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Linear and nonlinear measures of gait variability after anterior cruciate ligament reconstruction

Eduardo Afonso de Oliveira[a], Adriano O. Andrade[b], Marcus Fraga Vieira[a,*]

[a] Bioengineering and Biomechanics Laboratory, Federal University of Goiás, Goiânia, Goiás, Brazil
[b] Centre for Innovation and Technology Assessment in Health (NIATS), Faculty of Electrical Engineering, Federal University of Uberlândia, Uberlândia, Brazil

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ABSTRACT

The objective of this study was to assess gait variability after anterior cruciate ligament reconstruction (ACLR), as an indicative of possible altered gait pattern and a measure of recovery compared to control subjects. Forty subjects (32 male), divided into 4 groups of 10 participants, were enrolled in the study: a control group (CG), and observational groups OG-I (90 days), OG-II (180 days), and OG-III (360 days) after ACLR. All subjects underwent the same rehabilitation program for six months. For kinematic recording, each subject walked on a treadmill for 4 min at a preferred walking speed. Linear gait variability was assessed using average standard deviation (VAR) and normalized root mean square of medial–lateral (ML) trunk acceleration (RMS_{ratio}). Gait stability was assessed using the margin of stability (MoS) and local dynamic stability (LDS), and nonlinear variability was assessed using sample entropy (SEn). Compared to the CG, the VAR ML increased significantly in the OG-I group and decreased incrementally in OG-II and OG-III. MoS increased significantly in the OG-I group and tends to maintain in OG-II and OG-III, while LDS was greater in the CG and decreased incrementally in the OG groups. The SEn was higher in the OG groups than in the CG and increased in OG-II and OG-III. The results indicated that ACL reconstruction was followed by a progressive increase in stability and a progressive increase in variability over the postoperative rehabilitation period. In terms of stability and gait variability, six months of physiotherapy for rehabilitation after ACL reconstruction appears to be effective, but it is insufficient for a complete recovery as compared to healthy individuals.

1. Introduction

The gait pattern changes in patients after anterior cruciate ligament (ACL) reconstruction is relatively well established (Gao and Zheng, 2010; Hooper et al., 2002; Webster and Feller, 2011). The reconstruction technique itself induces changes in muscular function that in turn alter the gait pattern, so that most of the rehabilitation program is focused on muscular function recovery. Using motion capture and electromyography (EMG) from 16 lower limb muscles of individuals that undergone ACL reconstruction using hamstring autograft, Konrath et al. (2017) showed that semitendinosus and gracilis muscles exhibit altered muscle-tendon properties, resulting in reduced contribution to medial compartment contact loads, which is partially compensated by activity of semimembranosus muscle (Konrath et al., 2017). EMG analysis from bilateral hamstring muscles of individuals from 1 to 6 years after ACL reconstruction using hamstring autograft, during exercises involving eccentric knee flexor activity, revealed that such individuals may benefit from a rehabilitation program involving specifically strengthening of the medial hamstring muscles (Árnason et al., 2014). Such individuals exhibit inter-limb differences in hamstring activation patterns, a factor to be considered during their rehabilitation (Briem et al., 2016). In individuals with ACL rupture, EMG from lower limb muscles during a unilateral stance task has evidenced that some individuals present abnormal hamstring recruitment that may contribute to knee instability (non-coper individuals), whereas other individuals present normal tibial position that allows quadriceps activation without excessive anterior tibial translation (coper individuals) (Chmielewski et al., 2005).

Additionally, although restoration of nearly normal anterior translation of the knee occurs after ACL reconstruction, some residual impairments could remain, including pain, swelling or stiffness (Hooper et al., 2002), affecting gait variability (Moraiti et al., 2009). Furthermore, due to abnormal cartilage loading distribution and abnormal tibiofemoral kinematics, a long-term consequence of ACL reconstruction

* Corresponding author at: Bioengineering and Biomechanics Laboratory, Federal University of Goiás, Avenue Esperança, Campus Samambaia, 74690-900 Goiânia, Goiás, Brazil.
E-mail address: marcus@ufg.br (M.F. Vieira).

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Table 1
Rehabilitation Program followed by ACL reconstructed patients.

