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A biomechanical study of the relationship between running velocity and three-dimensional lumbosacral kinetics

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ABSTRACT

Faster running is not performed with proportional increase in all joint torque/work exertions. Although previous studies have investigated lumbopelvic kinetics for a single velocity, it is unclear whether each lumbopelvic torque should increase for faster running. We examined the relationship between running velocity and lumbopelvic kinetics. We calculated the three-dimensional lumbosacral kinetics of 10 male sprinters during steady-state running on a temporary indoor running track at five target velocities: 3.0 (3.20 ± 0.16), 4.5 (4.38 ± 0.18), 6.0 (5.69 ± 0.47), 7.5 (7.30 ± 0.41), and maximal sprinting (9.27 ± 0.36 m/s). The lumbosacral axial rotation torque increased more markedly (from 0.37 ± 0.06 to 1.99 ± 0.46 Nm/kg) than the extension and lateral flexion torques. The increase in the axial rotation torque was larger above 7.30 m/s. Conversely, the extension and lateral flexion torques plateaued when running velocity increased above 7.30 m/s. Similar results were observed for mechanical work. The results indicate that faster running required larger lumbosacral axial rotation torque. Conversely, the extension and lateral flexion torques were relatively invariant to running velocity above 7 m/s, implying that faster running below 7 m/s might increase the biomechanical loads causing excessive pelvic posterior tilt and excessive pelvic drop which has the potential to cause pain/injury related to lumbopelvic extensors and lateral flexors, whereas these biomechanical loads might not relate with running velocity above 7 m/s.

1. Introduction

Runners change their running velocities with changing kinetic behaviours. Gracovetsky et al. suggested that the movement of the lumbopelvic region may provide the primary drive for locomotive leg movements (Gracovetsky, 1985; Gracovetsky and Iacono, 1987). Recently, it has been revealed that the kinetic behaviour around the lumbopelvic region, which includes anatomically large muscles, has a critical role in running efficiently, along with trunk posture maintenance (Sado et al., 2016; Saunders et al., 2005). Thus, the understanding of the lumbopelvic kinetic demand in running with wide range of velocities would provide practical implications for performance improvement and injury prevention.

Previous studies (Dorn et al., 2012; Schache et al., 2015, Schache et al., 2011) revealed that increases in steady-state running velocities are not performed by proportional increases in all joint torque/work exertions. Schache et al. (2011) found that the hip flexion torque in initial swing and hip extension torque in terminal swing increase with velocity increments. In contrast, ankle plantar flexion torque and ankle joint work during the stance phase increase when running velocity increases in slower velocity range (<7 m/s), but they are invariant in the faster velocity range (>7 m/s). Dorn et al. (2012) analysed the muscle force/work itself and reported similar results. Schache et al. (2015) investigated the lower-limb joint power/work modulation with increasing running velocity and reported that the hip relative contribution to mean power and positive work significantly increased whereas the knee and ankle relative contributions did not significantly differ with running velocity increments. Thus, some torques around the lumbopelvic region may be also invariant to velocity increments, especially in the faster velocity range.

A three-dimensional (3D) analysis of the lower limbs (Schache et al., 2011) revealed that the sagittal torques markedly increase with running velocity increments while the changes in other planes are relatively invariant; however, it may not necessarily be true for the lumbopelvic region. Previous studies suggested that pelvic rotation, in the transverse plane, assists in a recovery motion from toe-off for the next step (Chapman and Caldwell, 1983; Sado

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et al., 2017a). The main cause of increases in lower limb energy after toe-off is the forward pulling force of the pelvis (i.e. forward hip joint force on thigh) (Chapman and Caldwell, 1983; Vardaxis and Hoshizaki, 1989) that increases simultaneously with transverse pelvic rotation towards the free leg side (moving stance leg hip forward) during the terminal stance phase (Sado et al., 2017a). In maximal sprinting, lumbosacral axial rotation torque mainly controls the pelvic rotation towards the free leg side (Sado et al., 2017a). Thus, changes in 3D lumbopelvic kinetics with velocity increments might differ from the lower limbs. However, previous studies (Sado et al., 2016; Seay et al., 2008) have obtained data in lumbopelvic kinetics for a single velocity. Although previous studies (Cappellini, 2006; Saunders et al., 2005) showed the increase in electromyographic (EMG) activities of some lumbopelvic muscles (such as the erector spinae and internal/external oblique) with running velocity increments, these studies are limited to the slow velocity range (<5 m/s). Thus, it remains unclear whether each lumbopelvic torque is required to increase for faster running, especially in the faster velocity range.

