

Article

Gait transition: activation of iliopsoas and major leg muscles in walking and running at the same speeds

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Abstract: Muscle activation changes may be one of the mechanisms that elicit the walk-to-run transition (WRT). The contribution of the iliopsoas muscle (IL) in gait transition has not been clarified and explored enough, despite its major role in hip flexion and controlling the swing leg. We aimed to determine the role of the IL and major leg muscles on the WRT. Twenty healthy young males walked and ran at 60%, 80%, 100%, 120% and 140% of their WRT-speed. To determine each individual WRT-speed, subjects started walking at 1.25 m/s and the speed was increased 0.1 m/s every 30 s, until they started running, and this speed was considered their WRT-speed. We analyzed surface electromyography (sEMG) of the IL, rectus femoris (RF), biceps femoris (BF), gluteus medius (GMed), tibialis anterior (TA), gastrocnemius (GAS) and soleus (SOL), in walking and running at the same speeds around the WRT. Major electromyography activity changes were found in IL, RF and TA for walking faster than the WRT-speed. Peaks of activity were significantly higher in walking than in running. Larger activation of IL, RF and TA is required to keep up with the increasing speed while maintaining walking pattern. These results underscore the importance of these muscles, especially the IL, to provide a powerful hip flexion to quickly bring the leg forward when walking at high speeds. However, when running at the same speeds, sEMG peaks of the IL are significantly lower. Such an activation pattern is also found in other muscles, suggesting that if certain muscles exceed an activation threshold this may trigger the transition from walk to run. Moreover, the abnormally high IL peaks at higher walking speeds could elicit high forces at the muscle insertions, which should be considered in the design of exercise or rehabilitation protocols.

Keywords: gait transition, walking, running, hip flexion, muscle activation.

1. Introduction

The most widespread gait forms in human locomotion, walking and running,



vary according to speeds, distinct biomechanical characteristics and muscle activity (Cappellini, et al., 2006; Gazendam & Hof, 2007; Ivanenko, et al., 2008).

Walking is a democratic, feasible, low-cost physical activity and transportation mode. Apart from its core functionality across a great variety of daily tasks, walking is also a popular exercise, which allows adjustment of intensity and level (Tudor-Locke & Rowe, 2012). As speed increases, individuals switch to running.

The mechanisms that determine the transition from walk to run (WRT) have been addressed from different perspectives (Kung, et al., 2018; Li & Ogden, 2012).

The energy minimization theory claimed that humans change gait to save energy, to keep gait cost low (Thorstensson & Rotherthson, 1987). However, studies found that the transition from walk to run occurs at a speed that is not optimal, energetically speaking (Hreljac, 1993; Minetti, Ardigo, & Saibene, 1994), therefore this theory has been mostly discredited nowadays (Kung et al., 2018).

Several other works have analyzed the dynamic stability of the system (Bruijn, et al., 2013.; Dingwell, et al. 2001; England & Granata, 2007; Jordan, Challis, & Newell, 2007; Raffalt, et al., 2020) within the dynamic non-linear behavior in human locomotion, as walking speed nears WRT-speed, the dynamic stability decreases as variability increases (Voigt & Hansen, 2021).

Recently, cadence has been emerging as a potential enhanced metric to predict the WRT (Chase et al., 2023; Hansen, et al., 2017; Tudor-Locke et al., 2018). Compared to other widely known predictors (Froude number and speed), cadence had not only the highest

accuracy but also highest sensibility and specificity in predicting the WRT (Chase et al., 2023). Nowadays, the constant development and technological upgrade in wearable devices, such as smartwatches, step-based data is often used as an important tool to measure health and exercise level (Akbari, Haghverd, & Behbahani, 2021) and cadence could be a great allied in this case.

The trigger theory then says that at a certain walking speed, muscle activity becomes extremely high, resulting in exertion and fatigue (Prilutsky & Gregor, 2001; Segers, et al., 2007), suggesting that gait transition is triggered at a muscular level. Other studies have shown that limitations of muscular effort, also related to overload prevention and potential injury (Bartlett & Kram, 2008; Hreljac, et al., 2001) have been already pointed out as determinants for WRT. Apparently, there are major changes in the activation of certain muscles related to the transition from walk to run (Hreljac et al., 2001; Neptune, Rick & Sasaki, 2005; Prilutsky & Gregor, 2001; Taira, Kawada, Kiyama, & Forner-Cordero, 2021).

When gait speed increases, the relative duration of gait phases change, such as reduction of double stance, thus decreasing the total stride duration while maintaining swing duration (Andersson, Nilsson, & Thorstensson, 1997). Further speed increase requires swinging the leg forwardly faster, demanding higher hip and knee flexors activity. In this respect, activation of the hip flexors, rectus femoris (RF) and iliopsoas (IL), could be critical to determine the WRT. On the other hand, some studies pointed to dorsiflexors effort or fatigue as determinant to the WRT (Hreljac et al., 2001; Pires, Lay, & Rubenson, 2014; Segers

et al., 2007) while others (Arnold, et al., 2013; Neptune, Rick & Sasaki, 2005) found ankle plantarflexors to contribute to the WRT.

