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Leg stiffness in unilateral transfemoral amputees across a range of running speeds

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ABSTRACT

Carbon fiber running-specific prostheses have allowed lower extremity amputees to participate in running activity by providing spring-like properties in their affected limb. It has been established that as running speed increases, stiffness of the leg spring (leg stiffness; k_{leg}) remains constant in non-amputees. Although a better understanding of k_{leg} regulation may be helpful for the development of spring-based prostheses, little is known about stiffness regulation in unilateral transfemoral amputees. The aim of this study was to investigate stiffness regulation at different running speeds in unilateral transfemoral amputees wearing a running-specific prosthesis. Nine unilateral transfemoral amputees performed running on an instrumented treadmill across a range of speeds (30, 40, 50, 60, and 70% of their maximum running speed). Using a spring-mass model, k_{leg} was calculated as the ratio of maximal vertical ground reaction force to maximum leg compression during the stance phase in both affected and unaffected limbs. We found a decrease in k_{leg} from the slower speed to 70% speed for the affected limb, whereas no change was present in the unaffected limb. Specifically, there was a significant differences in the k_{leg} between 30% and 70%, 40% and 70%, and 50% and 70%, and the magnitude of the k_{leg} difference between affected and unaffected limbs varied with variations in running speeds in unilateral TFAs with an RSP. These results suggest the k_{leg} regulation strategy of unilateral transfemoral amputees is not the same in the affected and unaffected limbs across a range of running speeds.

1. Introduction

Recent technical developments of running-specific prostheses (RSPs) with energy storing capabilities have allowed lower extremity amputees to participate in running activities by providing spring-like properties in their affected limb. To describe the spring-like leg behavior during running, the whole body is often modeled as a spring-mass model that consists of a body mass supported by a mass-less linear leg spring (Fig. 1; Blickhan, 1989; McMahon and Cheng, 1990). In this model, stiffness of the leg spring (leg stiffness; k_{leg}) is defined as the ratio of maximal vertical ground reaction force (F_{peak}) to maximum leg compression (ΔL) during the stance phase. Previous studies have demonstrated that non-amputees adjust the spring-like leg behavior at different running speeds by increasing the angle swept (θ) by the stance limb while keeping k_{leg} nearly constant (He et al., 1991; Farley et al., 1993; McMahon and Cheng, 1990), indicating that the constant-stiffness leg spring may be a basic and invariant characteristic of running. Spring-like leg behavior and stiffness regulation have long been considered the principal characteristics of bouncing gaits for legged animals including humans, but it is still unknown how unilateral transfemoral amputees (TFAs) who use a running-specific prosthesis regulate their spring-like leg behavior during bouncing gaits. A better understanding of spring-like leg behavior and stiffness regulation in this population will provide us with insight into the underlying biomechanics and control mechanism in humans and would be expected to aid in developing design parameters for spring-based prostheses for running (Farley and Gonzalez, 1996).

Despite several studies examining the spring-like leg behavior during running at different speeds in unilateral and bilateral transtibial amputees (Beck et al., 2017; Hobara et al., 2013; McGowan et al., 2012), little is known about stiffness regulation in TFAs. Comparing with unilateral transtibial amputees, running is a more demanding task for unilateral TFAs (Mensch and Ellis, 1986) due to the loss of both the biological knee and ankle joints in their affected limb. Indeed, Burkett et al. (2003) reported

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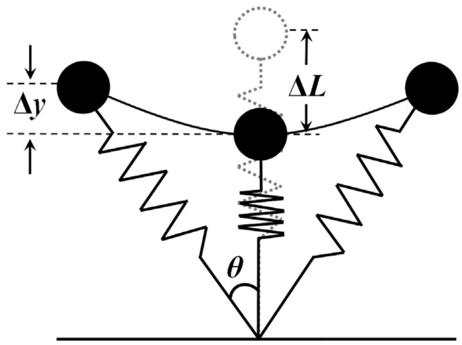

Fig. 1. Spring-mass model for running. The leg spring is compressed during the first half of the stance phase and rebounds during the second half. Maximal vertical displacement of the center of mass and leg spring compression during ground contact are represented by Δy and ΔL, respectively. Half of the angle swept by the leg spring during the ground contact is denoted by θ.

inter-limb asymmetry in kinetic variables during running at 2.47 and 3.22 m/s for four unilateral TFAs, but participants in their study used non-RSPs and two athletes adopted a "hop-skipping" movement pattern with a double-support phase during running. Although a recent study investigated k_{leg} regulation in unilateral TFAs (Sano et al., 2017), the study only analyzed maximal sprinting speed trials of each individual (5.79 ± 0.90 m/s). Therefore, it remains unclear how unilateral TFAs wearing RSPs regulate their spring-like leg behavior while running at different speeds.