The information regarding the increase in lumbopelvic kinetics with faster running would have implications for athletic training strategy in sprinters. Therefore, we aimed to clarify the 3D lumbopelvic kinetics at different running velocities, with *a priori* hypothesis that the lumbopelvic transverse torque particularly increases with running velocity increments and that other lumbopelvic torques do not increase in the faster velocity range.

2. Methods

2.1. Participants

Participants were 10 male sprinters [22.6 ± 1.2 years; 1.74 ± 0.05 m; 64.2 ± 4.9 kg; 100 m personal best time, 10.43–11.17 s (10.87 ± 0.24 s)]. All participants were members of the college track and field team and had participated in national-level athletic competitions in high school and/or college. They had performed regular sprint training 4–5 days per week for > 5 years. The purpose and experimental protocol were explained to the participants. All participants provided written informed consent. The Human Research Ethics Committee at The University of Tokyo, Japan, approved the study protocol (reference number: 356).

2.2. Procedure

The experiment was done in a temporary indoor running track setup in the biomechanics laboratory. All participants wore close-fitting clothing and their own running shoes with spikes. The 47 markers (Sado et al., 2017a), 20 mm in diameter, were secured to the trunk and limbs. After a self-directed warm-up, including jogging, dynamic stretching drills, and running at various velocities, each participant ran at approximately 3.0, 4.5 6.0, and 7.5 m/s and maximal sprinting. For practical reasons, the order of running velocities was incremental rather than randomised. Participants were instructed to maintain a steady-state velocity when running through the capture volume. To run at the specified steady-state velocity, each participant was allowed an acceleration distance of approximately 40 m. The passing time was recorded using photocell censors (E3JM-R4M4T, Omron, Kyoto, Japan) located at 2.5 m ahead and behind the force platform. Participants repeated the running trial until they completed two trials in which either foot contacted a force platform located 40 m from the start point without protruding off the edge, and the difference in passing time was less by 0.1 s from the required time for each target velocity (i.e. 3.0 m/s: 1.67 s; 4.5 m/s: 1.11 s; 6.0 m/s: 0.83 s; and 7.5 m/s: 0.67 s). As the force platform was difficult to see, foot contact was achieved by chance. The total number of trials for maximal sprinting condition (the sum of successful and unsuccessful trials) performed by the participant who performed most trials was five, until two successful trials were completed. Although the stance legs of two successful trials on each target velocity were the same, stance legs between some velocity conditions were different. Adequate recovery time (>3 min) was provided between trials to avoid fatigue. Kinetic analysis was performed in both trials for each velocity condition and each participant.

2.3. Data collection

A 13-camera motion capture system (Mac 3D, Motion Analysis Corporation, Santa Rosa, CA, USA) recorded the 3D coordinates of the position of the reflective markers (sampling rate, 200 Hz). Ground reaction force (GRF) was recorded using a force platform (Force Plate 9281E, Kistler, Winterthur, Switzerland) at a sampling rate of 1000 Hz; it was synchronised with the motion data. The $\boldsymbol{x}$, $\boldsymbol{y}$, and $\boldsymbol{z}$ axes of the global coordinate system defined the medial–lateral, anterior–posterior, and superior–inferior directions, respectively.

2.4. Data analysis

Data analyses were performed using MATLAB 2014a (MathWorks Inc., Natick, MA, USA). Position coordinates of the markers were smoothed using a Butterworth low-pass digital filter with a cutoff frequency of 15 Hz based on residual analysis (Winter, 2009). GRF data were smoothed using the same filter with that of the marker position data to prevent artefacts after contact (Bezodis et al., 2013; Bisseling and Hof, 2006).

The leg stepping on the force platform and the other leg were defined as 'stance leg' and 'free leg', respectively. We analysed a step cycle defined as the period from stance leg contact to free leg contact (assuming bilateral symmetry). The step cycle was then divided into the stance (from the stance leg contact on the force platform to the stance leg toe-off) and flight phases (from the stance leg toe-off to the free leg contact). The instances of stance leg contact and toe-off were identified from the onset of the vertical GRF signal. The threshold was 5 N. The free leg contact was identified with kinematic methods using vertical acceleration of the free leg toe (Nagahara and Zushi, 2013).

