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Fatigue influences the dynamic stability of the torso

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Fatigue in the extensor muscles of the torso affects neuromuscular recruitment and control of the spine. The goal of this study was to test whether fatigue influences stability of dynamic torso movements. A controlled laboratory experiment measured the change in the maximum finite-time Lyapunov exponent, λ_{max}, before and after fatigue of the extensor muscles. Non-linear analyses were used to compute stability from the embedding dimension and Lyapunov exponent recorded during repetitive dynamic trunk flexion tasks. Torso extensor muscles were fatigued to 60% of their unfatigued isometric maximum voluntary exertion force then stability was re-measured. Independent variables included fatigue, task asymmetry and lower-limb constraint. λ_{max} values increased with fatigue suggesting poorer dynamic stability when fatigued. Embedding dimension declined with fatigue indicating reduced dynamic complexity when fatigued. Fatigue-related changes in spinal stability may contribute to the risk of low-back injury during fatiguing occupational lifting tasks. The findings reported here indicate that one mechanism by which fatigue contributes to low back disorders may be spinal instability. This information may contribute to the development of ergonomic countermeasures to help prevent low back disorders.

Keywords: low back; fatigue; stability; dynamics

1. Introduction

Lifting-induced fatigue can influence neuromuscular control of spinal stability. Endurance and fatigue of the trunk extensor muscles have been identified as risk factors for low-back pain (Biering-Sorensen 1984, Luoto *et al*. 1995), but the mechanisms of this risk are not well understood. It may be associated with compensatory muscle recruitment and changes in spinal load when fatigued (Sparto and Parnianpour 1998). It may also be associated with decrements in the ratio of lift-strength vs. task requirements (Keyserling *et al*. 1980). Recent studies suggest that fatigue of the torso muscles can affect spinal stability (Granata *et al*. 2004). If fatigue impairs stability, then small kinematic disturbances or neuromuscular errors can cause brief uncontrolled intervertebral movement and subsequent tissue strain injury. However, there are no existing direct empirical measurements of stability to test whether fatigue influences stability during dynamic torso movements.

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Neuromuscular control factors that contribute to stability are influenced by fatigue. Control of stability is maintained by three biomechanical sub-systems (Panjabi 1992a). First, the passive sub-system describes stabilising contributions from spinal ligaments, discs and bones. Second, open-loop recruitment of torso and paraspinal muscles contributes the active visco-elastic stiffness necessary to maintain stability of the spinal column (Bergmark 1989). Stiffness of muscles increases with active recruitment (Morgan *et al.* 1978). For example, co-contraction can influence the bending stiffness of the torso (Lee *et al.* 2006). Therefore, appropriate open-loop recruitment may be used to control spinal stiffness and stability (Gardner-Morse and Stokes 1998, Granata and Marras 2000). Fatigue may influence both the intrinsic stiffness of actively contracting muscles (Gollhoffer *et al.* 1987) and the muscle recruitment patterns during steady-state exertions (Sparto and Parnianpour 1998). Hence, fatigue may influence the open-loop control of spinal stability. The third sub-system for control of stability is provided by active feedback from reflex and voluntary modulation of muscle recruitment (Hogan 1984). Recent studies demonstrate that the feedback provided by reflexes in the paraspinal muscles may account for up to 40% of the stabilising control of the torso (Moorhouse and Granata 2006). Fatigue influences the reflex response of torso control (Wilder *et al.* 1996). There is a disagreement whether fatigue causes reflex amplitude to increase or decrease (Avela *et al.* 1999, Qita and Kearney 2000) but most studies agree that fatigue causes increased myoelectric and electromechanical delay (Hotobagyi *et al.* 1991, Hagbarth *et al.* 1995). Changes in feedback delay can inhibit the neuromuscular control of spinal stability (Franklin and Granata 2006). Therefore, fatigue may influence the feedback control of spinal stability. Although fatigue may influence several control sub-systems, it is difficult to predict whether these effects are sufficient to affect overall stability. Therefore, empirical measures of torso stability are necessary.

