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QUANTIFYING COMPLEXITY AND VARIABILITY OF GAIT PHASE PORTRAITS

**Louis A. DiBerardino III (1), John D. Polk (2), Karl S. Rosengren (3),
Elizabeth T. Hsiao-Wecksler (1)**

(1) Department of Mechanical Science & Engineering
University of Illinois, Urbana, IL
Email: diberard@illinois.edu

(2) Department of Anthropology
University of Illinois
Urbana, IL

(3) Department of Psychology
Northwestern University
Evanston, IL

INTRODUCTION

Injuries to the lower extremity can limit joint motion and alter movement patterns of the limb segments during gait. Researchers have also found changes in variability and complexity as a function of injury. Injury or joint pain appears to decrease gait cycle variability [1]; however, the results on complexity are less clear. While some researchers have found a decrease in complexity due to disease and aging, e.g., [2], others have found an increase in complexity [3]. One of the goals of the current work was to quantify changes in the variability and complexity of motion patterns as assessed using phase portraits.

Much of the past research using phase portraits, which capture angular position versus angular velocity, has involved qualitative visual descriptions, e.g., [4]. Elliptical Fourier analysis (EFA) may provide more accurate descriptions of phase portrait shape, and has been used to describe closed loop contours with Fourier series [5]. Most recently, we adapted EFA to compare the gait of children with and without developmental coordination disorder [6].

The current work focused on how restrictive bracing affected the complexity and variability of segmental motions of the braced and contralateral limbs. Complexity was quantified via the number of harmonics needed for fitting the shape, relating to the regularity and smoothness of the motion patterns. Phase portrait size and centroid path characteristics were used to quantify variability between cycles and trials. Due to reduced range of motion caused by bracing, we hypothesized that segmental complexity and variability would be significantly affected in the braced leg. Specifically, we expected gait complexity to increase and variability to decrease in the braced leg. In addition, we hypothesized that motion complexity and variability of the contralateral (unbraced) leg would be oppositely affected to compensate for the braced leg. Finally, we expected a significant increase in gait asymmetry as a function of wearing the brace.

METHODS

Twenty healthy male subjects – mean age 23 (SD 2) years, height 1.79 (0.06) m, mass 73 (8) kg – walked on a treadmill at a self-selected speed with and without a brace on their right knee to resist flexion. Procedures were approved by the University of Illinois Institutional Review Board and participants gave informed consent. Kinematic data were captured using a six-camera motion capture system. Sagittal plane projections of thigh, shank, and foot segment angular data were calculated. One subject was omitted due to missing marker data.

Complexity was quantified by the number of harmonics of a Fourier fit characterizing 99.9% of the data (Figure 1). EFA was performed on (angular position vs. velocity) phase portraits for each segment, fitting the data using a 500 harmonic elliptical Fourier series (full fit). The maximum error (sum of squared errors, or SSE) was calculated in the full fit, SSE_{max}, as the sum of squared radii (distance between data points and mean centroid). Complexity was quantified by the number of harmonics, j, (of a reduced fit) satisfying:

$$SSE_j \leq 0.001 * SSE_{\max} \qquad (1)$$

$$SSE_j = \sum_i \left(\left(x_{full,i} - x_{j,i} \right)^2 + \left(y_{full,i} - y_{j,i} \right)^2 \right) \qquad (2)$$

where the point ($x_{j,i}$, $y_{j,i}$) is the i^{th} point on the reduced fit curve of j harmonics. For example, if an integer value was larger for a particular condition, then that phase portrait needed more harmonics (higher frequencies) to accurately describe its shape, indicating higher complexity.

Inter-cycle variability was assessed by measures used to quantify the fluctuations of the phase portrait centroid over multiple gait cycles. These measures were adapted from traditional center of pressure stabilogram analyses [7]: the 95% confidence ellipse area swept out by the centroid (**swept area**) and the total centroid **drift** (total Cartesian distance that the centroid traveled on the phase plane). The variability of segment orientation between trials was quantified by the mean centroid **location** for an entire trial, and the size variability was quantified by differences in the average **range** in each dimension (position [location-pos], velocity [location-vel]) of the phase portrait.

