THE THREE-DIMENSIONAL PELVIC MOTION IN THE ACCELERATION AND MAXIMUM VELOCITY PHASES

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The purpose of this study was to investigate the mechanism of pelvic motion in the acceleration and maximum velocity phases. Eleven male sprinters performed 30 m and 60 m sprints at maximal effort and captured sprint movement at 15 m and 50 m. The contact time was significantly longer, and the angular displacement of pelvic elevation on the frontal plane was significantly greater in the acceleration phase than in the maximum velocity phase. Additionally, the angular displacement of pelvic elevation on the frontal plane showed a significantly positive correlation with the contact time and the stance length in the acceleration phase. These findings suggest that the greater pelvic frontal plane motion in the acceleration phase could increase the contact time and longer stance length.

KEY WORDS: sprinter, step parameters, pelvic motion, three-dimensional

INTRODUCTION: The 100 m race time is significantly correlated with maximal sprinting speed, and the time during which the sprinter can accelerate with maximal effort is limited to 5−7 s. Because the maximal sprinting speed depends on the preceding increase in speed in the acceleration phase, the ability to accelerate is critical to 100 m race performance. In particular, the acceleration phase is an important phase for obtaining the maximum running speed. Many biomechanical sprinting studies examining the acceleration phase have been reported (Nagahara, Matsubayashi, Matsuo & Zushi, 2014), and the lower limb joint angle and GRF data (Morin et al., 2015) have been clarified. The pelvis, which is located at the center of the body, is vital to the transmission of mechanical energy (Bosch & Klomp, 2005). Pelvic motion contributes to the acquisition of stride frequency and stride length for determining the sprinting speed. Sado, Yoshioka & Fukashiro (2017) reported that the stride frequency is affected by rotating of the pelvis to the swing leg side in the stance phase. Additionally, Preece, Bramah & Mason (2016) reported that the pelvic elevation of the swing leg side in the stance phase acquires the flight time and contributes to increased stride length. However, while most of these previous studies focused on pelvic motion during the maximum velocity phase (Sado, Yoshioka & Fukashiro, 2017), the characteristics of pelvic motion during the acceleration phase and its relationship with stride parameters have not been clarified. Therefore, the purpose of this study was to investigate the mechanism of the pelvic motion in the acceleration and maximum velocity phases.

METHODS: The participants were 11 male college sprinters (mean ± standard deviation (SD) age, 19.82 ± 0.75 years; height, 1.75 ± 0.05 m; body mass, 66.21 ± 4.06 kg; personal best time in a 100 m sprint, 10.87 ± 0.27 s). All procedures undertaken in the study were approved by the Ethics Committee for the Institute of Health and Sport Sciences, University of Tsukuba, Japan. All participants wore close-fitting clothing and running spikes. Each participant was attached with 47 retro-reflective markers to their trunk and limbs for motion capture. After an appropriate warm-up session, the participants performed 30 m and 60 m sprints with maximal effort until two trials were completed, in which either foot established contact with a force platform (Kistler, Force Plate, 1000 Hz) located 15 m and 50 m from sprint commencement. All participants performed the two successful trials. Adequate recovery time was provided 15 minutes between trials to avoid fatigue. A 28-camera motion capture system (Vicon Motion Systems, Vicon T20 system, 250 Hz) was used to record the three-dimensional coordinates of the position of each reflective marker. The ground reaction

