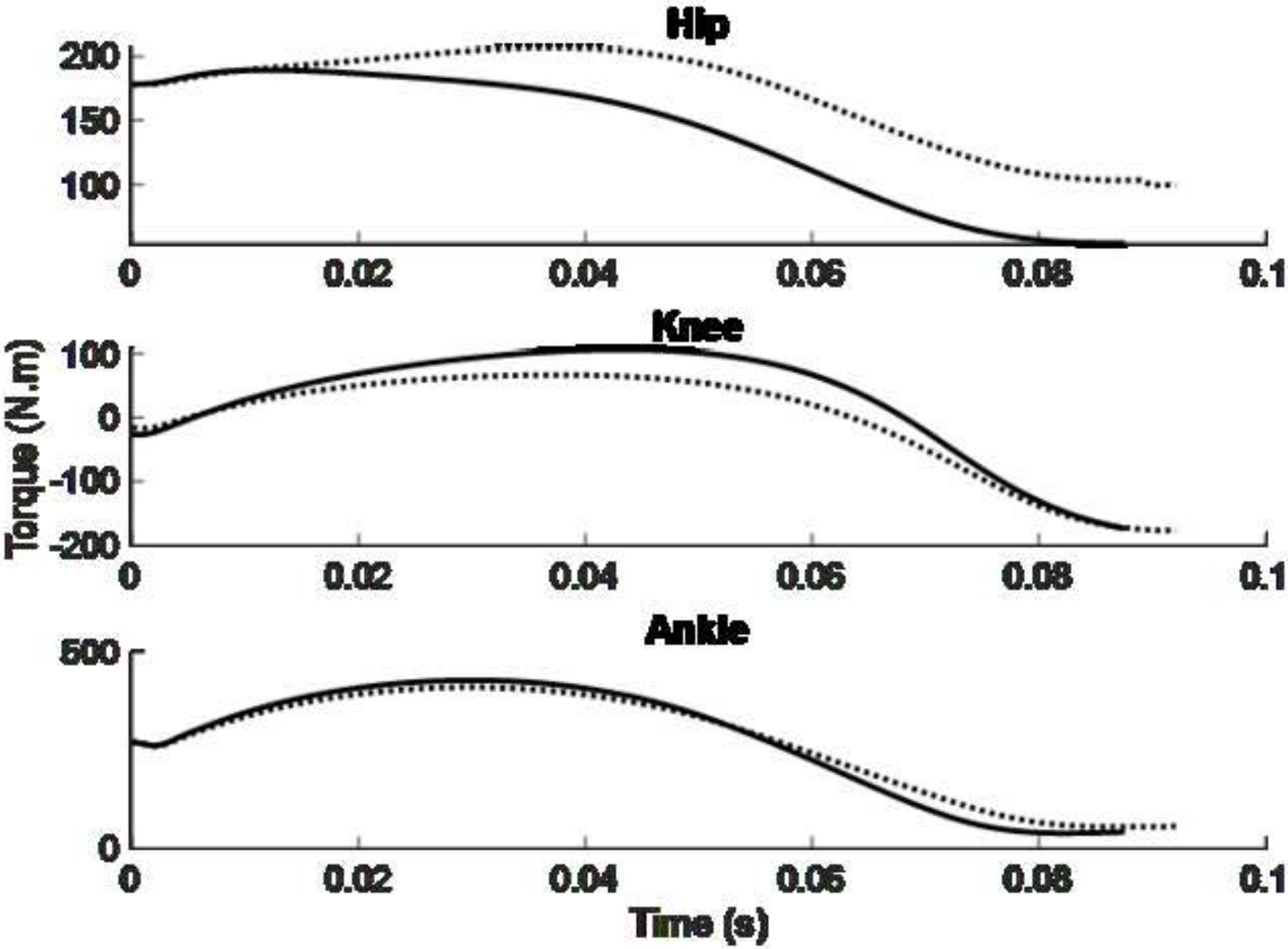
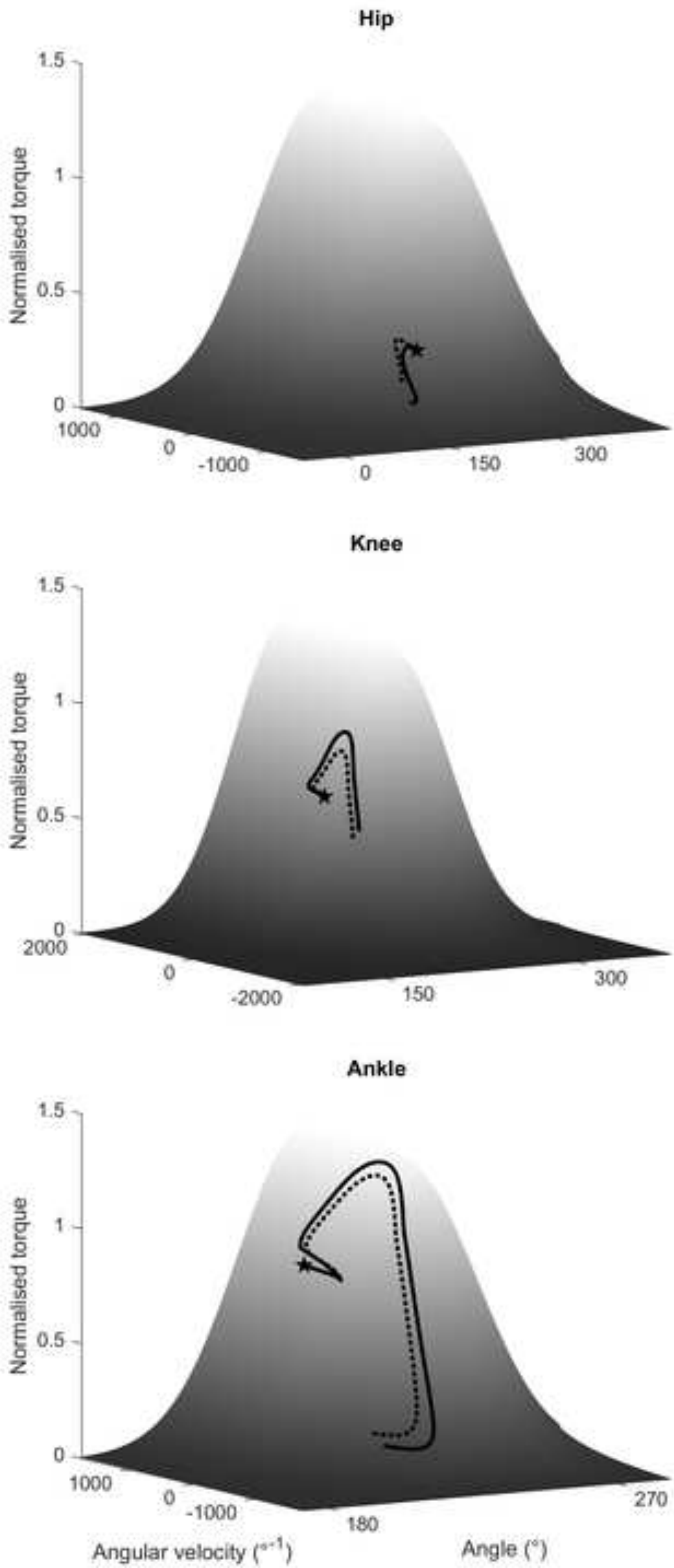


Figure 7



Hip
Normalised torque
1.5
1
0.5
0
1000
0
-1000
0
150
300
Knee
Normalised torque
1.5
1
0.5
0
2000
0
-2000
150
300
Ankle
Normalised torque
1.5
1
0.5
0
1000
0
-1000
180
270
Angular velocity (°s⁻¹)
Angle (°)

Matched simulation animation

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Supplementary Material

matching_resize.gif

Elite athlete optimised simulation animation

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Supplementary Material
sprinter_resize.gif

College athlete optimised simulation animation

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Supplementary Material
teamsport_resize.gif

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Conflict of interest statement:

In the manuscript entitled "The influence of swing leg motion on maximum running speed" submitted for publication in the Journal of Biomechanics there are no issues of conflict of interest arising from the personal or professional associations of any of the authors.

Yours sincerely,

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