All calculations were performed in MATLAB (R2008a). Significance was identified (SPSS v.15) through 2x2 (leg, brace) repeated measures MANOVAs for each segment ($\alpha=0.05$). Significant interactions were followed-up using paired *t*-tests ($\alpha=0.0125$) between specific cases (left leg: no brace vs. braced [L_nb-L_kb], right leg: no brace vs. braced [R_nb-R_kb], no brace: left vs. right [L_nb-R_nb] and braced: left vs. right [L_kb-R_kb]).

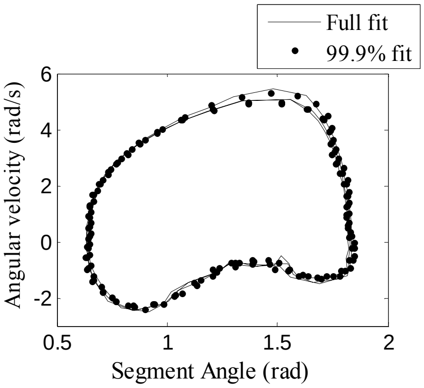

Figure 1. Sample shank segment phase portrait (only 3 gait cycles shown for clarity).

RESULTS AND DISCUSSION

MANOVA results for the shank segment revealed significant main effects for leg and brace conditions and a significant interaction. Follow-up ANOVAs indicated that the **complexity**, phase portrait **ranges** and mean centroid **locations** were significantly different for each leg-brace combination. For centroid **drift**, there was a significant effect of bracing and leg-by-brace interaction, but the elliptical **swept area** was affected only by bracing. Significant differences (as revealed by the *t*-tests) were found for L_nb-L_kb for **drift**, R_nb-R-kb and L_kb-R_kb for **complexity**, **ranges** and **locations**, and L_nb-R_nb for **location** (position only) (Table 1). Results for the thigh and foot were similar.

The overall shape **complexity** of the affected leg increased with introduction of the knee brace, but the brace did not alter the contralateral leg's complexity. In the braced condition, phase portrait **range** significantly decreased in both dimensions for the braced leg, but only in the velocity dimension for the non-braced leg. This can be explained by restricted motion of the shank due to knee immobilization. The mean centroid **location** increased for the right leg with bracing. This result demonstrates an overall change in segment orientation. The increases in centroid **drift** and **swept area** exhibit an overall increase in inter-cycle variability due to bracing. This is contrary to previous findings on variability and injury, e.g., [2]. The differences between our results and previous studies are likely due to the use of differing metrics, or the possibility that the knee brace could be affecting variability in a different manner than an actual injury or disease. For most measures, bilateral differences for the unbraced condition were not significant, while bilateral effects for the braced condition were significant, exhibiting a decrease in bilateral symmetry caused by the brace.

CONCLUSIONS

This study presented a novel approach to quantitatively assessing complexity and variability in gait. Our results suggest that this approach can successfully quantify gait changes induced by a simulated injury. Future studies should inspect the effectiveness of these phase portrait assessment metrics in examining actual injured or pathological gait.

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Table 1. Shank segment phase portrait metrics (Mean (SD)).

Measure	R_nb	R_kb	L_nb	L_kb
Complexity (# harmonics) 123bd	150 (18)	181 (25)	153 (16)	153 (14)
Range-pos (rad) 123bcd	1.18 (0.09)	0.86 (0.11)	1.20 (0.09)	1.17 (0.11)
Range-vel (rad/s) 123bd	7.67 (0.99)	4.42 (0.67)	7.77 (0.89)	7.90 (0.79)
Location-position (rad) 123bd	1.27 (0.03)	1.40 (0.04)	1.28 (0.03)	1.28 (0.03)
Location-velocity (rad/s) 123bd	0.0004 (0.0027)	-0.0028 (0.0015)	0.0007 (0.0020)	0.0002 (0.0022)
Centroid Drift (Cartesian distance) 23a	0.66 (0.31)	0.70 (0.18)	0.64 (0.23)	0.92 (0.39)
95% Swept Area (rad^2/s) 2	0.0062 (0.0047)	0.0073 (0.0031)	0.0057 (0.0043)	0.0109 (0.0076)

Significant MANOVA effects for leg (1), brace (2), and interaction (3).
Significant *t*-test results for L_nb-L_kb (a), R_nb-R_kb (b), L_nb-R_nb (c), and L_kb-R_kb (d).

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