Stability of the torso can be estimated using analyses of non-linear dynamics. A time-dependent kinematic reference trajectory represents the desired time-dependent movement path of the torso. It is considered stable if all trajectories that are sufficiently close to the reference path continue to remain close for all time (Strogatz 1994). These conditions ensure that small kinematic disturbances can be controlled and attenuated. By recording the time-dependent behaviour of kinematic variance, one can quantify stability from analyses of whether or not movement trajectories approach the reference trajectory. This concept was applied to record stabilising control of the torso in a seated posture (Cholewicki *et al.* 2000). Non-linear methods can also be used to quantify stability during dynamic movements. In dynamic tasks, Lyapunov exponents are used to describe whether local kinematic errors grow in time or decay toward the reference trajectory, i.e. kinematic expansion vs. kinematic contraction. For a system moving in n generalised dimensions there must be one Lyapunov exponent λ_i for each of n dimensions. Kinematic dispersion (contraction) of the i^{th} principal axis is proportional to $e^{\lambda it}$. The Lyapunov exponent is positive if there is kinematic expansion; it is negative if the kinematic perturbations grow smaller in time, i.e. contraction. For a dynamic system to be asymptotically stable (i.e. stable in a global sense), the sum of all the Lyapunov exponents must be less than zero. Unfortunately, it is infeasible to measure all of the Lyapunov exponents from experimental time series data. However, the largest Lyapunov exponent $\lambda_{\max}$ quickly dominates the dynamic behaviour of the system and is readily computed from measured data (Rosenstein *et al.* 1993). The kinematics of each flexion and extension movement of torso can be represented by a kinematic trajectory. Therefore, empirical measurement of $\lambda_{\max}$ has proven to be a useful estimate of stability during torso movements (Granata and England 2006).

The goal of this study was to test whether fatigue influences stability during dynamic torso movements. The task was repetitive trunk flexion and extension movements in the mid-sagittal plane to repeatedly touch a target in front of the subject. Two hypotheses were tested. First, stability of dynamic trunk flexion and extension movements decreases with fatigue. Second, task asymmetry influences fatigue-related changes in dynamic stability. Task asymmetry is considered a risk factor for occupationally related low-back pain (Fathallah *et al*. 1998). Evidence suggests that asymmetry influences stability during dynamic trunk movements (Granata and England 2006). Moreover, effects of asymmetry may influence compensatory muscle recruitment associated with fatigue (Van Dieen and Heijblom 1996). Therefore, fatigue-by-asymmetry interactions were expected in the stability of flexion tasks. To address the goals and hypotheses, non-linear time-series analyses were implemented to characterise stability during repetitive trunk flexion tasks both before and immediately following fatigue of the trunk extensor muscles.

2. Materials and methods

2.1. Subjects

A total of 10 healthy adults performed repetitive dynamic trunk flexion movements. Subjects included five men and five women with no self-reported history of low back pain (Table 1). Prior to participation in the experiment, each subject provided informed consent approved by the Virginia Tech institutional review board.

2.2. Stability assessment protocol

The stability assessment required subjects to repetitively touch a target placed in the anterior mid-sagittal plane (Figure 1) as per protocols described by Thomas *et al*. (2003). The target was located so that the subject could reach it by means of 60° of trunk flexion with elbow extended, shoulder flexed 90° and lower limbs vertical. Note that the target placement was determined by these postural constraints, but during data collection trials the subjects were free to move in a self-selected manner with no constraint on arm, shoulder or torso posture. The task required the subjects to touch the flexion target with their hands then return to the upright posture. To ensure that the subjects returned to an upright posture during each cycle, a sphere was suspended such that it made contact with the posterior thorax when the subject was fully upright. Subjects moved in time with a metronome tone so as to achieve 30 flexion cycles per min. The cyclic movements were performed repetitively and without pause for a trial duration of 1 min.

Movement kinematics were recorded throughout each dynamic stability trial. Two electromagnetic sensors were secured to the skin by double-sided tape over the S1 and T10

Table 1. Subject demographics and anthropometry.

	Males	Females
Number of subjects	5	5
Age (years)	29.60 (5.23)	23.60 (4.51)
Height (cm)	178.04 (9.35)	166.07 (3.10)
Body mass (kg)	72.30 (9.04)	61.24 (12.43)

Note: Values represent mean (standard deviation) of the subject pool.

