Journal of Sports Sciences

ISSN: (Print) (Online) Journal homepage: www.tandfonline.com/journals/rjsp20

Forces experienced at different levels of the musculoskeletal system during horizontal decelerations

Jasper Verheul, Damian Harper & Mark A. Robinson

To cite this article: Jasper Verheul, Damian Harper & Mark A. Robinson (15 Nov 2024): Forces experienced at different levels of the musculoskeletal system during horizontal decelerations, Journal of Sports Sciences, DOI: 10.1080/02640414.2024.2428086

To link to this article: https://doi.org/10.1080/02640414.2024.2428086

JOURNAL OF SPORTS SCIENCES
https://doi.org/10.1080/02640414.2024.2428086

Routledge
Taylor & Francis Group

SPORTS MEDICINE AND BIOMECHANICS

OPEN ACCESS Check for updates

Forces experienced at different levels of the musculoskeletal system during horizontal decelerations

Jasper Verheul[a], Damian Harper[b] and Mark A. Robinson[c]

[a]Cardiff School of Sport and Health Sciences, Cardiff Metropolitan University, Cardiff, UK; [b]School of Health, Social Work and Sport, University of Central Lancashire, Preston, UK; [c]School of Sport and Exercise Sciences, Liverpool John Moores University, Liverpool, UK

ABSTRACT

Horizontal decelerations are frequently performed during team sports and are closely linked to sports performance and injury. This study aims to provide a comprehensive description of the kinetic demands of decelerations at the whole-body, structural, and tissue-specific levels of the musculoskeletal system. Team-sports athletes performed maximal-effort horizontal decelerations whilst full-body kinematics and ground reaction forces (GRFs) were recorded. A musculoskeletal model was used to determine whole-body (GRFs), structural (ankle, knee, and hip joint moments and contact forces), and tissue (twelve lower-limb muscle forces) loads. External GRFs in this study, especially in the horizontal direction, were up to six times those experienced during accelerated or constant-speed running reported in the literature. To cope with these high external forces, large joint moments (hip immediately after touchdown; ankle and knee during mid and late stance) and contact forces (ankle, knee, hip immediately after touchdown) were observed. Furthermore, eccentric force requirements of the tibialis anterior, soleus, quadriceps, and gluteal muscles were particularly high. The presented loading patterns provide the first empirical explanations for why decelerating movements are amongst the most challenging in team sports and can help inform deceleration-specific training prescription to target horizontal deceleration performance, or fatigue and injury resistance in team-sports athletes.

ARTICLE HISTORY
Received 9 May 2024
Accepted 4 November 2024

KEYWORDS
Braking performance; muscle forces; joint loading; musculoskeletal demands

Introduction

Rapid horizontal decelerations are essential movements in team sports. A deceleration can be defined as a reduction of the whole-body momentum, which is required to slow down after a sprint or in preparation for a cutting manoeuvre. High-intensity decelerations ($>2.5\ \mathrm{m{\cdot}s^{-2}}$) especially, form a substantial part of the total amount of running activities performed by team-sports athletes (Akenhead et al., 2013; Dalen et al., 2016; Vigh-Larsen et al., 2018), and are more frequently performed than equivalent-intensity accelerations (D. J. Harper et al., 2019). Moreover, decelerations are a determining factor for performance (Martínez-Hernández et al., 2023; Rhodes et al., 2021), and are deemed to expose athletes to a high injury risk (Jaspers et al., 2018; Martins et al., 2023). An in-depth understanding of the kinematic and kinetic characteristics of horizontal decelerations is thus vital to understand the athletic requirements of team sports.

Decelerating is a critical skill for team-sport performance. In soccer, for example, horizontal decelerations have been shown to be one of the most important movements preceding goals (Martínez-Hernández et al., 2023), and a high frequency of decelerations is positively associated with match outcome (Rhodes et al., 2021). In addition, completing a high number of rapid decelerations can impair performance in sprinting, accelerating, and decelerating (Akenhead et al., 2013; Lakomy & Haydon, 2004). Therefore, the importance of examining and training the specific mechanisms and skills that enable athletes to rapidly decelerate have been highlighted as an opportunity for performance enhancement in team-sport athletes (D. J. Harper et al., 2022; McBurnie et al., 2022). Nevertheless, strong evidence for the effectiveness of targeted interventions to improve deceleration performance is currently missing (D. J. Harper et al., 2022; Marvin et al., 2023). A likely reason for this is that although a substantial amount of literature exists describing the distinct *kinematic* characteristics of decelerations, the *kinetic* demands are still largely unexplored (D. J. Harper et al., 2022). Consequently, potentially trainable athletic features that can be targeted to enhance horizontal deceleration skills are poorly understood.

Horizontal decelerations are commonly associated with injury. The large external forces (i.e., whole-body level) required to rapidly slow down the momentum of the body necessitate high internal forces (i.e., structural and tissue level) which have major damaging effects on individual structures and tissues (D. J. Harper & Kiely, 2018). Especially the eccentric nature (i.e., active lengthening) of muscle contractions that are required to rapidly brake and decelerate whole-body momentum is known to induce muscle damage (Armstrong et al., 1983; Proske & Morgan, 2001). An accumulation of such tissue damage over time can subsequently lead to an increased risk of injury for athletes. Indeed, a high intensity and large volume of deceleration efforts have been shown to be linked with elevated blood

CONTACT Jasper Verheul jpverheul@cardiffmet.ac.uk Cardiff School of Sport and Health Sciences, Cardiff Metropolitan University, Cyncoed Rd, Cardiff, CF23 6XD UK

markers of muscle damage (de Hoyo et al., 2016; Gastin et al., 2019; Young et al., 2012) and a higher overall injury risk (Jaspers et al., 2018; Martins et al., 2023) in team-sports athletes. For example, deceleration movements are known to be a high-risk task for sustaining anterior cruciate ligament injuries (Cochrane et al., 2007; Della Villa et al., 2020; Johnston et al., 2018). Nevertheless, the functional role of individual muscles and joints and the kinetic demands posed on these tissues and structures during horizontal decelerations are not well understood. Hence, a detailed kinetic assessment of the forces experienced across the various levels of the musculoskeletal system is required.

Given the high number of deceleration efforts performed during team sports, and their association with performance and injury risk, some kinetic demands of horizontal decelerations have been explored. These studies have either been qualitative in nature (Hewit et al., 2011) or are restricted to quantifications of the external ground reaction forces (GRFs) (Cross et al., 2024; Dos'Santos et al., 2021; Nedergaard et al., 2014), single-joint loading (Cassiolas et al., 2023; Cross et al., 2024; Peel et al., 2019), or individual muscle contributions (Mateus et al., 2020), separately. However, a unified quantification to understand the musculoskeletal forces of horizontal decelerations across the whole-body, structural, and tissue-specific level (Verheul et al., 2020), is yet lacking. The exact nature of the biomechanical loads experienced (e.g., magnitude, timing), and the primary locations of application within the musculoskeletal system, thus remain uncertain. This study, therefore, aims to provide a comprehensive description of the kinetic demands of horizontal decelerations at the whole-body, structural, and tissue-specific levels of the musculoskeletal system.

Methods

Participants

Fifteen recreational team-sport athletes participated in this study (12 males and 3 females, age 23 ± 4 years, height 178 ± 9 cm, body mass 73 ± 10 kg). Participants were all healthy, injury free for a minimum of six months, participated in team sports (e.g., football, handball) for at least three hours per week (7 ± 5 hours per week), and competed at a recreational level. Each participant completed a readiness to exercise questionnaire and provided written informed consent prior to participation. This study was approved by Liverpool John Moores University ethics committee (17/SPS/043).

Horizontal deceleration protocol

Following a warm-up of easy running and dynamic stretching, athletes performed a series of horizontal deceleration trials. Athletes were instructed to decelerate as quickly as possible from maximal sprinting to immediate standstill, while landing with one foot on a ground-embedded force platform for the first or second step of braking. Given the technically challenging nature of decelerations and to avoid athletes targeting the force platform, participants were instructed to initiate their deceleration from a visible line aligned with the edge of the force platform, leading to the first or second deceleration step to be on the force platform. A minimum of ten successful trials (i.e., full foot on the force platform; five on each leg) were completed by each participant. These tasks were performed as part of a broader assessment of team-sports movements – for full details of the movement protocol see (Verheul et al., 2019).

Kinematic and kinetic data collection

During the deceleration trials, three-dimensional kinematic and kinetic data were synchronously recorded with Qualisys Track Manager software (v2.16, Qualisys, Gothenberg, Sweden). Ten infrared cameras (Qualisys Qqus 300+, Gothenberg, Sweden) sampling at 250 hz were used, in combination with a marker set of seventy-six reflective markers, to collect full-body kinematics (for a detailed description of marker placement see (Verheul et al., 2019)). A ground-embedded force platform (Kistler 9287B, Winterthur, Switzerland) sampling at 3000 hz was used to record GRF. A static calibration trial was captured for each athlete at the start of the data-collection session. Marker trajectories and GRFs were filtered at 15 hz and 50 hz, respectively, using a second order lowpass Butterworth filter. Data were then exported to MATLAB (R2022a, MathWorks, Natick, USA) for further processing.