On top of that, there is evidence of possible hip mechanisms to influence this transition and/or how hip muscles behave when increasing gait speed, for instance, based on EMG muscle activation (Prilutsky & Gregor, 2001), peak hip flexion joint power (Pires, Lay, & Rubenson, 2014) or simulated muscle power (Neptune, Sasaki, & Kautz, 2008).

Hip flexors are widely known as responsible for moving the leg forward during high-speed locomotion, and their power contribution and coordination along with other hip spanning muscles are well-documented in the literature (Bogey & Barnes, 2017; Kim & Lee, 2016; Lewis, Sahrmann, & Moran, 2009; Neptune, Sasaki, & Kautz, 2008). Some past studies found them to have the main changes with increase of speed (Mann, Moran, & Dougherty, 1986; Montgomery, Pink & Perry, 1994).

IL, a major hip flexor, anatomically connects the spine, pelvis and lower limbs, contributing to stabilize and move them in several activities (Andersson, Nilsson & Thorstensson, 1997; Andersson, et al., 1995). Although IL is considered the most powerful hip flexor, it is not frequently measured in locomotion studies due to its deep location and thus difficult access.

IL has been assessed through invasive measurement (Andersson, Nilsson & Thorstensson, 1997; Bogey & Barnes, 2017) and through surface EMG (Cappellini et al., 2006; Gottschall & Kram, 2005; Jiroumaru, Kurihara, & Isaka, 2014; Kim & Lee, 2016; Lewis et al., 2009; Wang et al., 2019). However, IL has not been explored

sufficiently along with other muscle's activation regarding its contribution to gait transition. Consequently, there is a lack of studies so far that focus on IL as a potential contributor to gait transition, within the context of other muscle's activation.

We hypothesized that IL, a major hip flexor that controls swing leg speed, has a significant role in the WRT. It may be related to the mechanical limits in hip flexion torque to swing the leg forwardly, as previously reported in the recovery from stumbling (Forner-Cordero, Koopman & Van Der Helm, 2004). Hence, the maximal leg speed, limited by hip flexor power, constraints walking speed and forces a change in locomotion pattern.

Our hypothesis is that due to the hip flexion role of the IL in walking, its swing peak activity in walking should increase with velocity. After gait transition to run, IL peak activity would be smaller in running.

Therefore, the goal of this paper is to investigate IL and major leg muscles behavior in gait transition, walking and running at the same speeds around the walk-run transition speed (WRT-speed).

2. Materials and Methods

Twenty healthy male volunteers (age 23.9 ± 5.4 years, height 172.9 ± 6.3 cm, body mass 65.5 ± 11.4 kg) participated in the study. They were all young male students from the Faculty of Medicine, Kagoshima University. They volunteered to participate, and their athletic level was not considered as an inclusion or exclusion criterion. Some participants practiced recreational sports, but none were athletes. All of them signed an informed consent. This study was approved by the Clinical Research Review Board of the

Faculty of Medicine, Kagoshima University (No. 213).

Subjects were asked to walk and run on a treadmill (Pulsar, HP/ Cosmos Sports & Medical GmbH Germany) at different speeds. Firstly, there was a warm-up protocol, walking for 5 min at 1.25 m/s and then running for 5 min at 2.25 m/s. To determine their WRT-speed, they started walking at 1.25 m/s and the speed was increased 0.1 m/s every 30 s, until they started running, and this speed was considered their WRT-speed. The average WRT-speed was 1.83 ± 0.12 m/s.

We determined WRT-speed in one trial, according to a previous study (Gill, 2022). The WRT-speed is reported to fluctuate and according to the literature, it is within a reasonable range (Gill, 2022; Thorstensson, 1987; Mohler, 2007).

Surface EMG from seven major muscles of the lower limbs was recorded with disposable surface bipolar Ag/AgCl electrodes (Noraxon dual electrode, Scottsdale, USA, inter-electrode distance of 20 mm) connected to an EMG amplifier Myoresearch (2.11.13, Noraxon, USA). The skin was shaved and cleansed with alcohol before attaching a pair of electrodes aligned with the direction of the muscle fibers. The electrodes were placed on the muscle bellies of RF, tibialis anterior (TA), biceps femoris (BF), gastrocnemius (GAS), soleus (SOL) and gluteus medius (GMed) following SENIAM recommendations (Hermens, et al., 2000).

IL electrodes were placed medially to the RF, lateral to the femoral artery (detected by pulse) and inferior to the inguinal ligament, similarly reproducing previous experiments (Gottschall & Kram, 2005; Kim & Lee, 2016; Lewis et al., 2009).

We tested crosstalk by functional resistance test (FRT) (Talib et al., 2019), which consisted of voluntary hip abduction and adduction to confirm IL low activation. Sartorius, as an abductor, is active in hip abduction, when IL, as hip flexor, is not active. We confirmed that IL EMG signal did not increase when sartorius was activated by voluntary hip abduction. IL individual muscle traces for walking and running are shown in Figure 1.

