Parkinson’s Disease Influences the Structural Variations Present in the Leg Swing Kinematics

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This study investigated the nature of the structural variations found in the motor output of individuals with Parkinson’s disease (PD). Young (n = 21; 19.9 ± 1.3 yrs.), aged (n = 9; 74.8 ± 6.8 yrs.) and individuals with PD (n = 9; 73.4 ± 6.6 yrs.) swung their leg at a pendular frequency and frequencies that were 20% faster and slower. This study had three key findings. First, individuals with PD have greater variability in the leg swing angular kinematics and swing times. These variations appear to be related to the 0–15 Hz band of the angular displacement power spectrum. Second, changes in the structural variations appear to not be derived from a stochastic source. Third, the magnitude of the variations and the structure of the variations are influenced by the frequency that the leg is swung. These results are consistent with the viewpoint that changes in the magnitude of the variations and the regularity of the structural variations are dependent upon health and adaptability to the task dynamics.

Keywords: Nonlinear, Aging, Variability, Biomechanics, Complexity

Healthy physiological systems demonstrate subtle structural variations in their performance that are a result of interacting regulatory processes that are operating over multiple time scales (Lipsitz and Goldberger, 1992; Vaillancourt and Newell, 2002). The influential work of Lipsitz and Goldberger (1992) provided the initial framework for the hypothesis that the human body is less adaptive to stress and disease if these variations have a more regular pattern. Since the original conception of this hypothesis, it has been supported by a plethora of scientific data from a variety of human biorhythms (Goldberger, Rigney, Mietus, Antman, & Greenwald, 1998; Goldberger; Kaplan et al., 1991; Lehnerts & Elger, 1998; Pikkujamsa et al., 1999; Vaillancourt and Newell, 2000; Vaillancourt and Newell, 2002). For example, heart rhythms that have a more regular structural variations are associated with disease states and sudden death (Goldberger et al., 1998). Alternatively, a heart rhythm that has less predictable structural variations is associated with health. Similar changes in the structural variations present in other biorhythms of diseased and healthy physiological systems have been reported for brain waves.

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tremors, and systolic blood pressure (Lehnerts & Elger, 1998; Pikkujamsa et al., 1999; Vaillancourt and Newell, 2000). In spite of this overwhelming evidence, several studies have challenged this notion by showing that changes in the structure of the variability are also dependent upon the task dynamics. For example, several experiments that have shown that the structural variations present in the gait patterns of the aged and individuals with PD are less regular (Bartsch et al., 2007; Hausdorff, 2009; Hausdorff et al., 2000; Kurz et al., 2007; Pothakos et al., 2009). Hence, there appears to be an optimal amount of variability in the motor output that is dependent upon the interaction of neuromuscular health and the constraints imposed by the task dynamics (Stergiou et al., 2006; Harbourne & Stergiou, 2009).

Fourier analysis has also been used to explore what frequency bands may be responsible for the observed changes in the structure of the motor output (Morrison et al., 2008; Sosnoff et al., 2004; Vaillancourt and Newell, 2003). In patients with PD, it has been shown that the structure of the tremors present in the hand are related to change in the 4–7 Hz frequency band, while the structural of the postural sway patterns are related to changes in the 2–8 Hz frequency band (Morrison et al., 2008). It has been postulated that differences in the frequency structure reflect the inability to properly regulate the excitatory and inhibitory activity of the multiple neural oscillators found in the nervous system (Morrison et al., 2008; Sosnoff et al., 2004; Vaillancourt and Newell, 2003). Further exploration of the differences that are present in the spectral power distribution may provide additional means for identifying the role of the basal ganglia plays in the modulation of the neural oscillators that may be governing the frequency content of the motor output.