The OpenSim API (version 4.3, SimTK, Stanford, USA (Seth et al., 2018)) was used to determine kinematics, joint moments and contact forces, and muscle lengths and forces. A full-body musculoskeletal model consisting of 22 rigid bodies with 37 degrees of freedom, and 80 hill-type muscle-tendon units (Rajagopal et al., 2016), adapted to allow for high knee and hip joint flexion and calculating medial and lateral tibiofemoral joint contact forces (Bedo et al., 2020), was used. The athlete-specific static trials were used to scale the model's segment lengths to each athlete's dimensions. Whole-body kinematics were determined through inverse kinematics (i.e., minimising the sum of squared errors between the measured marker trajectories and virtual markers attached to the model). The centre of mass of the model was used to determine the initial velocity at the start of each deceleration step, the change in velocity between touchdown and take off, and the mean acceleration during the ground-contact phase. Joint moments were calculated from inverse dynamics, using the inverse kinematics results and measured GRFs, and muscle forces were estimated through static optimization (i.e., minimising the sum of the squared muscle activations). The maximum isometric force for each muscle within the model was multiplied by a factor of three to allow for the generation of sufficiently high forces required for decelerations, and minimise the contributions from the reserve torque actuators (Hagen et al., 2023; Swinnen et al., 2019). Hip, knee (medial and lateral), and ankle joint contact forces, and muscle lengths, were determined using the joint reaction analysis and muscle analysis respectively. For all OpenSim analyses, custom MATLAB scripts based on a publicly available toolbox (Bedo et al., 2021) were used.

Biomechanical loads

Several markers of biomechanical load across the different levels of the musculoskeletal system – whole-body, structural, and tissue level (Verheul et al., 2020) – were selected for further analysis.

To represent loads at a whole-body level, GRFs were examined. Forces were normalised to each athlete's body weight (BW). Peak GRFs were determined as the maximal force values in in anteroposterior and vertical direction. Vertical loading rates were calculated as the peak GRF divided by the time to reach this peak. Force impulses (i.e., area under the GRF curve) were calculated as the integral of the anteroposterior and vertical GRFs during the ground-contact phase. The ratio of forces was calculated as the anteroposterior force divided by the resultant force in the sagittal plane ($F_{ratio} = \frac{F_{ap}}{\sqrt{F_v^2 + F_{ap}^2}}$), averaged over the ground-contact phase and expressed as a percentage (Bezodis et al., 2021; Morin et al., 2011). The corresponding angle of body inclination was calculated as $\alpha = \sin^{-1}(F_{ratio})$ and expressed in degrees.

At a structural level, joint moments and contact forces for the hip, knee, and ankle were examined. Joint moments and contact forces were normalised to each athlete's body mass and weight, respectively. Peak moments were determined in both directions. Peak compressive contact forces were calculated as the maximal force during the ground-contact phase along the axial direction of the distal segment of each joint. Joint contact force impulses were determined as the area under the force curve throughout the ground-contact phase. For the knee joint, compressive contact forces for the medial and lateral tibiofemoral joints were determined separately, and the total anteroposterior shear forces at the proximal end of the tibia was calculated.

At a tissue level, muscle forces and lengthening velocities were examined for twelve muscles (gluteus maximus, gluteus medius, biceps femoris, semitendinosus, semimembranosus, vastus medialis, vastus lateralis, rectus femoris, gastrocnemius medialis, gastrocnemius lateralis, soleus, tibialis anterior). Peak forces and force impulse (i.e., integral of the force during ground contact, $\int_{t_{TD}}^{t_{TO}} F_T \cdot dt$) were calculated for each muscle and normalised to BW. Peak muscle velocities were determined as the first derivate of the muscle length and only positive peak values (i.e., lengthening) were considered. To allow for between-muscle comparisons, muscle forces were normalised to their maximum isometric force (F^0_{max}) as defined in Rajagopal et al. (2016). Since eccentric muscle contractions are well known to particularly induce muscle damage (Armstrong et al., 1983; Proske & Morgan, 2001), the velocities and timings of positive lengthening velocities (i.e., eccentric muscle contraction) were determined, to examine the potentially damaging nature of horizontal decelerations for individual muscles. Muscle lengthening velocities were expressed relative to each muscle's resting length (L^0_m).

Tendon forces on the patellar and Achilles tendons were calculated as the sum of the quadriceps (vastus lateralis, medialis, and intermedius, and rectus femoris) or calf (gastrocnemius lateralis and medialis, and soleus) muscle forces respectively, and normalised to each athlete's BW. Peak forces and force impulse were then calculated for the patellar and Achilles tendons. Moreover, to account for the non-linear relationship between cumulative tendon loading and damage (Edwards, 2018), a weighted force impulse was calculated according to $\left[\int_{t_{TD}}^{t_{TO}} F^b_{tendon} \cdot dt\right]^{\frac{1}{b}}$ – with b = 5.7 as the common weighting factor for the both tendons (Firminger & Edwards, 2021).

Results

A total of 161 deceleration trials were collected and included in the analysis. The horizontal approach velocity for the decelerations was 5.65 ± 0.65 m·s^{-1}, across all athletes and trials. The average duration of the ground-contact phase was 0.16 ± 0.04 s during which the change in whole-body centre of mass velocity was -0.8 ± 0.24 m·s^{-1}. The mean centre of mass acceleration during the

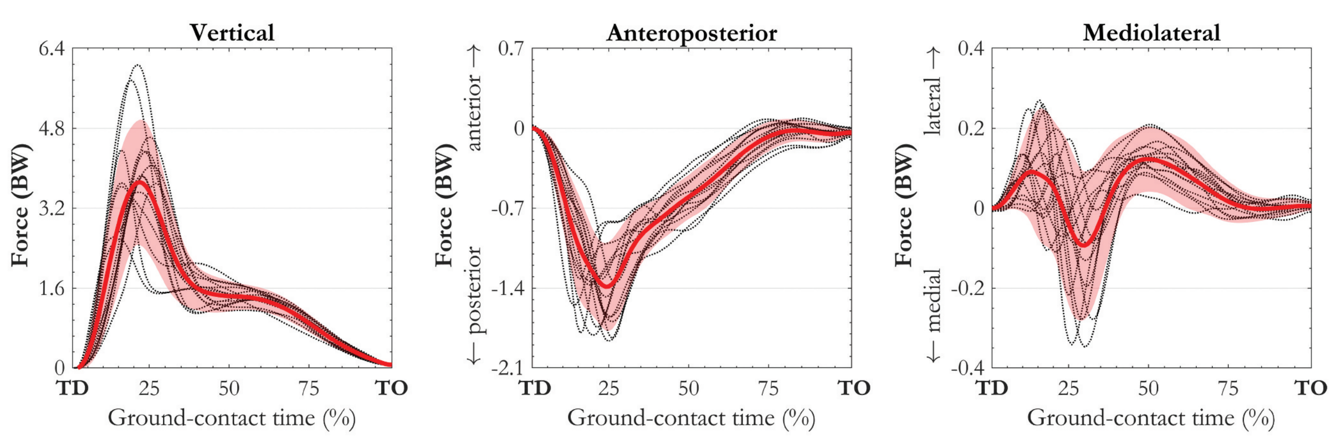

Figure 1. Ground reaction forces in vertical, anteroposterior, and mediolateral direction during the ground-contact phase of decelerating. Forces are normalised to body weight (BW). Black dotted curves represent the mean forces for each athlete. Red curves and shading represent the mean and standard deviations across all deceleration trials and athletes. TD = touchdown, to = take off.

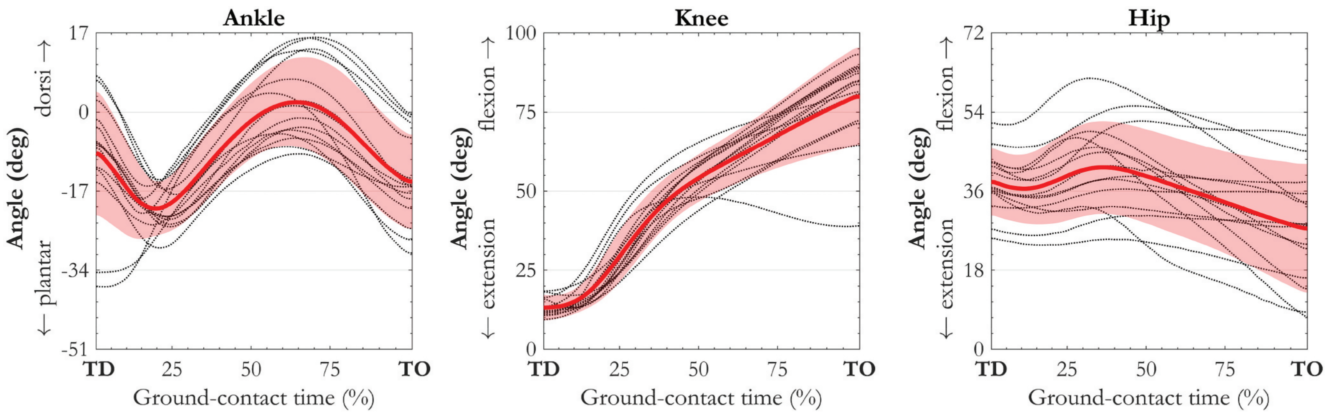

Figure 2. Joint angles for the ankle, knee, and hip joint during the ground-contact phase of decelerating. Black dotted curves represent the mean joint angles for each athlete, whereas red curves and shading represent the mean and standard deviations across all deceleration trials and athletes. TD = touchdown, to = take off.

