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

Peyman Aghaie Ataabadi1, Javad Sarvestan2, Fatemeh Alaei1, Fateme Yazdanbakhsh1, and Ali Abbasi1

1Department of Biomechanics and Sport Injuries, Faculty of Physical Education and Sport Sciences, Kharazmi University, Tehran, Iran; and 2Department of Natural Sciences in Kinanthropology, Faculty of Physical Culture, Palacky University Olomouc, Olomouc, Czech Republic

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, η² = .338), knee (p = .002, η² = .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 (Nochren 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, Meh dizadeh 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 vertebrae; Ekizos et al., 2017; Meh dizadeh et al., 2013).
By P. Aghaie Ataabadi et al.

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) 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 – β) 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 (4th order, zero-lag, cut-off frequency of 8 Hz). Every running cycle was identified according to the foot linear accelerations (Jasiwicz 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 (Jarvestan et al., 2020; Struzik et al., 2015).
Angle variability

Regarding the linear analysis of the lower limbs, the mean normalized three-dimensional 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):

\[
\text{MeanSD} = \{\text{SD}(i)\}, \quad i \in \{0--100\% \text{ running cycle}\},
\]

where SD(i) depicts the standard deviation of each value at i-th% running cycle, and || portrays the average over all i.

Lyapunov exponent

As for the non-linear analysis, the \( \lambda_3 \) 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:

\[
S(t) = [x(t), x(t + \tau), ..., x(t + (d_e - 1)\tau)],
\]

where the state vector is represented by \( S(t) \), \( x(t) \) is the original time-series, the time delay is presented as \( \tau \) 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:

\[
y(i) = 1/\Delta t \times (\ln(d_j(i))),
\]

where \( d(j) \) stands for the Euclidian 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 \( \lambda_3 \) 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 \( \lambda_j \) 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 \( \lambda_j \). The partial eta squared (\( \eta^2 \)) values were calculated to interpret the effect sizes. The \( \eta^2 \) of .01 was considered as a small effect size, while the \( \eta^2 \) 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 \( \lambda_3 \) and their differences between SSRs, PSRs and FSRs. The results demonstrated a significant effect of running speed in the \( \lambda_3 \) of the hip joint angles (\( p = .002, \eta^2_p = .338 \)) and the knee joint angles (\( p = .008, \eta^2_p = .249 \)) 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 \( \lambda_3 \) 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 \( \lambda_3 \) 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 \( \lambda_3 \) 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
significantly greater $\lambda_S$ 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 $\lambda_S$ 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 pattern of the stance phase would change so that lower 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 $\lambda_e$ 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.

References

Abbasi, A., Yazdanibakhsh, F., Taqji, M. K., Aghaei Ataabadi, P., Svboda, Z., Nazarpour, K., & Vieira, M. F. (2020). A comparison of coordination and its variability in lower extremity segments during treadmill and overground running at different speeds. Gait and Posture, 79, 139–144. https://doi.org/10.1016/j.gaitpost.2020.04.022

Abbasi, A., Zamanian, M., & Svboda, Z. (2019). Nonlinear approach to study the acute effects of static and dynamic stretching on local dynamic stability in lower extremity joint kinematics and muscular activity during pedalling. Human Movement Science, 66, 440-448. https://doi.org/10.1016/j.humov.2019.06.025

Bailey, C. A., Porta, M., Piloni, G., Aripoa, F., Cockett, J., & Pau, M. (2020). Does variability in motor output at individual joints predict stride time variability in gait? Interventions of age, sex, and plane of motion. Journal of Biomechanics, 99, Article 109574. https://doi.org/10.1016/j.jbiomech.2019.109574

Berner, K., Cockcroft, J., Morris, L. D., & Louw, Q. (2020). Concurrent validity and within-session reliability of gait kinematics measured using an inertial motion capture system with repeated calibration. Journal of Bodywork and Movement Therapies, 24(4), 251–260. https://doi.org/10.1016/j.jbmt.2019.109574

