

Breaking Down the Brakes: Mechanical Demands Across Successive Contacts During Maximal 5-m Deceleration

Vasileios Sideris,^{1 2} Joao Marques,¹ Martin Buchheit¹

¹Rehabilitation Department, Aspetar Orthopaedic and Sports Medicine Hospital, FIFA Medical Center, Doha, Qatar

²Department of Physical Education and Sport Science, University of Thessaly, Trikala, Greece

Horizontal deceleration | Braking mechanics | Successive foot contacts | Kinetic energy absorption | Horizontal braking power | Velocity reduction | Joint work distribution | Lower-limb biomechanics | Ground reaction forces | Return to sport

Headline

Rapid deceleration is a fundamental component of multidirectional sports, allowing athletes to stop, redirect momentum and respond to the demands of play. These actions are rarely completed within a single foot contact; instead, braking is distributed across successive contacts that collectively reduce forward momentum and absorb mechanical energy. Previous research has described whole-task and phase-level braking demands, but how these demands are distributed across individual braking contacts remains incompletely understood. In particular, measures of velocity reduction, kinetic-energy removal and joint-level energy absorption provide different mechanical perspectives on the stopping process. A contact-by-contact approach may therefore provide a more complete description of how athletes organise braking and where mechanical demands are expressed across the lower limbs.

Aim

This study aims to profile maximal 5-m deceleration across successive braking contacts by examining how forward velocity and anterior-posterior translational kinetic energy are reduced, how force and power are expressed at the limb level, and how negative work is distributed across the ankle, knee and hip. A secondary aim is to determine what contact-level profiling adds to a whole-body description of deceleration.

Design

This descriptive pilot study included four uninjured recreational male athletes (31.4 ± 3.9 years; 80.5 ± 4.6 kg; 1.84 ± 0.04 m). A convenience sample was used because this exploratory biomechanical pilot aimed to examine the value of contact-level analysis rather than test a specific hypothesis or estimate effects in the wider population. The study was approved by the Aspetar Institutional Review Board (E202501090), and all participants provided written informed consent. Each participant performed three maximal 5-m decelerations in a single session.

Methods

Deceleration task. Testing took place indoors on vinyl floor at the Riyadh Assessment and Movement Analysis Laboratory (RAMAL), Aspetar. After a dynamic warm-up, participants were instructed to cover the 5-m distance and come to a complete stop at the 5-m line as quickly as possible, using the right limb for the final braking contact. After stopping, participants backpedalled away from the line; this phase was not analysed.

A 30-s passive rest period was provided between trials.

Contacts and analysis windows. Vmax was defined as the instant of peak forward centre-of-mass velocity and V0 as the instant at which forward velocity reached zero; V0 occurred during the final stance. Foot contacts were numbered backwards from the final foot contact (FFC, contact 0; right limb): the penultimate (PFC, -1; left), antepenultimate (APFC, -2; right) and the preceding contact (FC-3; left) [1,2] (Figure 1).

For the APFC, PFC and FFC, foot strike and foot-off were identified from the unfiltered vertical ground-reaction force using a 40 N threshold. For contacts preceding the APFC, events were identified using the automated kinematic event-detection algorithm implemented in Vicon Nexus software. All events were verified against synchronised video. Two distinct, non-interchangeable analysis windows were then defined (Figure 2).

- **Interval** — foot strike to the next foot strike, the last running to V0, each named for the contact that starts it. All centre-of-mass quantities use this window. The four analysed intervals collectively cover the period from FC-3 foot strike to V0 without overlap. Contact-initiated interval outcomes represent whole-body changes and should not be interpreted as the isolated mechanical contribution of the limb initiating the interval, particularly during periods of double support.
- **Contact** — foot strike to foot off for that limb, truncated at V0 for the final contact, since force after V0 drives the backpedal rather than braking. All force-platform quantities use this window. Joint work uses the same contact; peak joint moments are taken over its absorption phase, defined below.