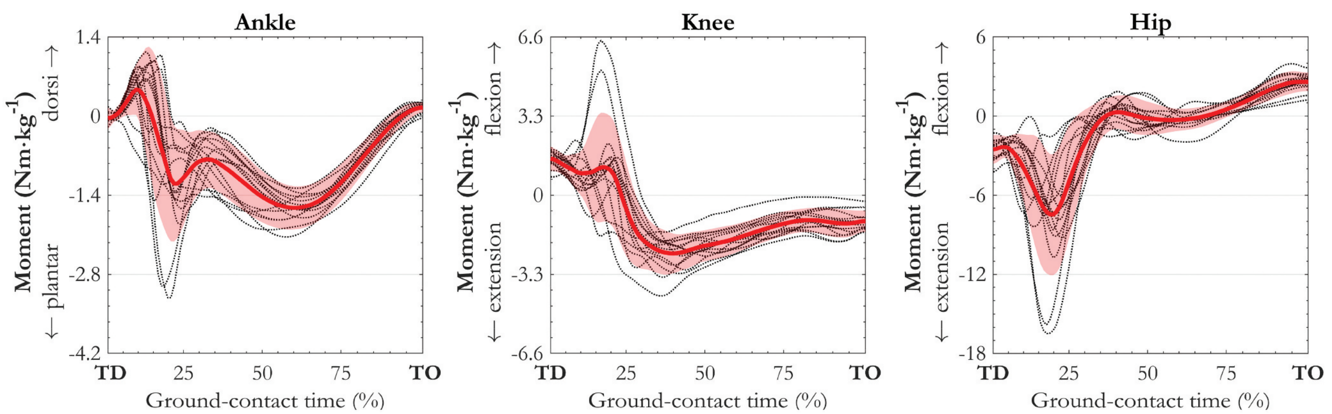

Figure 3. Joint moments for the ankle, knee, and hip joint during the ground-contact phase of decelerating. Moments are normalised to body mass. Black dotted curves represent the mean moments for each athlete. Red curves and shading represent the mean and standard deviations across all deceleration trials and athletes. TD = touchdown, to = take off.

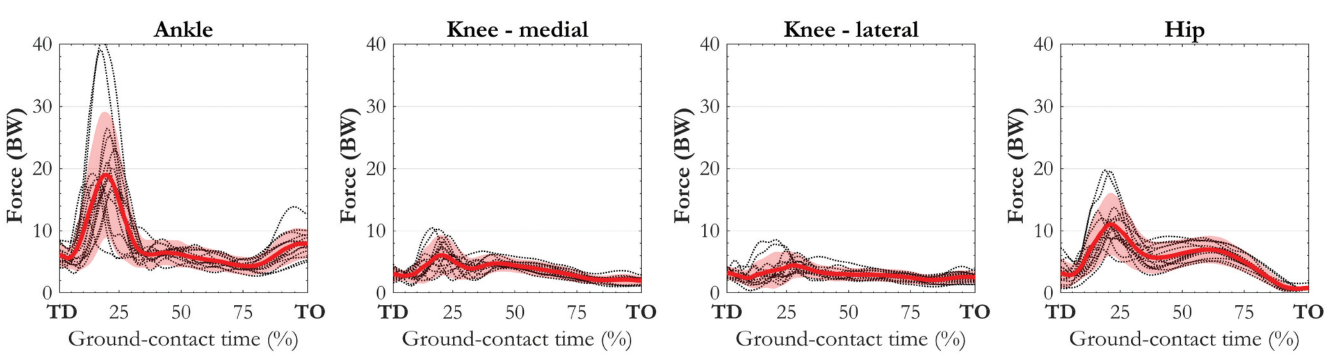

Figure 4. Joint contact forces for the ankle, knee (medial and lateral), and hip joint during the ground-contact phase of decelerating. Forces are normalised to body weight (BW). Black dotted curves represent the mean moments for each athlete. Red curves and shading represent the mean and standard deviations across all deceleration trials and athletes. TD = touchdown, to = take off.

ground-contact phase (i.e., the deceleration intensity) was -5.07 ± 1.5 m·s^{-2}.

Whole-body loading

At a whole-body level, the peak GRFs were 4.48 ± 1.11 BW and -1.67 ± 0.27 BW in vertical and anteroposterior direction, respectively (Figure 1). The average loading rates were 132.02 ± 45.7 BW·s^{-1} (vertical) and 49.24 ± 12.57 BW·s^{-1} (horizontal), and force impulses were 0.23 ± 0.04 BW·s (vertical) and -0.08 ± 0.02 BW·s (horizontal). The force ratio between the vertical and anteroposterior GRF components was $-29.79 \pm 9.93\%$, with an average angle of inclination α of -17.43 ± 5.98 degrees.

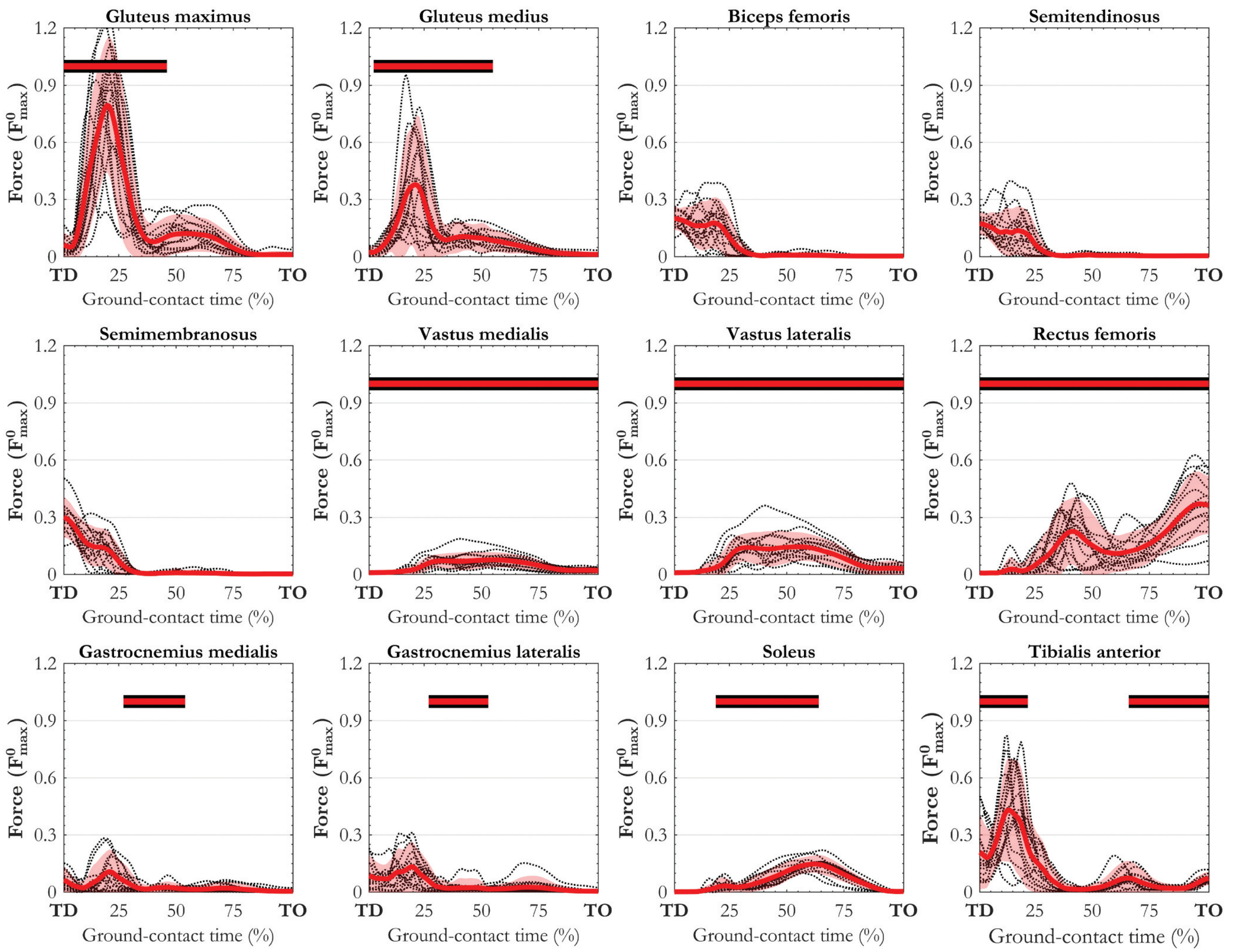

Figure 5. Muscle forces during the ground-contact phase of decelerating for twelve major muscles in the lower limbs. Forces are expressed as a percentage of the maximal isometric force for each muscle (F^0_{max}). Black dotted curves represent the mean forces for each athlete. Red solid curves and shading represent the mean and standard deviations respectively, across all deceleration trials and athletes. Horizontal bars represent the time during which the muscle is lengthening (i.e., eccentric action). TD = touchdown, to = take off.

Table 1. Muscle force and velocity characteristics. Forces are normalised to body weight (BW) and shown as mean ± standard deviation. Muscle lengthening velocities are expressed relative to each muscle's resting length (L^0_m). Muscle lengthening durations are expressed a percentage of the ground-contact time. Note that the three hamstring muscles only acted concentrically (shortening) during ground contact.