Foot switches (Norswitch, EM-134, Noraxon, Arizona, USA) were also attached to the heel and toes of the left foot, at the metatarsal heads, to determine walking and running cycles. Subjects walked and ran, gradually faster, at 60%, 80%, 100%, 120% and 140% of their WRT-speed.

They walked and ran at all these five speeds which were percentages of their preferred transition speed (WRT-speed). Keeping in mind that 100% WRT-speed is exactly when running becomes more comfortable and the transition from walking to running occurs naturally. Walking at 120% and 140% WRT-speed is challenging since it is unnatural for the respective subject to walk (as he would rather be running). The same is valid for running at 60% and 80% WRT-speed, when the subject would rather be walking.

Walking and running conditions were randomly ordered and separated by 2 min of rest, but speeds always followed an ascending order. Subjects had 30 s to get accustomed to each speed and data was taken for the consecutive 30 s.

Raw EMG data were bandpass-filtered (10-500 Hz), fourth-order Butterworth filter, performed automatically by Myoresearch software (2.11.13, Noraxon,

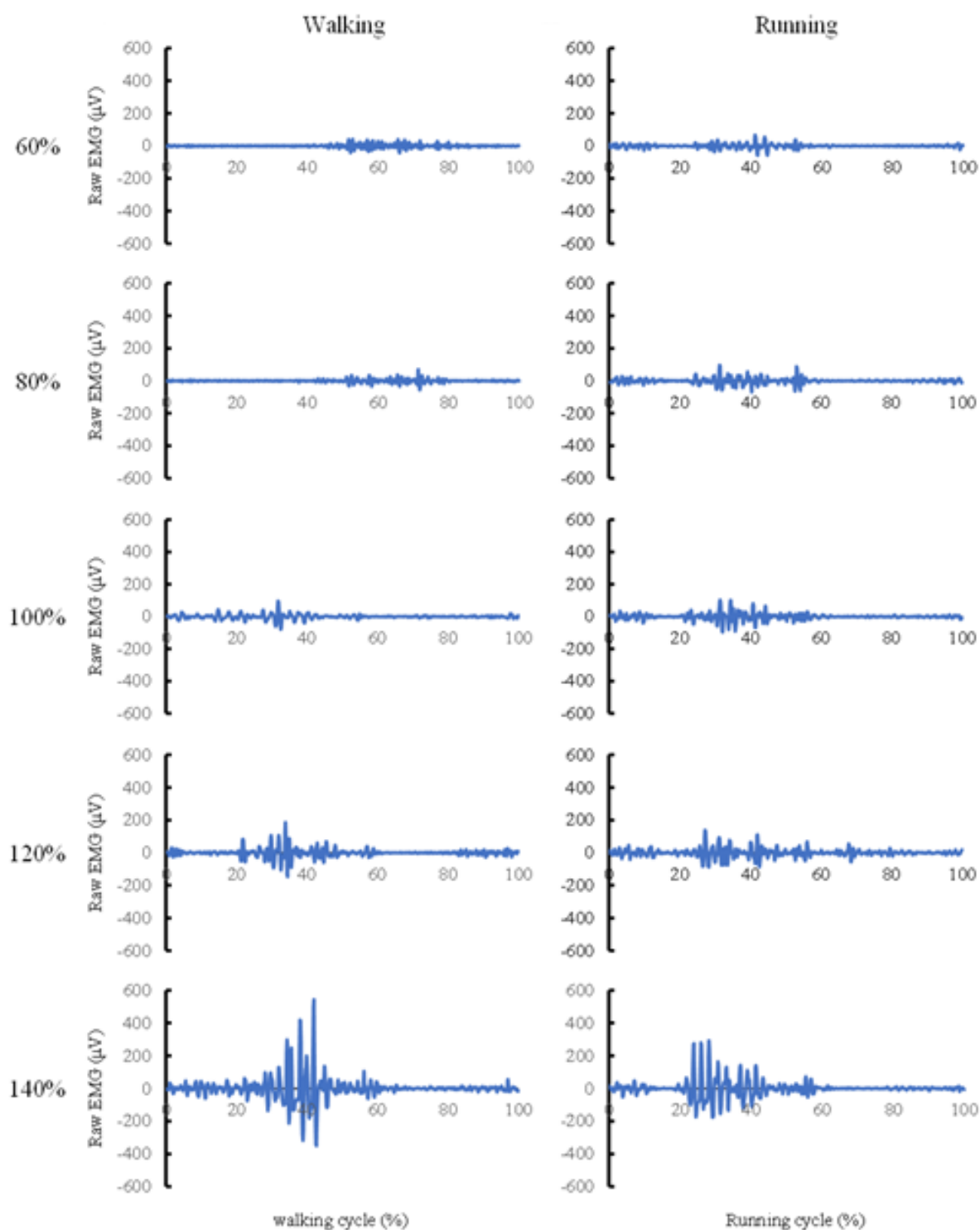


Figure 1. IL raw EMG data from the left leg muscles of one subject during walking and running at 60%, 80%, 100%, 120% and 140% of the walk-to-run transition.