The aim of this study was to investigate stiffness regulation at different running speeds in unilateral TFAs wearing RSPs. In treadmill running, previous studies showed that k_{leg} unchanged with speed in unaffected limbs, while it decreased in affected limbs of unilateral transtibial amputees wearing an RSP (McGowan et al., 2012). Further, a recent finding also demonstrated that k_{leg} during sprinting for the affected limb was approximately 12% lower than the unaffected limb in unilateral TFAs wearing RSPs (Sano et al., 2017). Therefore, we hypothesized that the k_{leg} regulation strategy of unilateral transfemoral amputees is not the same in the affected and unaffected limbs across a range of running speeds.

2. Methods

2.1. Participants

Nine TFAs volunteered to participate in the experiment (Table 1). Participants used their own RSPs. All participants belonged to a track and field team as sprinters who specialized in the 100-m sprint or long jump. They had been performing regular training between two and six days per week, for more than five years. All of the participants had competed at the national or international level within the preceding year, and their best recorded times in the 100-m sprint were on average 17.36 ± 2.36 s. The protocol was approved by the local ethical committee and was in accordance with the guidelines set out in the Declaration of Helsinki (1983).

2.2. Task and procedure

Participants ran on an instrumented treadmill (FTMH-1244WA, Tec Gihan, Kyoto, Japan) at incremental speeds of 30, 40, 50, 60 and 70% of their maximum speed. In this study, we calculated maximum running speeds for each individual by dividing the race distance (100-m) by the official personal best time

Table 1
Subject characteristics. * and ** indicated products by Ottobock (Duderstadt, Germany) and IMASEN Engineering Cooperation (Kakamigahara, Japan), respectively.

Subject	Sex	Age (years)	Height (m)	Total mass (kg)	Prosthetic knee	RSP model and category of stiffness	Time since amputation (years)	Etiology	Affected leg length (m)	Unaffected leg length (m)	100 m PR (s)	Speed range (m/s)
1	M	42	1.67	57.2	3S80*	1E90 Sprinter* #2	6	Cancer	0.89	0.88	17.66	1.71–3.96
2	M	54	1.70	65.8	3S80*	KATANA-β** #2	31	Trauma	0.95	0.89	16.25	1.85–4.30
3	M	23	1.68	55.7	3S80*	1E90 Sprinter* #3	20	Cancer	0.92	0.87	17.24	1.75–4.05
4	F	21	1.49	44.37	3S80*	1E90 Sprinter* #1	9	Sarcoma	0.83	0.76	20.66	1.44–3.39
5	F	19	1.56	58.9	3S80*	1E91 Runner* #3	5	Trauma	0.86	0.82	16.86	1.78–4.13
6	M	26	1.75	66.04	3S80*	1E90 Sprinter* #3	5	Trauma	0.95	0.90	14.08	2.10–4.95
7	M	34	1.61	58.67	3S80*	1E91 Runner* #5	21	Sarcoma	0.85	0.82	17.82	1.66–3.91
8	M	17	1.77	84.04	3S80*	1E90 Sprinter* #4	3	Congenital	0.94	0.92	14.59	2.06–4.81
9	F	21	1.52	51.59	3R95*	1E90 Sprinter* #2	13	Sarcoma	0.82	0.77	21.05	1.40–3.30
Mean		28.56	1.64	60.26			12.60		0.89	0.85	17.36	1.75–4.09
SD		12.42	0.10	11.13			9.51		0.05	0.06	2.36	

recorded in competition (Table 1). Participants started the series of trials at 30% of their maximum speed and the speed for each subsequent trial was increased by 10% until subjects approached 70% of their maximum speed. In all target speed, the belt speed was constantly accelerated at 0.84 m/s^2. On average, running speeds in the present study were 1.75 ± 0.24 m/s for 30%, 2.34 ± 0.32 m/s for 40%, 2.92 ± 0.39 m/s for 50%, 3.51 ± 0.47 m/s for 60%, and 4.09 ± 0.56 m/s for 70%. Based on a previous study (Zeni and Higginson, 2010), each subject performed running and walking at least 5 min as a familiarization period for instrumented treadmill running prior to the experiment. For each target speed, subjects performed running for 20 s. Subjects rested for as long as needed between speed conditions to minimize the effects of fatigue.

