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ORIGINAL RESEARCH

Linear and non-linear analysis of lower limb joints angle variability during running at different speeds

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Abstract

Background: Stability is one of the key demands in human locomotion including running. Various kinematical analytical approaches are adopted to investigate the running strategies; nevertheless, the impacts of running speeds on the variability of angles in individual lower limbs joints is still unclear. **Objective:** This study was aimed to investigate the impact of various running speeds on linear and non-linear variability of the hip, knee and ankle joints movement. **Methods:** Twenty-three collegiate athletes (13 females, 10 males, age 22.04 ± 3.43 years, body mass 62.14 ± 9.26 kg, height 168.29 ± 7.06 cm) ran at preferred running speed, 20% lower, and 20% higher than preferred running speed on a treadmill and their lower limbs joints kinematics were recorded using myoMOTION system at the sampling frequency of 200 Hz. The repeated measure analysis of variance test was adopted to investigate the linear (mean and standard deviation) and non-linear (Lyapunov exponent) variability of the hip, knee and ankle angle in sagittal, frontal and transverse planes throughout the running cycle. **Results:** No significant difference was observed between the lower limbs joint angles variability in linear analysis, while the Lyapunov exponent of the hip (p = .008, η_p^2 = .338), knee (p = .002, η_p^2 = .249) joints in the sagittal plane significantly increased as running speed increased. **Conclusions:** Findings of this study revealed that the hip and knee joints respond with more freedom of movement in the sagittal plane while walking speed increases, although nonlinear approaches were the only ones capable of detecting it. Given that speed changes might reduce body stability, it appears that these joints are attempting to maintain body stability by regulating internal body system perturbations by increasing their variability.

Keywords: Lyapunov exponent, kinematic variability, non-linear, running biomechanics

Introduction

Maintaining whole-body stability is the main functional strategy adopted by a neuromuscular in almost all routine tasks, particularly those that include locomotion (England & Granata, 2007). Running is associated with high demands on proper stability while various environmental and intrinsic constraints and disturbances constantly affect it (Chau et al., 2005). Fluctuations of the gait cycle during walking and running are not accidental, but they are self-similar and dependent on the long-term changes during locomotion (Jordan et al., 2009). Since a harmonized neuromuscular activity is in charge of decreasing fluctuations and enhancement of whole-body stability during running, analysis of whole-body patterning (using both non-linear and linear approaches) could provide precise information regarding the alterations in its pattern in different tasks (Noehren et al., 2014).

Of the non-linear analytical methods adopted for analysis of gait or running patterns, the largest Lyapunov (or short-term Lyapunov) exponent (LyE) is a subtle parameter in the analysis of stability during locomotion (Bruijn et al., 2013). This method could reflect any small perturbations in movement or chaotic behaviour of an individual during cyclic executions (Abbasi et al., 2019; van Schooten et al., 2014). It has been highlighted that the greater LyE values represent less stable the system (Bruijn et al., 2012). Adopting the LyE analysis, it was brought up that increment in walking speed results in a decrement in local dynamic stability (LDS; England & Granata, 2007). Stenum et al. (2014) also reported lower LDS while the pacing rate increased. More recently, Kibushi et al. (2018) revealed that LDS decreases while walking speed accelerates. Regarding the running performance, Mehdizadeh et al. (2014) reported lower LDSs while the running speeds increase, either in forward or in the backward running. Besides, it has been highlighted that either increment or decrement of walking speed results in increased nonlinear variability in lower limb joint range of motion (Kang & Dingwell, 2008). Since most studies are mainly focused on the effects of running speed on dynamic stability by investigation of nonlinear behaviour of center of mass (movement trajectories of markers or accelerometer placed on the sternum, C7 or L2 vertebras; Ekizos et al., 2017; Mehdizadeh et al.,

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2014); no explicit information exists about the strategies (i.e., changes in movement pattern at each joint) cause this irregularity in human locomotion.