force (GRF) was recorded using a force platform, and the values were synchronized with the motion data. The motion capture volume was 1.0 m × 8.0 m × 2.0 m, with a force platform located at the center of the captured volume. The x-, y- and z-axes of the global coordinate system (GCS) defined the medial-lateral, anterior-posterior, and superior-inferior directions, respectively. The position coordinates of the markers were smoothed using a Butterworth, low-pass digital filter. A residual analysis (Wells & Winter, 1980) was performed to identify the optimal cut-off frequency for the three-dimensional positions of each marker in each trial, and cut-off frequency between 15 and 25 Hz were ultimately used for the dataset. The leg stepping on the force platform and the other leg were defined as the ‘‘stance leg’’ and ‘‘swing leg,’’ respectively. Instants of stance leg touch-down (TD) and toe-off (TO) were identified from the onset of the GRF signal. The threshold of the GRF was 10N. The swing leg TD was identified via kinematic methods using the vertical acceleration of the marker on the swing leg toe, as described by Nagahara & Zushi (2013). The center of mass (CoM) and the inertial parameters were estimated based on the body-segment parameters of Japanese athletes (Ae, Tang & Yokoi., 1996). The hip joint centers were calculated based on recommendations by the Japanese Clinical Gait Analysis Forum (1992). We calculated the pelvic segment angles in the frontal, sagittal, and transverse planes (X−Z, Y−Z, and X−Y planes of the GCS, respectively). A pelvis segment coordinate system was defined using the hip joint centers, the mid-point of the posterior superior iliac spines, and the anterior superior iliac spines (Ota et al., 2020). Three-dimensional angular kinematics were calculated using the Cardan angle sequences, X→Y’→Z’’. Based on this definition, zero-degree pelvis angles were observed when the alignment of the pelvis matched the orientation of the GCS. The contact time and flight time were calculated based on the number of frames between TD and TO. The stride frequency was the reciprocal of the time required for one step. The stance length was the horizontal distance from the CoM at TD to the CoM at TO. The stride length was the horizontal distance from the CoM at TD to the CoM at the next TD. Sprint speed was defined as the product of stride frequency and stride length. In addition, the pelvic angular displacement (Δ) was calculated by subtracting the minimum from the maximum values of the pelvic angle in the stance phase. The time-series data of the stance phase were normalized to 0−100 %.
Means and SDs were calculated for descriptive analyses. A paired *t*-test was used to compare the characteristics of each phase for each variable. Cohen’s d-value effect size was calculated as the quotient of the difference between groups divided by the SD.

RESULTS: Table 1 shows the step parameters in the acceleration and maximum velocity phases. In the acceleration phase, subjects performed a significantly lower sprint speed, a significantly shorter stride length and a significantly longer flight length. However, the stride frequency and stance length between each phase did not differ significantly. In addition, the contact time in the acceleration phase was significantly longer than that in the maximum velocity phase. Table 2 shows the pelvic angle during the stance phase in the acceleration and maximum velocity phases. The pelvis in the sagittal plane was significantly greater pelvic

Table 1 Mean step parameters (±SD) in each phase.

	Acceleration	Maximum velocity	*p*	Effect size(*d*)
Sprint speed (m/s)	7.53 ± 0.25	9.32 ± 0.26	<0.01*	7.02
Stride frequency (Hz)	4.30 ± 0.36	4.23 ± 0.35	0.49	0.28
Stride length (m)	1.77 ± 0.19	2.22 ± 0.19	<0.01*	2.37
Stance length (m)	0.97 ± 0.09	0.98 ± 0.07	0.35	0.12
Flight length (m)	0.80 ± 0.13	1.24 ± 0.16	<0.01*	3.02
Contact time (s)	0.129 ± 0.008	0.106 ± 0.006	<0.01*	3.25
Flight time (s)	0.105 ± 0.015	0.132 ± 0.017	<0.01*	1.68

* : p < 0.05

anterior tilt at TD and TO in the acceleration phase. At initial ground contact, the degree of pelvic elevation on the swing leg side in the acceleration phase was significantly greater than that in the maximum velocity phase. In addition, the angular displacement of the pelvic elevation on the frontal plane in the acceleration phase was significantly greater than that in the maximum velocity phase. Figure 1 shows the relationship between the angular displacement of the pelvic elevation on the frontal plane and each parameter from each phase. The angular displacement of the pelvic elevation on the frontal plane showed a significantly positive correlation with the contact time and stance length in the acceleration phase.

Table 2 Mean pelvic angle (±SD) in each phase.