The whole-body model and definition of each joint centre position was consistent with those in a previous study (Sado et al., 2017a). In this model, the trunk was divided into two segments (thoracolumbar and pelvic segments) by the lumbosacral joint (Dumas et al., 2007). Right-handed local coordinate systems were defined for the thoracolumbar, pelvic, bilateral thighs, shanks, and feet segments (segment coordinate system). The joint coordinate system (JCS) was fixed at the lumbosacral joint (Grood and Suntay, 1983).

The centre of mass (CoM) of the whole body and each segment inertia parameter were calculated based on the study by Dumas et al. (2007). The whole-body velocity vector was calculated as the time derivative of the CoM position vector. Running velocity was defined as the average of the $\boldsymbol{y}$ axis component of the CoM velocity vector during the analysis phase. The step length was calculated as the anterior distance between CoM at the stance leg contact and that at the free leg contact. The step frequency was calculated as the reciprocal of the period from the stance leg contact to the free leg contact.

Newton–Euler equations were used to calculate the 3D internal joint torque at the lumbosacral joints (Winter, 2009), with joint torques transformed into the JCS (Desroches et al., 2010). To evaluate the inter-trial variability, we calculated the differences in peak torques between two separate maximal sprinting trials in each participant. To assess the effects of inter-condition changes in the stance leg on the trends of peak torques with running velocity increments, we examined whether the inter-condition changes induced the changes in peak torque excessively different from that without changes in the stance leg.

Joint power was calculated as the dot product of the joint torque and joint angular velocity (Sado et al., 2017b). The positive and negative work exerted at the lumbosacral joint at distinct phases throughout the step cycle was calculated by integrating the relevant portion of the power-versus-time curve.

2.5. Statistical analysis

The mean of the data from the two running trials was used as the representative value of each velocity condition of each participant. Each kinetic variable was compared for the five running velocity conditions by one-way (running velocity) analysis of variance with repeated measures. Partial Eta^2 (η^2) was used to measure the effect sizes. Values of 0.04, 0.25, and >0.64 were considered small, medium, and large, respectively (Ferguson, 2009). If the main effect was significant, pairwise comparisons were made for all pairs in the five running velocity conditions using a paired *t*-test. To control the family-wise error rate in each multiple comparison, the alpha level of each *t*-test was adjusted using Holm's method (Holm, 1979). Overall statistical significance was set at $\alpha < 0.05$. Statistical analyses were performed using SPSS 23 for Windows (SPSS, Inc., Chicago, IL, USA).

3. Results

The running velocities were 3.20 ± 0.16, 4.38 ± 0.18, 5.69 ± 0.47, 7.30 ± 0.41, and 9.27 ± 0.36 m/s. Table 1 shows the mean ± SD magnitudes of various variables, as well as the results of the statistical tests. From 3.20 ± 0.16 to 4.38 ± 0.18 m/s, step length increased greater (26%) than frequency (9%), whereas from 7.30 ± 0.41 to 9.27 ± 0.36 m/s, step frequency increased greater (22%) than length (4%) (Table 1).

Fig. 1 shows the ensemble average of the lumbosacral torques. In all velocity conditions, the lumbosacral extension and lateral flexion towards the free leg side torques were exerted during the stance phase, and axial rotation towards the stance leg side torque was developed from the terminal stance phase to the flight phase (Fig. 1).

Fig. 2 shows the ensemble average of the lumbosacral powers. Lumbosacral extension and lateral flexion torques exerted positive power during the terminal stance phase, and axial rotation torque exerted negative power during the terminal stance and initial flight phases and positive power during the terminal flight phase (Fig. 2).