Capture and model. A modified full-body Plug-in Gait marker set [3] was used, with additional sacral and medial knee and ankle markers, and clusters replacing the standard thigh and shank wands. Markers were used to track the head, trunk, pelvis, and bilateral thigh, shank, and foot segments; upper-limb segments were not tracked, and therefore no upper-limb markers were included. Trajectories were captured with 16 cameras (Vicon, Oxford, UK) at 250 Hz and forces with six embedded platforms (BMS600900, AMTI, USA) at 1000 Hz, and processed in Vicon Nexus 2.16 with the Conventional Gait Model 2.3 [4]. Marker trajectories and analog force signals were filtered using a fourth-order zero-lag (dual-pass) Butterworth filter with a matched 15 Hz cut-off before inverse-dynamics calculations, to minimise artefacts associated with mismatched filtering of marker and force data [5].



Fig. 1. The maximal 5-m deceleration task. Braking contacts are numbered backwards from the final foot contact (FFC, 0; right): penultimate (PFC, -1; left), antepenultimate (APFC, -2; right) and FC-3 (left). The black and orange lines show forward momentum and AP translational kinetic energy, respectively.

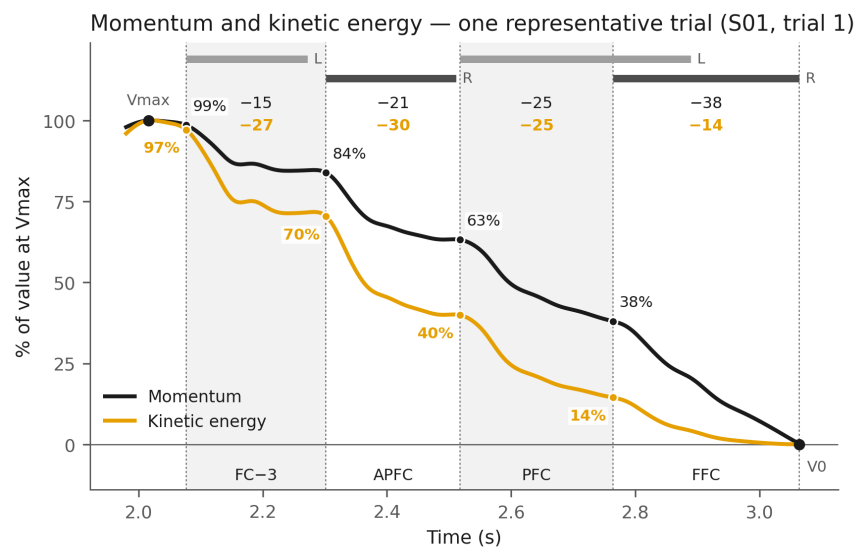


Fig. 2. The two windows, and the momentum and kinetic energy carried through them, on one representative trial (S01, trial 1). Mass-specific forward momentum of the centre of mass (black) and AP translational kinetic energy (orange), each expressed as a percentage of its value at Vmax. Because body mass is constant, mass-specific momentum is identical to forward COM velocity, so the black curve is also the forward velocity trace expressed as a percentage of Vmax. At APFC, a 21-percentage-point reduction in velocity removes 30 percentage points of kinetic energy, whereas at FFC a larger 38-point reduction removes only 14 points. This is because kinetic energy depends on the square of velocity, meaning that the same reduction in velocity corresponds to a greater loss of kinetic energy when it occurs at a higher speed. Alternating bands are the foot-strike-to-foot-strike intervals and the bars above are the foot contacts, which overlap in double support. Values at each foot strike give the percentage carried into that contact; values within each interval give its reduction in percentage points.

Quantities. All quantities are mass-normalised and reported as positive magnitudes. From the centre of mass alone, per interval and without force platforms, we calculated the forward velocity decrement ($\Delta v = v_{in} - v_{out}$) and mean deceleration ($\Delta v \div \text{interval duration}$), together with the two energy quantities defined next. Shares were expressed relative to the total velocity decrement from Vmax to V0 and the mass-specific anteroposterior translational kinetic energy present at Vmax. The four analysed intervals accounted for 97.4% and 95.4% of those totals, respectively; the remainder was removed between Vmax and FC-3 foot strike.

Anteroposterior translational kinetic energy and braking power. For each contact-initiated interval, mass-specific anteroposterior translational kinetic-energy reduction was calculated as:

$$\Delta KE_{AP} = \frac{1}{2} (v_{AP,in}^2 - v_{AP,out}^2)$$

Where $v_{AP,in}$ and $v_{AP,out}$ are the forward COM velocities at the start and end of the interval. For brevity, anteroposterior translational kinetic-energy reduction is referred to as **kinetic-energy reduction** throughout the remainder of the paper. Mean net AP translational braking power was calculated as:

$$P_{brake,AP} = \frac{\Delta KE_{AP}}{\Delta t}$$

where Δt is the interval duration. The outcomes were reported as positive magnitudes in $\text{J} \cdot \text{kg}^{-1}$ and $\text{W} \cdot \text{kg}^{-1}$, respectively.