Traditionally, the irregular structural variations present in the performance of the muscular system have been viewed as a result of stochastic processes that are present in the nervous system (Faisal et al., 2008; Riley and Turvey, 2002). Based on this viewpoint, the major challenge of the motor command is to overcome these stochastic processes to achieve the desired goal. An alternative perspective is what appears to be stochastic is actually a deterministic process that arises from the higher-order couplings that are present in the nervous system. As such, changes in the regularity of the structural variations may represent changes in the coordinative output of the excitatory and inhibitory neural oscillators that govern the motor command. Previous experimental work has provided initial evidence that the structural variations that are present in the walking patterns of individuals with PD have a deterministic pattern (Kurz & Hou, 2010). This implies that the variations found in the gait patterns of individuals with PD are not an amplification of stochastic noise in the nervous system; rather these variations most likely a result from faulty computations by the nervous system of the relevant movement parameters. Further exploration of these concepts will provide additional insight on what factors may be governing the changes in the regularity of the structural variations seen in the motor performance of individuals with PD.

Gait represents a complex series of interlinked motor commands that control the stance and swing phase dynamics. It has been suggested that the swing phase kinematics are controlled by the higher brain centers, and the stance phase kinematics are controlled by a spinal central pattern generator (Frenkel-Toledo et al., 2005a; Gabell and Nayak, 1984). This logic is based on experimental data that has shown that the magnitude of the variability present in the stance phase kinematics is speed dependent, while the magnitude of the variability present in the swing
phase kinematics is speed independent. Based on these experimental outcomes, it has been recommended that the variability present in the swing phase kinematics provides a better means to identify how PD influences the motor control of gait (Frenkel-Toledo et al., 2005a). In this investigation, we used a simplified leg swing task to further explore this concept. Our experiment was directed at addressing the following questions: (1) Does PD influence the magnitude and regularity of the variations found in the leg swing kinematics?, (2) Does PD have differences in the power spectrum frequency components that comprise the leg swing kinematics?, (3) Are the structural variations found in the leg swing kinematics a result of a deterministic or stochastic neural process? (4) Does the speed the leg is swung influence the magnitude and regularity of the structural variations found in the leg swing kinematics?

Materials and Methods

Participants
Young \((n = 21; \text{Age} = 19.9 \pm 1)\), aged \((n = 9; \text{Age} = 77.8 \pm 6)\), and individuals with PD \((n = 9; \text{Age} = 73.4 \pm 7)\) were recruited to participate in the study. The inclusion criteria for the young and aged groups with no neuromuscular disorders, no musculoskeletal injuries, and no history of falls as reported by a modified Physical Activity Readiness Questionnaire (PAR-Q; CSEP 2002). The inclusion criteria for the PD group were (a) diagnosis of PD by a neurologist, (b) a stable regimen of anti-Parkinsonian medications, and (c) early or moderate PD onset—stage 2 or 3 of the Hoehn and Yahr disability scale (Hoehn and Yahr, 1967).

This study conformed to all regulations pertaining to research on human subjects, and was approved by the University’s Committee for the Protection of Human Subjects. All subjects read and signed a comprehensive informed consent form before participating in the study. Medical record information was released with written consent of the participants and their physicians.

Procedures
The participants performed the experimental procedures while standing strapped by a Velcro belt in a leg suspension apparatus (Champion Barbell BSNCP, Dallas, TX) retrofitted with a custom-built footrest that isolated the swinging leg (Figure 1). The footrest was adjustable along the vertical axis to accommodate variations in participants’ height and leg length. The footrest was attachable to either sides of the apparatus to accommodate variations in participant’s leg dominance or PD affectedness. A similar apparatus was used by Doke and colleagues (2005) to isolate how much metabolic energy is expended to swing the leg during the swing phase of gait. A commercial knee brace (Donjoy DJO Inc., Vista, CA) was used to fully immobilize the knee of the swing leg to isolate the source of angular displacement to the hip joint. The leg angular displacement was measured at 100 Hz with a twin-axis electronic goniometer (Biometrics Ltd., Ladysmith, VA) that was placed at the hip of the swinging leg. Power spectrum analysis from our pilot data indicated that 99.99% of the cumulative power was represented up to 24 ± 11 Hz. Hence, collecting the data at 100 Hz was well above the Nyquist frequency.
While supporting their body weight with one leg and both arms, the participants swung their unsupported leg at the prescribed frequency. The young and the aged participants performed the task with their nondominant leg. Participants with PD performed the task with the most affected leg—i.e., the one showing more severe PD symptoms. Level of leg affectedness was determined based on the Hoehn-Yahr scale descriptors (Hoehn and Yahr, 1967) and physician reports. Leg dominance was determined by having each participant kick a soccer ball. The participants were provided a three-minute warm up period to practice the pendular leg swings and matching leg swings to the metronome beat.