The running velocity condition had significant main effects on the peak lumbosacral torques and mechanical works ($p < 0.001$, Table 1). For the peak lumbosacral extension and lateral flexion torque, there was no significant difference between 7.30 m/s (2.89 ± 0.64 Nm/kg for extensors and 1.88 ± 0.64 Nm/kg for lateral flexors) and 9.27 m/s (2.58 ± 0.60 Nm/kg for extensors and 2.01 ± 0.71 Nm/kg for lateral flexors) (Table 1). The peak lumbosacral axial rotation torque significantly increased for all running velocity increments (from 0.37 ± 0.06 in 3.20 ± 0.16 m/s to 1.99 ± 0.46 Nm/kg in 9.27 ± 0.36 m/s) (Table 1) and had a peak value range for maximal sprinting of 1.27–2.74 Nm/kg. In mechanical work, only the lumbosacral axial rotation negative work during the initial flight phase (from −0.01 ± 0.00 J/kg in 3.20 ± 0.16 m/s to −0.15 ± 0.10 J/kg in 9.27 ± 0.36 m/s) and positive work during the terminal flight phase (from 0.01 ± 0.01 J/kg in 3.20 ± 0.16 m/s to 0.19 ± 0.06 J/kg in 9.27 ± 0.36 m/s) were significantly different for all running velocity increments (Table 1).

Table 1
Mean ± standard deviation magnitudes of the step length, step frequency, lumbosacral peak torques and works.

Variables			Velocity 1		Velocity 2		Velocity 3		Velocity 4		Velocity 5		Main Effect	Effect size
			3.20 ± 0.16	m/s	4.38 ± 0.18	m/s	5.69 ± 0.47	m/s	7.30 ± 0.41	m/s	9.27 ± 0.36	m/s	F (4,36)	(partial η^2)
Step	Length	(m)	1.24 ± 0.08	2,3,4,5	1.56 ± 0.08	1,3,4,5	1.87 ± 0.08	1,2,4,5	2.07 ± 0.13	1,2,3,5	2.16 ± 0.15	1,2,3,4	203.88*	0.96
	Frequency	(Hz)	2.54 ± 0.11	2,3,4,5	2.78 ± 0.14	1,3,4,5	2.99 ± 0.21	1,2,4,5	3.48 ± 0.29	1,2,3,5	4.22 ± 0.30	1,2,3,4	134.37*	0.94
Extension	T_{peak}	(Nm/kg)	1.48 ± 0.31	3,4,5	2.04 ± 0.50		2.66 ± 0.68	1	2.89 ± 0.64	1	2.58 ± 0.60	1	12.67 *	0.59
	$pW_{terminal\text{-}stance}$	(J/kg)	0.08 ± 0.03	4,5	0.10 ± 0.04	3,4,5	0.14 ± 0.05	2	0.15 ± 0.06	1,2	0.15 ± 0.07	1,2	13.31 *	0.60
Lateral-Flexion	T_{peak}	(Nm/kg)	1.10 ± 0.26	4,5	1.36 ± 0.35		1.44 ± 0.48		1.88 ± 0.64	1	2.01 ± 0.71	1	7.84 *	0.47
	$pW_{terminal\text{-}stance}$	(J/kg)	0.08 ± 0.05	5	0.10 ± 0.03	5	0.12 ± 0.06		0.18 ± s	1	0.22 ± 0.10	1	11.03 *	0.55
Axial Rotation	T_{peak}	(Nm/kg)	0.37 ± 0.06	2,3,4,5	0.60 ± 0.13	1,3,4,5	0.86 ± 0.19	1,2,4,5	1.27 ± 0.27	1,2,3,5	1.99 ± 0.46	1,2,3,4	85.31 *	0.91
	$nW_{terminal\text{-}stance}$	(J/kg)	−0.04 ± 0.02	4,5	−0.05 ± 0.02	5	−0.04 ± 0.02	4,5	−0.07 ± 0.03	1,3,5	−0.12 ± 0.04	1,2,3,4	26.45 *	0.75
	$nW_{initial\text{-}flight}$	(J/kg)	−0.01 ± 0.00	2,3,4,5	−0.02 ± 0.01	1,3,4,5	−0.04 ± 0.02	1,2,4,5	−0.07 ± 0.03	1,2,3,5	−0.15 ± 0.10	1,2,3,4	19.41 *	0.68
	$pW_{terminal\text{-}flight}$	(J/kg)	0.01 ± 0.01	2,3,4,5	0.03 ± 0.01	1,3,4,5	0.06 ± 0.03	1,2,4,5	0.12 ± 0.04	1,2,3,5	0.19 ± 0.06	1,2,3,4	57.31 *	0.86

1, 2, 3, 4, or 5 : Significantly different from running speed 1, 2, 3, 4, or 5 ($p < 0.05$).
T: Torque, *pW* or *nW*: positive or negative Work.
* The main effect was significant ($p < 0.001$).

