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Short communication

Sensor positioning and experimental constraints influence estimates of local dynamic stability during repetitive spine movements

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ABSTRACT

Application of non-linear dynamics analyses to study human movement has increased recently, which necessitates an understanding of how dependent measures may be influenced by experimental design and setup. Quantifying local dynamic stability for a multi-articulated structure such as the spine presents the possibility for estimates to be influenced by positioning of kinematic sensors used to measure spine angular kinematics. Oftentimes researchers will also choose to constrain the spine's movement by physically restraining the pelvis and/or using targets to control movement endpoints. Ten healthy participants were recruited, and asked to perform separate trials of 35 consecutive cycles of spine flexion under both constrained and unconstrained conditions. Electromagnetic sensors that measure three-dimensional angular orientations were positioned over the pelvis and the spinous processes of L3, L1, and T11. Using the pelvic sensor as a reference, each sensor location on the spine was used to obtain a different representation of the three-dimensional spine angular kinematics. Local dynamic stability of each kinematic time-series was determined by calculating the maximum finite-time Lyapunov exponent (λ_{max}). Estimates for λ_{max} were significantly lower (i.e. dynamically more stable) for spine kinematic data obtained from the L3 sensor than those obtained from kinematic data using either the L1 or T11 sensors. Likewise, λ_{max} was lower when the movement was constrained. These results emphasize the importance of proper placement of instrumentation for quantifying local dynamic stability of spine kinematics and are especially relevant for repeated measures designs where data are obtained from the same individual on multiple days.

1. Introduction

The prevalence of non-linear dynamics analyses to quantify local dynamic stability has recently increased in studies of human movement. Originally these analyses were applied to temporal joint and segmental angular kinematic patterns of the lower limb during gait (Dingwell et al., 2001; Bruijn et al., 2009a); however, Granata and England (2006) used a non-linear dynamics approach to quantify local dynamic stability of spine angular kinematics during cyclical trunk flexion/extension. Studies to date have investigated the effects of movement speed, fatigue, external load, and low back pain on local dynamic stability of spine angular kinematics (Granata and England, 2006; Granata and Gottipati, 2008; Graham et al., 2012a; Beaudette et al., 2013; Graham et al., 2014). While the computational algorithms for calculating local dynamic stability are identical for any time-varying quantity, there are additional methodological considerations when applying these techniques to spine angular kinematic data.

Local dynamic stability analysis of joint angular kinematics requires kinematic information from the proximal and distal segments that span the joint. For example, local dynamic stability of knee kinematics during gait is determined using the time-series of relative orientations between the femur and shank. Regarding the spine, studies have primarily focused on determining the local dynamic stability of the lumbar spine angular kinematics. Contrary to the knee joint (a single articulation between the shank and femur), the lumbar spine consists of five vertebrae and the pelvis that span five joints. To obtain three-dimensional kinematics of the lumbar spine, sensors are often positioned over the pelvis and near the level of the twelfth thoracic vertebra (Cholewicki and McGill, 1996). Since these sensors span multiple segments, the magnitudes of lumbar spine angular kinematics are directly influenced by the number of vertebral segments that are spanned. Therefore, larger angular deviations are measured when the sensors span a larger number of segments.

Given that sensor placement can effectively alter the magnitude of spine angular kinematics, it is also important to know if sensor

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Table 1
The average age, height, and weight of participants used in this study. Values in parentheses represent the standard error of the mean.

	Females	Males	Overall
N	5	5	10
Age(years)	24.6 (3.4)	21.2 (0.7)	22.9 (1.7)
Height(m)	1.71 (0.03)	1.83 (0.04)	1.77 (0.04)
Weight(kg)	65.4 (3.1)	78.5 (4.6)	72.0 (3.4)

placement could influence estimates of local dynamic stability of spine movement. The purpose of the current investigation was to determine the influence of sensor positioning on estimates of local dynamic stability of spine movements. In addition, this study compared estimates of local dynamic stability for spine movements both with and without the pelvis being restrained.

2. Methods

2.1. Participants

A total of 10 participants (5 male and 5 female) were recruited for this investigation (Table 1). All participants were free from lower back pain at the time of data collection. Prior to data collection, each participant read and signed an informed consent document outlining the experimental protocols that were approved by the research ethics board at Nipissing University.

2.2. Instrumentation

Spine angular kinematic data were obtained by using electromagnetic sensors that measured three-dimensional angular orientations (3D Guidance trakSTAR, Ascension Technology Corporation, Milton, VT, USA). A single sensor was placed over the pelvis. Three additional sensors were positioned over the spinous processes at the levels of the first lumbar (L1), third lumbar (L3), and eleventh thoracic (T11) vertebrae. An investigator identified each of the spinous processes by manual palpation. Angular kinematic data were digitally sampled from each of the sensors at 240 Hz.

2.3. Protocol

Following instrumentation, participants were asked to perform repetitive spine flexion under two conditions. During the first condition the pelvis was constrained. Participants were instructed to keep their arms outstretched with their hands together and to touch targets positioned directly in front of them at shoulder height and 0.5 m anterior to the knee by flexing and extending their spine (Fig. 1A). The participant's pelvis was also restrained during this condition. For the second condition the protocol was repeated without pelvic constraint and participants were instructed to maximally flex and extend their spine. The order of the two conditions was randomly determined for each participant. All participants completed two consecutive trials of 35 consecutive cycles of forward spine flexion followed by extension to an upright posture for each condition. Thus, each participant completed a total of four trials. The rate of each spine flexion/extension cycle for all trials was controlled at 0.25 Hz by a digital metronome.