Table 1. Mechanical demands across contact-initiated intervals and instrumented contacts, mean $\pm$ SD across four athletes. Window identifies the analysis period; Needs identifies the required equipment. Platform variables are reported for the three contacts captured completely in every athlete. Force peaks were taken from the 15 Hz filtered signal; one body weight is $9.81 \text{ N}\cdot\text{kg}^{-1}$. FFC contact variables, including contact time, end at V0. Phase rows are whole-body values over the braking phase (Vmax to V0) and the early and late sub-phases used by Harper et al., and span the contact columns.

Variable	Window	Needs	FC-3 (L)	APFC (R)	PFC (L)	FFC (R)
Entry velocity ($\text{m}\cdot\text{s}^{-1}$)	interval	COM	4.10 ± 0.06	3.51 ± 0.21	2.71 ± 0.30	1.87 ± 0.52
Velocity decrement ($\text{m}\cdot\text{s}^{-1}$)	interval	COM	0.60 ± 0.19	0.80 ± 0.15	0.84 ± 0.24	1.86 ± 0.53
share of total (%)	interval	COM	14.1 ± 4.4	19.0 ± 3.5	20.0 ± 5.8	44.3 ± 12.7
AP translational kinetic-energy reduction ($\text{J}\cdot\text{kg}^{-1}$)	interval	COM	2.25 ± 0.66	2.46 ± 0.37	1.85 ± 0.36	1.85 ± 1.10
share of total (%)	interval	COM	25.4 ± 7.0	27.9 ± 4.3	21.0 ± 4.3	21.1 ± 12.6
Mean deceleration ($\text{m}\cdot\text{s}^{-2}$)	interval	COM	3.26 ± 1.33	4.50 ± 0.46	3.78 ± 0.80	6.10 ± 0.97
Mean net AP translational braking power ($\text{W}\cdot\text{kg}^{-1}$)	interval	COM	12.28 ± 4.59	13.98 ± 1.84	8.46 ± 1.37	5.89 ± 2.61
Horizontal braking power, whole phase ($\text{W}\cdot\text{kg}^{-1}$)	phase	COM	9.05 ± 0.43			
early sub-phase, Vmax to 50% Vmax	phase	COM	11.49 ± 0.53			
late sub-phase, 50% Vmax to V0	phase	COM	5.68 ± 1.08			
Velocity change attributable to this limb ($\text{m}\cdot\text{s}^{-1}$)	contact	plates	—	0.81 ± 0.18	1.00 ± 0.17	1.64 ± 0.43
share of total (%)	contact	plates	—	19.3 ± 4.3	23.8 ± 4.2	39.1 ± 10.4
Mean AP braking force ($\text{N}\cdot\text{kg}^{-1}$)	contact	plates	—	4.23 ± 0.40	2.90 ± 0.64	5.39 ± 0.70
Peak AP braking force ($\text{N}\cdot\text{kg}^{-1}$)	contact	plates	—	7.92 ± 0.57	5.97 ± 1.42	7.53 ± 1.72
peak-to-mean ratio	contact	plates	—	1.89 ± 0.05	2.07 ± 0.23	1.39 ± 0.14
Mean vertical force ($\text{N}\cdot\text{kg}^{-1}$)	contact	plates	—	10.24 ± 1.34	6.05 ± 1.06	10.67 ± 0.12
Peak vertical force ($\text{N}\cdot\text{kg}^{-1}$)	contact	plates	—	20.15 ± 2.55	13.22 ± 2.58	14.64 ± 1.34
Effective braking velocity ($\text{m}\cdot\text{s}^{-1}$)	contact	plates	—	3.05 ± 0.27	2.12 ± 0.31	0.83 ± 0.25
Mean net limb AP braking power ($\text{W}\cdot\text{kg}^{-1}$)	contact	plates	—	12.94 ± 2.04	6.06 ± 1.22	4.61 ± 2.02
Contact time (s)	contact	plates	—	0.196 ± 0.054	0.348 ± 0.036	0.304 ± 0.043