Muscles	*Peak force (BW)*	*Force impulse (BW·s)*	*Peak lengthening velocity ($L^0_m \cdot s^{-1}$)*	*Lengthening duration (%)*
Gluteus maximus	5.18 ± 1.36	0.15 ± 0.05	0.8 ± 0.39	45.9 ± 23.5
Gluteus minimus	1.57 ± 1.09	0.04 ± 0.02	1.21 ± 0.61	55.7 ± 21.6
Biceps femoris	1.49 ± 0.48	0.04 ± 0.02	-	-
Semitendinosus	0.58 ± 0.25	0.01 ± 0.01	-	-
Semimembranosus	2.63 ± 0.76	0.06 ± 0.02	-	-
Vastus medialis	1.06 ± 0.41	0.07 ± 0.05	1.6 ± 0.23	95.7 ± 13.1
Vastus lateralis	3.9 ± 1.46	0.26 ± 0.17	1.55 ± 0.21	95.7 ± 13.1
Rectus femoris	3.78 ± 1.25	0.19 ± 0.08	1.05 ± 0.13	99.8 ± 1.4
Gastrocnemius medialis	2.17 ± 1.41	0.05 ± 0.03	0.39 ± 0.22	26.8 ± 16.3
Gastrocnemius lateralis	1.46 ± 0.87	0.04 ± 0.02	0.4 ± 0.24	25.8 ± 16.9
Soleus	3.86 ± 1.06	0.21 ± 0.1	1.03 ± 0.27	46.5 ± 6.7
Tibialis anterior	2.95 ± 1.07	0.08 ± 0.04	1.36 ± 0.38	57 ± 8.3

Joint kinematics and loading

At touchdown, the ankle plantarflexed for the first ~ 20% of ground contact, after which a gradual dorsiflexion occurred, before plantarflexing again prior to take off (Figure 2). Peak ankle dorsi- and plantarflexion angles were 4 ± 10 and −24 ± 7 degrees, respectively.

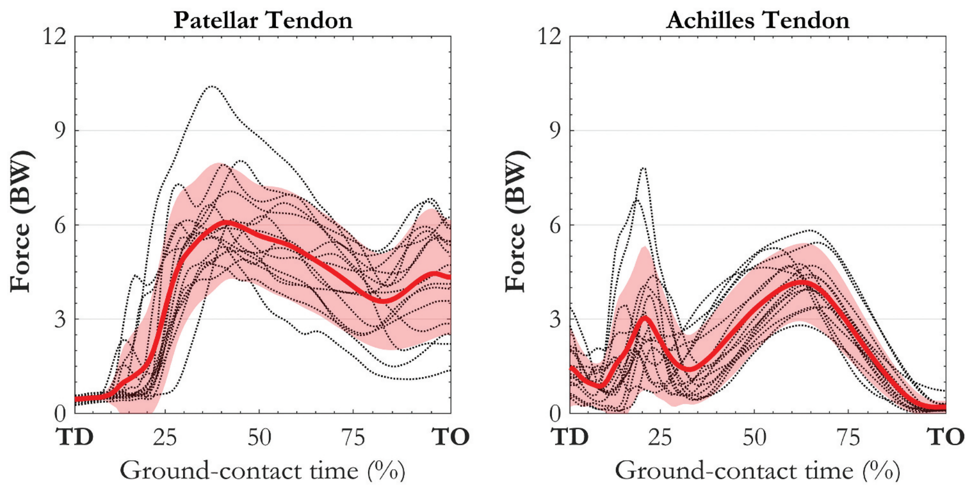

Figure 6. Patellar and achilles tendon forces during the ground-contact phase of decelerating. Forces are normalised to body weight (BW). Black dotted curves represent the mean forces for each athlete. Red curves and shading represent the mean and standard deviations across all deceleration trials and athletes. TD = touchdown, to = take off.

Throughout the ground-contact phase the knee was in flexion which increased from 12.8 ± 3.6 at touchdown to 81 ± 13 degrees at take off. Likewise, the hip was flexed throughout ground contact with peak angles ranging on average from 25 to 44 degrees.

Joint moments and contact forces showed similar profiles across athletes (Figures 3 and 4). Ankle dorsiflexion moments (peak 0.87 ± 0.34 Nm·kg^{-1}) during the first ~ 15% were followed by a plantarflexion moment (peak −2.19 ± 0.81 Nm·kg^{-1}) for the remainder of ground contact. Peak knee moments ranged from 2.63 ± 1.81 Nm·kg^{-1} (flexion) immediately after touchdown, to −2.99 ± 0.87 Nm·kg^{-1} (extension). In contrast, peak hip extension moments (−9.93 ± 4.07 Nm·kg^{-1}) occurred during the first ~ 25% of ground contact, whereas the peak hip flexion moments (2.74 ± 0.78 Nm·kg^{-1}) were in preparation for take off.

Ankle joint contact forces were the largest of all joints with average peak forces of 24.49 ± 9.49 BW and contact force impulses of 1.25 ± 0.26 BW·s. The total peak compressive knee contact forces were 12.86 ± 3.76 BW across all athletes and were higher on the medial (8.38 ± 2.85 BW) compared to the lateral (6.69 ± 2.47 BW) tibiofemoral joint. Knee contact force impulses were 0.6 ± 0.19 BW·s (medial) and 0.49 ± 0.14 BW·s (lateral). The total peak anterior shear force on the knee joint was 1.06 ± 0.73 BW. Peak hip joint contact forces were 13.76 ± 4.97 BW, with an average force impulse on the hip joint of 0.91 ± 0.32 BW·s.

Muscle and tendon loading

At a tissue level, nine of the twelve examined muscles contracted eccentrically during the ground-contact phase of decelerating (Figure 5; Table 1). The hip extensors and abductors (gluteus maximus/medius) produced the largest forces eccentrically during the first ~ 45-55% of ground contact. The hamstrings (biceps femoris, semitendinosus, semimembranosus) were the only muscles that did not show any active lengthening, whereas the quadriceps (vastus medialis/lateralis, rectus femoris) acted eccentrically throughout the entire ground-contact phase. Of the shank musculature, the tibialis anterior produced its largest forces whilst lengthening during the first ~ 20% of ground contact. This was followed by an active lengthening phase of the calf muscles (gastrocnemius medialis/lateralis, soleus) after which the tibialis anterior acted eccentrically during the final ~ 30% before take off (Figure 5; Table 1). Across the nine eccentrically acting muscles, the average peak lengthening velocity ranged between 0.39 and 1.6 L_m^0·s^{-1} (Table 1).

Peak forces on the patellar and Achilles tendons (Figure 6) were 7.42 ± 1.84 BW and 5.58 ± 1.8 BW, respectively. The tendon force impulses were 0.65 ± 0.24 BW·s (patellar) and 0.37 ± 0.16 BW·s (Achilles), whereas the weighted tendon force impulses were 3.92 ± 0.97 BW·s (patellar) and 2.71 ± 0.77 BW·s (Achilles).

From the reported loading patterns presented above, three general phases of ground contact were distinguished. These phases were defined as being from 1) touchdown to ~ 30% of ground contact, 2) ~30-80% of ground contact, and 3) 80% to take off (Figure 7).

Discussion

The aim of this study was to provide a comprehensive description of the kinetic demands of horizontal decelerations at the whole-body, structural, and tissue-specific levels of the musculoskeletal system. To the best of our knowledge, this is the first unified quantification and in-depth examination of deceleration-specific loading patterns across different levels.

The peak vertical GRFs occurred during the first 30% of ground contact (phase 1) and were ~ 1.5-3 times (up to 6.3 BW) the peak vertical forces experienced during other common running tasks in team sports, such as accelerated or constant-speed running (Cavanagh & Lafortune, 1980; Nagahara et al., 2018; Verheul et al., 2019, 2021). Peak horizontal GRFs on the other hand, were up to four times (1.9 BW) the external forces experienced during