From the studies conducted on the impacts of locomotion speeds on linear variability of human movement, it has been presented that the movement variability has a quadratic relationship with walking speed, where it increases by either decreasing or increasing walking velocities (Dingwell & Marin, 2006; Jordan et al., 2007). It was highlighted that movement variability is less at self-selected walking or running velocity (Jordan et al., 2007; Jordan & Newell, 2008). As the running speed increases, lower limb segments have to deal with greater applied torque magnitudes, a wider range of motions, and greater ground reaction forces (Fukuchi et al., 2017; Schache et al., 2011).

In this regard, a plethora of research studies have adopted the linear method to investigate lower limb linear variability, mostly focused on the variability of stride time, stride length, cadence, etc. (Bailey et al., 2020; Wang et al., 2021). Nevertheless, since these traditional approaches mainly focused on variables at discrete time points, they failed to identify the details regarding the changes of variability during the specific moment of running cycles (as a time series). To this end, investigating the continuous time-series could provide a detailed view of the underlying mechanisms leading to changes in running variability.

According to the literature covered above, the LDS decreases by the increment of the running speeds, while the movement variability could be increased by either increasing or decreasing locomotor speeds. This highlights the fact that changes in velocities could decrease stability, and consequently increase fall risks during walking (Bizovska et al., 2018). Thus, identification of the impacts of different running speeds on the movement variability of lower limb joints could provide more precise information about the impacts of different running speeds on individual joints. To this end, this study was aimed to investigate the effects of different running speeds on linear and non-linear lower limb joints angle variability among healthy active runners. We speculated that running at preferred running speeds would result in better stability due to faster and easier adaptation of the neuromuscular system to the preferred circumstance. Hence, it was hypothesized that lower limb joints angle variability increases by the increment of the running speed.

Methods

Study design

This study adopted a crossover study design. Independent variables were the running speeds and the dependent variables were the largest or short-term LyE (λ_S) and angle variability of the hip, knee and ankle joints in three dimensions.

Participants

Twenty-three collegiate athletes (13 females, 10 males, age 22.04 ± 3.43 years, body mass 62.14 ± 9.26 kg, height 168.29 ± 7.06 cm) have voluntarily participated in this study. A priori power analysis, using G*Power 3.1.9, indicated that a sample size of 21 would be sufficient with power ($1 - \beta$) of .85, an α of .05, and the Cohen's *d* of 0.3 (Erdfelder et al., 1996). Participants were active collegiate athletes from Faculty of Physical Education and Sport Sciences, Kharazmi University, Tehran, Iran (three training sessions per week). It has been formerly claimed that treadmill running affects the dynamics of the natural gait or running cycle (Dingwell et al., 2001). Participants of this study were familiar with treadmill running as they performed treadmill running in their training sessions. Participants had no severe injuries or surgery in lower extremities, including muscle or ligament rupture, joint laxation and bone fracture within the last 12 months of the measurement procedure. The entire test protocol was comprehensibly explained to the participants and they signed the informed consent prior to the measurement procedure. The procedures followed the 1964 Helsinki Declaration's ethical principles (and its later amendments) for human experiments. The Kharazmi University Institutional Review Board ethically approved the protocols of this study.

Experimental procedure

Prior to the warm-up and testing procedure, participants' dominant leg was identified by the ball-kicking test (van Melick et al., 2017). Thereafter, participants ran through a 10-m pathway at their preferred speed six times. Their running speed was then calculated by dividing the covered distance by time at each walking trial. Eight myoMOTION sensors (Noraxon, Scottsdale, AZ, USA) were located on the sacrum, pelvic, femurs, shanks and feet by an experienced laboratory assistant. In the measurement procedure, they randomly ran on a treadmill at one of the three following speeds: 20% slower than preferred speed (slow-speed running, SSR), preferred-speed (PSR), and 20% faster than preferred speed (fast-speed running, FSR; Nakayama et al., 2010). Running tests were conducted for 2 minutes interspersed with 5 minutes of rest intervals, and every participant started the tests randomly in one of the SSR, PSR or FSR conditions. Hence, a minimum of 50 strides for each participant was recorded and used for further data analysis (Myers et al., 2009). The 3D kinematics data of dominant side segments and joints were recorded using the myoMOTION system at the sampling frequency of 200 Hz (Abbasi et al., 2020). The validity and reliability of the Noraxon myoMOTION sensors have been already documented (Berner et al., 2020).