	Acceleration	Maximum velocity	*p*	Effect size(*d*)
Sagittal				
Anterior/Posterior tilting angle at TD (deg)	17.7 ± 6.9	5.2 ± 4.5	<0.01*	2.14
Anterior/Posterior tilting angle at TO (deg)	20.3 ± 5.3	9.1 ± 4.7	<0.01*	2.23
Maximum anterior/posterior tilting angle (deg)	21.4 ± 5.9	9.8 ± 4.9	<0.01*	2.14
Minimum anterior/posterior tilting angle (deg)	16.7 ± 6.4	5.0 ± 4.5	<0.01*	2.10
Δ Anterior/Posterior tilting angle (deg)	4.8 ± 2.5	4.8 ± 1.6	0.99	0.01
Frontal				
Elevation/Depression angle at TD (deg)	−5.0 ± 2.4	−3.8 ± 2.5	0.01	0.53
Elevation/Depression angle at TO (deg)	4.3 ± 2.3	4.1 ± 2.9	0.77	0.06
Maximum elevation/depression angle (deg)	4.3 ± 2.3	4.1 ± 2.9	0.75	0.06
Minimum elevation/depression angle (deg)	−7.8 ± 2.0	−5.0 ± 2.4	<0.01*	1.38
Δ Elevation/Depression angle (deg)	12.3 ± 2.0	9.1 ± 2.4	<0.01*	1.44
Transverse				
Swing/Stance leg side rotation angle at TD (deg)	−1.1 ± 2.8	−3.5 ± 4.5	0.04	0.66
Swing/Stance leg side rotation angle at TO (deg)	−8.0 ± 4.6	−8.9 ± 4.6	0.28	0.20
Maximum swing/stance leg side rotation angle (deg)	−1.0 ± 2.8	−3.2 ± 4.3	0.07	0.59
Minimum swing/stance leg side rotation angle (deg)	−17.3 ± 3.0	−17.2 ± 3.8	0.95	0.02
Δ Swing/Stance leg side rotation angle (deg)	16.2 ± 2.4	14.0 ± 3.0	0.03	0.81

* : $p < 0.05$

DISCUSSION: The purpose of this study was to investigate the mechanism of the pelvic motion in the acceleration and maximum velocity phases. These results of the step parameters were similar to the changes in stride frequency and stride length from sprint commencement to 100m in previous studies (Haneda, Ae, Enomoto, Hoga & Fujii, 2003; Mackala, 2007). The angular displacement of pelvic elevation on the frontal plane in the acceleration phase was significantly greater than that in the maximum velocity phase (Table 2). Additionally, the angular displacement of pelvic elevation on the frontal motion showed a significantly positive correlation with contact time and stance length in the acceleration phase (Figure 1). Kugler and Janshen (2010) suggested that a greater forward lean of the body during sprint in the acceleration phase can be achieved by a longer contact time. Therefore, these results indicate to obtain a large impulse product in the acceleration phase, the angular displacement of pelvic elevation on the frontal plane during the stance phase may be greater, and consequently the contact time and the stance length in the acceleration phase may increase. The flight time in the acceleration phase was significantly shorter than that in the maximum velocity phase (Table 1). Additionally, the pelvic rotation angle at TO shows a similar tendency in each phase (Table 2). Sado, Yoshioka & Fukashiro. (2017) suggest that stride frequency is affected by rotating of the pelvis to the swing leg side in the stance phase.

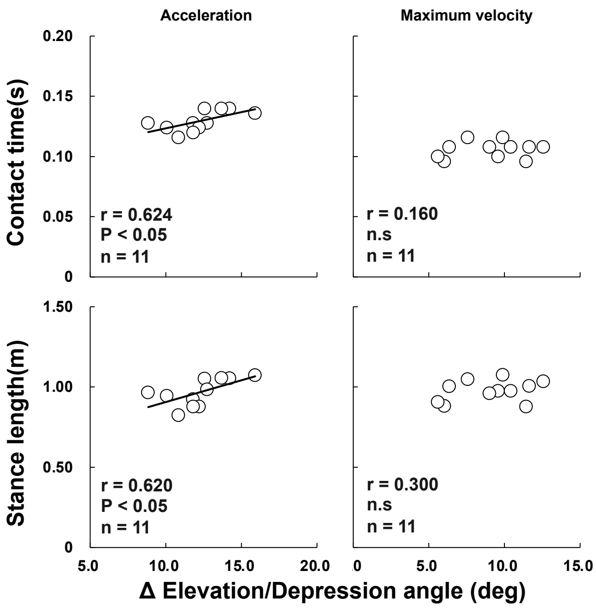

Figure 1: Relationship between the angular displacement of pelvic elevation on the frontal plane and each parameter in each phase.

These results indicate that pelvic rotation during the stance phase contributes to recovery motions during sprinting and plays an important movement to obtain the stride frequency in each phase.

CONCLUSION: In the present study, the contact time was significantly longer, whereas the flight time was significantly shorter in the acceleration phase. Additionally, the angular displacement of the pelvic elevation on the frontal plane in the acceleration phase was significantly greater than that in the maximum velocity phase. Moreover, the angular displacement of the pelvic elevation on the frontal plane showed a significantly positive correlation with the contact time and stance length in the acceleration phase. Therefore, the results in the acceleration phase suggest that the greater pelvic frontal plane motion could increase the contact time and a longer stance length and pelvic transverse plane motion could maintain stride frequency.

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