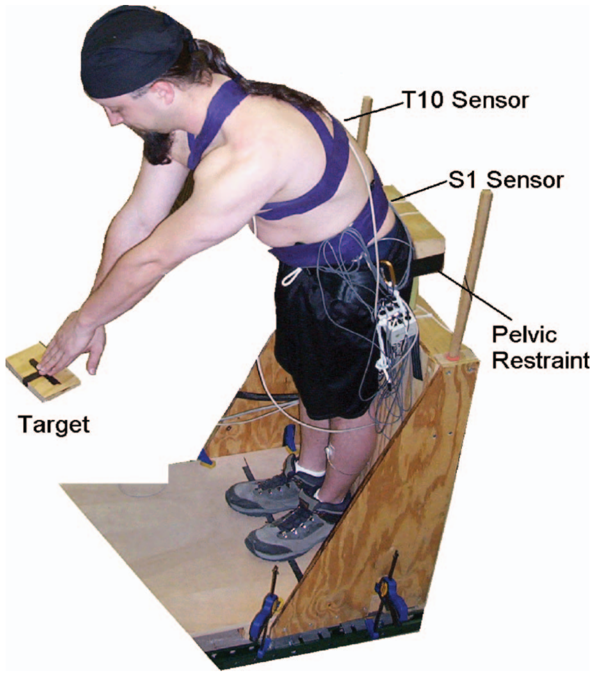

Figure 1. Stability Assessment Protocol. The task was to touch a target placed near knee level with their hands then return to the upright posture in time with a metronome tone. The task required continuously repetitive movement patterns at a rate of 30 cycles per min.

spinous processes (Motionstar ERT; Ascension Technology, Natick MA, USA). Lumbar angles were computed from the Euler rotation matrices recorded from the T10 sensor with regard to the S1 sensor at a data collection rate of 100 Hz (Granata and Sanford 2000). These provided time-series signals of 3-D lumbar angles including sagittal flexion, lateral flexion and twist during the repetitive dynamic movements.

Experimental conditions included task asymmetry, lower-limb constraint and fatigue. During symmetric conditions, subjects were required to touch the target with both hands simultaneously. During asymmetric trials they touched the target with their dominant hand only. Pilot measurements illustrated that one-handed reaching tasks produced asymmetric movement characterised by simultaneous torso flexion and axial rotation. Conditions of lower-limb constraint describe whether the subjects' legs and pelvis were constrained by strapping to a pelvic support structure. During constrained conditions, the pelvic support structure was designed to restrict the motion of the lower body, thereby isolating movement to the torso and arms (Figure 1). In the unconstrained trials the legs and pelvis were free to move normally, but the subjects were instructed to keep their knees straight. Each experimental combination of asymmetry and lower-limb constraint was performed twice, i.e. a total of eight trials before fatigue and eight trials after fatigue. Subjects were allowed to practise the movements until they were comfortable with the movement trajectory and movement pace before data collection of each trial. Experimental combinations of asymmetry and lower-limb constraint were presented in random order with at least 1 min rest between trials.

2.3. Fatigue protocol

Stability was assessed before and immediately following a fatigue protocol. The fatigue protocol was performed immediately after the non-fatiguing protocol. The protocol required subjects to perform dynamic trunk extension exercises until the maximum voluntary trunk extension strength declined to 60% of the unfatigued strength. Subjects completed warm-up exercises including lumbar stretching and back extension movements on a 45° Roman chair. Following the warm up, isometric maximum voluntary exertion (MVE) force was recorded. To record the MVE, subjects were positioned on the Roman chair and they applied a maximum isometric trunk extension force against a cable strung between the thorax and the ground (Figure 2). The cable was connected to the subject via a modified construction harness so as to resist isometric trunk extension force when the torso was in a neutral posture. A load cell joined the cable to the ground so as to measure cable tension during the exertion (Interface Force, Scottsdale, AZ, USA). Two isometric maximum exertions were performed with 1 min rest between exertions. The MVE force was recorded as the greatest peak of the two isometric trials. Following the MVE measurements, the cable was detached and fatigue was induced by performing dynamic trunk extension exercises on the Roman chair. During the initial exercise period subjects completed 10 repetitions of torso flexion–extension at a rate of 30 cycles per min in time with a metronome. After 2 min exercise, the MVE extension force was re-recorded. With each measurement of maximum force, a 5% decline in maximum force was expected. This ensures a final fatigued value of 60% MVE after 16 min exercise. If the recorded force decayed less than 4.5% during any 2-min exercise period, then the number of exercise repetitions was increased by two repetitions per min. If the decline in force was greater than 5.5%, then the number of exercise repetitions was reduced by two repetitions per min. This ensured a continuous and approximately linear fatigue onset. Detailed description and validation of the fatigue protocol is described by Davidson *et al*. (2004).