Data analysis

A three-dimensional skeletal avatar and streaming of anatomical joint angles, orientation angles, and segment acceleration data were reconstructed using the myoMOTION software package (Abbasi et al., 2020). Prior to data calculation, the kinematics data were filtered using a low-pass Butterworth filter (4^{th} order, zero-lag, cut-off frequency of 8 Hz). Every running cycle was identified according to the foot linear accelerations (Jasiewicz et al., 2006). The joint angles were calculated considering the relative angles between the pelvis-femur, femur-shank and the shank-foot with the positive values for flexion, internal rotation and adduction (Sarvestan et al., 2020; Struzik et al., 2015).

Angle variability

Regarding the linear analysis of the lower limbs, the mean normalized threedimensional angles of the hip, knee and ankle joints were analyzed during SSR, PSR and FSR throughout the entire running cycles. For determining the selected joints angle variability throughout the running cycle (normalized to 100 data points), the MeanSD of each joint angle in three dimensions were calculated as follows (Sarvestan et al., 2021):

$$\text{MeanSD} = \{\text{SD}(i)\}_i,\ i \in \{0\text{–}100\%\ \text{running cycle}\},$$

where SD(i) depicts the standard deviation of each value at ith% running cycle, and $\{\}_i$ portrays the average over all i.

Lyapunov exponent

As for the non-linear analysis, the λ_S of the hip, knee and ankle joints angles was calculated throughout each running cycle. At each condition, we reduced the sampling frequency to 100 Hz, and afterwards, trimmed the data to 50 strides and then normalized the data to 5000 time-points to keep the same number of strides and data points across all individuals and running circumstances (Kao et al., 2015). In order to estimate the time delays, the first minimum of the average mutual information function was computed (Fraser, 1986). Due to the fact that each participant ran at a different speed, the time delay and embedding dimension were calculated based on the individual's preferred running speed and applied to all other conditions. The median embedding time delay for the entire test was 20 samples for the kinematics (Graham & Brown, 2014). The d_E was computed from the global false nearest neighbours analysis and the d_E of 5 to 6 was chosen for further calculations (Dingwell & Cusumano, 2000; Kennel et al., 1992). Then, the state-spaces were reconstructed from all sagittal, frontal and transverse plane angles of each joint using the delay-coordinate embedding methods (Takens, 1981), as follows:

$$S(t) = [z(t), z(t+\tau), ..., z(t+(d_E-1)^*\tau)],$$

where the state vector is represented by $S(t)$, $z(t)$ is the original time-series, the time delay is presented as τ and d_E is the embedding dimension. We determined the Euclidian distances between neighbouring trajectories as a function of time after the state-space construction process. Then, the mean of the entire pairs of nearest neighbours was used to calculate the average logarithmic rate of divergence, using the following equation:

$$y(i) = 1/\Delta t\,(\ln(d_j(i))),$$

where $d_j(i)$ stands for the Euclidean distance between the pairs of nearest neighbours at i discrete time steps. Then, the calculated slope of the resulting divergence curves was considered as an estimation of the maximum finite-time LyE (Rosengren et al., 2009). The λ_S was calculated from the slope of 0 to 0.5 strides (Bruijn et al., 2009).

Statistical analysis

The normality of data distribution was checked and confirmed using the Shapiro-Wilk test (for both discrete and time-series). The repeated measure analysis of variance was used to determine significant differences between the λ_S of the hip, knee and ankle joints angles at three different running speeds. Bonferroni post-hoc test was employed to identify the significant differences between every two different running speeds in the λ_S. The partial eta squared (η_p^2) values were calculated to interpret the effect sizes. The η_p^2 of .01 was considered as a small effect size, while the η_p^2 of .06 and .14 were considered as moderate and large effect sizes, respectively (Sink & Mvududu, 2010). In order to determine the significant differences of the time series data (3D joints angle variability at different speeds), the repeated measure analysis of variance test of the spm1d package (Version 0.4.3; www.spm1d.org) was employed ($\alpha < .05$), and MATLAB software (Version 2020b; MathWorks, Natick, MA, USA) was employed to perform the entire data and statistical analyses described in data and statistical analysis sections.