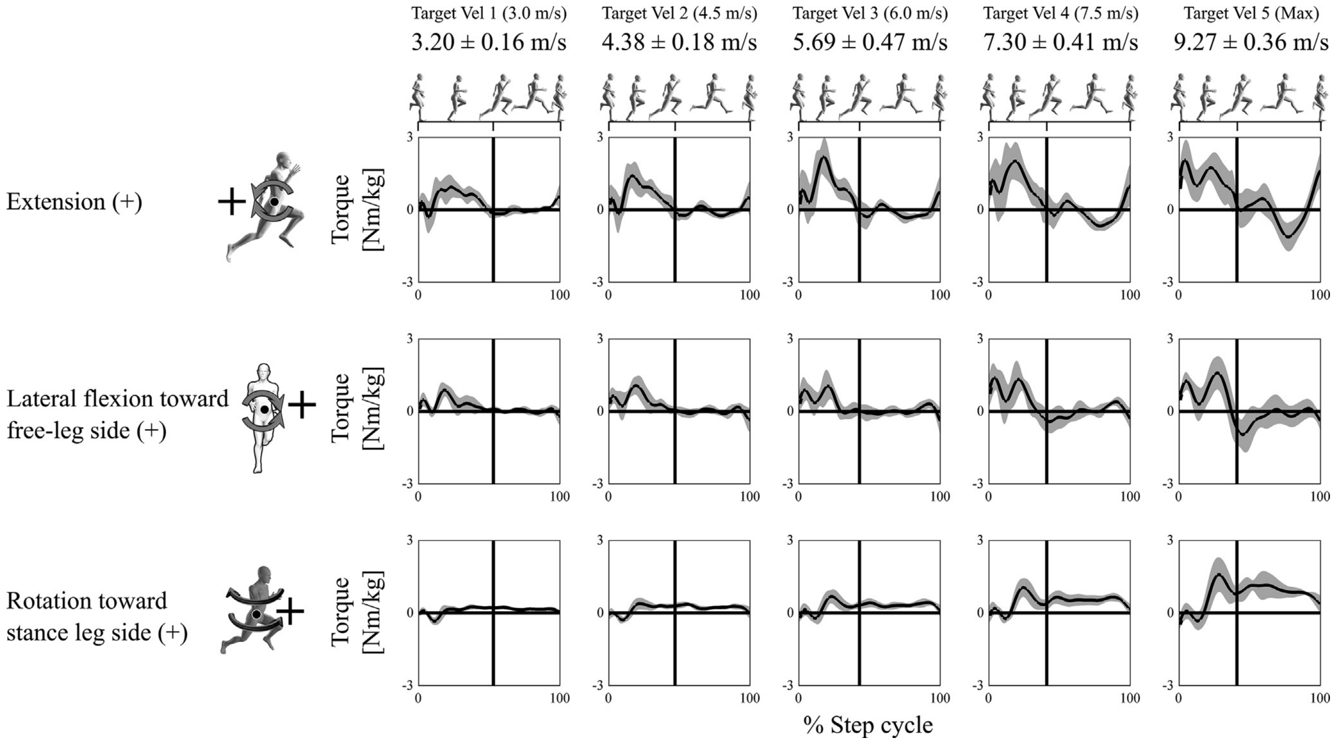

Fig. 1. Ensemble averages of the lumbosacral joint torques.