Time after ACL reconstruction	Exercises
2nd to 5th day	Passive exercises to increase knee AOM, knee active flexion/extension, isometric contractions of quadriceps muscle.
1st week	Aerobic exercise on a level bike without load. Focus on knee extension.
2nd week	Focus on knee muscle strength: active-assisted exercises in all planes. Focus on complete knee extension
3rd week	Transition to mechanotherapy, using some strength training apparatus: high number of repetitions with low load
4th week	Mechanotherapy with a low/moderate load
5th to 12th week	Focus on progressive increase of muscle resistance and, in minor extension, strength, with progressive load increase, with the same number of repetitions.
12th week	Transition from resistance to strength muscle training: reduction of the number of repetitions with increased load. Increased aerobic exercise on a bike
12th to 24th week	Focus on muscle strength with progressive increase of load
16th week	Addition of proprioceptive exercises, and running in straight line only
20th week	More intense proprioceptive and plyometric exercises.
24th week	If the muscle strength of the operated side is equal or higher than 80% of the non-operated side, and the functional tests were well executed, the patient is discharged from physiotherapy

is the development of knee osteoarthritis (Lohmander et al., 2007; Vignos et al., 2018). In addition to biomechanical alterations, the absence of proprioceptive input from the reconstructed ACL alters the function of neuromuscular system, raising, therefore, the necessity to assess the postoperative period.

Studies that have assessed linear or nonlinear measures of gait variability after ACL reconstruction are scarce, limited to only one specific postoperative period or extracting only one measure, focusing on gait stride-to-stride variability (Moraiti et al., 2009), knee joint flexion-extension time series (Georgoulis et al., 2006), or accelerometer data (Tsivgoulis et al., 2011). However, none of the cited studies have focused on medial-lateral trunk variability. As the most of body mass is located in upper body, maintaining stability of the trunk is critical for human locomotion, allowing a proper visual inflow, a more effective vestibular input (MacKinnon and Winter, 1993; Prince et al., 1994), and a consequent control of balance. In addition, anterior-posterior (AP) perturbations can be stabilized by means of passive dynamics of musculoskeletal system, whereas medial–lateral (ML) perturbations not: they require active intervention by neuromotor strategies, so that motor control in frontal plane seems to play an important role in gait balance and stability (Allet et al., 2012). Furthermore, trunk data have produced more reliable estimation of gait stability (Hamacher et al., 2015). Therefore, this study has focused in linear and nonlinear measures of gait variability extracted from ML trunk kinematics in patients after ACL reconstruction.

Gait variability is influenced by the ability to control gait optimally from one stride to the next. In biomechanics, variability refers to the ability of the motor system to perform a wide variety of tasks under environmental constraints (Stergiou and Decker, 2011). Variability is quantified using measures derived from linear statistics, such as the standard deviation of a time series, or using measures derived from nonlinear dynamics, such as entropy and local dynamic stability (LDS), estimated by maximum Lyapunov exponent. The variations in human movement are distinguishable from noise; they have a deterministic origin, being neither random nor independent. The variations display a temporal structure, captured by nonlinear tools. In this approach, the manner in which gait evolves over time becomes important (Stergiou and Decker, 2011). Therefore, linear and nonlinear measures of variability express different aspects of a time series: the magnitude of the variations presented in a time series, an amplitude variability, and the structure/organization of these variations, a temporal variability, respectively.

For optimal operation of the motor system, a moderate amount of variability is expected as a necessary condition for adaptability, since a system with a total absence of variability will lack the repertoire to negotiate even subtle perturbations (van Emmerik et al., 2016). In contrast, a large amount of variability may be detrimental to the motor control system; greater variability can mean that it is incapable of accurately achieving the desired target. Thus, higher levels of variability are evidence of poor task performance (van Emmerik et al., 2016). In this sense, entropy was higher two years after ACL reconstruction (Moraiti et al., 2009), whereas LDS was lower (higher maximum Lyapunov exponent) in patients with neurological disorders when compared to healthy individuals (Reynard et al., 2014); it decreases from 40 to 50 years old (Terrier and Reynard, 2015); it was lower after general fatigue (Vieira et al., 2016).

Therefore, linear and nonlinear measures of gait variability were estimated to assess how gait variability of patients after ACL reconstruction behaves during three periods of the postoperative phase, and whether it would be an indicative of progressive recovery of normal biomechanical levels. We hypothesize that gait variability increases immediately after surgery, and, gradually, return to the control values.

2. Methods

2.1. Subjects

The study included 40 healthy non-athletes participants (32 males, 8 females; 26.83 ± 3.68 years; 72.5 ± 2.62 kg; 1.73 ± 0.05 m). They were organized into four different groups with 10 subjects (8 males and 2 females): control group (CG, with no ACL reconstruction), observational group I (OG-I, 90 days after ACLR), observational group II (OG-II, 180 days after ACL reconstruction), and observational group III (OG-III, 360 days after ACL reconstruction). Results from a 2-year follow-up group has been previously reported (Moraiti et al., 2009).