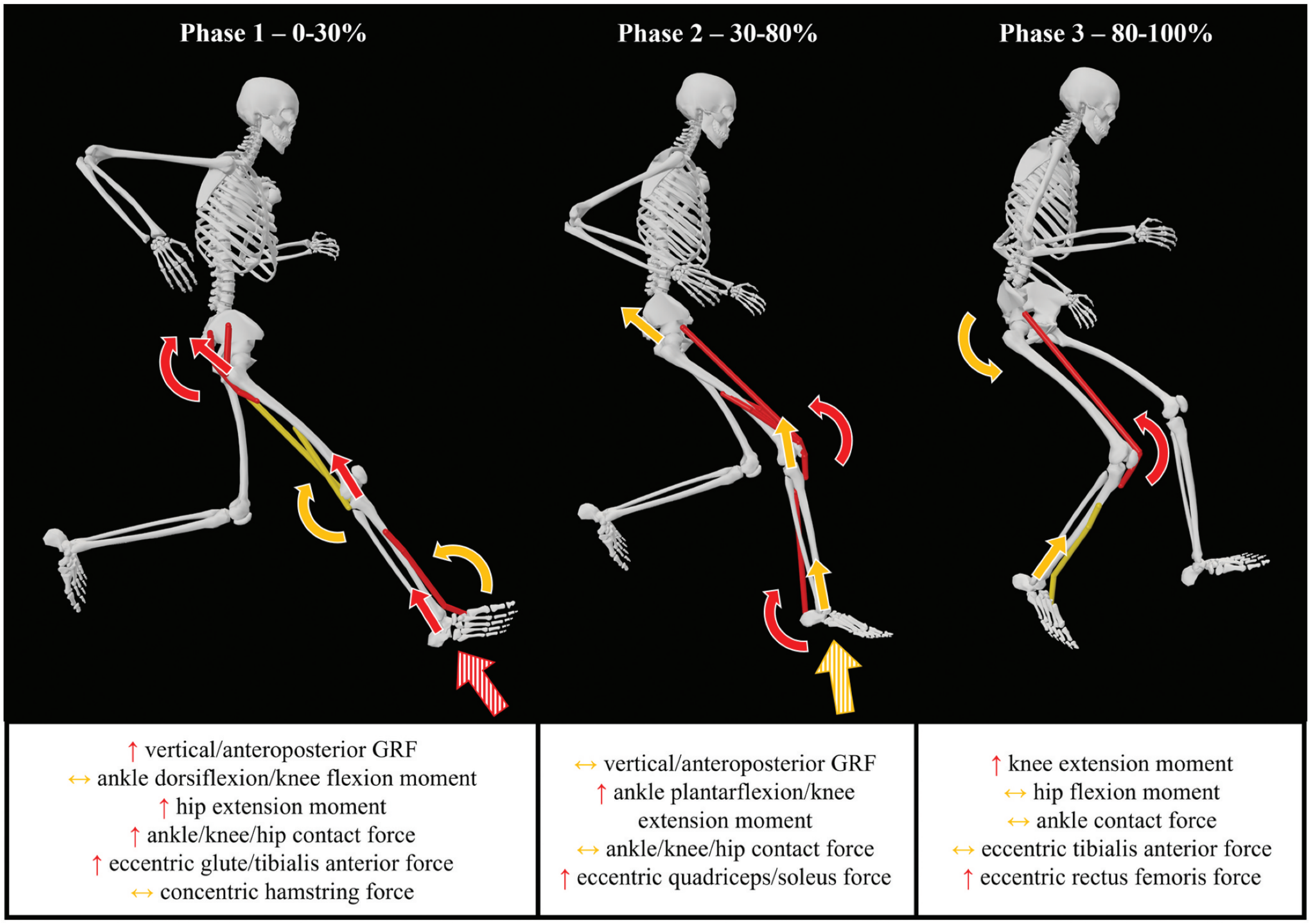

Figure 7. Visual representation of the three phases of ground contact during horizontal deceleration. The key loading mechanisms for each phase are highlighted in red (high load) or yellow (moderate load) for the joints and muscles. Striped straight arrows represent the external ground reaction force (GRF), solid straight arrows represent joint contact forces, solid curved arrows represent joint moments.

other running tasks (Cavanagh & Lafortune, 1980; Nagahara et al., 2018). This indicates that increases in the external force demands when rapidly decelerating are more pronounced in the horizontal component. This is not surprising, since the horizontal GRF impulse is essential for slowing down the horizontal velocity of the body and can thus be regarded as a key contributor to deceleration performance. An important question, however, is which characteristics of the horizontal GRF profile should be targeted to improve deceleration performance. For example, mean horizontal forces have previously been found to be associated with *acceleration* performance, but not the peak horizontal force (Nagahara et al., 2021). Similarly, horizontal force impulse, vertical-horizontal force ratio, or the angle of inclination, rather than peak horizontal force, may be important performance indicators when decelerating. Although the presented data cannot answer this question directly, it adds further quantification of GRFs to the limited body of knowledge on external forces experienced during horizontal decelerations and can function as a benchmark for future investigations. Such investigations may focus on identifying specific horizontal force characteristics that are associated with deceleration performance at various approach speeds, different deceleration steps, the state of fatigue, or distinct force profiles between males and females.

Immediately after landing (phase 1), a short ankle dorsiflexion moment is required to slow down the foot and rapid plantarflexing movement. To achieve this, a substantial eccentric force is required of the tibialis anterior muscle (peak 2.95 ± 1.07 BW). These forces are up to six times the peak tibialis anterior forces produced during running and sprinting, which are required during the swing phase (i.e., in the absence of external forces) and primarily concentric in nature (Almonroeder et al., 2013; Dorn et al., 2012; Hamner & Delp, 2013). This combination of large forces and eccentric contraction are likely to put the tibialis anterior at increased risk of muscle damage and fatigue when performing a high volume of horizontal decelerations, and contribute to an increased risk of shank injuries, such as tibial stress fractures or chronic anterior compartment syndrome (Mayor, 1956; Tweed & Barnes, 2008). In addition, the peak ankle joint contact forces, which coincided with the peak GRFs during the first 30% of ground contact, were ~ 2-2.5 times greater compared to constant-speed running (at 4.3 m·s^{-1}) (Rooney & Derrick, 2013). Large ankle forces may contribute to the high ankle injury burden observed in court and team sports (Fong et al., 2007) during which decelerations are known to be common (D. J. Harper et al., 2019). A focus on training the eccentric tibialis anterior strength, as well as the other ankle-stabilising musculature, is thus likely to be beneficial for athletes who frequently perform horizontal decelerations.

Forces produced by the calf muscles were not considerably large. The peak gastrocnemius and soleus forces were substantially smaller than the peak forces required of the calves for accelerated running or sprinting (Dorn et al., 2012; Pandy et al., 2021). Indeed, the gastrocnemius and soleus do not appear to play a major role in decelerating the horizontal velocity of the body, but rather in providing vertical support during ground contact (Hamner & Delp, 2013; Maniar et al., 2019; Mateus et al., 2020). Calf muscle strength capacity does, therefore, not appear to be a major factor for deceleration performance beyond the typical requirements of team-sports movements. However, the coincidence of peak soleus forces and a relatively high lengthening velocity, may predispose the soleus to accumulating muscle damage and fatigue due to repeated deceleration efforts. Eccentric training is thus likely to be beneficial to enhance fatigue (and possibly injury) resistance of the soleus muscle.

For a short period after touchdown (phase 1), a knee flexion moment is required. This moment is produced by the hamstring muscles to reduce the anterior velocity of the shank. After this, a knee extension moment is primarily required to resist knee collapse from ~ 25-80% (phase 2) of ground contact. The peak knee extension moments in this study (-2.99 ± 0.87 $Nm{\cdot}kg^{-1}$) are in line with previous horizontal deceleration studies (Fitzwilliam et al., 2024; Nedergaard et al., 2014; Peel et al., 2019) and are accounted for by the rectus femoris and vasti muscles – making the quadriceps the key muscle group that reduce the horizontal centre of mass velocity (Hamner & Delp, 2013; Maniar et al., 2019; Mateus et al., 2020). It is important to note that the knee continues to flex throughout ground contact, leading to the quadriceps muscles to produce forces whilst lengthening only (i.e., eccentric action), with potentially damaging effects to the muscles and increased risk of fatigue or injury. Furthermore, the compressive knee joint contact forces in this study were found to be higher compared to previous investigations. Peak tibiofemoral joint contact forces were between 20–45% (medial) and up to 36% (lateral) greater compared to constant-speed running (Saxby, Modenese, et al., 2016), decelerations (Cassiolas et al., 2023) (*note*: approach speeds were not reported in this study and a direct comparison cannot be made), and 45° and 90° changes of direction (Cassiolas et al., 2023; Saxby, Modenese, et al., 2016). However, the peak anterior shear forces in the knee (mean 1.06 BW) were similar (0.97–1.16 BW) compared to running, decelerations, and 45° sidestepping (Moon et al., 2023; Peel et al., 2019). This highlights the emphasised loading posed on the knee joint in axial direction during decelerating at a high intensity. Rapid horizontal decelerations may thus also elevate the risk of sustaining injuries that have been associated with high compressive tibiofemoral joint contact forces, including rupture of the anterior cruciate ligament (Beaulieu et al., 2023; Wojtys et al., 2016) and post-traumatic knee osteoarthritis (Buckwalter et al., 2013; Lohmander et al., 2007; Saxby, Bryant, et al., 2016).

A substantial hip extension moment is required immediately after landing (phase 1). This large moment is primarily produced by the gluteus maximus muscle (Mateus et al., 2020) and is essential to stabilise the pelvis, prevent anterior collapse of the trunk, and maintain a relatively low hip range of motion of ~ 19° throughout ground contact (compared to, e.g., >50° during running (Hamner et al., 2010)). As a result, the gluteus maximus produced the largest forces from the muscles examined in this study, which were 2.5–5 times the maximal gluteus maximus force required during running and sprinting (Dorn et al., 2012; Hamner & Delp, 2013). Accordingly, the hip joint contact forces also were found to be substantial – 36–43% higher than those reported for running at 4.3 $m{\cdot}s^{-1}$ (Rooney & Derrick, 2013). Although the gluteus maximus forces in this study were largely eccentric in nature, its lengthening velocity and duration was considerably smaller compared to the vasti, and thus less likely to have a major damaging, fatiguing, or injurious effect. From a performance perspective, however, these findings warrant a focus on training the maximal force capacity of the gluteus maximus to enable the production of sufficiently high hip extension moments to achieve appropriate trunk stabilisation.

Eccentric forces were produced at the highest lengthening velocities by the vasti muscles, followed by the tibialis anterior, gluteus minimus, rectus femoris, and soleus muscles. Moreover, the quadriceps muscles lengthened for the entire ground-contact phase. Other muscles, however, contracted eccentrically for only half (46–57%; tibialis anterior, soleus, and glute muscles) or a quarter (~26%; gastrocnemius muscles) of ground contact. These findings are in line with previous studies that have shown the quadriceps to be the primary moderators of braking (Hamner & Delp, 2013; Maniar et al., 2019; Mateus et al., 2020). The observed combination of substantial force production, high muscle-lengthening velocities, and long duration of eccentric muscle contraction, exposes the quadriceps to the highest risk of accumulating muscle damage and fatigue due to horizontal decelerations. Although this may lead to muscle soreness or injury to the quadriceps, it can also elevate the injury risk for other musculoskeletal structures. For example, the quadriceps are an important stabiliser of the knee joint and reduced muscle functioning (e.g., due to fatigue) may expose athletes to an increased risk of knee injuries, such as the anterior cruciate ligament. Decelerations have indeed been found to be one of the foremost movements during which anterior cruciate ligament injuries are sustained (Cochrane et al., 2007; Della Villa et al., 2020; Johnston et al., 2018). Eccentric conditioning of the quadriceps, in particular in the range of the reported muscle-lengthening velocities, may thus be useful to help prevent injuries that are associated with decelerations. Future work should investigate the most appropriate movements and exercises during which such muscle-lengthening velocities are achieved.