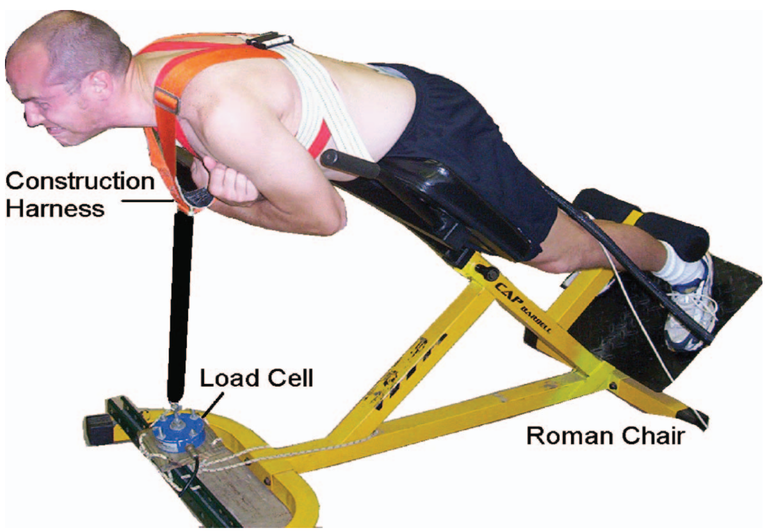

Figure 2. Fatigue Protocol. Subjects were asked to perform dynamic trunk extensions at a rate of 30 repetitions per min on a Roman chair. Fatigue was documented by recording the change in maximum trunk extension strength throughout the exercise protocol. Once strength declined to 60% of the unfatigued level, the stability assessment was repeated.

Immediately following the fatigue protocol the subjects repeated the dynamic stability assessment. The stability assessment protocol was identical to the protocol performed prior to fatigue, but there was no rest between trials so as to avoid fatigue recovery. Pilot studies indicated that there could be partial recovery after four trials of the stability assessment. Therefore, the subjects repeated the fatigue protocol after four trials of stability assessment. The re-fatigue was performed until the MVE force declined to less than 60% MVE. After completing the remainder of the stability measurement tasks trunk extension MVE was again recorded to document fatigue at the end of the stability assessment.

2.4. Analysis

Stability was quantified from the maximum finite-time Lyapunov exponent, i.e. time-dependent expansion rate of kinematic variance. Kinematic data were filtered with a 10 Hz, low-pass, second-order Butterworth filter in preparation for calculation of dynamic stability. Non-linear techniques often do not filter raw data to avoid removing physical artefact. However, the natural frequency of torso dynamics is approximately 1 Hz (Moorhouse and Granata 2005), so torso movement artefact at frequencies greater than 10 Hz are attributable to noise and thereby removed from the signal. The first five and last five cycles of each trial were removed so as to focus on the steady-state dynamic movements. The data were re-sampled to 6000 data points per 20 movement cycles so that each cycle had a mean value of 300 data points. However, note that this re-sampling process retains cycle-to-cycle variation of cycle duration.

Expansion of kinematic variability in one dimension may be compensated by contraction in another dimension. Therefore, stability analyses were performed on a time signal representing the Euclidean norm of the three Euler angles at each time interval. This resulted in a 1-D time-series vector of kinematics data, $x(t)$. However, the system under consideration must be represented with number of dimensions 'n' sufficient to represent the complete dynamics of the task (Takens 1981). An n-dimensional Euclidean space, $\mathbf{Y}(t)$, can be created by reconstructing the dynamics of a system from the single time-series vector by method of delays:

$$\mathbf{Y}(\mathrm{t}) = [x(t), x(t+\mathrm{T_d}), x(t+2\mathrm{T_d}) \ldots\ x(t+(n-1)\cdot \mathrm{T_d})] \tag{1}$$

where $\mathrm{T_d}$ is a constant time delay (Figure 3). Reconstructed dynamics represent information associated with previously unmeasured state dimensions of the complex system. The reconstruction delay, $\mathrm{T_d}$, was estimated using the average mutual information function (Fraser and Swinney 1986). This provides an appropriate time delay such that the information contained in vectors $x(t)$ and $x(t+\mathrm{T_d})$ are maximally uncorrelated (Rosenstein *et al.* 1994). Based upon these results, a constant $\mathrm{T_d} = 200$ ms was used to ensure that all trials were treated similarly. The embedding dimension, n, was determined from a global false nearest neighbours analyses (Kennel and Isabelle 1992). This method incrementally increases embedding dimension until the number of trajectory intersections is a minimum (England and Granata 2006). It was found that an average embedding dimension for the analysed data was n = 6. This indicates that at least six state variables are needed (three lumbar angles and three angular velocities) to describe the complete dynamics of the torso movements in the present study.