USA), with a sampling frequency of 1000 Hz, rectified and smoothed by root mean square with 100 ms moving window and normalized to the maximum amplitude found at the WRT-speed during walking, based on previous studies (Andersson, Nilsson & Thorstensson, 1997, Gottschall & Kram, 2005).

Each trial duration was normalized to the percentage of the gait or running cycle between consecutive ipsilateral foot contacts to obtain an ensemble average of ten locomotion cycles for each subject. Normalized average EMG activation patterns for each subject were retained to select the peak amplitude value and its timing (in gait percentage) for every muscle at each condition.

The goal was to compare muscle activation in walking and running at the same speeds to investigate their contribution to gait transition. We focused on muscles that had their activation peak near the transition speed and after the transition to run, this activation decreased. The muscles selected are from both anterior and posterior chain, and it is expected that they present different peaks according to their position and respective to their antagonist.

The interval for peak selection considered the role of different muscles during gait and that some muscles may have two activation peaks. Therefore, we selected the peaks during pre-swing and swing phase of the IL, RF and TA, and peaks of the RF, BF, GMed, SOL and GAS during the stance phase. Finally, the BF peak before foot contact was also chosen.

Mauchly's test was applied to test the assumption of sphericity, and a Greenhouse-Geisser correction was used if

the data violated this assumption. Multivariate normality was assessed using Kolmogorov-Smirnov. A two-way repeated measures ANOVA was applied to test significant differences in EMG activation patterns between the two locomotion modes (walking vs running) and the five speeds (60%, 80%, 100%, 120% and 140% WTR-speed). ANOVA effect size was calculated using Eta squared (η^2).

For the relevant significant interactions, contrasts were performed comparing each speed to the lowest walking speed. We also plotted the interaction graphs and performed specific post-hoc comparisons to assess differences between speeds within locomotion mode (one-way ANOVA) or differences at the same speed between locomotion modes (paired t-tests). Statistics were performed using SPSS (Version 20.0 for Windows: Chicago, USA). It was considered significant a p-value of 0.05 with the appropriate corrections for multiple comparisons.

When significant interaction was observed, one-way ANOVA was used to test for significant effects of speeds, and the differences between the two conditions were tested by paired t-test. We focused on differences in muscle activity caused by locomotion patterns, thus, post hoc test was not performed to analyze differences among speeds. As there were only two conditions (walking and running), we considered using t-test to be appropriate. Other studies have applied similar statistical method (Salvatore et al., 2016, Matsuura et al., 2007).

3. Results

Normalized muscle traces averaged across all subjects for walking and running are shown in Figure 2.

Our results show that muscles increased activation patterns as speed increased (Figure 3). Besides, there are higher peaks of activity (near swing phase) from IL, RF (2nd peak) and TA (2nd peak) muscles for walking at speeds above the WRT-speed (Figure 2, 3).

The repeated measures multiple factor ANOVA revealed significant interactions for all the muscles between walking and running at different speeds, as expected ($F = 4.67-44.7$, $P < 0.019$, $\eta^2 = 0.018-0.352$). All Mauchly's tests indicated a violation of sphericity so the Greenhouse-Geisser correction was applied.

IL, TA and RF showed an increase in their activation along with increasing walking speed and this effect was more pronounced above the WRT-speed (Figure 2, 3). The most remarkable changes in EMG activity between walking above the WRT-speed and running were seen on IL, RF and TA peaks related to the swing phase in walking and BF prior to heel strike (Figure 2; Table 1). Therefore, we analyzed in more detail results for these muscles.

IL was active approximately in the middle of the gait cycle (Figure 2), which corresponds to the transition from hip extension in late stance advancing to hip flexion preparing for swing, and it has higher