Peak patellar and Achilles tendon loading was found on average to be 7.42 BW and 5.58 BW, respectively. Previous studies have reported peak patellar tendon forces to range between ~ 6-9.22 BW for constant-speed running at 4.4–9 $m{\cdot}s^{-1}$ (Dorn et al., 2012; Hagen et al., 2023; Hamner & Delp, 2013), and peak Achilles tendon forces from ~ 6.3-11.9 BW for running at 5–9 $m{\cdot}s^{-1}$ (Dorn et al., 2012; Hamner & Delp, 2013; Scott & Winter, 1990; Starbuck et al., 2021) (*note*: if tendon forces were not reported, individual muscle forces were summed as in this study to get an estimate of tendon loading for comparison). Moreover, Achilles tendon force impulses have been found to be 0.47–0.85 BW·s for running at a relatively easy speed of ~3.6 $m{\cdot}s^{-1}$ (Almonroeder et al., 2013; Rice & Patel, 2017), compared to 0.37 ± 0.16 BW·s in this study. Both tendons thus do not appear to experience biomechanical loads during decelerations beyond those typically experienced during other common running tasks. Hence, the initial steps of horizontal decelerations are unlikely to pose an excessive risk of patellar- or Achilles-tendon injuries beyond that of constant-speed running.

Despite recent emphasis on the importance of training deceleration skills (D. Harper et al., 2024; D. J. Harper et al., 2022; McBurnie et al., 2022), there is a current lack of strong longitudinal evidence for the effectiveness of targeted interventions to enhance deceleration performance (D. J. Harper et al., 2022; Marvin et al., 2023). One possible reason is that eccentric training and adaptation mechanisms are yet not well understood (Hody et al., 2019; LaStayo et al., 2003). Another likely reason is that the kinetic characteristics which contribute to deceleration ability, have not previously been specified or comprehensively quantified in detail. Hence, it is our desire that the biomechanical loads at a whole-body, structural, and tissue level, as described in this study will help inform training prescription. Although several recommendations are highlighted above, we encourage future investigation to thoroughly examine the effectiveness of targeting specific loading mechanisms, to promote enhanced deceleration performance or aid injury prevention.

Similar to most musculoskeletal modelling studies, a Hill-type muscle model (Millard et al., 2013; Zajac, 1989) was used to describe muscle mechanics. It has been well documented that the Hill-model does not model history-dependent muscle dynamics, such as residual force enhancement due to active muscle lengthening, well (Ettema & Meijer, 2000; McGowan et al., 2013; Yeo et al., 2023). Since eccentric contractions are common in the lower-limb muscles during horizontal decelerations, we did not examine the underlying muscle dynamics (e.g., fibre lengths, pennation angles) predicted by the Hill-type model for each muscle. Nevertheless, it is likely that the absence of history-dependency in the Hill-type model is compensated for by adjustments in muscle excitations (McGowan et al., 2013). The overall forces produced and experienced by each muscle-tendon unit as presented in this study are thus likely to be truthful estimates of the actual forces experienced by individual muscles during horizontal decelerations.

The biomechanical loads presented in this study are for the initial steps of decelerating after a straight sprint. Given the high deceleration intensity (i.e., change of centre of mass velocity) of the first deceleration steps, it is likely that the musculoskeletal system also experiences the highest forces during these steps. These loading patterns are expected to differ for subsequent steps and are affected by various other factors, such as approach speed, change of direction or pivot angle, or anticipating follow on tasks (e.g., kicking, tackling, etc). Moreover, how such movement variations affect the loads experienced are likely to vary across the whole-body, structural, and tissue level. We thus recommend that future work examines the musculoskeletal demands of deceleration efforts with whilst introducing movement variations, using the loads presented in this study as a reference point.

Some caution should be exercised in the interpretation and generalisation of the results presented in this study. For example, it is important to acknowledge the relatively small sample size ($n = 15$) from a homogenous athletic background (i.e., recreational team sports) used to describe the distinct kinetic demands of horizontal decelerations. Moreover, specific individual characteristics (e.g., sex or athletic experience and proficiency) can be expected to affect the loading patterns experienced during horizontal decelerations. Therefore, we recommend future studies to extend this foundational descriptive work by examining larger sample sizes across a higher diversity of athletic populations. Although the loading patterns described in this study can be used to understand the general deceleration-specific demands on the musculoskeletal system, detailed individualised training recommendations cannot be made.

Conclusions

In this study we examine the biomechanical loads across the whole-body, structural, and tissue level during horizontal decelerations. This is a further step towards the understanding of deceleration performance characteristics and provides the first empirical explanations for why decelerating movements can be particularly damaging and injury-inducing. The external GRFs, lower-limb joint moments and contact forces, and eccentric force requirements of the tibialis anterior, soleus, quadriceps, and gluteal muscles are particularly high during horizontal decelerations. Although the generalisation of these findings to other athletic groups remains to be established, the identified loading patterns in this study may help inform deceleration-specific training prescription, to target horizontal deceleration performance, or fatigue and injury resistance in team-sports athletes.

Disclosure statement

No potential conflict of interest was reported by the author(s).

Funding

The author(s) reported there is no funding associated with the work featured in this article.

ORCID

Jasper Verheul http://orcid.org/0000-0002-2939-8046

Data availability statement

The data that support the findings of this study are openly available on GitHub (https://github.com/JasperVerheul/horizontal-deceleration-forces).

References

Akenhead, R., Hayes, P. R., Thompson, K. G., & French, D. (2013). Diminutions of acceleration and deceleration output during professional football match play. *Journal of Science & Medicine in Sport*, *16*(6), 556–561. https://doi.org/10.1016/j.jsams.2012.12.005

Almonroeder, T., Willson, J. D., & Kernozek, T. W. (2013). The effect of foot strike pattern on achilles tendon load during running. *Annals of Biomedical Engineering*, *41*(8), 1758–1766. https://doi.org/10.1007/s10439-013-0819-1

Armstrong, R. B., Ogilvie, R. W., & Schwane, J. A. (1983). Eccentric exercise-induced injury to rat skeletal muscle. *Journal of Applied Physiology*, *54*(1), 80–93. https://doi.org/10.1152/jappl.1983.54.1.80

Beaulieu, M. L., Ashton-Miller, J. A., & Wojtys, E. M. (2023). Loading mechanisms of the anterior cruciate ligament. *Sports Biomechanics*, *22*(1), 1–29. https://doi.org/10.1080/14763141.2021.1916578

Bedo, B. L. S., Catelli, D. S., Lamontagne, M., & Santiago, P. R. P. (2020). A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions. *Computer Methods in Biomechanics and Biomedical Engineering*, *23*(10), 658–663. https://doi.org/10.1080/10255842.2020.1757662

Bedo, B. L. S., Mantoan, A., Catelli, D. S., Cruaud, W., Reggiani, M., & Lamontagne, M. (2021). BOPS: A matlab toolbox to batch musculoskeletal data processing for OpenSim. *Computer Methods in Biomechanics and Biomedical Engineering*, *24*(10), 1104–1114. https://doi.org/10.1080/10255842.2020.1867978

Bezodis, N., Colyer, S., Nagahara, R., Bayne, H., Bezodis, I., Morin, J.-B., Murata, M., & Samozino, P. (2021). Ratio of forces during sprint acceleration: A comparison of different calculation methods. *Journal of Biomechanics*, *127*, 110685. https://doi.org/10.1016/j.jbiomech.2021.110685

Buckwalter, J. A., Anderson, D. D., Brown, T. D., Tochigi, Y., & Martin, J. A. (2013). The roles of mechanical stresses in the pathogenesis of osteoarthritis: Implications for treatment of joint injuries. *Cartilage*, *4*(4), 286. https://doi.org/10.1177/1947603513495889

Cassiolas, G., DiPaolo, S., Marchiori, G., Grassi, A., Della Villa, F., Bragonzoni, L., Visani, A., Giavaresi, G., Fini, M., Zaffagnini, S., & Lopomo, N. F. (2023). Knee joint contact forces during high-risk dynamic tasks: 90° change of direction and deceleration movements. *Bioengineering*, *10* Article(2), 179. https://doi.org/10.3390/bioengineering10020179

Cavanagh, P. R., & Lafortune, M. A. (1980). Ground reaction forces in distance running. *Journal of Biomechanics*, *13*(5), 397–406. https://doi.org/10.1016/0021-9290(80)90033-0

Cochrane, J. L., Lloyd, D. G., Buttfield, A., Seward, H., & McGivern, J. (2007). Characteristics of anterior cruciate ligament injuries in Australian football. *Journal of Science & Medicine in Sport*, *10*(2), 96–104. https://doi.org/10.1016/j.jsams.2006.05.015