2.3. Data collection and analysis

Two six-degree-of-freedom piezoelectric force platforms (each with dimensions of 40 cm × 120 cm for a total measurement area of 80 cm × 120 cm) were embedded in the instrumented treadmill to collect vertical ground reaction force (VGRF) at 1000 Hz. The VGRFs were filtered using a fourth order, zero lag, low-pass Butterworth filter with a cut-off frequency of 25 Hz (Kram et al., 1998).

From VGRF data, we determined step frequency (f_{step}), ground contact time (t_c), and peak VGRF (F_{peak}) in both the unaffected and affected limbs. In the present study, k_{leg} (N/m) was calculated as the ratio of F_{peak} to peak leg spring compression (ΔL) during ground contact:

$$k_{leg} = F_{peak}/\Delta L \quad (1)$$

ΔL was calculated using the initial length of the leg spring (L_0), the distance from greater trochanter to the ground during standing upright, half of the angle swept by the leg spring while it was in contact with the ground (θ), the average forward velocity of the body (u) and ground contact time (t_c) at each step (He et al., 1991; Farley and Gonzalez, 1996; McMahon and Cheng, 1990):

$$\theta = \sin^{-1}(\mathrm{ut_c}/2\mathrm{L_0}) \quad (2)$$

Next, we determined ΔL using peak vertical displacement of the center of mass (COM) during ground contact (Δy) calculated by twice integrating the vertical acceleration of the COM with respect to time (Cavagna, 1975):

$$\Delta L = \Delta y + L_0(1 - \cos\theta) \quad (3)$$

Finally, k_{vert} (N/m) was calculated as the ratio of F_{peak} to Δy during ground contact:

$$k_{vert} = F_{peak}/\Delta y \quad (4)$$

We calculated k_{leg} as the ratio of F_{peak} and ΔL at the instant of F_{peak} (Hobara et al., 2008, 2009, 2010, 2012 and 2013). Since body mass influences the stiffness (Farley et al., 1993), k_{leg} was divided by the subject's body mass (kg).

2.4. Statistics

A two-way repeated measures ANOVA with two factors, running speed (5 levels) and limbs (2 levels), was performed to compare the unaffected and affected legs at five running speeds. In order to assess the assumptions of variance, Mauchly's test of sphericity was performed using the ANOVA results. If an assumption was violated, a Greenhouse-Geisser correction was performed to adjust the degrees of freedom, and if a significant main effect was observed, a Bonferroni post hoc multiple comparison was performed. Tests of simple main effect results were calculated for each dependent variable using a Bonferroni post hoc multiple comparison if a speed-by-limb interaction effect was present in the repeated-measures ANOVA. Statistical significance was set at $p < 0.05$. Statistical analysis was executed using SPSS (IBM SPSS Statistics Version 19, SPSS Inc., Chicago, IL).

3. Results

Fig. 2 shows vGRF-COM displacement curves for one subject during the ground contact while running at 30% to 70% of their maximum speed. Both affected and unaffected legs compressed after touchdown, and vGRF increased with increased COM displacement. The VGRF peaked at the moment of maximum leg compression (middle of the stance phase), and subsequently, the VGRF decreased with leg extension until take-off.

For all test subjects, there was no significant main effect of running speed ($F_{(1.66,\ 13.27)} = 0.328$, $p = 0.69$) on k_{leg}, but there was a significant main effect for limbs ($F_{(1,\ 8)} = 11.48$, $p < 0.05$) and an interaction effect ($F_{(4,\ 32)} = 12.58$, $p < 0.01$). Therefore, we examined the simple main effect, which indicated a decrease in k_{leg} from the slower to 70% speed for the affected limb, whereas no change was present in the unaffected limb. Specifically, there was a significant difference in the k_{leg} between 30% and 70%, 40% and 70%, and 50%