The participants of the observational groups were enrolled in the same hospital. They were submitted to unilateral ACL reconstruction with the same technique (autologous hamstring autograft with the semitendinosus tendon accompanied by the gracilis tendon) (Shaerf et al., 2014) by the same surgeon, within 2 months of rupture, and underwent the same rehabilitation program (Table 1), from the second/fifth day to the 24th week after surgery, when the participant was discharged from the rehabilitation program. Exclusion criteria were the presence of any impairment in the non-operated knee, and the presence of any post-operative complication.

All participants were instructed to avoid any type of exercise for 24 h prior to the experiment. After written informed consent has been obtained, the participants were submitted to the protocols previously approved by the local Research Ethics Committee.

2.2. Protocols

Each participant had 39 reflective markers attached to his or her body, in accordance with the Vicon Full Body Plug-In Gait, and one additional marker placed on spinous process of first thoracic vertebrae (T1 marker). The markers were used for movement registration with a

3-D motion-capture system that used 10 infrared cameras operating at 100 samples/s (Vicon Nexus; Oxford Metrics; Oxford, UK).

Each participant walked on a level treadmill at their preferred walking speed (PWS) (Devita et al., 1997) for 8 min, including 3 min for warm-up, 4 min for data collection, and a final minute at gradually reducing speed. The PWS was evaluated following a previously reported protocol (Dingwell and Marin, 2006).

2.3. Data analysis

Before data analysis, all kinematic data, except for calculation of LDS and SEn, were low-pass filtered with a fourth-order, zero-lag Butterworth filter with a cut-off frequency of 12 Hz. Some authors do not recommend filtering the signal for nonlinear features calculation (LDS and SEn) (Mees and Judd, 1993; Stergiou et al., 2004). All parameters were calculated for 150 strides. First, the initial and final 15 s of data collection interval (4 min) were discarded (Hak et al., 2013). Next, all steps were detected as the zero-cross of the heel markers' velocity (Souza et al., 2017). Then, 150 contiguous strides were selected by discarding strides from both the beginning and the end until 150 contiguous strides were retained. Data analysis used a customized Matlab code.

2.3.1. Linear gait variability

Two descriptors were used to evaluate linear gait variability: VAR ML (Dingwell and Marin, 2006) and RMS_{ratio} (Terrier and Reynard, 2015). To compute VAR, each stride was time normalized (0–100%). At each of the 101 normalized time points, the SD of the ML trunk acceleration (T1 marker) over the 150 strides was calculated. Next, the average SD of these 101 standard deviations was calculated (Dingwell and Marin, 2006).

To compute RMS_{ratio} of trunk acceleration we first calculate the vector norm of the 3D trunk acceleration (Ta) for each sample n (Eq. (1)),

$$Ta_n = \sqrt{ax_n^2 + ay_n^2 + az_n^2} \quad (1)$$

where (ax, ay, az) is the 3D trunk acceleration. Next, the RMS of Ta is calculated as follows:

$$Ta_{RMS} = \sqrt{\frac{1}{N}\sum_{n=1}^{N}(Ta_n)^2} \quad (2)$$

The same procedure is used to calculate ML_{RMS}. The normalized RMS is $RMS_{ratio} = ML_{RMS}/Ta_{RMS}$, which quantifies the proportion of trunk acceleration variability that occurs in the ML direction compared to the total acceleration variability (Terrier and Reynard, 2015).

2.3.2. Gait stability

Gait stability was assessed using MoS (Hof et al., 2005). The MoS was calculated using the adapted method proposed by Hak et al. (2013). The center of mass (CoM) was estimated from the full-body plug-in gait markers, and the maximal height of the estimated CoM was used as the length of the inverted pendulum (l). The extrapolated center of mass (XCoM) was calculated as the CoM plus its velocity multiplied by the factor $\sqrt{l/g}$, where g is the gravity acceleration. The MoS was calculated as the distance between the XCoM and the lateral malleolus marker (estimated border of support base) of the leading foot for ML MoS. The minimum value of the MoS within each step was averaged over the 150 strides. The closer the extrapolated center of mass to the border of the base of support, the more likely to occur a loss of balance. Thus, a decrease in MoS is detrimental to stability.