The participants swung their leg at three frequencies: pendular, 20% faster and 20% slower than pendular. The leg swing conditions were presented in a random order and each condition lasted for two minutes. The participants were given three minutes to practice each condition, and a minimum of one minute of rest was given between each condition. The subjects were required to match the maximum

Figure 1 — Schematic representation of the experimental apparatus and protocol.
forward swing angle with the first metronome beat, the maximum backward swing angle with the second metronome beat. The swing frequencies were determined based on the natural pendular frequency of the leg to normalize leg swings based on geometric and dimensional properties as indicated by the dynamic similarity hypothesis (Alexander, 2003). The pendular frequency of the whole leg was calculated based on the Equation 1:

$$\tau = 2\pi \sqrt{\frac{m \cdot r^2}{m \cdot g \cdot l}}$$

Where $r$ is the radius of gyration around greater trochanter, $m$ is the mass of the whole leg from greater trochanter to medial malleolus, $g$ is gravity and $l$ is the length of the leg. The length of the leg was calculated based on the distance between the greater trochantor and the floor. The anthropometric values presented in Winter (2005) were used to estimate the mass of the limb based on the subject’s total body mass, and the leg’s the radius of gyration based on the subject’s leg length. Using Equation 1, the average pendular frequency for the young was 1.05 ± 0.04 Hz, the aged was 1.02 ± 0.02 Hz, and PD groups was 1.06 ± 0.05 Hz.

**Variability Analyses**

A ninth order low pass digital filter with a 6 Hz cut-off was used to smooth the leg swing kinematics. The times between the local maximums of the leg swing kinematics were used to determine the period of each of the respective leg swings. The standard deviation and coefficient of variation (e.g., 100 x [standard deviation of swing time/mean swing time]) of the leg swing times were used to quantify the magnitude of the variations that were present in the leg swing performance. Separate mixed ANOVAs (Group X Leg Swing Condition) were used to determine if there were differences in the standard deviations and coefficient of variations. Tukey post hoc tests were used to identify differences in the magnitude of the variability present in the respective groups and leg swing frequencies. All statistical tests were performed at alpha level of 0.05.

The regularity of the structural variations present in the unfiltered leg swing angular kinematics were evaluated using Approximate Entropy (ApEn) (Pincus, 1991; Stergiou, 2004; Vaillancourt and Newell, 2003)—a measure of the regularity of occurrence of self-similar structures of signal within a time series. ApEn calculations yield a score in the range of 0–2, where 0 reflects a highly repeatable and regular signal and 2 a random signal with no regularity. Highly regular signals are very predictable and indicate high likelihood of reoccurrence of self-similar patterns. Irregular signals have low predictability indicating low likelihood of reoccurrence of self-similar patterns. ApEn is calculated using Equation 2:

$$Apen(N, m, r) = \ln \left[ \frac{C_m(r)}{C_{m+1}(r)} \right]$$

Where $N$ is a number of data points $\{u(1), u(2), \ldots, u(N)\}$ in the time series, $m$ is the number of consecutive $u$ data points $\{u(i)\}$ that define the length of vectors $x(i) = \{u(i) \ldots u(i+m-1)\}$ to be compared over the data set $\{x(1) \ldots x(N-1-m)\}$,
and \( r \) is radius of acceptance (i.e., a similarity criterion between compared vectors) based on the standard deviation of the data series. To calculate the similarity of consecutive vectors \( x(i), x(j), \) etc., the largest distance \( d [x(i), x(j)] \) between their scalar components is calculated and compared with \( r \). \( C \) then is the number of self similar vectors defined by \( m \) points based on the \( r \) criterion (i.e., number of \( x(j) \) such that \( d [x(i), x(j)] \leq r \)). Similar to previous investigations of the structural variability present in a physiological time series, \( m \) was set to 2, and \( r \) was set to 20% of the standard deviation of the time series (Pincus, 1991; Vaillancourt and Newell, 2000; Vaillancourt and Newell, 2003).