The largest Lyapunov exponent $\lambda_{\max}$ was used to quantify and compare the local dynamic stability of the repetitive trunk movement represented in the reconstructed

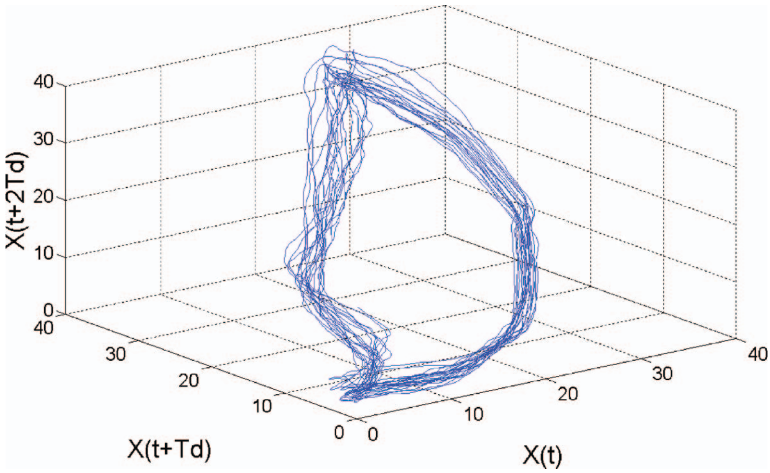

Figure 3. Reconstructed dynamics of repetitive torso flexion represented in three dimensions with $T_d = 200$ ms. Although the movement data were analysed with $n = 6$, for purposes of demonstration three embedding dimension is the largest that can be illustrated.

state-space matrix, $\mathbf{Y}(t)$. Local perturbations to a system cause displacements between kinematic trajectories. For every data point on a given trajectory (movement cycle) there is a data point on another trajectory that lies closest to it in the n-dimensional state-space. These are defined as nearest neighbours and will approach each other in time if the system is stable. The distance between each pair of nearest neighbours, $d_i(t)$, was recorded for all data points $i = 1, \ldots, p$. Kinematic dispersion can be measured by noting the change in distance $d_i(t)$ to $d_i(t + \Delta t)$ as the trajectories moved forward in time by Δt. Two randomly selected neighbouring trajectories diverge at a rate described by the largest Lyapunov exponent, λ_{max} (Rosenstein *et al.* 1993). Therefore, the maximum Lyapunov exponent was approximated as the slope of the linear best-fit line created by the equation:

$$\ln\{d_i(t + t)/d_i(t)\} = \lambda_{max} t. \tag{2}$$

The term '$\ln\{d_i(t + \Delta t)/d_i(t)\}$' represents the logarithmic of dispersion from, $d_i(t)$, averaged across all pairs of nearest neighbours, i (Figure 3). The slope of this line is λ_{max} and quantifies the mean rate of divergence of initially neighbouring trajectories along the least stable dimension. λ_{max} was recorded for each experimental trial.

Statistical analyses examined the effect of fatigue, asymmetry and lower-limb constraint on the dependent variables of λ_{max} and embedding dimension, n. Preliminary analyses revealed no significant gender differences in λ_{max} ($p = 0.960$). Therefore, data were pooled across gender. Remaining independent variables were treated as within-subject effects in a repeated measures ANOVA. Analyses were performed using commercial software (Statsoft, Inc., Tulsa, OK, USA) using a significance level of $\alpha < 0.05$.

3. Results

Fatigue was characterised by the decline in MVE recorded during isometric trunk extension tasks. The mean MVE force ($\pm$ SD) of all the subjects after the first fatigue protocol was 64.24 $\pm$ 3.92% of the unfatigued trunk extension strength. Following the fatigue protocol four stability assessment trials were performed and the MVE trunk extension strength was then re-tested. A significant strength recovery was observed after these four stability trials ($p = 0.0001$, $F(1, 9) = 52.890$). The mean of the MVE force after four trials was 75.01 $\pm$ 4.26%. At this time, subjects were re-fatigued, after which they completed the remaining four stability assessment trials. After the final stability trial, a significant recovery was again noted ($p = 0.0001$, $F(1, 9) = 49.726$). The MVE force after these final stability assessments was 73.05 $\pm$ 5.13%. Therefore, the authors are confident that the protocol successfully maintained a level of fatigue equivalent to at least 25% decrement in trunk extension strength throughout the post-fatigue stability assessments.