Bivozovska, L., Svboda, Z., Janura, M., Bisi, M. C., & Vullerme, N. (2018). Local dynamic stability during gait for predicting falls in elderly people: A one-year prospective study. PLoS ONE, 13(5), Article e0197091. https://doi.org/10.1371/journal.pone.0197091

Bruij, S. M., Bregman, D. J., Meijer, O. G., Beek, P. J., & van Dieen, J. H. (2012). Maximum Lyapunov exponents as predictors of global gait stability: A modelling approach. Medical Engineering & Physics, 34(4), 428–436. https://doi.org/10.1016/j.medengphy.2011.07.024

Bruij, S. M., Meijer, O., Beek, P., & van Dieen, J. H. (2013). Assessing the stability of human locomotion: A review of current measures. Journal of the Royal Society Interface, 10(83), Article 20120999. https://doi.org/10.1098/rsif.2012.0999
Dingwell, J. B., & Marin, L. C. (2006). Kinematic variability and local dynamic stability versus kinematic variability of continuous overground and treadmill walking. Journal of Biomechanical Engineering, 128(1), 27–32. https://doi.org/10.1115/1.1367298

Dingwell, J. B., & Marin, L. C. (2006). Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. Journal of Biomechanics, 39(3), 444–452. https://doi.org/10.1016/j.jbiomech.2004.12.014

Ekizios, A., Sautze, A., & Arampatzis, A. (2017). Transition from shod to barefoot alters dynamic stability during running. Gait and Posture, 56, 31–36. https://doi.org/10.1016/j.gaitpost.2017.04.055

England, S. A., & Granata, K. P. (2007). The influence of gait speed on local dynamic stability of walking. Gait and Posture, 25(2), 172–178. https://doi.org/10.1016/j.gaitpost.2006.03.011

Erdfelder, E., Faul, F., & Buchner, A. (1996). GPOWER: A general power analysis program. Behavior Research Methods, Instruments, & Computers, 28(1), 1–11. https://doi.org/10.3758/BF03203630

Fraser, A. (1986). Using mutual information to estimate metric entropy. In G. Mayer-Kress (Ed.), Dimensions and entropies in chaotic systems (pp. 82–91). Springer. https://doi.org/10.1007/978-3-642-84701-9_11

Fukuchi, R. K., Fukuchi, C. A., & Duarte, M. (2017). A public dataset of running biomechanics and the effects of running speed on lower extremity kinematics and kinetics. PeerJ, 5, e3296. https://doi.org/10.7717/peerj.3296

Graham, R. B., & Brown, S. H. (2014). Local dynamic stability of spine muscle activation and stiffness patterns during repetitive lifting. Journal of Biomechanical Engineering, 136(12), Article 121006. https://doi.org/10.1115/1.4028818

Hamill, J., & Knutzen, K. M. (2006). Biomechanical basis of human movement (2nd ed.). Lippincott Williams & Wilkins.

Jasiewicz, J. M., Allum, J. H., Middleton, J. W., Barriskill, A., Condle, P., Purcell, B., & Li, C. T. (2006). Gait event detection using linear accelerometers or angular velocity transducers in able-bodied and spinal-cord injured individuals. Gait and Posture, 24(4), 502–509. https://doi.org/10.1016/j.gaitpost.2005.12.007

Jordan, K., Challis, J. H., Cusumano, J. P., & Newell, K. M. (2009). Stability and the time-dependent structure of gait variability in walking and running. Human Movement Science, 28(1), 113–128. https://doi.org/10.1016/j.humov.2009.09.001

Jordan, K., Challis, J. H., & Newell, K. M. (2007). Walking speed influences on gait cycle variability. Gait and Posture, 26(1), 128–134. https://doi.org/10.1016/j.gaitpost.2006.08.010

Kang, H. G., & Dingwell, J. B. (2008). Effects of walking speed, strength and range of motion on gait stability in healthy older adults. Journal of Biomechanics, 41(14), 2899–2905. https://doi.org/10.1016/j.jbiomech.2008.08.002