Results

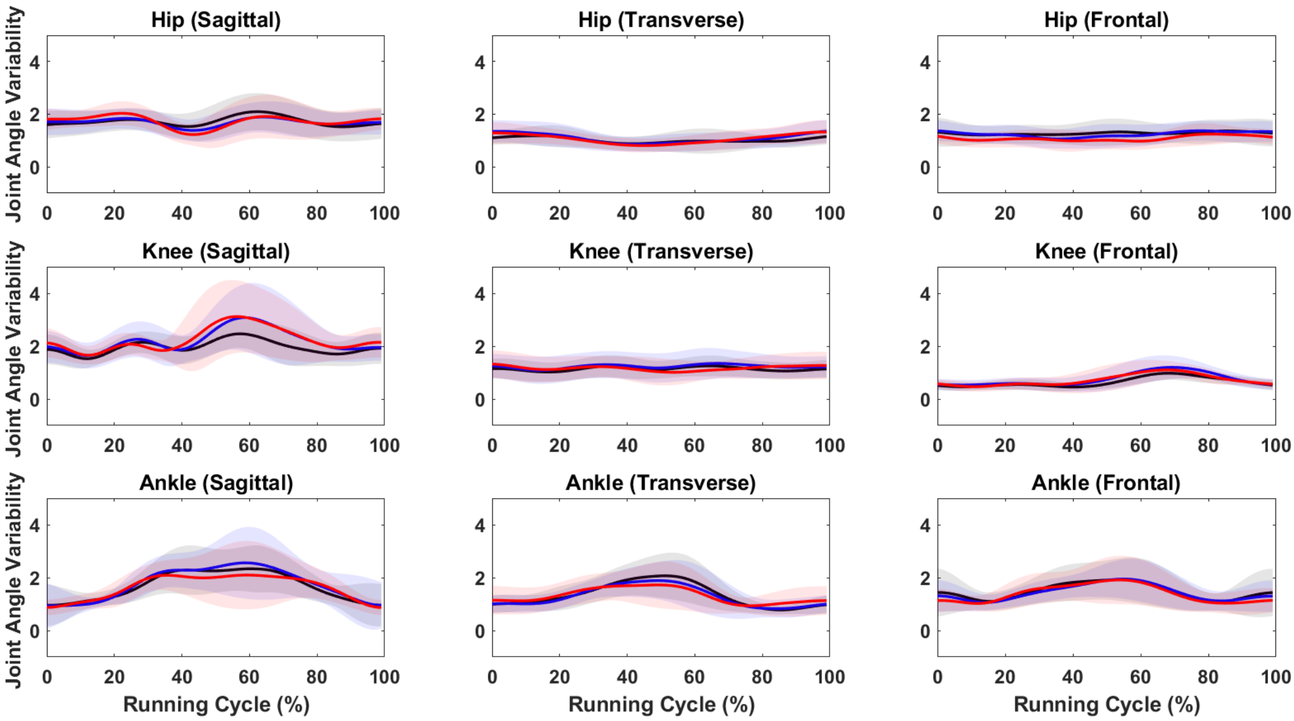
The average preferred running speed (on the ground) was 8.48 ± 0.88 km·h^{-1}. Hence, the participants were running at 6.78 ± 0.64 km·h^{-1} in their SSRs and 10.17 ± 0.72 km·h^{-1} in their FSRs. As for the hip angle variability, Figure 1 portrays no significant difference between the hip angle variability in three sagittal, frontal and transverse planes of SSRs, PSRs and FSRs. Similar to the hip joint, no significant difference was observed between the angle variability of the knee and the ankle joints at different running speeds in all three planes.

Regarding the joints angle non-linear variability, Table 1 illustrates the descriptive measures of λ_S and their differences between SSRs, PSRs and FSRs. The results demonstrated a significant effect of running speed in the λ_S of the hip joint angles (p = .002, η_p^2 = .338) and the knee joint angles (p = .008, η_p^2 = .249) in the sagittal plane. The Bonferroni post-hoc test determined significant differences between SSR, PSR and FSR in the hip joint angles in the sagittal plane, where the λ_S were greater in FSR in comparison with PSR (p = .006), and SSR (p = .015). Besides, it was portrayed that the knee sagittal angle variability was significantly greater in FSR in comparison with PSR (p = .037) and SSR (p = .009), while no significant difference was observed between PSR and SSR. In the ankle joint angles, no significant difference was observed between the λ_S values in different running speeds.