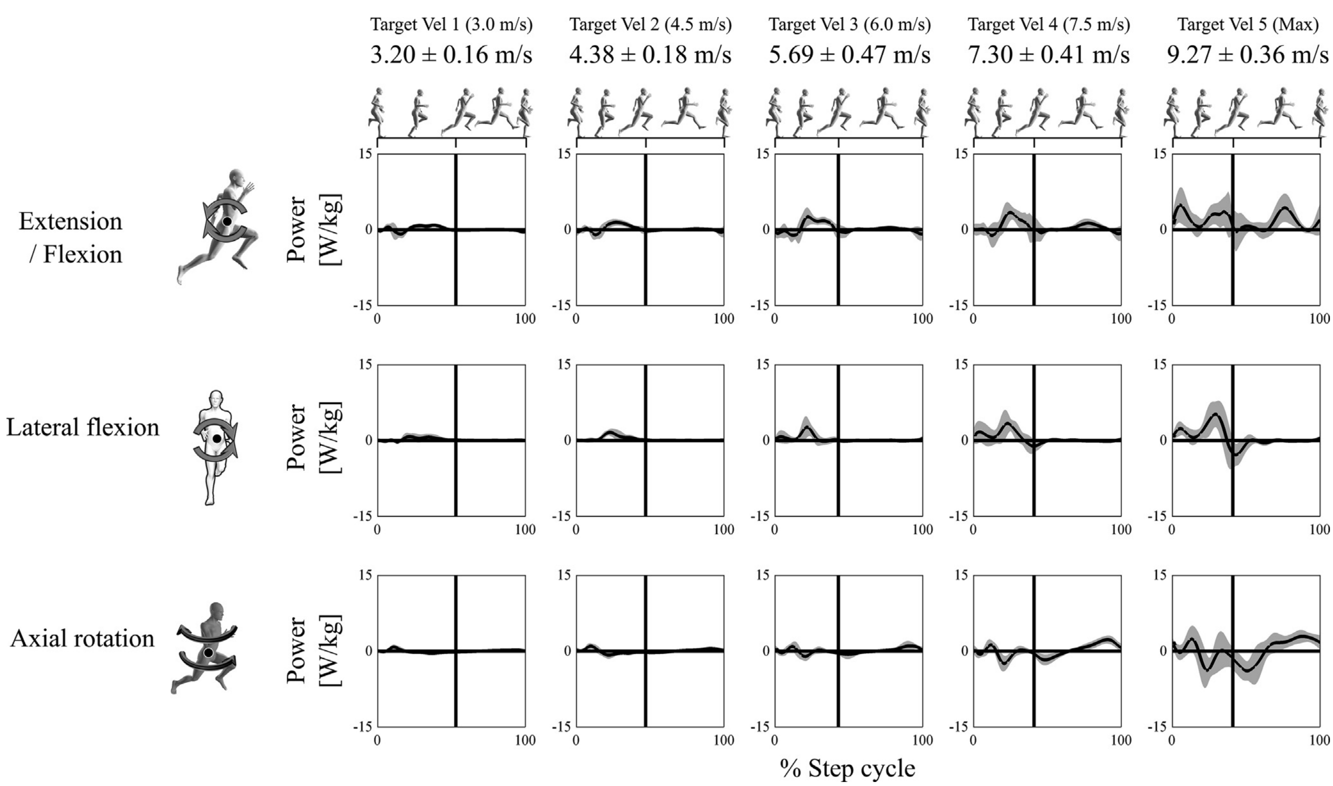

Fig. 2. Ensemble averages of the lumbosacral joint powers.

The means ± SDs and medians of the differences in peak torques between separate maximal sprinting trials were 0.51 ± 0.39 Nm/kg and 0.46 Nm/kg for the extension torque, 0.30 ± 0.33 Nm/kg and 0.18 Nm/kg for the lateral flexion torque, and 0.26 ± 0.27 Nm/kg and 0.17 Nm/kg for the axial rotation torque, respectively. Although some participants showed large differences (>1 Nm/kg) in extension (one participant) and lateral flexion (one participant) torques, the differences shown by other participants were small and no participant showed large difference in axial rotation torque.

The inter-condition changes of the stance leg did not induce the changes in peak torques with running velocity increments excessively different from that without the inter-condition change (Appendix).

4. Discussion

We examined lumbosacral torques at different running velocities. The main findings were as follows: the lumbosacral axial rotation torque increased with running velocity increments and the lumbosacral extension and lateral flexion torques were invariant to running velocity in the high-velocity range. The findings regarding the variance of axial rotators and invariances of extensors and lateral flexors are novel in this study.