When using ApEn to analyze the structural variations of a time series, one assumes that the distance between to vectors is due to changes in the behavior of the system, and not due to the vectors being neighbors in time (Provenzale et al., 1992). For example, a slow moving leg swing task will appear to have more vectors that are similar because there is not as much change in the position of the limb. Alternatively, a fast leg swing task will have greater change in the data samples. We used a method of delays to overcome the differences in the movement speeds of the respective leg swing tasks (Abarbanel, 1996; Kantz and Schreiber, 2003). An autocorrelation function was used to select a delay that created vectors that contained unique information about the evolving leg swing dynamics (Equation 3).

\[
\rho(T) = \frac{\sum_{n=1}^{N} (y_n + T - \bar{y})(y_n - \bar{y})}{\sum_{n=1}^{N} (y_n - \bar{y})^2}
\]

where \( \bar{y} \) was the sample mean, and \( y \) was the data of the time series, and \( T \) was the time delay. The smallest possible value of \( T \) which meet the criteria of \( \rho(T) \leq 0 \) was used for the delay. Selecting a \( T \) that was very close to zero created vectors that contained unique information about the dynamics of the leg swing task (Abarbanel, 1996).

ApEn scores were analyzed using a mixed ANOVA design with 2 factors (Group X Leg Swing Condition). Tukey post hoc tests were used to determine differences in the ApEn for the groups and leg swing frequencies. All statistical tests were performed at alpha level of 0.05.

**Frequency Structure Analysis**

Fourier analysis was used to evaluate what frequencies contributed to the structure of the unfiltered leg swing kinematics. An auto-spectrum analysis was performed in Matlab (Mathworks, Natick, MA) using a Welch’s averaged periodogram method with a window size of 254. The resulting power spectrum was normalized to the maximum power of the spectral distribution, which accounted for any amplitude differences between the subject’s leg swing conditions. We examined the sum of the power present in five Hz bins (e.g., 0–5, 5–10, 10–15 etc.) that were created from 0 to 30 Hz. These bins represented 99.99% of the power in the spectrum.

A mixed design ANOVA (Group X Leg Swing Condition X Frequency Bin) was used to determine if there were significant differences in the amount of spectral power of the leg swing kinematic of the respective groups and leg swing conditions. Tukey post hoc tests were used to determine if there was a significant difference in the amount of spectral power found in each of the groups x condition x bin interactions.
Swing Phase Structural Variations

Surrogation Analysis

We used a pseudo periodic surrogation (PPS) algorithm to determine if the calculated structural variability resulted from a deterministic or a stochastic process. Complete details of the algorithm are found in Small & Tse (2002) and Miller et al. (2006). Conceptually, the PPS algorithm generates a surrogate of the original time series that preserves the inherent periodic components while destroying the nonlinear structure. The structure of the surrogate follows the same vector pattern as the original leg angle time series, but was contaminated with white noise. If fluctuations in the original time series had deterministic structural variations, these features would be destroyed in the surrogate. Alternatively, if the structural variations in the time series were a result of a stochastic motor process, the surrogate would be no different from the original time series. Dependent t tests with a Bonferroni corrected alpha level were used to determine if the ApEn values for the original leg kinematic time series of the respective groups were statistically different from the ApEn values from the surrogate time series. The structural variations were considered to be deterministic if ApEn of the original data were statistically different from the surrogate.