Kao, P.-C., Higgison, C. I., Seymour, K., Kadermez, M., & Higgison, J. S. (2015). Walking stability during cell phone use in healthy adults. Gait and Posture, 43(4), 947–953. https://doi.org/10.1016/j.gaitpost.2015.03.047

Kempen, I. G., & van Schooten, K. S., Rispens, S. M., Elders, P. J., van Dieën, J. H., & Pijnappels, M. (2009). Differences in gait complexity and variability between children with and without developmental coordination disorder. Gait and Posture, 29(2), 225–229. https://doi.org/10.1016/j.gaitpost.2008.08.005

Kirk, R., & Dunn, M. (2000). The largest Lyapunov exponent of gait in young and elderly individuals: A systematic review. Gait and Posture, 60, 242–250. https://doi.org/10.1016/j.gaitpost.2017.12.014

Koistinen, M., & Newell, K. M. (2009). Gait variability is altered in patients with peripheral arterial disease. Journal of Vascular Surgery, 49(4), 924–931. https://doi.org/10.1016/j.jvs.2008.11.020

Koizumi, Y., Kudo, K., & Ohotsuki, T. (2010). Variability and fluctuation in running gait cycle of trained runners and non-runners. Gait and Posture, 31(3), 331–335. https://doi.org/10.1016/j.gaitpost.2009.12.003

Kojo, O., Schmitz, A., Hempel, R., Westlake, C., & Black, W. (2014). Assessment of strength, flexibility, and running mechanics in men with iliotibial band syndrome. Journal of Orthopaedic and Sports Physical Therapy, 44(3), 217–222. https://doi.org/10.2519/jospt.2014.4911

Kooy, P. W., van der Valk, W., & van Cingel, R. E. (2017). How to determine leg dominance: The agreement between self-reported and observed performance in healthy adults. Gait and Posture, 56, 31–36. https://doi.org/10.1016/j.gaitpost.2017.02.003

Kromer, F., Molyneaux, J., & Newell, K. M. (2007). The impact of cell phone use on walking stability and invariable orbital stability in the activation of muscle synergies in locomotion. Physiological Reviews, 86(1), 89–154. https://doi.org/10.1152/physrev.00028.2005

Kronilas, M., Krajewski, T., Bubel, J., Staszek, M., Nepomuceno, T., & Lemos, N. (2013). Local dynamic stability of walking during cell phone use in healthy adults. Gait and Posture, 37(2), 225–229. https://doi.org/10.1016/j.gaitpost.2012.07.003

Kronilas, M., Krajewski, T., Bubel, J., Staszek, M., Nepomuceno, T., & Lemos, N. (2013). Local dynamic stability of walking during cell phone use in healthy adults. Gait and Posture, 37(2), 225–229. https://doi.org/10.1016/j.gaitpost.2012.07.003

Kronilas, M., Krajewski, T., Bubel, J., Staszek, M., Nepomuceno, T., & Lemos, N. (2013). Local dynamic stability of walking during cell phone use in healthy adults. Gait and Posture, 37(2), 225–229. https://doi.org/10.1016/j.gaitpost.2012.07.003

Kronilas, M., Krajewski, T., Bubel, J., Staszek, M., Nepomuceno, T., & Lemos, N. (2013). Local dynamic stability of walking during cell phone use in healthy adults. Gait and Posture, 37(2), 225–229. https://doi.org/10.1016/j.gaitpost.2012.07.003

Kronilas, M., Krajewski, T., Bubel, J., Staszek, M., Nepomuceno, T., & Lemos, N. (2013). Local dynamic stability of walking during cell phone use in healthy adults. Gait and Posture, 37(2), 225–229. https://doi.org/10.1016/j.gaitpost.2012.07.003

Kronilas, M., Krajewski, T., Bubel, J., Staszek, M., Nepomuceno, T., & Lemos, N. (2013). Local dynamic stability of walking during cell phone use in healthy adults. Gait and Posture, 37(2), 225–229. https://doi.org/10.1016/j.gaitpost.2012.07.003