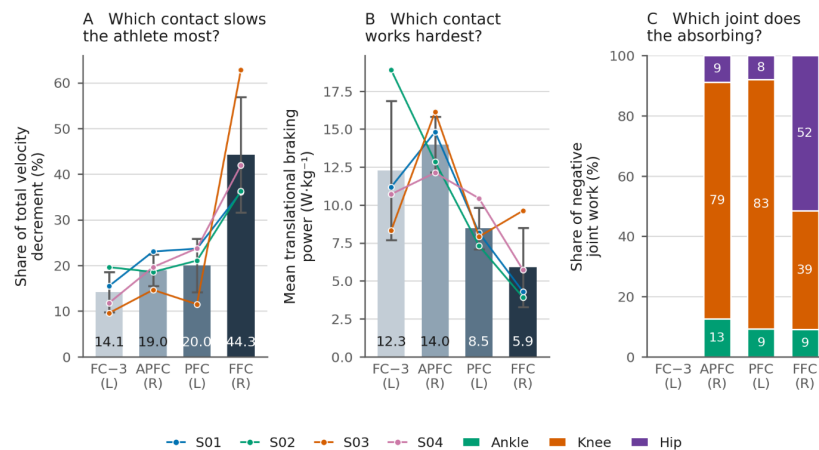


Fig. 3. The three answers. (A) Share of the total forward velocity decrement. (B) Mean net AP translational braking power — the whole-body quantity, $\Delta KE_{AP} \div \text{interval duration}$; the limb quantity and its decomposition are in Figure 4. (C) Composition of the limb's negative joint work, stacked ankle $\rightarrow$ knee $\rightarrow$ hip. Panels A and B rank the contacts differently. Individual athletes are shown in A and B; C shows group means, with the spread given in Table 2. Panels A and B are centre-of-mass quantities and cover all four intervals; panel C leaves the FC-3 slot empty because that contact was not modelled at joint level.

Both measures were calculated over non-overlapping foot-strike-to-next-foot-strike intervals, with the final interval ending at V0, to avoid double-counting COM energy changes during double support. This approach adapts the whole-body energy framework of Zamparo et al. [6,7] to the anteroposterior translational component and successive contact-initiated intervals.

Mean net anteroposterior translational braking power corresponds to the horizontal braking power (HBP) reported by Harper et al. [2], calculated as horizontal braking force (mass $\times$ deceleration) $\times$ forward velocity. Because this product is

the instantaneous rate of change of anteroposterior translational kinetic energy, its mean over any window equals the kinetic-energy reduction divided by the window duration. To allow comparison with that work, HBP was also calculated over the whole braking phase (Vmax to V0) and over the early (Vmax to 50% of Vmax) and late (50% of Vmax to V0) sub-phases used by Harper et al. [2], with the 50% of Vmax instant identified from the COM velocity trace. Air resistance, which Harper et al. included, was not modelled; it was estimated at approximately 1% of braking power in this protocol.

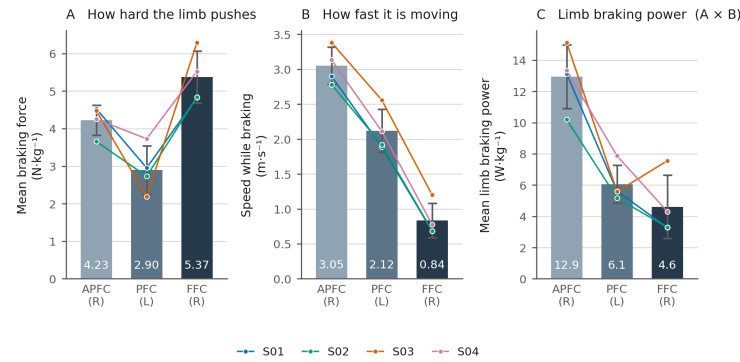


Fig. 4. Why mean AP braking force and mean net limb AP braking power rank the contacts differently. All panels are platform-derived and limb-specific. (A) Mean AP braking force. (B) Effective braking velocity, calculated as the force-weighted mean forward COM velocity over the contact. (C) Their product, mean net limb AP braking power. Bars show means with SD whiskers; coloured lines show individual athletes.

Table 2. Joint-level demands at each contact: negative joint work over the contact and peak moments over the absorption phase, mean $\pm$ SD across four athletes. FC-3 was not modelled at joint level.