Results

Magnitude of the Variations

There were significant differences in the standard deviation main effects for group ($p = .015$) and leg swing frequency ($p < .001$) (Figure 2). However, no significant interaction effects were found ($p = .087$). Post hoc analysis indicated that the grand mean for the PD group was significantly different from the young ($p = .032$) and aged ($p = .024$) group grand means. No significant differences ($p > .05$) were noted between the young and aged groups’ standard deviation grand means (Figure 2A). These results indicate that the magnitude of the variations found in the PD leg swing timings were greater than those found in the aged and young. Our post hoc on the frequency grand means indicated that the magnitude of the variations found in the slow were significantly different from the pendular ($p < .001$) and fast frequencies ($p < .001$). Furthermore, the standard deviation of the pendular condition was significantly different from the fast condition ($p < .001$) (Figure 2B). This result indicated that moving the leg faster or slower than the leg pendular velocity influenced the magnitude of the structural variations.

The results from the coefficient of variation complemented what was found for the standard deviation (Figure 3). There were significant differences in the coefficient of variation main effects for group ($p = .005$) and leg swing frequency ($p < .001$). Once again, there were no significant interaction effects ($p = .253$). Post hoc analysis indicated that the grand mean for the PD group was significantly different from the young ($p = .011$) and aged ($p = .006$) group grand means. No significant differences ($p > .05$) were noted between the young and aged groups’ grand means (Figure 3A). Our post hoc on the frequency grand means indicated that the coefficient of variation for the slow frequency was significantly different from the pendular ($p = .015$) and fast frequencies ($p < .001$). Furthermore, the coefficient of variation of the pendular condition was significantly different from the fast condition ($p < .001$) (Figure 3B).
Figure 2 — A) Mean ± SE (SE) of the standard deviations for the young, aged and individuals with Parkinson’s disease; B) Mean ± SE coefficient of variations (%) for the young, aged and PD. * indicates $p < .05$. 
Figure 3 — A) Mean ± SE (SE) standard deviations for the slow, pendular and fast leg swing frequencies; B) Mean ± SE coefficient of variations (%) for the slow, pendular and fast leg swing frequencies. * indicates $p < .05$. 
Regularity of the Structural Variations

There were significant differences in the main effects for group \( (p = .01) \) and leg swing frequency \( (p < .001) \). However, no significant interaction effects were found \( (p = .5) \). Post hoc analysis indicated that the ApEn grand mean for the PD group was significantly different from the young \( (p = .01) \) and aged \( (p = .01) \) ApEn group grand means (Figure 4A). No significant differences \( (p > .05) \) were noted between the young and aged groups’ ApEn grand means. Our post hoc on the leg swing frequency ApEn grand means indicated that the regularity of the structural variations found in the slow and pendular conditions were not significantly different \( (p > .05) \). However, the regularity of the structural variations seen in the slow \( (p = .01) \) and pendular \( (p = .01) \) conditions were significantly different from the fast condition (Figure 4B).

Frequency Structure

There were significant main effects for condition \( (p < .001) \), group \( (p = .002) \) and bin \( (p < .001) \). Furthermore, there was a significant Leg Swing Condition x Group x Bin interaction effect \( (p < .001) \). For the slow leg swing condition (Figure 5A), our post hoc analysis determined that there were significant differences in the 0–5 Hz and 5–10 Hz frequency bins between the individuals with PD and the aged \( (p = .01) \), and the individuals with PD and the young \( (p = .01; \text{Figure 5A}) \). Additional significant differences \( (p = .01) \) were present in the 5–10 Hz frequency bins of the aged and young. No significant \( (p > .05) \) differences between the respective groups were found above 10 Hz for the slow leg swing condition.

For the pendular leg swing condition (Figure 5B), our post hoc analysis determined that there were significant differences in the 0–5 Hz and 5–10 Hz bins between the individuals with PD and the aged \( (p = .01) \). Our post hoc analysis also indicated that there were significant differences in the 0–5 Hz, 5–10 Hz and 10–15 Hz frequency bins of the individuals with PD and the young \( (p = .01; \text{Figure 5A}) \). Furthermore, there were additional significant differences in the 5–10 Hz and 10–15 Hz frequency bins of the aged and young \( (p = .01) \). No significant \( (p > .05) \) differences between the respective groups were found above 15 Hz for the pendular leg swing condition.