Cross, J., Kung, S. M., & Welch, N. (2024). Assessing the reliability of biomechanical variables during a horizontal deceleration task in healthy adults. *Journal of Sports Sciences*, *42*(3), 263–269. https://doi.org/10.1080/02640414.2024.2327874

Dalen, T., Jørgen, I., Gertjan, E., Havard, H. G., & Ulrik, W. (2016). Player load, acceleration, and deceleration during 45 competitive matches of elite soccer. *The Journal of Strength & Conditioning Research*, *30*(2), 351–359. https://doi.org/10.1519/JSC.0000000000001063

de Hoyo, M., Cohen, D. D., Sañudo, B., Carrasco, L., Álvarez-Mesa, A., Del Ojo, J. J., Domínguez-Cobo, S., Mañas, V., & Otero-Esquina, C. (2016). Influence of football match time–motion parameters on recovery time course of muscle damage and jump ability. *Journal of Sports Sciences*, *34*(14), 1363–1370. https://doi.org/10.1080/02640414.2016.1150603

Della Villa, F., Buckthorpe, M., Grassi, A., Nabiuzzi, A., Tosarelli, F., Zaffagnini, S., & Della Villa, S. (2020). Systematic video analysis of ACL injuries in professional male football (soccer): Injury mechanisms, situational patterns and biomechanics study on 134 consecutive cases. *British Journal of Sports Medicine*, *54*(23), 1423–1432. https://doi.org/10.1136/bjsports-2019-101247

Dorn, T. W., Schache, A. G., & Pandy, M. G. (2012). Muscular strategy shift in human running: Dependence of running speed on hip and ankle muscle performance. *Journal of Experimental Biology*, *215*(11), 1944–1956. https://doi.org/10.1242/jeb.064527

Dos'santos, T., Thomas, C., & Jones, P. A. (2021). How early should you brake during a 180° turn? A kinetic comparison of the antepenultimate, penultimate, and final foot contacts during a 505 change of direction speed test. *Journal of Sports Sciences*, *39*(4), 395–405. https://doi.org/10.1080/02640414.2020.1823130

Edwards, W. B. (2018). Modeling overuse injuries in sport as a mechanical fatigue phenomenon. *Exercise and Sport Sciences Reviews*, *46*(4), 224–231. https://doi.org/10.1249/JES.0000000000000163

Ettema, G. J. C., & Meijer, K. (2000). Muscle contraction history: Modified hill versus an exponential decay model. *Biological Cybernetics*, *83*(6), 491–500. https://doi.org/10.1007/s004220000190

Firminger, C. R., & Edwards, W. B. (2021). Effects of cyclic loading on the mechanical properties and failure of human patellar tendon. *Journal of Biomechanics*, *120*, 110345. https://doi.org/10.1016/j.jbiomech.2021.110345

Fitzwilliam, E., Steventon-Lorenzen, N., Opar, D., Schache, A. G., & Maniar, N. (2024). Lower limb joint mechanics during maximal accelerative and decelerative running. *Medicine & Science in Sports and Exercise*, *56*(9), 1655. https://doi.org/10.1249/MSS.0000000000003445

Fong, D. T.-P., Hong, Y., Chan, L.-K., Yung, P. S.-H., & Chan, K.-M. (2007). A systematic review on ankle injury and ankle sprain in sports. *Sports Medicine*, *37*(1), 73–94. https://doi.org/10.2165/00007256-200737010-00006

Gastin, P. B., Hunkin, S. L., Fahrner, B., & Robertson, S. (2019). Deceleration, acceleration, and impacts are strong contributors to muscle damage in professional Australian football. *The Journal of Strength & Conditioning Research*, *33*(12), 3374. https://doi.org/10.1519/JSC.0000000000003023

Hagen, M., Vanmechelen, A., Cloet, E., Sellicaerts, J., Van Welden, K., Verstraete, J., Catelli, D. S., Verschueren, S., & Vanrenterghem, J. (2023). Increasing step frequency reduces patellofemoral joint stress and patellar tendon force impulse more at low running speed. *Medicine & Science in Sports and Exercise*, *55*(9), 1555–1563. https://doi.org/10.1249/MSS.0000000000003194.

Hamner, S. R., & Delp, S. L. (2013). Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds. *Journal of Biomechanics*, *46*(4), 780–787. https://doi.org/10.1016/j.jbiomech.2012.11.024

Hamner, S. R., Seth, A., & Delp, S. L. (2010). Muscle contributions to propulsion and support during running. *Journal of Biomechanics*, *43*(14), 2709–2716. https://doi.org/10.1016/j.jbiomech.2010.06.025

Harper, D., Cervantes, C., Dyke, M. V., Evans, M., McBurnie, A., Santos, T. D., Eriksrud, O., Cohen, D., Rhodes, D., Carling, C., & Kiely, J. (2024). The braking performance framework: Practical recommendations and guidelines to enhance horizontal deceleration ability in multi-directional sports. *International Journal of Strength and Conditioning*, *4* Article(1) https://doi.org/10.47206/ijsc.v4i1.351

Harper, D. J., Carling, C., & Kiely, J. (2019). High-intensity acceleration and deceleration demands in elite team sports competitive match play: A systematic review and meta-analysis of observational studies. *Sports Medicine*, *49*(12), 1923–1947. https://doi.org/10.1007/s40279-019-01170-1

Harper, D. J., & Kiely, J. (2018). Damaging nature of decelerations: Do we adequately prepare players? *BMJ Open Sport and Exercise Medicine*, *4*(1), 1–3. https://doi.org/10.1136/bmjsem-2018-000379

Harper, D. J., McBurnie, A. J., Santos, T. D., Eriksrud, O., Evans, M., Cohen, D. D., Rhodes, D., Carling, C., & Kiely, J. (2022). Biomechanical and neuromuscular performance requirements of horizontal deceleration: A review with implications for random intermittent multi-directional sports. *Sports Medicine*, *52*(10), 2321–2354. https://doi.org/10.1007/s40279-022-01693-0

Hewit, J., Cronin, J., Button, C., & Hume, P. (2011). Understanding deceleration in sport. *Strength & Conditioning Journal*, *33*(1), 47. https://doi.org/10.1519/SSC.0b013e3181fbd62c

Hody, S., Croisier, J.-L., Rogister, B., & Leprince, P. (2019). Eccentric muscle contractions: Risks and benefits. *Frontiers in Physiology*, *10*, 442082. https://doi.org/10.3389/fphys.2019.00536

Jaspers, A., Kuyvenhoven, J. P., Staes, F., Frencken, W. G. P., Helsen, W. F., & Brink, M. S. (2018). Examination of the external and internal load indicators' association with overuse injuries in professional soccer players.

Journal of Science & Medicine in Sport, *21*(6), 579–585. https://doi.org/10.1016/j.jsams.2017.10.005

Johnston, J. T., Mandelbaum, B. R., Schub, D., Rodeo, S. A., Matava, M. J., Silvers-Granelli, H. J., Cole, B. J., ElAttrache, N. S., McAdams, T. R., & Brophy, R. H. (2018). Video analysis of anterior cruciate ligament tears in professional American football athletes. *American Journal of Sports Medicine*, *46*(4), 862–868. https://doi.org/10.1177/0363546518756328

Lakomy, J., & Haydon, D. T. (2004). The effects of enforced, rapid deceleration on performance in a multiple sprint test. *The Journal of Strength & Conditioning Research*, *18*(3), 579. https://doi.org/10.1519/1533-4287(2004)18<579:TEOERD>2.0.CO;2

LaStayo, P. C., Woolf, J. M., Lewek, M. D., Snyder-Mackler, L., Reich, T., & Lindstedt, S. L. (2003). Eccentric muscle contractions: Their contribution to injury, prevention, rehabilitation, and sport. *Journal of Orthopaedic & Sports Physical Therapy*, *33*(10), 557–571. https://doi.org/10.2519/jospt.2003.33.10.557

Lohmander, L. S., Englund, P. M., Dahl, L. L., & Roos, E. M. (2007). The long-term consequence of anterior cruciate ligament and meniscus injuries: Osteoarthritis. *American Journal of Sports Medicine*, *35*(10), 1756–1769. https://doi.org/10.1177/0363546507307396

Maniar, N., Schache, A. G., Cole, M. H., & Opar, D. A. (2019). Lower-limb muscle function during sidestep cutting. *Journal of Biomechanics*, *82*, 186–192. https://doi.org/10.1016/j.jbiomech.2018.10.021

Martínez-Hernández, D., Quinn, M., & Jones, P. (2023). Linear advancing actions followed by deceleration and turn are the most common movements preceding goals in male professional soccer. *Science and Medicine in Football*, *7*(1), 25–33. https://doi.org/10.1080/24733938.2022.2030064

Martins, F., Marques, A., França, C., Sarmento, H., Henriques, R., Ihle, A., de Maio Nascimento, M., Saldanha, C., Przednowek, K., & Gouveia, É. R. (2023). Weekly external load performance effects on sports injuries of male professional football players. *International Journal of Environmental Research and Public Health*, *20* Article(2), 1121. https://doi.org/10.3390/ijerph20021121

Marvin, L. A., Read, P., McLean, B., Palmer, S., & Fransen, J. (2023). *Training interventions for improved deceleration ability in adult team-based field sports athletes: A systematic review and meta-analysis of the literature*. SportRxiv. https://doi.org/10.51224/SRXIV.352

Mateus, R. B., Ferrer-Roca, V., João, F., & Veloso, A. P. (2020). Muscle contributions to maximal single-leg forward braking and backward acceleration in elite athletes. *Journal of Biomechanics*, *112*, 110047. https://doi.org/10.1016/j.jbiomech.2020.110047