2.3.3. Local dynamic stability

LDS evaluation was based on the maximum Lyapunov exponent (λ) estimation using Rosenstein's algorithm (Rosenstein et al., 1993). Briefly, the ML velocity of the T1 marker was calculated from the raw T1 marker data using the 3-point method (Robertson et al., 2014). Next, the velocity signal was time-normalized to 15,000 samples, preserving differences in stride time between strides (Bruijn et al., 2013). A high-dimension attractor was constructed using the normalized T1 marker velocity and its delayed copies. A delay of 34 samples was used based on the mean value of the minimum of the mutual information function across all data, and an average dimensionality of 5 across all participants was found to be sufficient based on global false-nearest-neighbor analysis. For each point in state-space, a nearest neighbor was found, and the Euclidean distance between these points was tracked for 10 strides, resulting in as many time-distance curves as time points in state-space. The divergence curve was calculated as the mean of the natural log of the time-distance curves. The λ was calculated as the slope of a linear fit to the first 50 samples (average time needed for 0.5 stride or 1 step) of the divergence curve. Thus, λ indicates the relative rate of divergence over 0.5 stride that results from a small difference in initial conditions.

The concept of LDS assumes that the motor control system ensures a dynamically more stable gait if the divergence exponent, λ, remains lower between trajectories in a reconstructed state space that reflects gait dynamics over a period of 0.5 strides. Thus, an increase in λ implies a decrease of LDS. Unlike traditional measures of gait variability that treat each gait cycle as independent, the LDS evaluates the evolution of stability over the course of several strides.

2.3.4. Gait regularity

The temporal structure of variability was assessed using SEn (Ramdani et al., 2009). SEn was calculated to quantify the amount of regularity and the predictability of variations over time regarding ML T1 marker velocity time series. SEn is the negative natural logarithm of the conditional probability that two m-dimensional delayed vectors that are close within a tolerance r, remain close in the $(m + 1)$-dimensional state space, without allowing self-matches (Ramdani et al., 2009). Parameter values of m = 2 and r = 0.2 were selected based on previous studies (Vieira et al., 2017a; Vieira et al., 2017b). SEn reflects the likelihood that a pattern of similar observations will not be followed by additional similar observations. A time series containing many repetitive patterns, (i.e., one that is more predictable), has a relatively small SEn, whereas a less predictable process has a higher SEn and smaller regularity.

Entropy can be viewed as a measure of system adaptability and regularity, which refers to the presence of nonrandom fluctuation on time in the apparently irregular dynamics of a system (Manor et al., 2010). Entropy has been used in a variety of studies in biomechanics, to quantify the dynamical structure of human postural sway (Ramdani et al., 2009), the time and frequency structure of force output in adult humans (Vaillancourt et al., 2004), and the stride-to-stride fluctuations in gait (Georgoulis et al., 2006). A movement with high regularity (low entropy) or a very random movement (high entropy) would reveal poor performance, inferring the motor control process as either unyielding or too imprecise, respectively (Stergiou and Decker, 2011).

2.4. Statistical analysis

Analysis of variance (one-way ANOVA) was applied to assess intergroup differences, followed by Tukey post-hoc tests for all variables (normal distribution Shapiro-Wilk test, $p > 0.05$, Levene's homogeneity test $p > 0.05$). A Pearson correlation analysis was performed to check a possible relationship between the variables. The correlation coefficient was classified according to: 0.00–0.29 as very weak, 0.30–0.49 as weak, 0.50–0.69 as moderate, 0.70–0.89 as strong, and 0.90–1.00 as very strong (Hinkle et al., 2003). The statistical analysis was performed using SPSS software version 17 (SPSS Inc.; Chicago, IL), with a significance level set at $p < 0.05$.