Variable	APFC (R)	PFC (L)	FFC (R)	FFC – PFC (95% CI)
Summed negative joint work (J·kg ⁻¹)	2.24 $\pm$ 0.27	2.08 $\pm$ 0.46	1.73 $\pm$ 0.53	–0.35 (–1.13 to 0.44)
knee (J·kg ⁻¹)	1.75 $\pm$ 0.26	1.74 $\pm$ 0.51	0.69 $\pm$ 0.27	–1.05 (–1.80 to –0.29)
hip (J·kg ⁻¹)	0.21 $\pm$ 0.21	0.15 $\pm$ 0.09	0.89 $\pm$ 0.32	0.74 (0.17 to 1.31)
ankle (J·kg ⁻¹)	0.28 $\pm$ 0.22	0.19 $\pm$ 0.07	0.15 $\pm$ 0.04	–0.04 (–0.13 to 0.06)
Knee share of negative joint work (%)	78.7 $\pm$ 15.7	82.9 $\pm$ 8.4	39.4 $\pm$ 8.3	–43.6 (–58.4 to –28.7)
Hip share of negative joint work (%)	8.7 $\pm$ 7.9	7.9 $\pm$ 5.0	51.5 $\pm$ 7.2	43.7 (29.4 to 58.0)
Ankle share of negative joint work (%)	12.5 $\pm$ 8.3	9.2 $\pm$ 3.7	9.1 $\pm$ 4.0	–0.1 (–2.9 to 2.7)
Peak knee moment (N·m·kg ⁻¹)	2.67 $\pm$ 0.40	2.68 $\pm$ 0.23	1.61 $\pm$ 0.22	–1.08 (–1.67 to –0.48)
Peak hip moment (N·m·kg ⁻¹)	2.18 $\pm$ 0.70	2.01 $\pm$ 0.43	3.11 $\pm$ 0.14	1.09 (0.51 to 1.68)
Peak ankle moment (N·m·kg ⁻¹)	1.00 $\pm$ 0.24	0.72 $\pm$ 0.20	0.97 $\pm$ 0.12	0.25 (–0.16 to 0.66)

Force-platform variables. For each instrumented contact, we calculated mass-normalised net AP braking impulse (reported as the velocity change attributable to the limb), mean and peak AP braking force, mean net limb AP braking power, effective braking velocity (the force-weighted mean forward COM velocity over the contact), mean and peak vertical ground-reaction force, the peak-to-mean AP braking-force ratio, and contact time. Force peaks were obtained from the 15 Hz filtered signal.

Joint-level variables. Sagittal-plane ankle, knee and hip angles and net internal moments were exported as stance-normalised curves. Negative joint work was calculated over the contact as the integral of the negative portion of each joint's sagittal-plane power, summed across joints, with each joint expressed as a percentage of the limb total. Peak joint moments were extracted over the absorption phase, from foot strike until summed sagittal-plane joint power became positive and remained so for at least 10% of stance, or to foot-off if it did not. FC-3 was not modelled at joint level.

Analysis. The three valid trials were averaged within each athlete, and the athlete was treated as the unit of analysis ($n = 4$). Outcomes are presented as mean $\pm$ SD across athletes. Contact contrasts were calculated from within-athlete differences and are reported as the mean difference with 95% confidence intervals. Given the exploratory design and very small sample, no null-hypothesis significance tests were per-

formed, and the confidence intervals should be interpreted descriptively rather than as confirmatory evidence.

Results

Which contact slows the athlete down most?

Athletes reached 4.20 ± 0.05 m·s⁻¹ and stopped in 0.98 ± 0.07 s over 2.36 ± 0.11 m, with a mean deceleration of 4.31 ± 0.24 m·s⁻².

Braking was distributed unevenly across the four intervals (Table 1; Figure 3A). The final interval accounted for the largest share of the total velocity decrement ($44.3 \pm 12.7\%$) and showed the greatest mean deceleration (6.10 ± 0.97 m·s⁻²). It produced the largest velocity decrement in every athlete and all 12 trials. The ante-penultimate interval had greater mean deceleration than the penultimate interval despite contributing slightly less to the total velocity decrement, indicating that the rankings differed between these measures.