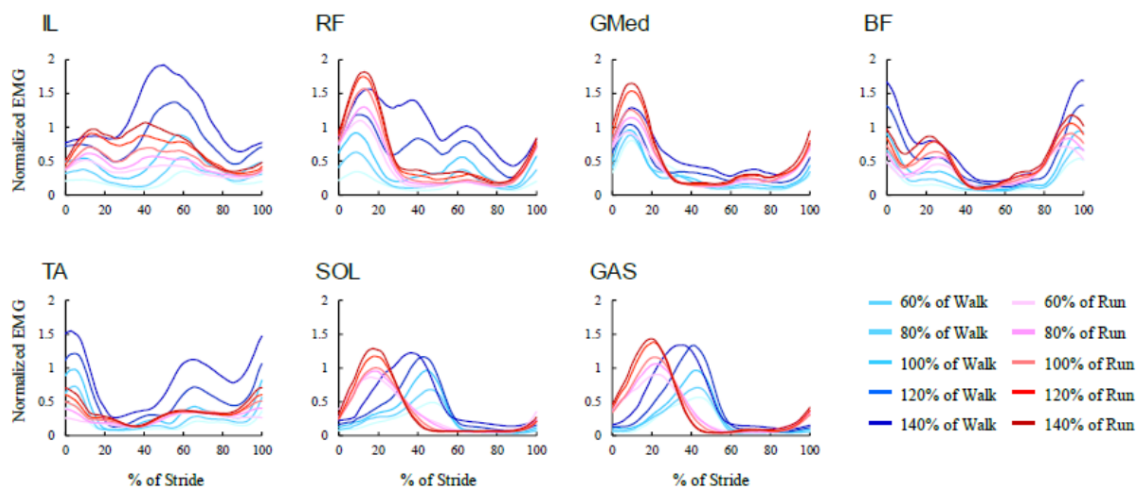


Figure 2. Normalized averaged muscle activation (EMG) of the left leg muscles during walking and running at 60%, 80%, 100%, 120% and 140% of the walk-to-run transition speed (WRT-speed). EMG waveforms activation are represented by ensemble average of all subjects, throughout the whole stride duration (normalized to percentage of the gait or running cycle).

Peak activation increased along with speed in both locomotion modes for all muscles, except for TA in running (Figure 3, 6).

B

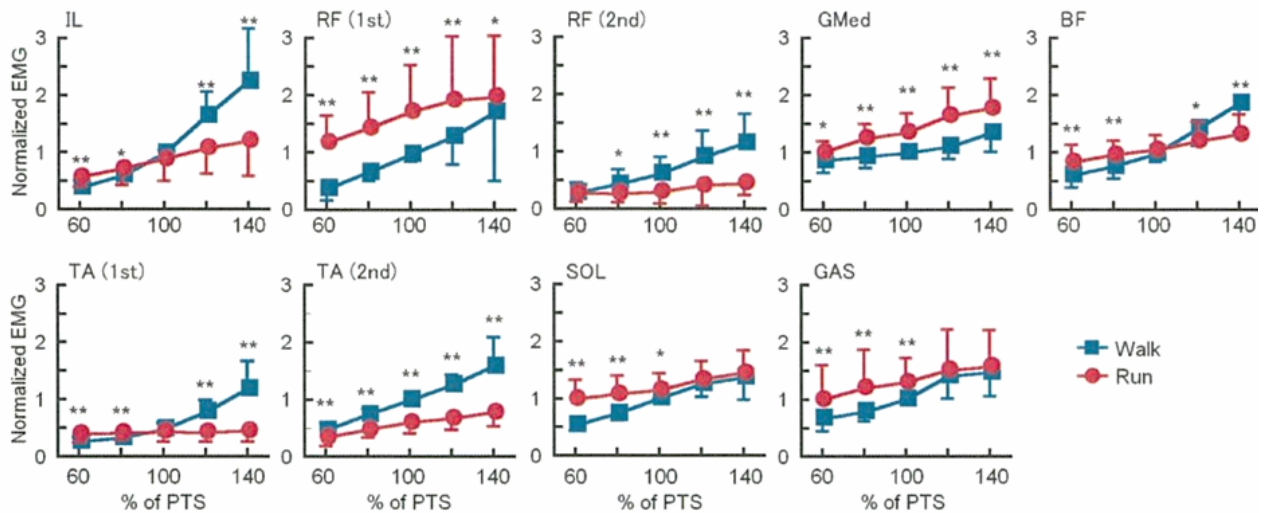


Figure 3. Activation peak values are presented across percentages of the walk-to-run transition speed (WRT-speed) velocities. An asterisk indicates statistically significant differences between walking and running conditions ($P < 0.05$), and double asterisk indicates ($P < 0.01$).

swing-related activation in walking above WRT-speed. Raw EMG data of IL for walking and running across speeds (one subject) are shown in Figure 1. The pairwise comparison at each speed between walking and running revealed significantly lower peaks during running for the speeds below WRT (Table 1).

There were no significant differences for IL at the WRT-speed but the peaks were

significantly higher in walking at speeds above the WRT-speed (Figure 4a).

RF had in walking and running at every speed, one activation peak at the beginning of the stance phase (0~20% of the cycle) and close to heel strike, related to extensor torque to foot contact and body weight support. More interestingly, there is

Table 1. Paired t-tests (t values and significance, $df=19$) of the IL, RF, TA and BF EMG peaks comparing walking and running at the same speeds expressed as a percentage of the WRT. Significant differences ($p < 0.001$) are marked in bold.

WALK- RUN	IL		RF		TA		BF	
	t	Sig.	t	Sig.	t	Sig.	t	Sig.
60%	5.99	.000	.79	.438	-4.58	.000	-3.94	.001
80%	2.29	.034	4.04	.001	-3.86	.001	-3.97	.001
100%	2.07	.053	6.02	.000	2.29	.033	-1.11	.283
120%	6.21	.000	6.49	.000	8.11	.000	2.7	.014
140%	8.03	.000	6.57	.000	7.21	.000	5.98	.000