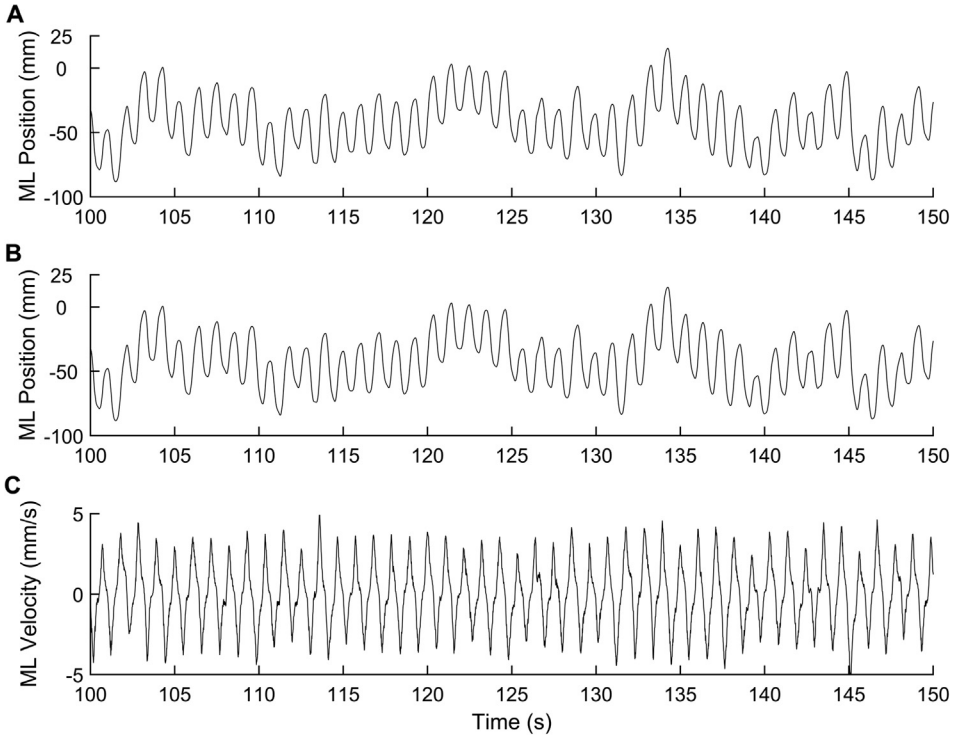

Fig. 1. Typical ML trunk kinematic signals. A: raw signal. B: filtered signal. C: derived velocity signal.

3. Results

Typical raw and processed ML trunk kinematic signals are presented in Fig. 1. Note that trunk velocity signal (Fig. 1-C) presents reduced non-stationarities compared to position trunk signal (Fig. 1-A), a requirement for non-linear analytical techniques applied here (Dingwell and Marin, 2006).

Table 2 summarizes the results related to the extracted linear and nonlinear measures of gait variability. Although not significant, OG-I group walked slightly slower than the other groups.

VAR ML increased in the OG-I group compared to the control group (CG) and decreased in groups OG-II and OG-III compared to OG-I group (Fig. 2). Thus, the group 90 days post-ACL reconstruction presented the highest linear variability.

RMS_{ratio} increased significantly for all OG groups compared to the CG, with OG-I, the group 90 days post-ACL reconstruction, having the highest value (Fig. 3).

The λ decreased significantly for all OG groups compared to the CG (Fig. 4). Hence, the local dynamic stability increased after ACL reconstruction. The MoS increased significantly in OG-I and tended to maintain in OG-II and OG-III (Fig. 5), also an indicative of increased stability after ACL reconstruction.

SEn significantly increased for OG-II and OG-III groups compared to

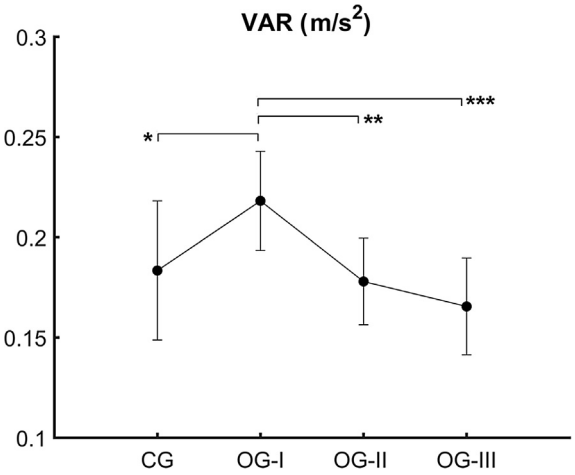

Fig. 2. VARML, averaged standard deviation of trunk acceleration behaviour. CG: control group, OG-I: observational group I, OG-II: observational group II, OG-I: observational group III. Tukey test: *(p = 0.031), **(p = 0.010), ***(p = 0.001).

Table 2
Linear and nonlinear measures of gait variability for each group.

	CG	OG-I	OG-II	OG-III	p	Levene's test
VAR ML (m/s^2) (cm/s^2)	0.18 ± 0.03[a]	0.22 ± 0.02[a,b,c]	0.18 ± 0.02[b]	0.17 ± 0.02[c]	**0.001**	0.153
RMS_{ratio}	0.44 ± 0.07[d,e,f]	0.56 ± 0.09[d]	0.55 ± 0.08[e]	0.55 ± 0.07[f]	**0.004**	0.705
λ	0.52 ± 0.05[h,i,j]	0.43 ± 0.07[h]	0.40 ± 0.07[i]	0.39 ± 0.09[j]	**0.001**	0.210
MoS (cm)	2.89 ± 1.26[g]	4.49 ± 1.26[g]	3.64 ± 1.26	3.75 ± 0.98	**0.044**	0.724
SEn	0.28 ± 0.03[k,l]	0.29 ± 0.03	0.31 ± 0.02[k]	0.31 ± 0.02[l]	**0.005**	0.307

p: one-way ANOVA. The superscripts [a,b,c,d,e,f,g,h,i,j,k,l] indicate significant pairwise comparisons $p < 0.05$. CG: control group, OG-I: observational group I, OG-II: observational group II, OG-I: observational group III. VAR: average standard deviation along each stride sample; ML: medial-lateral; RMS_{ratio}: proportion of trunk acceleration variability that occurs in the ML direction compared to the total acceleration variability; λ: maximum Lyapunov exponent; MoS: margin of stability; SEn: sample entropy.