We have shown that faster running required larger axial rotation torque around the lumbopelvic region. Lumbosacral axial rotation torque mainly rotated the pelvis towards the free leg side, which assisted leg recovery by pulling the stance leg forward (Sado et al., 2017a). As the relationship between lumbosacral torque and running velocity was unclear, there was a possibility that the lumbosacral axial rotation torque did not increase linearly with the changes in the running velocity, which is similar to the ankle plantar flexion torque observed by Schache et al. (2011). However, the peak lumbosacral axial rotation torque towards the stance leg side increased as the running velocity increased and to a greater extent at higher velocities. This results in higher lumbosacral axial rotation positive and negative work during the initial and terminal flight phases, respectively. Thus, the present findings have rejected the possibility of the plateau of increase in axial rotation torque.

The peak lumbosacral axial rotation torque towards the stance leg side during maximal sprinting in this study (1.99 ± 0.46 Nm/kg) was slightly larger than the measurements of isometric maximal voluntary trunk contraction in normal men in previous studies (1.83 ± 0.56 Nm/kg in 30° rotation (Kumar, 1997) and 1.83 Nm/kg in 40° rotation (Torén and Öberg, 1999)). When maximal axial rotation torque was observed, it exerted a negative power. Saunders et al. (2005) also observed that the lumbar axial rotator (external and internal oblique) muscles activated eccentrically, decreasing the difference between pelvic and lumbar rotations during slow running. The axial rotator muscles achieved a slightly larger torque than the maximal strength with eccentric contraction during sprinting. Although the participants and the type of contraction differed between the present study and previous studies (Kumar, 1997; Torén and Öberg, 1999), we speculated that near maximal torque is required for maximal sprint. To clarify this speculation, further study needs to examine the axial rotator strength in sprinters under dynamic (isokinetic or isotonic) conditions.

Previous EMG reports showed that the lumbar extensor (erector spinae) and lateral flexor (internal/external oblique) muscle activities increased with running velocity increments ranging 1.4–3.3 m/s (Cappellini, 2006) and 2.0–5.0 m/s (Saunders et al., 2005). The lumbosacral extension and lateral flexion toward free leg side torques in the present study increased with running velocity increments in <7.30 m/s; however, at a joint level, increases in lumbosacral extension and lateral flexion torques plateaued above 7.30 m/s. While sprinting, the athlete needs to exert a lumbosacral extension torque almost equal to the stance leg hip extension torque to maintain the anterior pelvic tilt (Sado et al., 2016). Schache et al. (2011) reported that the hip extension torque did not increase significantly above 7.0 m/s, as with the lumbosacral extension torque observed in this study. Thus, lumbosacral extension torque plateaus at high velocities are, in effect, plateaus of the stance leg hip extensors, implying that the lumbosacral extension torque might be adjusted to be similar to stance leg hip extension torque during running at various velocities. In the frontal plane, the torque produced by the upward force of the stance leg must be cancelled to prevent pelvic drop. Hip abductors of the stance leg are mainly responsible for maintaining pelvic position (Petrofsky, 2001; Westhoff et al., 2006). The lumbosacral lateral flexion torque likely assists hip abduction torque to maintain pelvic position. The vertical GRF did not increase above 7 m/s (Dorn et al., 2012; Weyand et al., 2010). Thus, lateral flexion torque plateaus would be the result of plateaus of the vertical GRF. Sprinters can increase running velocities in three ways: pushing the ground more forcefully, increasing the frequency of ground pushing, or a combination (Schache et al., 2014); however, the force exerted on the ground is limited by increases in running velocity, because as the running velocity increases, the contact time with the ground decreases (Dorn et al., 2012; Weyand et al., 2010). Therefore, we suggest that lumbosacral extension and lateral flexion torque plateaus at high velocities results from limitations in leg extensor force developments caused by the shorter ground contact time.

Our findings have practical implications for sprinters and distance runners. Strength trainings on the axial rotators can be recommended to improve sprinting performance. Strength exercises on the extensors and lateral flexors may be recommended when undesirable movements due to their weakness (e.g. excessive pelvic posterior tilt for extensors and excessive pelvic drop for lateral flexors) are observed. Future studies should conduct training experiments to show the effect of strengthening each lumbopelvic torque exertion on sprinting performance. In contrast, faster running at distance running velocity range (<7 m/s) required larger extension and lateral flexion torque exertions. The excessive pelvic posterior tilt (loss of moderate lumbar lordosis) (Chun et al., 2017) and pelvic drop (Lavine, 2010) would be implicated with several types of pain/injuries, such as low back pain and iliotibial band friction syndrome. Our findings imply that faster running at distance running velocity range might increase the biomechanical load, which has the potential to cause pain/injury related to lumbopelvic extensors and lateral flexors. However, we did not directly examine the biomechanical mechanisms regarding the relationship between pain/injury and running velocity, which is an important future theme.