Because the final interval ends at zero forward velocity, its velocity decrement is determined by the velocity remaining at FFC onset and therefore depends on braking completed during the preceding intervals; it should not be interpreted as a direct measure of mechanical intensity. Despite similar Vmax between athletes (4.14 – 4.26 m·s⁻¹), entry velocity at FFC varied from 1.53 to 2.64 m·s⁻¹, indicating substantial variation in how braking was distributed before the final contact.

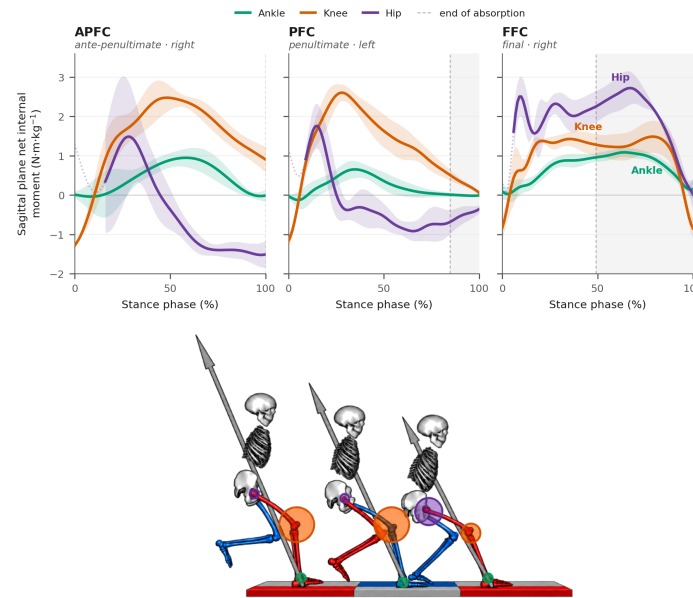


Fig. 5. Sagittal-plane net internal ankle, knee and hip joint moments across each of the three modelled contacts, presented as mean $\pm$ SD across the four athletes. The dashed vertical line and shaded grey region indicate the mean end of the absorption phase; at the APFC, absorption continued to, or close to, foot-off, so no region is shaded. The hip moment is shown dotted during the first 30 ms after foot strike, where it reflects the impact transient rather than the stance-phase moment. Representative sagittal-plane limb postures for each contact are shown below, together with the corresponding ground-reaction-force vector in grey. The coloured bubbles show the relative contribution of each joint to total negative joint work, with larger bubbles indicating a greater contribution.

Where is braking power greatest?

The pattern changed when braking was expressed in energetic rather than kinematic terms. Despite the final interval producing the largest velocity decrement, the two earliest intervals accounted for the greatest reductions in kinetic energy ($25.4 \pm 7.0\%$ and $27.9 \pm 4.3\%$; Table 1). The distinction was greater for braking power. Mean net AP translational braking power was highest in the two earliest intervals and lowest in the final interval (Figure 3B), with the mean of the two earliest values approximately 2.2 times the final-interval value. Consistent with this, horizontal braking power over the whole braking phase was $9.05 \pm 0.43 \text{ W}\cdot\text{kg}^{-1}$ and approximately twofold greater in the early than in the late sub-phase (11.49 ± 0.53 vs $5.68 \pm 1.08 \text{ W}\cdot\text{kg}^{-1}$); 50% of V_{max} was reached at $59 \pm 5\%$ of the braking-phase duration. Limb-level force-platform measures showed a similar pattern, with the final contact producing the greatest mean AP braking force but the lowest mean net limb AP braking power because it occurred at the lowest effective braking velocity. (Figure 4).

How is negative joint work distributed across contacts?

Negative joint work distribution shifted markedly across the three modelled contacts (Table 2; Figure 3C). The knee contributed approximately 79% and 83% at the ante-penultimate and penultimate contacts, respectively, whereas at the final contact the hip became the largest contributor (52%), with the knee contribution falling to 39%. From the penultimate to the final contact, knee negative work decreased by $1.05 \text{ J}\cdot\text{kg}^{-1}$ (95% CI 0.29 to 1.80), whereas hip negative work increased by $0.74 \text{ J}\cdot\text{kg}^{-1}$ (95% CI 0.17 to 1.31).

Joint moments were consistent with this proximal shift (Figure 5), as peak hip moment increased by $1.09 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ (95%

CI 0.51 to 1.68) from the penultimate to the final contact, whereas peak knee moment decreased by $1.08 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ (95% CI 0.48 to 1.67). The knee-to-hip shift in negative joint work occurred in every athlete, although the hip was the larger contributor at the final contact in only three of four. The exception was the athlete who entered FFC with the greatest remaining forward velocity.