The average embedding dimension was $n = 6.00 \pm 0.91$. This value was used for reconstruction of the state-space and calculation of the maximum finite-time Lyapunov exponent. However, the embedding dimension for the recorded trials ranged from 3 to 7, indicating that some conditions contained greater dynamic complexity than others (Table 2). Main effect of fatigue significantly influenced embedding dimension ($p = 0.041$, $F(1, 9) = 5.709$). The embedding dimensions of the dynamic movements before fatigue, $n = 6.18 \pm 0.91$, was greater than after fatigue, $n = 5.83 \pm 0.88$ (Figure 4). This indicated reduced dynamic complexity with fatigue. However, a trend was observed in the fatigue by lower-limb constraint interaction ($p = 0.082$, $F(1, 9) = 3.830$). Post-hoc analyses suggest that the fatigue effects on embedding dimension may be limited to conditions wherein subjects were fixed to the pelvic restraint structure ($p = 0.037$). In this condition the number of embedding dimensions was greater before fatigue, $n = 6.38 \pm 0.93$, than after the fatiguing exercises, $n = 5.78 \pm 1.00$. When the flexion task was performed without lower-limb constraint the embedding dimension was not significantly affected by fatigue ($p = 0.943$).

Dynamic stability of the torso was estimated from maximum Lyapunov exponents. Large values of λ_{max} indicate less stable control of movement. λ_{max} was significantly influenced by main effect of fatigue ($p = 0.005$, $F(1, 9) = 13.950$). Mean kinematic rate of expansion before fatigue was $\lambda_{max} = 0.87 \pm 0.12$. After the fatigue protocol it was 0.94 ± 0.11, indicating poorer stability when fatigued (Figure 5). There was no significant main effect of asymmetry ($p = 0.317$, $F(1, 9) = 1.124$) or lower-limb constraint ($p = 0.282$, $F(1, 9) = 1.308$). Contrary to the second hypothesis, there was no significant interaction between fatigue and asymmetry. However, a significant interaction between asymmetry and lower-limb constraint was observed ($p = 0.047$, $F(1, 9) = 5.270$). When the movement was isolated to the torso and arms, asymmetric tasks were associated

Table 2. Statistical table for embedding dimension (n) and largest Lyapunov exponent (λ_{max}).

	Embedding dimension n		Lyapunov exponent λ_{max}	
	F	*p* level	F	*p* level
Fatigue	5.709	**0.041**	13.950	**0.005**
Lower-limb constraint	0.663	0.437	1.308	0.282
Asymmetry	0.109	0.749	1.124	0.317
Fatigue × constraint	3.830	0.082	0.457	0.516
Fatigue × asymmetry	0.159	0.708	1.103	0.321
Constraint × asymmetry	3.641	0.089	5.270	**0.047**

Note: Numbers represent the type I statistical error, i.e. *p*-value. Significant effects ($p < 0.05$) are highlighted in bold. Three-way interactions were not statistically significant.

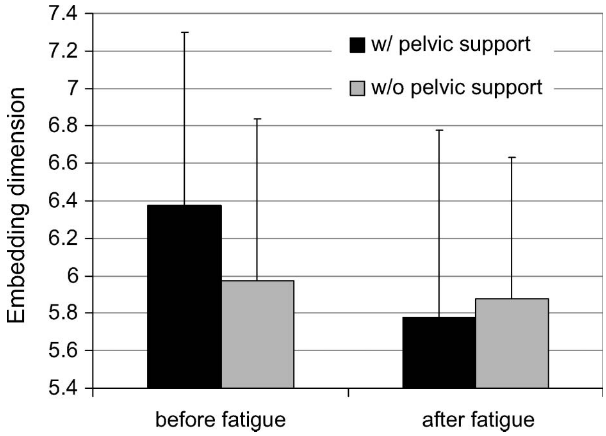

Figure 4. The embedding dimension 'n' indicates the dynamic complexity of process. The number of embedding dimensions 'n' decreased with fatigue. (The errors bars are the standard deviations of the mean.)

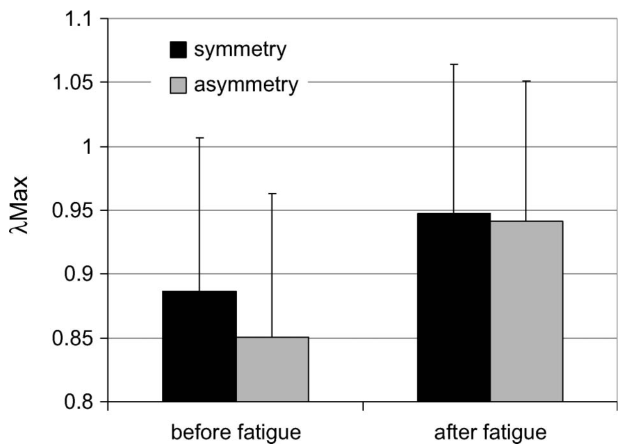

Figure 5. Large λ_{max} values indicate poorer stability. λ_{max} values were greater for the fatigued movements than for the unfatigued movements. (The errors bars are the standard deviations of the mean.)

with a smaller $\lambda_{max} = 0.90 \pm 0.15$ than symmetric tasks, 0.94 ± 0.14 ($p = 0.053$). Furthermore, λ_{max} for the symmetric movements with lower-limb constraint, 0.94 ± 0.14, was greater ($p = 0.017$) than symmetric movements without lower-limb constraint, 0.89 ± 0.10.