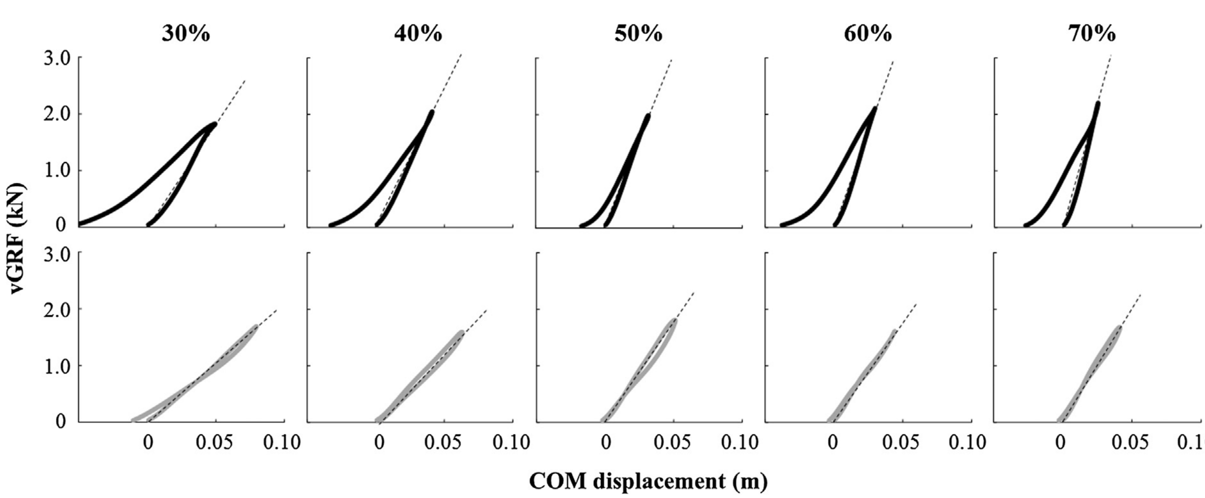

Fig. 2. Time-normalized VGRF-COM displacement curves during the ground contact during running at 30% to 70% of the maximum speed. Black thick (unaffected limb) and gray thick (affected limb) curves for one subject. The slopes (dotted lines) of these curves represent vertical stiffness (k_{vert}). k_{vert} is the slope of the VGRF-COM displacement curve in the leg compression phase.

and 70%. Further, the k_{leg} was significantly smaller in the affected limb than the unaffected limb at 40%, 50%, 60% and 70% speed (Fig. 3-A). The magnitude of the k_{leg} difference between affected and unaffected limbs varied with running speeds (23.0% at 40% speed, 37.1% at 50% speed, 40.2% at 60% speed, and 45.6% at 70% speed). F_{peak} showed significant main effects from running speed ($F_{(1.25,\ 9.96)} = 5.92$, $p < 0.05$) and limbs ($F_{(1,\ 8)} = 10.21$, $p < 0.05$). Further, statistical analysis revealed that there was a significant interaction effect on F_{peak} ($F_{(1.29,\ 10.30)} = 8.07$, $p < 0.05$). The simple main effect revealed a significant limb effect on the F_{peak}, while no significant speed effect was observed. F_{peak} was smaller in the affected limb compared to the unaffected limb at 40%, 50%, 60% and 70% (Fig. 3-B). Further, the magnitude of the differences in the F_{peak} between the limbs was greater in slower speed than 70% speed.

We also found significant main effects of running speed ($F_{(4,\ 32)} = 18.63$, $p < 0.01$), limbs ($F_{(1,\ 8)} = 10.74$, $p < 0.05$), and interaction effects ($F_{(4,\ 32)} = 6.64$, $p < 0.01$) on ΔL (Fig. 3-C). Simple main effects testing demonstrated an increase in ΔL from the slower to 70% speed for the affected limb, whereas no change was present in the unaffected limb. ΔL was significantly greater in the affected limb than the unaffected limb at 50%, 60%, and 70%, and the magnitude of the differences in the ΔL between the limbs was greater in 70% speed than slower speed.