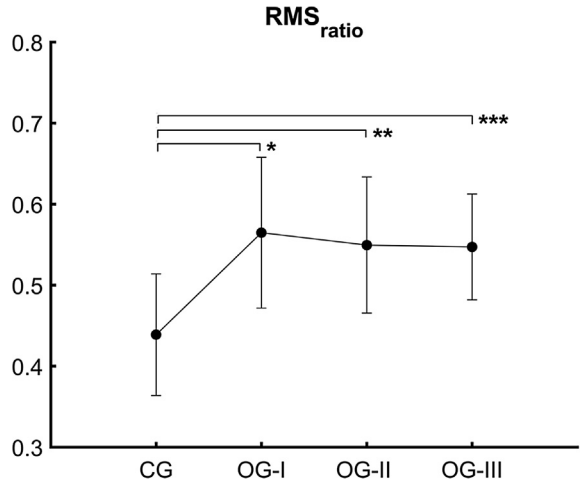

Fig. 3. RMS_{ratio}, normalized root mean square behaviour. CG: control group, OG-I: observational group I, OG-II: observational group II, OG-I: observational group III. Post-hoc Tukey test: *(p = 0.006), **(p = 0.019), ***(p = 0.023).

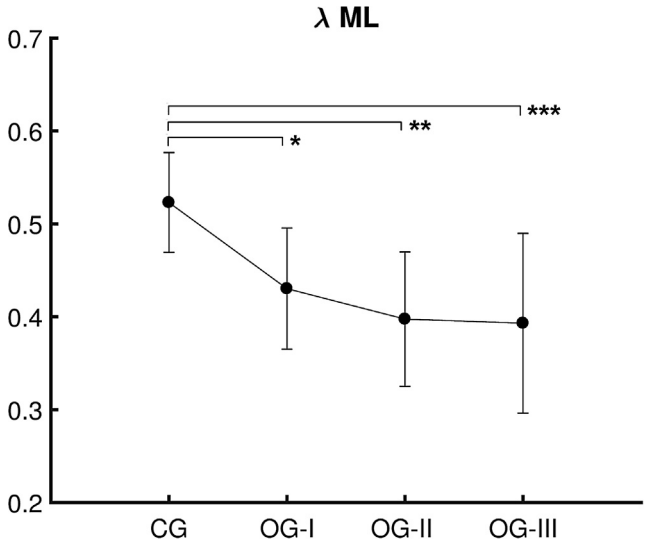

Fig. 4. λ, maximum Lyapunov exponent behaviour. CG: control group, OG-I: observational group I, OG-II: observational group II, OG-I: observational group III. Tukey test: *(p = 0.038), **(p = 0.003), ***(p = 0.002).

CG (Fig. 6). This indicates an increase in temporal variability from 6 to 12 months after ACL reconstruction.

Correlation analysis revealed that RMS_{rate} and MoS exhibited a significant, although weak, correlation (r = 0.462, p = 0.003). Since RMS_{rate} would explain only 21% of the variance in MoS, this correlation may be not functional. Additionally, SEn and λ exhibited a significant and strong negative correlation (r = −0.777, p < 0.001), which, however, may only be due to a limitation of the algorithms to estimate them.

4. Discussion

The results of the present study demonstrated increased stability in the postoperative period in biomechanical aspect (MoS), and in terms of nonlinear dynamic system (λ). Supporting our hypothesis, increased linear (VAR ML and RMS_{ratio}) and nonlinear (SEn) variability were mainly observed in the first postoperative period, OG-I (90 days). In addition, progressive recovery of local stability (λ) and slight progressively decreasing linear variability were observed over the subsequent postoperative periods (OG-II and OG-III) from 180 to 360 days,