4. Discussion

Workplace and personal factors that affect spinal load and stability contribute to mechanisms of biomechanical risk. Damage to the spine occurs when biomechanical loads exceed injury tolerance. Spinal loads during occupational lifting tasks are recognised as an important biomechanical factor risk for low-back pain (Norman *et al.* 1998). Fatigue causes muscle recruitment to shift from the paraspinal muscles to more laterally situated muscles of the torso but it remains unclear how this influences spinal load (Van Dieen and Heijblom 1996, Sparto and Parnianpour 1998). However, spinal load must be considered with regard to the loading tolerance. Multi-segment structural loading tolerance of the spine is measured in terms of its stable buckling load (Bergmark 1989). If spinal load exceeds this stability threshold then the spine cannot maintain its alignment (Crisco and Panjabi 1992, Crisco *et al.* 1992) and a small disturbance or muscle recruitment error would cause brief uncontrolled intervertebral movement (Panjabi 1992b). This may cause passive tissue strain injury and/or nerve-root impingement. Therefore, biomechanical tolerance to spinal overload injury can be attributed, in part, to the ability of the neuromuscular system to maintain spinal stability. Fatigue influences the neuromuscular control factors that contribute to spinal stability (Hagbarth *et al.* 1995, Sparto and Parnianpour 1998). Moreover, recruitment errors associated with fatigue may contribute to disturbance perturbations that precipitate unstable intervertebral movement. However, methods to empirically test the hypothesis that fatigue affects stability have only recently been developed.

Rosenstein *et al.* (1993) developed a method to calculate the maximum Lyapunov exponent, λ_{max}, from experimental time-series data. This non-linear dynamic analysis has previously been used to quantify dynamic stability of walking (Dingwell and Cusumano 2000). In measurements of walking, λ_{max} increases with subjective fatigue assessment (Yoshino *et al.* 2004). These non-linear analyses can also be used to quantify the stability of the torso during dynamic movements (Granata and England 2006) but the effects of fatigue have not been tested. The Lyapunov exponent represents the rate at which kinematic disturbances change with time. Clearly, the trunk flexion movements remained stable because the subjects successfully completed the experimental tasks. Therefore, during repetitive torso movements the stabilising neuro-control system will cause the dynamics to be attracted toward the reference movement trajectory. This guarantees that the sum of Lyapunov exponents was less than zero, i.e. n-dimensional volume of kinematic dispersion contracts with time. However, while disturbances are attenuated in one dimension, they may grow in another, as noted by positive values of λ_{max}. These unstable dimensions may be characterised as a member of the uncontrolled manifold (Scholz and Schoner 1999). Specifically, the neuromuscular controller does not need to ensure stability in all of the dynamic movement dimensions to achieve stability of the repetitive task (Nayfeh and Balachandran 1995). In fact, theoretical analyses suggest that optimal neuromotor behaviour may allow kinematic variance in uncontrolled dimensions (Todorov and Jordan 2002). Results suggest that the number of controlled dimensions was affected by fatigue.

The number of embedding dimensions decreased with fatigue of the trunk extensor muscles. Simple dynamical systems can be described with few state dimensions. Conversely, a larger number of embedding dimensions are required to describe complex dynamic systems, i.e. the number of state variables of a system increases with the dimensions. The average integer value of state dimensions, $n = 6.0$, agrees with previous measurements of torso dynamics (Granata and England 2006). It suggests that the dynamic state at any instant in time requires a set of three lumbar angles and three angular velocities. Hence, sagittal plane dynamic analyses overlook important mechanical coupling. Fatigue appears to influence this coupling. When the extensor muscles were fatigued, the reduction of embedding dimensions indicates the reduction in complexity of the system (Figure 4). This may simplify the control of fatigued movements because systems with less complexity are intuitively easier to control. This effect may be explained by changes in muscle recruitment. When the paraspinal muscles become fatigued the recruitment shifts to greater co-activation of the lateral torso muscles to generate the trunk extension movement (Van Dieen and Heijblom 1996). The oblique line of action of these muscles clearly provides the capacity to couple sagittal with lateral and axial control. Recruitment of these muscles as a consequence of fatigue may have resulted in constraining the torso movements to a lower dimensional state space, causing the reduction in embedding dimension. However, analyses were limited to integer dimensions. Fractal dimensions are often observed in non-linear dynamic systems, indicating deterministic chaos and strange attractor stability (Nayfeh and Balachandran 1995). In fact, fractal dimensions are common in physiological signals and provide predictive risk assessment (Goldberger *et al.* 2002). Further analyses should investigate the fractal characteristics of torso dynamics and its association with fatigue.