Discussion

This study was designated to investigate the effects of different running speeds on linear and non-linear variability of the lower limb joints angle among healthy active runners. We, therefore, compared the hip, knee and ankle joints angle variability (time-series analysis) and the λ_S in three different running speeds, including SSR, PSR and FSR. One finding of the study was no significant difference between the lower limb joints angles variability (using linear methods) in different running speeds, while on the other hand, the increment of running speeds resulted in

Figure 1 Differences between hip, knee and ankle sagittal, transverse and frontal angle variability in running at 80% of preferred speed (black lines), at preferred speed (blue lines), and at 120% of preferred speed (red lines).



Note. For the hip joint, the positive values represent flexion, abduction, and external rotation. In the knee joint, the positive values represent the flexion, abduction, and external rotation. As for the ankle joint, the positive values represent dorsiflexion, inversion, and external rotation.

Table 1 Descriptive measures of λS and their differences in SSRs, PSRs and FSRs

Variable	SSR	PSR	FSR	*F*	*p*	η^2_p
HIP_{FL-EX}	1.244 (0.198)[a]	1.293 (0.201)[a]	1.379 (0.184)[a]	7.433	.002	.338
HIP_{AB-AD}	1.404 (0.284)	1.383 (0.254)	1.471 (0.226)	1.831	.172	.077
HIP_{IR-ER}	1.331 (0.286)	1.350 (0.236)	1.379 (0.267)	0.360	.699	.016
KNE_{FL-EX}	0.916 (0.156)[b]	0.945 (0.169)[b]	1.045 (0.153)[b]	5.476	.008	.249
KNE_{AB-AD}	1.269 (0.266)	1.242 (0.250)	1.261 (0.183)	0.143	.867	.007
KNE_{IR-ER}	1.230 (0.244)	1.236 (0.215)	1.321 (0.249)	2.291	.113	.104
ANK_{DF-PF}	1.265 (0.173)	1.293 (0.229)	1.302 (0.236)	0.440	.647	.020
ANK_{IN-EV}	1.426 (0.197)	1.442 (0.257)	1.479 (0.302)	0.315	.732	.014
ANK_{IR-ER}	1.252 (0.152)	1.268 (0.186)	1.286 (0.239)	0.194	.824	.009

Note. SSR = slow-speed running; PSR = preferred-speed running; FSR = fast-speed running; KNE = knee; ANK = ankle; FL-EX = flexion-extension; AB-AD = abduction-adduction; IR-ER = internal rotation-external rotation; DF-PF = dorsal flexion-plantar flexion; IN-EV = inversion-eversion. [a]Significantly different ($p < .05$) between all three running speeds using Bonferroni post-hoc test. [b]Significantly different ($p < .05$) between FSR with PSR and SSR using Bonferroni post-hoc test.

significantly greater λ_S values in the hip and the knee joints in the sagittal plane.

Achieving whole-body stability during running tasks is one of the relatively simple challenges for the neuromuscular system (Rossignol et al., 2006). Narrower base-of-support (compared with walking), longer swing phase duration, greater muscle contractions in a coordinated pattern, higher locomotion velocities, and consequently shorter response times to the internal and external perturbations are the main challenges that individuals are dealing with during the running performance (England & Granata, 2007; Mehdizadeh et al., 2014). A combination of all mentioned factors, in turn, could result in greater values of divergence of kinematic trajectories in the reconstructed state-space, and due to which, greater LyE could be observed in whole-body stability status (Mehdizadeh et al., 2014). In a study focused on the impacts of different walking speeds on whole-body stability, Dingwell and Marin (2006) revealed that walking around at comfortable walking speed is more stable than walking at either lower or higher walking speeds. On the contrary, England and Granata (2007) portrayed a monotonic trend between the lower limbs joints variability and the walking velocity, where increment in walking speed resulted in lower stability. Likewise, outcomes of this study revealed significantly greater variability in the hip and the knee angles in the sagittal plane when the running speeds increased.