Statistical analysis revealed that there were significant main effects of running speed ($F_{(1.56,\ 12.49)} = 31.93$, $p < 0.01$) and limb ($F_{(1,\ 8)} = 15.00$, $p < 0.01$) on k_{vert}, as well as the existence of a significant interaction effect ($F_{(1.50,\ 11.98)} = 10.93$, $p < 0.01$). Simple main effects testing demonstrated an increase in the k_{vert} from the

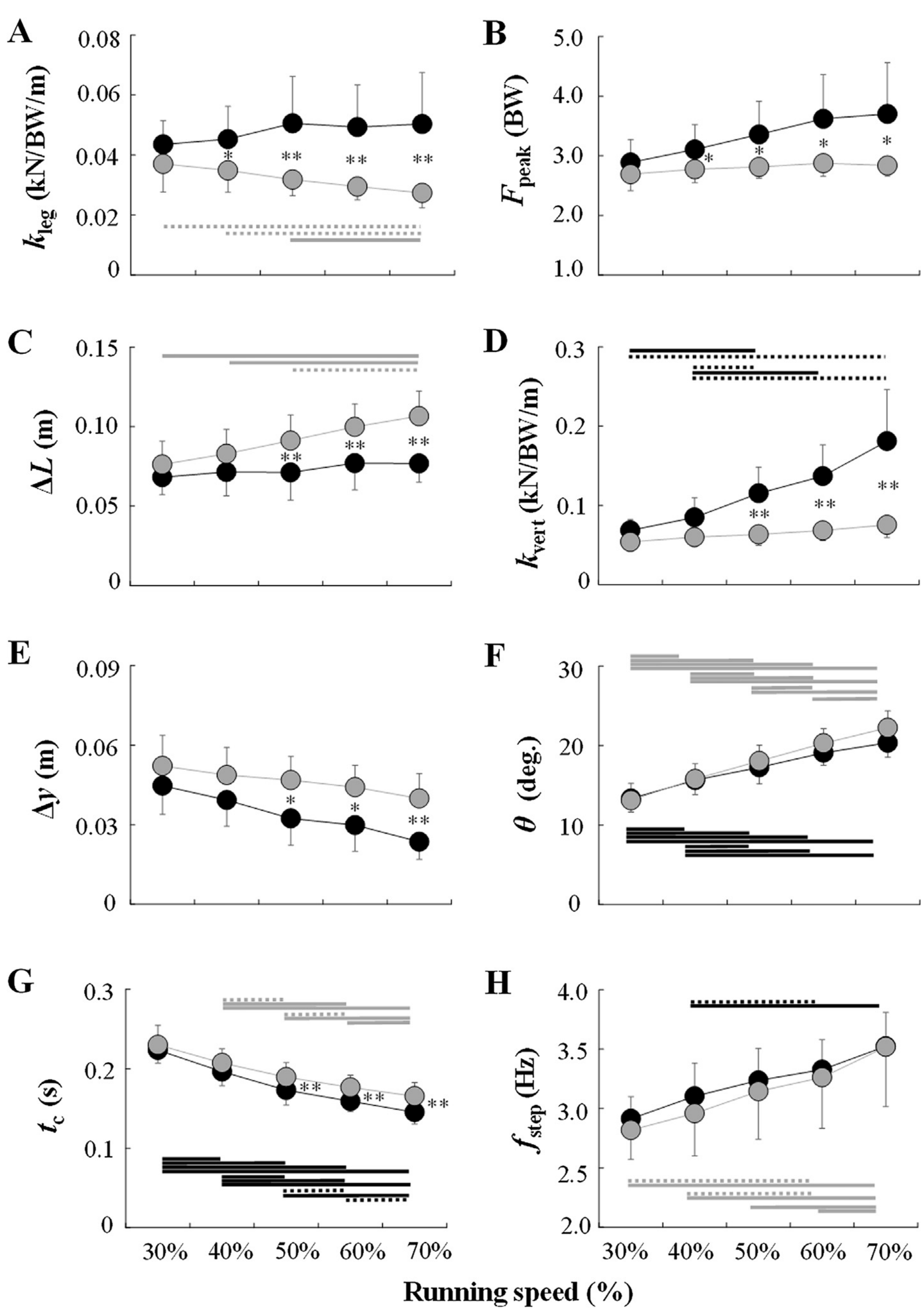

Fig. 3. Comparisons of (A) k_{leg}, (B) F_{peak}, (C) ΔL, (D) k_{vert}, (E) Δy, (F) θ, (G) t_c, and (H) f_{step} across a range of running speeds. Black (unaffected leg) and gray (affected leg) circles are means of nine subjects. Asterisks (*, **) indicate significant differences between the unaffected and affected legs at $p < 0.05$ and 0.01, respectively. Black (unaffected leg) and gray (affected leg) horizontal lines indicate significant differences at $p < 0.05$ (dotted lines) and 0.01 (solid lines), respectively.