For the fast leg swing condition (Figure 5C), our post hoc analysis determined that there were significant differences in the 0–5 Hz, 5–10 Hz, 10–15 Hz frequency bins between the individuals with PD and the aged \( (p = .01) \), and between the individuals with PD and the young \( (p = .01) \). No significant \( (p > .05) \) differences between the respective groups were found above 15 Hz for the fast leg swing condition. In addition, no significant differences \( (p > .05) \) were found between the aged and young for any of the fast leg swing condition bins.

Surrogation Analysis

There was a significant difference in the ApEn of the original data and the surrogates for the PD \( (p < .001) \), Aged \( (p < .001) \) and young \( (p < .001) \) groups. Hence, indicating that the structural variations found in the leg kinematics of the respective groups were not a result of stochastic noise.
Figure 4 — A) Mean ± SE (SE) of the ApEn for the slow, pendular, and fast leg swing conditions. B) Mean ± SE of the ApEn for the young, aged and Parkinson’s disease. * indicates $p < .05$. 


Figure 5 — Normalized power spectrum for the young (Bold Line), aged (Thin Line) and individuals with Parkinson's disease (Dashed Line) while swinging the leg at A) slow frequency, B) pendular frequency and C) fast frequency.
Discussion

Influence of Parkinson’s Disease

Our results show that individuals with PD have greater variations in the leg swing timings, and angular kinematics with less regular structural variations. Although our movement task is far more simplified than walking, somewhat similar results have been presented by Frenkel-Toledo et al. (2005b) where they found that the magnitude of the variations seen in the swing time are greater in individuals with PD. Taken together, these outcomes infer that PD influences the ability to control the rhythmicity of the leg swing kinematics. Since PD patients have been shown to have a loss of kinesthesia (Maschke et al., 2003), we suggest that the noted differences in the regularity of the structural variations were due to the inability of the individuals with PD to effectively integrate the available sensory information to correct the ongoing limb trajectories. The loss of kinesthesia may also be the reason why there were differences in the 0–15 Hz region of the power spectrum across all the leg swing conditions. Changes in the frequency domain may represent a reduced capacity to effectively adapt the slow and fast time scale sensorimotor processes for maintenance of the leg swing kinematics.

An alternative explanation for the less regular structural variations seen in the PD group may be due to the medication taken by the participants. This notion is supported by experimental results that have shown that the regularity of structural variations seen in the postural sway pattern and walking kinematics are influenced by levodopa therapy (Kurz and Hou, 2010; Morrison et al., 2008). Hence, it is plausible that the medication taken by the patients may partly explain the noted changes seen in this investigation. Further studies on the effect of PD medications on the regularity of the structural variations of the motor output are warranted. These investigations may provide a means to clinically identify if current drug therapies provide beneficial improvement in the control of movement.

Influence of Aging

Our results suggest that age does not affect magnitude of the variations present in the swing timings and regularity of the structural variation present in the leg swing angular kinematics. Contrary to our results, a number of studies have shown that the regularity of the structural variations present in rhythmic fine motor tasks and gait are less regular in the aged (Hausdorff et al., 1997; Kurz and Stergiou, 2003; Vaillancourt and Newell, 2002; Vaillancourt and Newell, 2003). We suspect that our inability to detect differences in the variations of the leg swing kinematics were due to the fact that our aged participants were highly active and participated regularly in physical activity. Based on this notion, our results may infer that physical activity can prevent age related changes in the variations present in the motor system. On the other hand, we did find that there were differences in the 5–15 Hz frequency bins between the aged and young for the pendular and slow leg swing conditions. Previous investigations have used the frequency structure to identify reliance on specific sensory-motor processes (Sosnoff and Voudrie, 2009). We suspect that these results infer that the aged have more difficulty integrating the fast time scale sensory motor processes that are necessary for controlling the leg
swing kinematics. However, these fast time scale sensorimotor impairments appear to not be sufficient to influence the magnitude and the structural regularity of the variations seen in the healthy aged leg swing patterns.