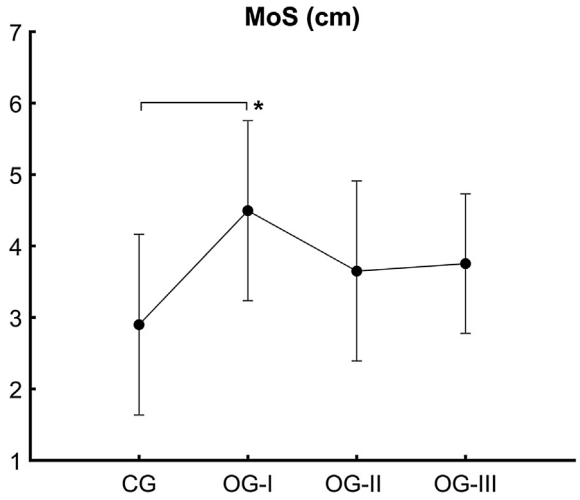

Fig. 5. MoS, margin of stability behaviour. CG: control group, OG-I: observational group I, OG-II: observational group II, OG-I: observational group III. Tukey test: *(p = 0.025.

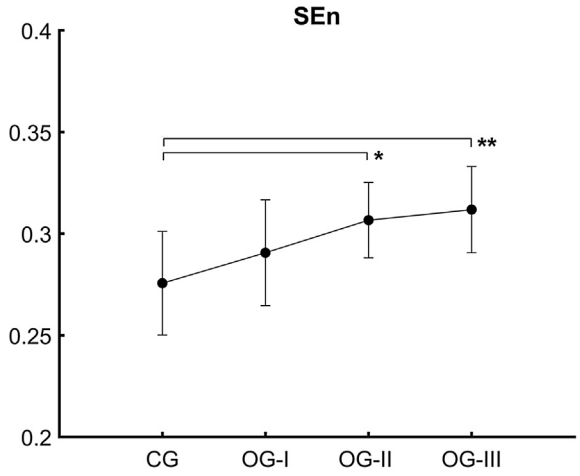

Fig. 6. SEn, sample entropy behaviour. CG: control group, OG-I: observational group I, OG-II: observational group II, OG-I: observational group III. Tukey test: *(p = 0.024), **(p = 0.006).

however the nonlinear variability (SEn) increased.

Increased linear variability in the first postoperative period, evidenced by increased average values for RMS_{ratio} and VAR ML (Table 2), suggested decreased performance of the task and increased risk of a new lesion in this period (Moraiti et al., 2009). However, from 180 days after surgery, both OG-II and OG-III showed reduction of variability over time, demonstrated by the slight reduction in RMS_{ratio} and a significant reduction in VAR ML. This recovery is probably due to rehabilitation program, indicating a progressive improvement in the performance of the task and a reducing risk of injury, given the characteristics of the group studied. Similar result has also been found in other studies that compared individuals six months after ACL reconstruction with healthy controls (Tsivgoulis et al., 2011).

Increased gait stability in the first postoperative period was evidenced by greater MoS (Fig. 5). In addition, a lower λ value (Fig. 4) in this period suggests increased LDS. These results can be explained by a lower PWS, an observed greater cognitive effort, and the adoption of a conservative gait due to insecurity and fear related to the injured limb in this period. After 180 days, the OG-II group showed decreased λ values (Fig. 4) and maintained MoS values (Fig. 5), indicating a progressive increase in stability. Decreased λ values and maintained MoS values in the OG-III group suggest that this group gradually recover

stability. MoS appears to show a gradual tendency to approximate the CG values, suggesting recovery of a dynamic gait pattern a year after surgery.

While older adults appear to prioritize stability, healthy young adults optimize forward progression, mobility, and efficiency while walking (Rogers et al., 2008). Therefore, the gait patterns of healthy young adults, characterized by phases of instability that lead to forward progression and lateral shifting of the body's CoM at each step, are detrimental to stability. This comparison can be extended to young patients after ACL reconstruction: they appear to prioritize stability, unlike healthy young adults.

With regard to the correlation found between RMS_{ratio} and MoS, the increased ML acceleration variability (denoted by increased RMS_{ratio} after ACL reconstruction), and indirectly force variability, deteriorates movement precision, and may have induced the adoption of a strategy that preserves stability, increasing MoS. However, this assumption needs to be further investigated. Furthermore, since the correlation coefficient was weak, the correlation between RMS_{ratio} and MoS may be not functional.

The decreased regularity observed in the postoperative periods, evidenced by increased SEn values (Fig. 6), indicates increased structural/temporal movement variability and randomness. An increase beyond optimal variability makes the system noisy and imprecise, and is considered detrimental to motor system (Stergiou and Decker, 2011). The SEn values did not approximate those of the CG, even after 360 days, and indicate a decreased movement complexity, where complexity is defined as a rich behavioral state that exhibits chaotic properties. These results could be explained by the altered neuromuscular activity in individuals who underwent ACL reconstruction, including both knee extensor and knee flexor strength deficit, and changes in agonist/antagonist balance around the reconstructed knee (Hiemstra et al., 2000).