The values of λ_S considerably increased in the sagittal plane with each degree of running speed increment, indicating greater hip and knee joint angle variabilities during low-speed running. There could be two different reasons for this: first, by decreasing the running speed, the patterning of the stance phase would change so that longer foot-ground contact duration occurs (England & Granata, 2007; Mehdizadeh, 2018). This increment in the ground contact phase could provide the lower limbs neuromuscular with sufficient time to counterbalance any external perturbations by the increment of sensory integration (i.e. applied forces, surface changes, and cognitive conditions; Schniepp et al., 2012). Therefore, it seems by the increment of running speeds, the stance phase duration and the base of support decreases, the body-weight loading rate on one leg increases (at each step), the neuromuscular cannot properly and sufficiently respond to the external perturbations, and consequently, the whole-body stability could decrease. Nevertheless, since no falling was recorded during running at different speeds, we cannot firmly claim that the increment in λ_S values of the hip and the knee joints is associated with poorer whole-body stability during running.

To this effect, as the secondary explanation, it could be interpreted that higher hip and knee angle variability aids the whole-body stability. Although previous studies linked higher values of LyE to poorer whole-body stability, the same inference for individual joints may be deceptive. Hence, it could be claimed that the hip and the knee joints, as unique sub-systems working in a greater system, constantly adopt a different range of angles (at each running cycle) in response to external perturbations to maintain the whole-body stability at higher running speeds. Furthermore, no significant differences between LyE values of ankle joints could be associated with the location of this joint. Forasmuch as the ankle joint is the closest joint to the ground, and since the structure of this joint is less stable than the knee and the hip joints, higher angle variability could end up with a weaker basement for planting the body structure, and consequently, end up in falling (Hamill & Knutzen, 2006). Accordingly, the ankle joint structure stiffness remains constant to provide a safe cornerstone for whole-body structure while the running speed increases.

An increment of walking speed (from 80% to 120% of preferred speed) significantly changes the kinematics of the gait cycle (Dingwell & Marin, 2006; Jordan et al., 2007). The outcomes of this study portrayed that neither decrement nor increment in running speeds significantly changed the lower limb joints angle linear variability. One main reason for the lack of differences could be high values of standard deviations between participants. As Figure 1 depicts, the standard deviations of angle variability are relatively high in each speed, which highlights high inter-subject variability. This might explain that this linear method of calculation of movement variability in running could not reveal differences with high precision since it is dependent on standard deviation values. It is also noteworthy to mention that the standard deviation values in LyE were relatively small in comparison to the joint angle variability, which depicts that within-participants differences had no impact on it.

This study, nevertheless, was included with limitations. Only three different running speeds were sampled. We speculate that higher running velocity (> 120% of normal running speed) could result in significantly higher angle variability and higher LyE values in lower limbs joints. To this effect, we recommend further studies to investigate the impacts of higher ranges of running velocities on linear and non-linear variability of individual joint angles. Further, given that we aimed to investigate the impacts of running speed on individual joint variability, whole-body stability was not measured in this study. To that end, we suggest that further research be done to create a connection between whole-body stability and individual joint variability.

Conclusions

This study demonstrated that the hip and knee joints non-linear variability in the sagittal plane is considerably higher when the running speed increases. Due to the nature of high speed running, the whole-body stability could decrease (i.e., narrower base-of-support or relatively shorter adaptation time). To this end, the central nervous system, in response to the applied perturbations, increases the hip and the knee joints angle variability (due to greater muscles acting around these joints and wider range of motion) to regulate the whole-body structure and decrease fall risks.

These outcomes indicate that the ankle joint, as the closest joint (distal joint) to the ground contact, had relatively lower adaptations to the changes in the running speed in order to provide the structure with a firm base to maintain dynamic balance. Furthermore, decrements in the LyE values of the hip and knee joints in the sagittal plane due to the transition from lower to higher running speeds highlight the fact that these joints constantly try to regulate the whole-body dynamic balance by increasing angle variability.

Conflict of interest

The authors report no conflict of interest.

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