Discussion

This pilot study shows that the mechanical account used determines which part of a maximal 5-m deceleration appears most demanding. Profiling contact by contact, the same stop yielded two main findings: 1) the final interval removed the most forward velocity, whereas the two earliest intervals removed the most kinetic energy at the highest braking power; and 2) negative joint work shifted from the knee towards the hip at the final contact.

Velocity and energy rank the contacts differently

Because kinetic energy depends on the square of velocity, the same reduction in velocity removes more energy at higher speed. The large final-interval velocity decrement (44% of the total) therefore reflects the velocity remaining at FFC entry, which ranged from 1.5 to $2.6 \text{ m}\cdot\text{s}^{-1}$ despite similar V_{max} , rather than greater mechanical intensity. The force-platform data showed the same distinction at limb level: the FFC produced the greatest mean braking force but, applied at the lowest velocity, the lowest braking power. This is consistent with the greater early-phase horizontal braking power reported by Harper et al. [2] and with the energy-based approach of Zamparo et al. [6,7], and locates that early demand within the FC-3- and APFC-initiated intervals.

Joint demand shifts from knee to hip

The knee contributed ~80% of negative joint work at the APFC and PFC, whereas the hip became the largest contributor (52%) at the FFC, with corresponding changes in peak joint moments. This adds a contact-specific view to the large knee and hip demands reported in musculoskeletal modelling of deceleration [8]. However, because the PFC and FFC were performed with different limbs, the shift cannot be attributed to contact order alone.

Implications for assessment

As in hop testing after ACL reconstruction, where symmetrical performance can mask reduced knee work [9,10], completing the stop does not reveal how energy absorption is distributed across the joints; whether contact-level profiling detects such compensations after injury remains to be tested. COM-derived interval measures require only whole-body kinematics and may transfer to wearable or markerless systems [11,12], although their validity in that setting is untested; limb- and joint-level outcomes still require force platforms and inverse dynamics.

Conclusion

Contact-by-contact analysis showed that velocity, energy and joint-level measures give distinct accounts of the same stop, with the final interval removing the most velocity, the earlier intervals removing the most kinetic energy at the highest braking power, and negative joint work shifting from the knee towards the hip at the final contact. Neither the stopping outcome nor the final contact alone therefore describes how braking demand is distributed across the deceleration. Given four athletes and a fixed limb sequence, these findings are preliminary and hypothesis-generating.

Practical applications

- **Match the metric to the question.** Velocity measures describe how much of the stop is completed; kinetic-energy reduction and braking power describe energetic demand, which is greatest in the earlier contacts.
- **Look beyond the final contact.** Earlier contacts carried the highest braking power and knee-dominant absorption; assessment or training limited to the final stop may under-represent them.
- **Choose equipment for the level of question.** COM-derived interval measures may support field-based assessment; limb braking and joint-work distribution require force platforms and inverse dynamics.

Limitations

Four healthy recreational men performed one task at one approach speed with a prescribed final-contact limb; the findings are descriptive, not normative. Because the PFC was performed with the left limb and the APFC and FFC with the right, contact-order effects cannot be separated from inter-limb differences; future studies should counterbalance the final-contact limb. Energy was quantified in the AP direction only; including the gravitational potential energy released as the COM lowered (0.19 ± 0.04 m) would increase the energy removed by ~20%, mostly in the final two intervals, without changing the ranking of braking power. Arm motion was not tracked, so COM estimates excluded arm motion.

AI Use Statement

Artificial intelligence tools (ChatGPT and Claude) were used to assist with language editing and refinement, as well as the development of Python code for specific aspects of data anal-

ysis. All AI-assisted code was reviewed and executed by the authors, who independently verified the analytical outputs. AI tools were also used to enhance the visual presentation and clarity of Figures 1 and 5 without altering their scientific content. All AI-assisted content and figures were critically reviewed and verified by the authors. The authors retained full responsibility for the scientific content, methodological decisions, execution and accuracy of the analyses, and interpretation of the findings. No AI tools were used to generate, fabricate, or manipulate the underlying research data.