slower to 70% speed for the unaffected limb, whereas no change was present in the affected limb. Specifically, there was a significant difference in the k_{leg} between 30% and 70%, 40% and 70%, and 50% and 70% (Fig. 3-D). k_{vert} was smaller in the affected limb compared to the unaffected limb at 50%, 60% and 70% (Fig. 3-D). It was also found that both running speed ($F_{(2.10,\ 16.79)} = 33.91$, $p < 0.01$) and limb ($F_{(1,\ 8)} = 11.11$, $p < 0.05$) have a significant main effect on the Δy. However, there was no significant interaction between running speeds and limb on the Δy ($F_{(4,\ 32)} = 1.71$, $p = 0.17$). Although it did not reach statistical significance, Δy tended to decrease with increasing running speeds. Further, it was found that Δy was significantly greater in the affected limb than the unaffected limb at 50%, 60%, and 70% (Fig. 3-E).

There were also significant main effects of running speed ($F_{(1.52,\ 12.18)} = 141.98$, $p < 0.01$) on θ, but not for limb ($F_{(1,\ 8)} = 2.46$, $p = 0.16$). However, there was a significant interaction effect on θ ($F_{(4,\ 32)} = 25.21$, $p < 0.01$). The post hoc analysis of simple main effect of running speed revealed that θ increased with running speed in both affected and unaffected limbs, but θ in the affected limb increased more than the unaffected limb (Fig. 3-F). Further, we also identified that both running speed ($F_{(1.52,\ 12.13)} = 69.57$, $p < 0.01$) and limb ($F_{(1,\ 8)} = 8.84$, $p < 0.05$) have a significant main effect on the t_c, as well as the existence of a significant interaction between running speeds and limb ($F_{(4,\ 32)} = 5.35$, $p < 0.01$). The simple main effects showed that t_c decreased with increasing running speed in both affected and unaffected limbs, but t_c was significantly longer in the affected limb than the unaffected limb at 50%, 60%, and 70% (Fig. 3-G). A significant main effect of running speed on f_{step} ($F_{(4,\ 32)} = 39.24$, $p < 0.01$) was identified where f_{step} increased with running speed in both limbs (Fig. 3-G). However, significant main effects of limb ($F_{(1,\ 8)} = 0.82$, $p = 0.39$, ES = 0.09) and interaction effect ($F_{(1.40,\ 11.16)} = 0.62$, $p = 0.50$) were not observed for f_{step}.

4. Discussion

The aim of this study was to investigate stiffness regulation at different running speeds in unilateral TFAs wearing an RSP. As shown in Fig. 3-A, we found a decrease in k_{leg} from the slower speed to 70% speed for the affected limb, whereas no change was present in the unaffected limb. Specifically, there was a significant differences in the k_{leg} between 30% and 70%, 40% and 70%, and 50% and 70%, and the magnitude of the k_{leg} difference between affected and unaffected limbs varied with variations in running speeds in unilateral TFAs with an RSP. Therefore, the results of the present study support our initial hypothesis that the k_{leg} regulation strategy of unilateral transfemoral amputees is not the same in the affected and unaffected limbs across a range of running speeds.

According to a previous study (Farley et al., 1993), the spring system is adjusted to operate at higher speeds by increasing θ while it is in contact with the ground rather than by increasing k_{leg}. Consequently, at higher speeds, the leg spring experiences greater vertical forces, which result in larger ΔL. Furthermore, it sweeps through a greater angle and the COM follows a flatter trajectory. The combination of these effects results in an increased k_{vert} and a decreased t_c. However, current results indicated that affected limb of unilateral TFAs could not maintain a constant k_{leg} with running at different speeds. A possible explanation for this phenomenon may be due to compensatory k_{leg} regulation during running. In the present study, unilateral TFA might try to compensate for the decrease in k_{leg} by increasing θ in their affected limb more than in the unaffected leg, with increases in k_{vert} and therefore decreases in t_c. However, as shown in Fig. 3-D and G, k_{vert} was significantly smaller and t_c was significantly longer in the affected limb than those of the unaffected limb at relatively faster speeds, indicating that the compensatory strategy of extra increase in θ might be insufficient. Consequently, the longer t_c in the affected limb could then be compensated by a decrease in flight time to induce an equal increase of f_{step} with increasing speed compared with unaffected limb (Fig. 3-F).