2.4. Data processing and analysis

Using the pelvis sensor as a common reference for each of the three vertebral sensors, three different representations for the spine's angular positions were calculated (T11 vs. Pelvis, L1 vs. Pelvis, and L3 vs. Pelvis). In each case, three-dimensional angles were extracted through Euler rotation matrices of the corresponding low back sensor relative to the pelvis using a flexion-extension (FE), lateral-bending (LB), and axial twist (AT) rotation sequence (Graham et al., 2012a). Then, in order to get an overall depiction of the spine's movement (Granata and England, 2006), the Euclidean norm (EN) of the three Euler angles for each angular representation was calculated at each sample (i) as

$$EN_i = \sqrt{FE_i^2 + LB_i^2 + AT_i^2}. \tag{1}$$

Furthermore, to ensure that any observed stability differences were not simply due to magnitude effects (i.e. the T11 vs. Pelvis EN angle would be greater than the L3 vs. Pelvis EN angle because a greater number of vertebrae were spanned), each EN angle was normalized to unit variance as

$$\overline{EN_i} = \frac{EN_i}{\mathrm{var}(EN)}. \tag{2}$$

The maximum finite-time Lyapunov exponent (λ_{max}) was then used to quantify local dynamic stability of each representation of spine movement. A brief description of the mathematical process used to calculate λ_{max} from a time-series of angular kinematic data is presented here. Interested readers are referred elsewhere for more detailed explanations (Rosenstein et al., 1993; Dingwell and Cusumano, 2000; Granata and England, 2006; Graham et al., 2011). The 35 cycles of repetitive spine flexion were identified for each of the kinematic time-series. Each EN time-series was adjusted so that a total of 33600 (35 cycles × 4 s/cycle × 240 Hz) samples fell between the defined start and end points to account for potential bias that may occur due to differences in time-series duration (Bruijn et al., 2009b). The n-dimensional state space for the time-series representing the EN angle was then reconstructed using the method of delays (Abarbanel et al., 1993). Based on previous work, a delay of 96 samples (~10% of the average number of samples per cycle), and a reconstruction dimension of 6 were chosen (Granata and England, 2006; Graham et al., 2012a; Graham and Brown, 2012; Howarth et al., 2013). The exponential rate of divergence between initially neighboring trajectories in the reconstructed state space was used to estimate λ_{max}. Specifically, λ_{max} was equivalent to the slope of the linear fit

$$y(i) = \frac{1}{\Delta t} < \ln\ d_j(i) > , \tag{3}$$

where $y(i)$ is the average logarithm of divergence ($d_j(i)$) for all pairs of nearest neighbors (j) throughout a certain number of time steps (i). The slope was determined across the first 0.5 cycles (480 samples) (Bruijn et al., 2009a, 2009b; Graham et al., 2012a, 2012b.

2.5. Statistical analysis

Dependent measures of λ_{max} and peak spine angles that were both derived from each of the time-series of spine angular kinematics were first averaged across the two trials for each of the constrained and unconstrained conditions and then analyzed with two factor (SENSOR POSITION and CONSTRAINT CONDITION) repeated measures analyses of variance. Post hoc analyses for any significant main or interaction effects were conducted using paired t-tests. Intraclass correlation coefficients ($ICC_{2,1}$) were also determined between values of λ_{max} for each of the three combinations of spine angular kinematic time series. For $ICC_{2,1}$ analyses, values for λ_{max} from both trials under each constraint condition were used. All statistical analyses were performed in SPSS (SPSS 21, SPSS Corporation, Chicago, IL, USA). The level of significance for all analyses was set to 0.05 a priori.

3. Results

The interaction between constraint condition and sensor placement for both peak spine angle, and λ_{max} was not statistically significant (p=0.371 for peak spine angle; p=0.406 for λ_{max}); however, there were significant main effects of both constraint condition (p=0.017 for both peak spine angle and λ_{max}), and sensor placement (p=0.001 for peak spine angle; p=0.004 for λ_{max}). Increasing the height of sensor placement increased the peak spine angles (L3=25.1 ± 3.6°; L1=42.6 ± 2.8°; T11=49.9 ± 4.1°). In particular, the peak spine angles recorded from the L3 sensor were significantly lower than those recorded from either the L1 (p=0.010) or T11 (p=0.006) sensors. Positioning the second sensor at L3 produced a 13% lower estimate for λ_{max} than positioning the second sensor at either L1 (p=0.008) or T11 (p=0.031) (Table 2). There was no difference between λ_{max} values determined from the sensors placed at L1 and T11 (p=0.961). Maximum finite-time Lyapunov exponents derived from the constrained trials were 11% lower on average than those derived from the unconstrained trials (p=0.017) (Table 2).

Intraclass correlation coefficients for both unconstrained and constrained conditions showed the best correspondence in λ_{max} determined from the spine angular kinematic data obtained using the L1 and T11 locations for the second sensors ($ICC_{2,1}$=0.789 and 0.946) (Table 3). Conversely, the lowest correspondence in λ_{max} was observed between spine angular kinematic data obtained using the L3 and T11 (furthest separation distance) locations for the second sensors ($ICC_{2,1}$=0.255 and 0.559) (Table 3).