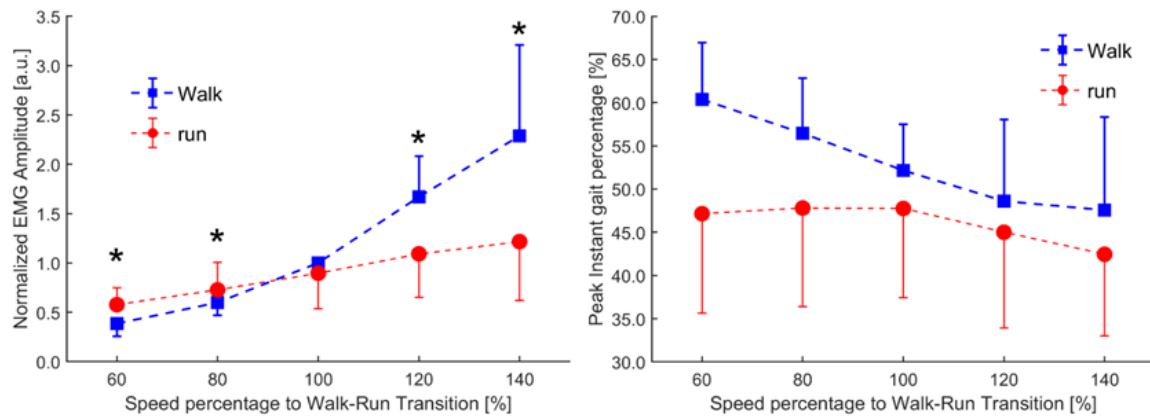


Figure 4. Interaction plots of the Iliopsoas: a) peak amplitudes and b) peak instants for the two locomotion modes (walk and run) and the five speeds. The asterisks indicate the significant differences between locomotion modes at each speed.

another RF peak related to swing around 60% of the cycle (Figure 2) and when walking speed increases, this peak also increases significantly (Figure 5, Table 1). It must also be noted that there is an additional peak for the two walking speeds above the WRT-speed around 40% of the cycle (Figure 2).

TA muscle activation was higher when walking at high speeds, and it also had two peaks of activation. One for the swing phase (around 60%~80% of the gait or running cycle), that above the WRT-speed is larger in walking than in running (Figure 6a). The other peak was at late-swing loading

response with walking having the highest activation across all speeds (Figure 3). The interaction plots of TA EMG peaks during swing showed a significant increase when walking faster than the WRT-speed (Figure 5). BF showed peaks prior to foot contact that also increased with walking speed (Figure 2).

GAS and SOL showed similar behavior with a peak prior to push off and higher variability in running and walking at higher speeds (Figure 2). GMed showed peaks after heel-strike and they were higher in running at all speeds (Figure 3).

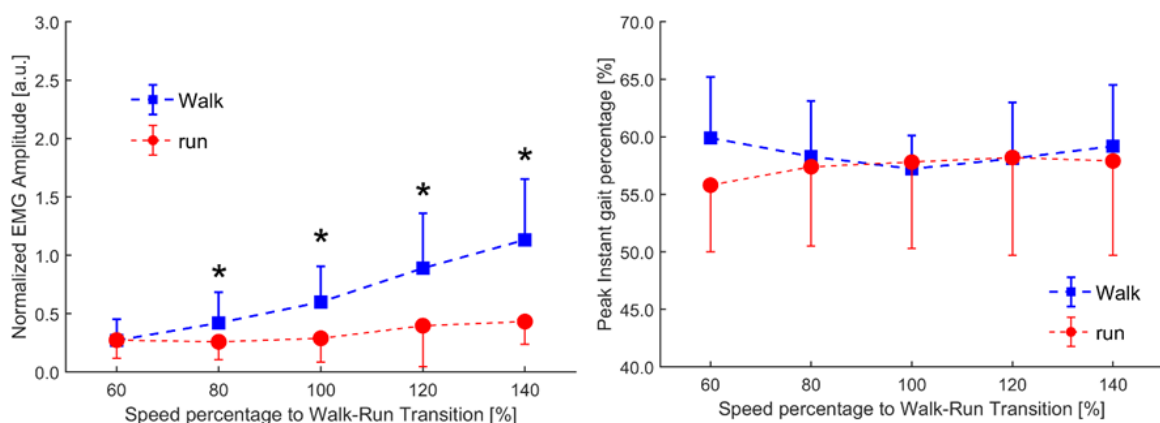


Figure 5. Interaction plots of Rectus Femoris: a) peak amplitudes and b) instants related to the swing phase for the two locomotion modes (walk and run) and the five speeds. The asterisks indicate the significant differences between locomotion modes at each speed.