This study has some limitations. First, the thoracolumbar region is not a rigid structure, which may affect the lumbosacral power/work calculations. However, the effect of running velocity increments on joint power/work is similar to that on the joint torque, suggesting that modelling the thoracolumbar region into multiple segments would not change our conclusion. Second, only two trials were analysed per velocity for each participant to avoid the fatigue with increasing number of trials. However, the inter-trial variabilities of kinetic values were small in most participants. A previous kinetic study using a single trial for each velocity also reported high inter-trial repeatability (Schache et al., 2015). Third, the order of running velocities was incremental rather than randomised for practical reason, implying the possibility of order effect. Fourth, we did not control the stance leg to avoid increasing number of trials, which may have effects on the results. However, the inter-condition changes of stance leg did not critically affect the trends of peak torques (Appendix). Fifth, the sample size was limited to 10 participants. However, a larger sample size would not alter our conclusion regarding variance/invariance of lumbosacral torque exertions, because (1) the mean value of the peak extension torque in 9.29 m/s was smaller than that in 7.30 m/s and (2) the difference of the mean values of the peak lateral flexion torque between 7.30 m/s and 9.29 m/s was small compared with the standard deviation. Sixth, as we performed a joint level analysis, we cannot quantify the individual muscle activities/forces. Even if the net joint torque remains unchanged, muscle activities can change. Moreover, some lumbopelvic muscles have multiple plantar actions. Future studies need to examine individual muscle activities/forces to understand how human controls each muscle to perform different adaptation of each planer torque to faster running.

In conclusion, we investigated lumbosacral kinetics at different running velocities to examine whether an increase in each lumbopelvic torque is required for faster running. We found that the running velocity increments from slow to maximal sprinting required an increase in the lumbosacral axial rotation torque, although sprinting is generally believed to be a sagittal movement. Conversely, the lumbosacral extension and lateral flexion torques were invariant to running velocity in the high-velocity range. Our findings imply that strength training for axial rotation torque may be recommended to improve sprinting performance and that the load with the potential to cause pain/injury related on extensors and lateral flexors may increase with faster running in distance running velocity range. These findings would be useful for sprinters and distance runners to design their training programs.

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Declaration of Competing Interest

There are no conflicts of interest to declare.

Appendix

See Fig. A1.

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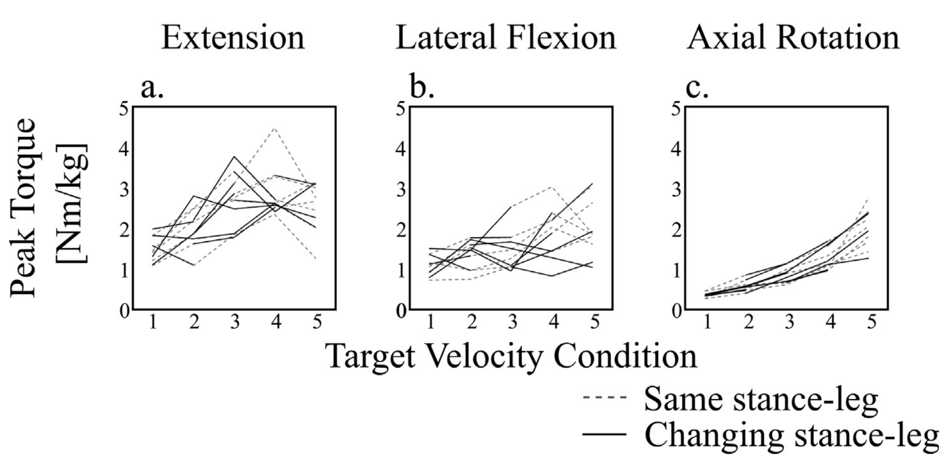

Fig. A1. Effects of the inter-condition changes in the stance leg on lumbosacral torques. The solid and dashed lines represent the data with the same stance leg and changing stance leg, respectively.

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