Mayor, G. E. (1956). The anterior tibial syndrome. *The Journal of Bone and Joint Surgery British Volume*, *38-B*(2), 513–517. https://doi.org/10.1302/0301-620X.38B2.513

McBurnie, A. J., Harper, D. J., Jones, P. A., & Dos'santos, T. (2022). Deceleration training in team sports: Another potential 'vaccine' for sports-related injury? *Sports Medicine*, *52*(1), 1–12. https://doi.org/10.1007/s40279-021-01583-x

McGowan, C. P., Neptune, R. R., & Herzog, W. (2013). A phenomenological muscle model to assess history dependent effects in human movement. *Journal of Biomechanics*, *46*(1), 151–157. https://doi.org/10.1016/j.jbiomech.2012.10.034

Millard, M., Uchida, T., Seth, A., & Delp, S. L. (2013). Flexing computational muscle: Modeling and simulation of musculotendon dynamics. *Journal of Biomechanical Engineering*, *135*(2), 1–11. https://doi.org/10.1115/1.4023390

Moon, J., Koo, D., Kim, S., & Panday, S. B. (2023). Effect of sprinting velocity on anterior cruciate ligament and knee load during sidestep cutting. *Frontiers in Bioengineering and Biotechnology*, *11*, 1033590. https://doi.org/10.3389/fbioe.2023.1033590

Morin, J.-B., Edouard, P., & Samozino, P. (2011). Technical ability of force application as a determinant factor of sprint performance. *Medicine & Science in Sports and Exercise*, *43*(9), 1680–1688. https://doi.org/10.1249/MSS.0b013e318216ea37

Nagahara, R., Kanehisa, H., Matsuo, A., & Fukunaga, T. (2021). Are peak ground reaction forces related to better sprint acceleration performance? *Sports Biomechanics*, *20*(3), 360–369. https://doi.org/10.1080/14763141.2018.1560494

Nagahara, R., Mizutani, M., Matsuo, A., Kanehisa, H., & Fukunaga, T. (2018). Association of sprint performance with ground reaction forces during acceleration and maximal speed phases in a single sprint. *Journal of Applied Biomechanics*, *34*(2), 104–110. https://doi.org/10.1123/jab.2016-0356

Nedergaard, N. J., Kersting, U., & Lake, M. J. (2014). Using accelerometry to quantify deceleration during a high-intensity soccer turning manoeuvre. *Journal of Sports Sciences*, *32*(20), 1897–1905. https://doi.org/10.1080/02640414.2014.965190

Pandy, M. G., Lai, A. K. M., Schache, A. G., & Lin, Y.-C. (2021). How muscles maximize performance in accelerated sprinting. *Scandinavian Journal of Medicine & Science in Sports*, *31*(10), 1882–1896. https://doi.org/10.1111/sms.14021

Peel, S. A., Schroeder, L. E., Sievert, Z. A., & Weinhandl, J. T. (2019). Comparing anterior cruciate ligament injury risk variables between unanticipated cutting and decelerating tasks. *Journal of Applied Biomechanics*, *35*(2), 101–106. https://doi.org/10.1123/jab.2017-0343

Proske, U., & Morgan, D. (2001). Muscle damage from eccentric exercise: Mechanism, mechanical signs, adaptation and clinical applications. *Journal of Physiology, Paris*, *537*(2), 333–345. https://doi.org/10.1111/j.1469-7793.2001.00333.x

Rajagopal, A., Dembia, C. L., DeMers, M. S., Delp, D. D., Hicks, J. L., & Delp, S. L. (2016). Full-body musculoskeletal model for muscle-driven simulation of human gait. *IEEE Transactions on Biomedical Engineering*, *63*(10), 2068–2079. https://doi.org/10.1109/TBME.2016.2586891

Rhodes, D., Valassakis, S., Bortnik, L., Eaves, R., Harper, D., & Alexander, J. (2021). The effect of high-intensity accelerations and decelerations on match outcome of an elite English league two football team. *International Journal of Environmental Research and Public Health*, *18*(18), 9913. https://doi.org/10.3390/ijerph18189913

Rice, H., & Patel, M. (2017). Manipulation of foot strike and footwear increases achilles tendon loading during running. *American Journal of Sports Medicine*, *45*(10), 2411–2417. https://doi.org/10.1177/0363546517704429

Rooney, B. D., & Derrick, T. R. (2013). Joint contact loading in forefoot and rearfoot strike patterns during running. *Journal of Biomechanics*, *46*(13), 2201–2206. https://doi.org/10.1016/j.jbiomech.2013.06.022

Saxby, D. J., Bryant, A. L., Modenese, L., Gerus, P., Killen, B. A., Konrath, J., Fortin, K., Wrigley, T. V., Bennell, K. L., Cicuttini, F. M., Vertullo, C., Feller, J. A., Whitehead, T., Gallie, P., & Lloyd, D. G. (2016). Tibiofemoral contact forces in the anterior cruciate ligament–reconstructed knee. *Medicine & Science in Sports and Exercise*, *48*(11), 2195. https://doi.org/10.1249/MSS.0000000000001021

Saxby, D. J., Modenese, L., Bryant, A. L., Gerus, P., Killen, B., Fortin, K., Wrigley, T. V., Bennell, K. L., Cicuttini, F. M., & Lloyd, D. G. (2016). Tibiofemoral contact forces during walking, running and sidestepping. *Gait & Posture*, *49*, 78–85. https://doi.org/10.1016/J.GAITPOST.2016.06.014

Scott, S. H., & Winter, D. A. (1990). Internal forces at chronic running injury sites. *Medicine & Science in Sports and Exercise*, *22*(3), 357–369. https://doi.org/10.1249/00005768-199006000-00013

Seth, A., Hicks, J. L., Uchida, T. K., Habib, A., Dembia, C. L., Dunne, J. J., Ong, C. F., DeMers, M. S., Rajagopal, A., Millard, M., Hamner, S. R., Arnold, E. M., Yong, J. R., Lakshmikanth, S. K., Sherman, M. A., Ku, J. P., Delp, S. L., & Schneidman, D. (2018). OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. *PLOS Computational Biology*, *14*(7), e1006223. https://doi.org/10.1371/journal.pcbi.1006223

Starbuck, C., Bramah, C., Herrington, L., & Jones, R. (2021). The effect of speed on achilles tendon forces and patellofemoral joint stresses in high-performing endurance runners. *Scandinavian Journal of Medicine & Science in Sports*, *31*(8), 1657–1665. https://doi.org/10.1111/sms.13972

Swinnen, W., Hoogkamer, W., De Groote, F., & Vanwanseele, B. (2019). Habitual foot strike pattern does not affect simulated triceps surae muscle metabolic energy consumption during running. *Journal of Experimental Biology*, *22*(23), 1–8. https://doi.org/10.1242/jeb.212449

Tweed, J. L., & Barnes, M. R. (2008). Is eccentric muscle contraction a significant factor in the development of chronic anterior compartment syndrome? A review of the literature. *The Foot*, *18*(3), 165–170. https://doi.org/10.1016/j.foot.2008.06.005

Verheul, J., Gregson, W., Lisboa, P. J., Vanrenterghem, J., & Robinson, M. A. (2019). Whole-body biomechanical load in running-based sports: The validity of estimating ground reaction forces from segmental accelerations. *Journal of Science & Medicine in Sport*, *22*(6), 716–722. https://doi.org/10.1016/j.jsams.2018.12.007

Verheul, J., Nedergaard, N. J., Pogson, M., Lisboa, P. J., Gregson, W., Vanrenterghem, J., & Robinson, M. A. (2021). Biomechanical loading during running: Can a two mass-spring-damper model be used to evaluate ground reaction forces for high-intensity tasks? *Sports Biomechanics*, *20*(5), 571–582. https://doi.org/10.1080/14763141.2019.1584238

Verheul, J., Nedergaard, N. J., Vanrenterghem, J., & Robinson, M. A. (2020). Measuring biomechanical loads in team sports – from lab to field. *Science and Medicine in Football*, *4*(3), 246–252. https://doi.org/10.1080/24733938.2019.1709654

Vigh-Larsen, J. F., Dalgas, U., & Andersen, T. B. (2018). Position-specific acceleration and deceleration profiles in elite youth and senior soccer players. *The Journal of Strength & Conditioning Research*, *32*(4), 1114–1122. https://doi.org/10.1519/JSC.0000000000001918

Wojtys, E. M., Beaulieu, M. L., & Ashton-Miller, J. A. (2016). New perspectives on ACL injury: On the role of repetitive sub-maximal knee loading in causing ACL fatigue failure. *Journal of Orthopaedic Research*, *34*(12), 2059–2068. https://doi.org/10.1002/jor.23441

Yeo, S.-H., Verheul, J., Herzog, W., & Sueda, S. (2023). Numerical instability of Hill-type muscle models. *Journal of the Royal Society Interface*, *20*(199), 20220430. https://doi.org/10.1098/rsif.2022.0430

Young, W. B., Hepner, J., & Robbins, D. W. (2012). Movement demands in Australian rules football as indicators of muscle damage. *The Journal of Strength & Conditioning Research*, *26*(2), 492–496. https://doi.org/10.1519/JSC.0b013e318225a1c4

Zajac, F. E. (1989). Muscle and tendon: Properties, models, scaling, and application to biomechanics and motor control. *Critical Reviews in Biomedical Engineering*, *17*(4), 359–411.