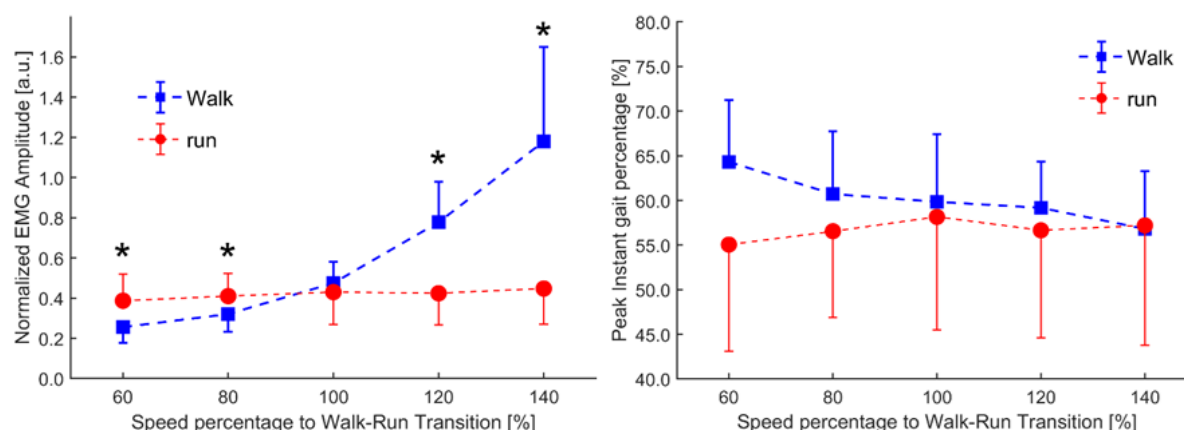


Figure 6. Interaction plots of Tibialis Anterior: a) peak amplitudes and b) peak instants for the two locomotion modes (walk and run) and the five speeds. The asterisks indicate the significant differences between locomotion modes at each speed.

4. Discussion

We hypothesized that IL, a primary hip flexor, controls swing leg speed and it is of paramount importance in the WRT, and the data presented here support this hypothesis.

All muscles increased activation patterns as speed increased, in agreement with previous works in the literature (Cappellini et al., 2006; Gazendam & Hof, 2007; Neptune, Sasaki, & Kautz, 2008; Prilutsky & Gregor, 2001). EMG activity of the IL, RF and TA increased for walking speeds above the WRT-speed with peaks significantly higher compared to running at the same speeds, indicating their crucial role in the WRT.

We expected that muscles that contribute to the WRT would show higher peaks in walking than in running at speeds beyond the WRT (Pires, Lay, & Rubenson, 2014; Prilutsky & Gregor, 2001). This is exactly what we found in RF, TA and even more remarkably, in IL.

IL activity increased with walking speed above the WRT-speed, well above the

values obtained in running at the same speeds, suggesting a large hip flexion effort to swing faster the leg while increasing walking speed. This fact supports the hypothesis that IL is a key factor in gait transition because there is a limit in the maximum IL activity and its increasing effort to maintain walking patterns above the WRT-speed can lead to fatigue, discomfort or even injury.

This not only agrees with our hypothesis but it is also aligned with other studies that pointed out to the same muscle to have a large contribution to increase gait speeds (Neptune, Sasaki, & Kautz, 2008; Mann & Hagy, 1980). Another result worth mentioning was that IL peak was also occurring earlier when walking speed increased. This is related to the reduction in double stance time, as a consequence of walking speed increases, which underscores the importance of the IL in controlling the start of the swing phase.

In fast walking, it is necessary to extend the hip joint during the late stance phase to increase stride length and quickly swing the lower limb while keeping it

extended. Because both are achieved through the activity of the hip flexors, it is likely that the iliopsoas muscle presents such significant activity. In addition, the shortening of the preparation phase for the swing due to the reduced double support phase is likely to accelerate the activity of the iliopsoas muscles.

IL was found to exert the largest peak muscle force (among hip spanning muscles) in a model prediction in running across speeds, especially at high speeds. The model has also shown that, in order to increase stride frequency, larger forces were generated by IL, hamstrings and gluteus maximus as they contributed to hip and knee acceleration, driving the leg forward quicker in running at high speeds (Dorn, Schache, & Pandy, 2012).

RF and TA activation during swing also increased after WRT-speed and they were significantly higher compared to running at the same speeds, suggesting that these muscles may also play an important role to determine the WRT, in agreement with previous research (Hreljac et al., 2001; Prilutsky & Gregor, 2001; Segers et al., 2007).

In this respect, the limitation to increase speed while maintaining gait patterns may be related to the maximal speed to swing the leg forward, related to IL and RF activities. Interestingly, this mechanism is similar to the strategies to recover from a trip at early swing, in which the swing leg must be advanced rapidly to avoid falling (Forner-Cordero, Koopman, & Van der Helm, 2003; Forner-Cordero et al., 2004).

Another work that used enhanced static and dynamic optimizations, IL peak powers were found to be 10 and 3 times higher than RF peak powers, respectively

(Trinler, et al., 2018).

Our results from TA, which controls foot dorsiflexion to avoid foot scuffing during swing, showed higher activation levels when walking at high speeds, agreeing with previous studies (Hreljac et al., 2001; Segers et al., 2007). BF, that shows a peak prior to the foot contact related to decelerating the leg, also increases activity for speeds above the WRT.

In another approach to analyze the WRT, based on the Dynamic Systems Theory, the locomotion mode shifts when a non-equilibrium state is reached and a bifurcation leads the transition to another attractor. Walking and running are different attractors and the WRT occurs when the behavior gets closer to the attractor for running than for walking (Diedrich & Warren, 1995). It has been pointed out that stride frequency is the variable that captures the details of transition (Hansen et al., 2017). Our results on EMG activity present data supporting the importance of the muscles controlling leg swing, which is related to stride frequency, in agreement with this approach (Hansen et al., 2017).