Moraiti et al. (2009) reported that, two years after ACL reconstruction, the individuals showed higher entropy values compared to those estimated for healthy subjects. These results are similar to those of the present study, although there is a difference in the postoperative follow-up period (two years versus one year), a difference in the method of calculating entropy (approximate entropy versus sample entropy) and using a different biomechanical variable (knee flexion/extension versus T1 marker velocity). Moraiti et al. (2009) attribute their results to changes in muscular performance and the related neural components, such as the lack of afferent proprioceptive input due to absence of the natural ACL, and changes in mechanoreceptors around the injured knee.

SEn and λ exhibited a significant and strong negative correlation. However, this result should be interpreted with caution. It would be not advisable to stablish a linear correlation between two nonlinear variables. Lyapunov exponent (λ) is estimated from a reconstructed high-dimensional state-space, whereas SEn is estimated from a one-dimensional time series, i.e. the original signal. Besides, the negative correlation found between these two variables can only be due to limitations of the algorithms to estimate them, since it has been shown that Lyapunov exponent is lower for more noisy human walking signals (Mehdizadeh and Sanjari, 2017). Therefore, the correlation between SEn and λ found here needs to be further investigated.

In the present study, gait variability was assessed using linear and nonlinear methods. Although linear methods, such as used here, are easy to generate and interpret, they only assess the overall amplitude of variability, providing the variability information only in a single time scale. Linear measures do not account for the signal dynamics, not considering potentially valuable information for movement control. In contrast, nonlinear methods assess how a signal evolves and changes over time by quantifying its underlying structure. Extraction of this information provides additional insights into the changes in movement control that occur after ACL reconstruction.

The clinical impact of nonlinear dynamical methods requires further investigation with large populations, prolonged follow-ups, and with longitudinal-designed studies. However, it seems that these methods are a powerful prognostic tool for assessing gait stability and variability after ACL reconstruction, important characteristics for optimal operation of the motor system, and probably after other major orthopedic surgeries, requiring a simpler patient preparation than a complete kinematics gait analysis. Furthermore, a six-month period of rehabilitation normally adopted for individuals who have undergone ACL reconstruction appears to be effective, but it is not sufficient: for complete gait stability and variability recovery, the rehabilitation period should be extended further.

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Conflict of interest

The authors declare that there are no conflicts of interest.

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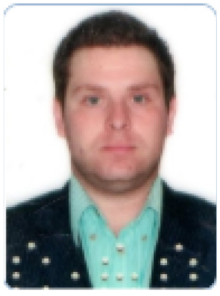

Eduardo Afonso de Oliveira is graduated in physiotherapy at the State University of Goiás, Brazil. He received his MSc in Health Sciences from the Federal University of Goiás, Brazil. He is post-graduated in Traumato-Orthopedic Physiotherapy by the Center for Scientific Development in Health and Social Sciences (CDCS). Member and Specialist Member of the National Society of Sports Physiotherapy, Brazil. His research interests include clinical practice in the areas of Sports Biomechanics, Traumato-Orthopedic Physiotherapy and Sports Physiotherapy.

Adriano de O. Andrade is graduated in Electrical Engineering at the Federal University of Goiás (Brazil) in 1997. He received his MSc and PhD in Biomedical Engineering respectively from the Federal University of Uberlândia (Brazil) in 2000 and from the University of Reading (UK) in 2005. He completed a one year Post-Doctoral Fellowship awarded by the DFAIT (Foreign Affairs and International Trade Canada) at the Institute of Biomedical Engineering of the University of New Brunswick (Canada) in 2010. Currently, he is Professor in the Faculty of Electrical Engineering of the Federal University of Uberlândia. His research interests include Biomedical Signal Processing and Modelling, Assistive Technology, Rehabilitation Engineering and Neuroengineering.

Marcus Fraga Vieira is graduated in Electrical Engineering (1984) and Physical Education (1994) at the Federal University of Goiás, Brazil. He received a MSc (1996) and a PhD (2002) in Biomedical Engineering from University of São Paulo, Brazil. He is currently a Full Professor of Bioengineering and Biomechanics at the Federal University of Goiás, Brazil. Head of the Bioengineering and Biomechanics Laboratory, his current research topics are CPG modeling, biomechanics in motor control, including postural control, transitory tasks and gait dynamic stability.