**Influence of Leg Swing Frequency**

Our results indicated that swinging the leg at a slow or pendular frequency had similar structural variations. However, the regularity of these structural variations were different if the leg was swung at a speed that was 20% faster than the pendular speed. We suspect that this may be partially explained as a result of over-reliance on mechanical properties of the leg for passive control (Mochon & McMahon, 1980; McGeer, 1990, 1990). Moving at a faster frequency may challenge the ability of the nervous system to use kinesthetic feedback to correct the trajectory of the limb. For example, the nervous system may have sufficient time to use a closed-loop control scheme to correct the limb’s movement pattern when it is swung slow. Conversely, at a faster speed the nervous system may have less time to use the necessary feedback to correct the limb position and may rely on passive dynamics of the limb for control (i.e., limb’s inertia and viscoelastic properties of the musculotendon units).

These inferences are further supported by the magnitude of the variations that were seen in the leg swing timings, where our results show that swinging the leg slower resulted in greater variations in the swing timings, while swinging the leg faster resulted in less variation in the swing times.

Previously it has been noted that the magnitude of the variations seen in the swing phase timings during gait are relatively consistent and are not dependent on walking speed (Frenkel-Toledo et al., 2005a). Our results are not consistent with this finding and suggest that the variations are speed dependent. Prior investigations have shown that a change in the loading of the foot during the terminal portion of the stance is involved in the control of the onset of the swing phase during gait (Harkema et al., 1997; Pang & Yang, 2000). It is possible the lack of such feedback during our leg swing task may be the reason our results were speed dependent and had a larger amount of variability in the swing phase timings. Future investigations should challenge our results by further testing how sensory feedback influences the magnitude and structure of variations seen in healthy and pathological rhythmical movement patterns.

**Surrogation Analysis**

The surrogation analysis indicated that differences in the structural variations present in the rhythmical leg swing pattern were deterministic and not a result of a stochastic process. These results support the concept that the structural variations that are present in the motor performance are a result of how inhibitory and excitatory neural oscillators cooperate for an effective motor command (Vaillancourt and Newell, 2002). The surrogation analysis tested whether the structural variations seen in the movement pattern are different from white noise that is superimposed on top of the attractor. White noise is a mathematical construct where the components of the signal are not correlated such that changes from one data point to the next cannot be predicted. Often white noise is used as a means to model the variability that is present in the nervous system by superimposing it on top of the control signal (Su
and Dingwell, 2007). Our surrogation results indicate that this may not be the best method for modeling the structural variations that are present in the nervous system.

Stergiou and colleagues (2006) proposed a theory that there is an optimal amount of variability for health and motor adaptability. Our results appear to support this notion since PD and leg movement speed influenced the magnitude and regularity of the structural variations present in the leg swing kinematics. The optimal movement variability theory states that the structural variations should ideally have a chaotic structure. A chaotic structure means that the variations that are present in a biological signal are not stochastic and arise from a deterministic source (Abarbanel, 1996; Kantz and Schreiber, 2003). Based on the results from our surrogation analysis, it is possible that the variations seen in this investigation may have a chaotic pattern. However, the surrogation analysis used here cannot rule out the possibility that these variations may have result from periodic dynamics with colored noise (Small et al., 2001), which would have a persistent pattern but would not be chaotic. Although we agree that there should be an optimal amount of variability for health and motor adaptability, the structure of this variability remains to be defined and deserves further exploration.

Summary

This study had three key findings. First, individuals with PD have greater variations in the leg swing kinematics, and these variations appear to be related to the 0–15 Hz band of the of power spectrum. Second, changes in the structural variations appear to not be derived from a stochastic source. Third, the magnitude and structure of the variations are influenced by the frequency that the leg is swung. These results are consistent with the viewpoint that there is an optimal amount of variability that is influenced by health and adaptability to the task dynamics.

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References