Other studies reported that there may be an association between different muscles and mechanisms at different joints that could explain the changes involved with increasing speeds, leading to gait transition (Bartlett & Kram, 2008; Neptune, Sasaki, & Kautz, 2008; Pires, Lay, & Rubenson, 2018).

In this study we reported that, along with RF and TA, IL showed large increases in muscle activation as walking speeds became higher. As IL is a large and deep muscle, capable of acting on the trunk, pelvis and legs, key not only to leg swing, flexing the leg to advance the limb, but also as a potent

trunk stabilizer (Andersson et al., 1995), it was hypothesized and corroborated by our data the determinant contribution of the IL to gait transition.

Our results are consistent with previous studies that assessed IL activation during gait through both experimental (Andersson, Nilsson & Thorstensson, 1997; Cappellini et al., 2006; Gottschall & Kram, 2005; Mann, Moran, & Dougherty, 1986; Neptune, Sasaki, & Kautz, 2008) and simulated activity (Dorn, Schache, & Pandey, 2012; Souza et al., 2022; van der Krogt, Delp, & Schwartz, 2012). In this context, our results of muscular activation patterns across running and walking speeds agree with results from previous studies (Gazendam & Hof, 2007; Neptune, Sasaki, & Kautz, 2008; Prilutsky & Gregor, 2001).

We have tested muscle behavior within increases of speeds to investigate the hypothesis that IL, as a major hip flexor, is an important factor contributing to speed changes and ultimately to the WRT. Our results support this hypothesis as IL, RF and TA activations not only represent major changes between walking and running at the same speeds but also their behavior in walking above the WRT-speed was significantly higher than in running at the same speeds, showing that activation changes in these muscles may be crucial to gait transition.

Limitations to this study are that crosstalk in activation of synergists and antagonists cannot be completely ruled out. However, careful electrode placement and the noticeable difference between the EMG activity of synergistic muscles guarantee the validity of these results.

5. Practical Applications

Iliopsoas has a paramount importance clinically and functionally, as it may be involved in several abnormal orthopedic conditions of human movement, gait, daily activities and sports. Due to its location, it is a core muscle group, connecting the spine and lower limbs, acting as a primary hip flexor (Lifshitz, et al., 2020).

Invasive techniques have been used to assess iliopsoas (Andersson, Nilsson & Thorstensson, 1997; Mann, Moran, & Dougherty, 1986; Willenborg et al., 2002; Katsavrias, Primetis & Karandreas, 2005). However, intramuscular assessment may not only be painful but it also requires highly trained people for signal acquisition (Bogey & Barnes, 2017). Thus, this procedure is not suitable to measure gait within the average population nor appropriate to analyze dynamic motion such as in exercises or daily activities such as walking and running (Jiroumaru, Kurihara, & Isaka, 2014). Therefore, a non-invasive method for assessing IL is important to evaluate its contribution in locomotion control.

In pelvic rehabilitation protocols, iliopsoas is crucial since it acts as a stabilizer of the pelvis and hip. (Anderson, 2016; Lifshitz et al., 2020). Moreover, due to the overlap of anatomical parts, hip dysfunctions are often related to low back pain (Hicks, Sions, & Velasco, 2018).

Walking at high speeds and running at low speeds usually increase metabolic cost as well as peak effort of certain muscles. In this way, the motor control system may change locomotion patterns to avoid both larger individual muscle effort and larger metabolic cost of a forced gait speed. Hence, the contribution of our work was to identify

iliopsoas participation in gait transition along with other main lower limbs muscles, better understanding energy consumption and metabolic cost of both gait modes, which helps to design rehabilitation protocols that involve both gait modes, targeting these muscles work. Besides, once we know which speeds they are highly active, we can also predict and possibly adjust their response according to what we need. On top of that, walking at speeds above the WRT may result in high forces pulling from the muscle insertion points, potentially causing discomfort, pain and injury.

Hip pain can be challenging to diagnose and differentiate from other possible causes of pain, due to the anatomical complex in this area and overlapping. For example, Iliopsoas syndrome involves iliopsoas tendinopathy, iliopsoas bursitis and hip internal snapping. It can be a result of overuse, hip arthroplasty, arthritis rheumatoid and osteoarthritis, thus a variety of possible pain sources (Tyler, Fukunaga, & Gellert, 2014).

The incidence and prevalence of iliopsoas syndrome is unknown, however, specific types of athletes such as dancers, footballers and runners are more prone to develop iliopsoas pathology (Lewis, 2010; Tyler, Fukunaga, & Gellert, 2014). Physical therapy is crucial to iliopsoas syndrome treatment, with high response rate from patients with this conservative type of treatment (Lifshitz et al., 2020).

Besides, even though iliopsoas is often neglected in diagnosis processes, it should be investigated in lower back and lower limbs impairments due to its central position and core functionality, in several

daily routine activities and basic locomotion, to several sports modalities.

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Conflicts of Interest: None

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