References

1. T. Dos'Santos, C. Thomas, and P. A. Jones, "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," *J. Sports Sci.*, vol. 39, no. 4, pp. 395–405, Feb. 2021, doi: 10.1080/02640414.2020.1823130.
2. D. J. Harper *et al.*, "Biomechanical and Neuromuscular Performance Requirements of Horizontal Deceleration: A Review with Implications for Random Intermittent Multi-Directional Sports," *Sports Med.*, vol. 52, no. 10, pp. 2321–2354, Oct. 2022, doi: 10.1007/s40279-022-01693-0.
3. R. B. Davis, S. Öunpuu, D. Tyburski, and J. R. Gage, "A gait analysis data collection and reduction technique," *Hum. Mov. Sci.*, vol. 10, no. 5, pp. 575–587, Oct. 1991, doi: 10.1016/0167-9457(91)90046-Z.
4. F. Leboeuf, R. Baker, A. Barré, J. Reay, R. Jones, and M. Sangeux, "The conventional gait model, an open-source implementation that reproduces the past but prepares for the future," *Gait Posture*, vol. 69, pp. 235–241, Mar. 2019, doi: 10.1016/j.gaitpost.2019.04.015.
5. E. Kristianslund, T. Krosshaug, and A. J. van den Bogert, "Artefacts in measuring joint moments may lead to incorrect clinical conclusions: the nexus between science (biomechanics) and sports injury prevention!," *Br. J. Sports Med.*, vol. 47, no. 8, pp. 470–473, May 2013, doi: 10.1136/bjsports-2012-091199.
6. P. Zamparo *et al.*, "Mechanical work in shuttle running as a function of speed and distance: Implications for power and efficiency," *Hum. Mov. Sci.*, vol. 66, pp. 487–496, Aug. 2019, doi: 10.1016/j.humov.2019.06.005.
7. P. Zamparo, G. Pavei, F. Nardello, D. Bartolini, A. Monte, and A. E. Minetti, "Mechanical work and efficiency of 5 + 5 m shuttle running," *Eur. J. Appl. Physiol.*, vol. 116, no. 10, pp. 1911–1919, Oct. 2016, doi: 10.1007/s00421-016-3443-6.
8. J. Verheul, D. Harper, and M. A. Robinson, "Forces experienced at different levels of the musculoskeletal system during horizontal decelerations," *J. Sports Sci.*, vol. 42, no. 23, pp. 2242–2253, Dec. 2024, doi: 10.1080/02640414.2024.2428086.
9. A. Kotsifaki *et al.*, "Single leg hop for distance symmetry masks lower limb biomechanics: time to discuss hop distance as decision criterion for return to sport after ACL reconstruction?," *Br. J. Sports Med.*, p. bjsports-2020-103677, 2021, doi: 10.1136/bjsports-2020-103677.
10. A. Kotsifaki *et al.*, "Symmetry in Triple Hop Distance Hides Asymmetries in Knee Function After ACL Re-

construction in Athletes at Return to Sports,” *Am J Sports Med*, vol. 50, no. 2, pp. 441–450, Feb. 2022, doi: 10.1177/03635465211063192.

11. J. B. Marques, V. Sideris, R. Whiteley, P. J. Read, M. M. Gomes, and P. R. P. Santiago, “Wearable technology identifies differences in change of direction kinetics and kinematics in soccer players with a history of anterior cruciate ligament reconstruction,” *Knee Surg. Sports Traumatol. Arthrosc.*, vol. 34, pp. 402–414, 2026, doi: 10.1002/ksa.12679.

12. J. A. Turner, C. R. Chaaban, and D. A. Padua, “Validation of OpenCap: A low-cost markerless motion capture

system for lower-extremity kinematics during return-to-sport tasks,” *J. Biomech.*, vol. 171, p. 112200, Jun. 2024, doi: 10.1016/j.jbiomech.2024.112200.

Copyright: The article published on Sport Performance and Science Reports are distributed under the terms of the Creative Commons Attribution 4.0 International License (<http://creativecommons.org/licenses/by/4.0/>), which permits unrestricted use, distribution and reproduction in any medium, provided you give appropriate credit to the original author(s) and the source, provide a link to the Creative Commons license, and indicate if changes were made. The Creative Commons Public Domain Dedication waiver (<http://creativecommons.org/publicdomain/zero/1.0/>) applies to the data made available in this article, unless otherwise stated.