In the present study, k_{leg} was significantly smaller in the affected limb than the unaffected limb at 40%, 50%, 60% and 70% (Fig. 3-A). Current results are in accordance with a recent finding that k_{leg} during sprinting for the affected limb was approximately 12% lower than the unaffected limb in unilateral transfemoral amputees wearing RSPs (Sano et al., 2017). Further, as shown in Fig. 3-B, the smaller k_{leg} in the affected limb was associated with smaller F_{peak} than that of the unaffected limb. Several studies have demonstrated that vGRFs of the affected leg of sprinters with an amputation using RSPs are smaller than those of the unaffected leg during running and sprinting (Grabowski et al., 2010; Hobara et al., 2013; McGowan et al., 2012). Further, recent studies also showed that unaffected limbs of unilateral TFAs produced greater GRFs than affected limbs with RSPs from the start and the first two steps up to the maximum speeds (Makimoto et al., 2017; Strutzenberger et al., 2018; Willwacher et al., 2016). As suggested by previous studies, limited force production capability in affected limb than that of the unaffected limb observed in the present study may be due to the RSP's mechanical properties (Brüggemann et al., 2009; Noroozi et al., 2014; Baum et al., 2013) and/or muscle weakness/impairment derived from atrophy after amputation (Isakov et al., 1996).

As shown in Fig. 3-C, the smaller k_{leg} in affected limbs was also due to greater ΔL than that of unaffected limbs at 50%, 60%, and 70%. Further, the magnitude of the differences in the ΔL between the limbs was greater in 70% speed than slower speed. Current results were in accordance with a previous finding that leg compression in affected limb increased, while that of unaffected limb remained nearly constant in subjects with a unilateral transtibial amputation with increasing running speeds (McGowan et al., 2012). McGowan et al. (2012) also demonstrated that the increase in ΔL of the affected limb is predominantly owing to an increase in the leg angle at foot contact and the resulting half angle swept. Indeed, in our study, θ in the affected limb increased more than the unaffected limb (Fig. 3-F). Further, as shown in Table 1, the leg spring length (L_0) in the affected limb was about 5% longer than the unaffected limb. Since ΔL is a function of θ and L_0 when the foot is on the ground, a longer L_0 could contribute greater ΔL. Furthermore, a recent study suggested that the use of less stiff RSPs in athletes with bilateral transtibial amputations would result in decreased overall leg stiffness and increased ΔL during running (Beck et al., 2017), indicating that mechanical properties of RSPs (carbon fiber reinforced plastic) could explain the greater ΔL than intact limb of unilateral TFAs. In addition, a recent study suggested that socket-stump interface or proximal joint structure accompanied with as less stiffness of the hip joint in the frontal plane (larger pelvic obliquity) could contribute to the k_{leg} regulation during running (Oudenhoven et al., 2017). Therefore, greater ΔL in the affected limb compared to the unaffected limb may be attributed to differences in θ, L_0, mechanical properties of RSPs, other residual structures or any combination of these factors.

In the present study, we found a decrease in k_{leg} from the slower to 70% speed for the affected limb, whereas no change was present in the unaffected limb. Current results agree with a previous finding, which demonstrated that k_{leg} decreased with speed in affected limbs, while it remained constant in unaffected limbs of unilateral transtibial amputees wearing an RSP (McGowan et al., 2012). As with current results (Fig. 3A–C), the previous study also demonstrated that a decrease of k_{leg} in the affected limb was associated with both smaller F_{peak} and greater ΔL than that of unaffected limbs with increasing running speed (McGowan et al., 2012).

Therefore, the results of the present study suggest that spring-like leg behavior and k_{leg} regulation during running using RSPs may not necessarily depend on amputation levels.

In summary, the current results indicated that (1) affected limb of unilateral TFA cannot maintain a constant k_{leg} with running at different speeds, and (2) the k_{leg} regulation strategy of unilateral transfemoral amputees was not the same in the affected and unaffected limbs across a range of running speeds. Differences in the k_{leg} regulation strategy during running in unilateral TFAs might be due to the limited force production capability and mechanical constraints of their affected limb. Unilateral TFAs in the present study used different types of prosthetic knees with varied time since amputation, which might indirectly influence the running mechanics (Baum et al., 2013; Noroozi et al., 2014). Thus, caution needs to be taken regarding the interpretation and generalization of these findings.

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Conflict of interest statement

None of the authors have any conflicts of interest associated with this study.

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