The maximum Lyapunov exponent, λ_{max}, increased with fatigue. This suggests that the torso was less stable after the fatiguing protocol than during unfatigued measurements. The result agrees with the hypothesis. This effect may be attributed to previously reported fatigue-related changes in open-loop recruitment and changes in feedback neuro-control. However, results are limited to the interpretation of the λ_{max} coefficient. Specifically, the maximum Lyapunov exponent, λ_{max}, characterises the maximum time rate of expansion for the *n*-dimensional volume that describes kinematic variability. In other words, this value represents the least stable aspect of the movement dynamics (Rosenstein *et al.* 1993). Recognising that kinematic dispersion of the i^{th} principal axis is proportional to $e^{\lambda it}$, the positive values of λ_i quickly dominate the system dynamics. Unfortunately, with current mathematical techniques it is not possible to determine the direction of the least stable axis of movement. The results cannot guarantee that the least stable dimensions were along the same axes in fatigued and unfatigued conditions. For example, it is reasonable that the least stable dimension during symmetric unfatigued movements may exist along the twisting axis of the torso; whereas, increased lateral muscle recruitment may improve axial control and the least stable dimension in fatigued conditions may have been aligned with the sagittal flexion axis. Moreover, the least stable direction may change with flexion angle throughout the dynamic movement trajectory. The authors hope to pursue further studies that will attempt to identify the local vector characteristics of instabilities, thereby providing potential insight into the weakest control direction. Nonetheless, results demonstrate that the ability of the neuromuscular system to attenuate kinematic errors is impaired by fatigue of the trunk extensor muscles.

Contrary to the second hypothesis, task asymmetry did not influence the fatigue-related change in stability. There were no significant main effects of task asymmetry. This was in contrast to previous studies wherein movements in the mid-sagittal plane were less stable than when moving in a combined sagittal and twist trajectory (Granata and England 2006). However, the asymmetric movement protocols in these two studies were different. Specifically, the asymmetric movement in the current protocol required touching a target in the mid-sagittal plane, whereas previous experimental designs placed the target to the left or right of the mid-sagittal plane. Moreover, in previous studies, lower-limb constraint was imposed in all experimental conditions by strapping subject's legs and pelvis to a rigid structure so as to limit movement solely to the torso and arms. In this regard, results from the current study agree with previous analyses. A significant interaction between asymmetry and lower-limb constraint was observed in the current results. When the movement was isolated to the torso and arms, there was a trend wherein asymmetric tasks were more stable than symmetric tasks. No effect of asymmetry was observed when subjects were free standing. This interaction indicates that dynamic coupling between the legs and torso contributes to the control of stability in asymmetric movements. Similarly, pair-wise analyses of embedding dimension revealed that fatigue-related changes in dynamic complexity were observed only if the legs were constrained. Future studies should investigate the dynamic coupling and control between the lower limbs and torso stability.

Results provide insight into the mechanisms of fatigue, but limitations of the experimental design and analyses must be considered. The protocol rapidly introduced a very high level of fatigue. The protocol caused isometric MVE strength reduction to 60% of unfatigued strength within 16 min of exercise. This was done to investigate the fundamental hypotheses of fatigue and stability. However, workplace tasks are more likely to produce lower levels of fatigue over long time durations, e.g. accumulation of fatigue throughout an 8 h work shift. Extrapolation of results to workplace conditions requires further experimental studies. Similarly, the experimental protocol was a single task, paced movement without a load in the hands. Manual materials handling often require multiple task components with a lifted load in the hands. Moreover, previous analyses showed that movement pace influences stability (Dingwell and Marin 2006, Granata and England 2006). The dynamic analyses can be extended to investigate multiple component movement patterns. The effect of lifting a box must be considered for a range of lifted weights. Small variations in embedding dimension, n, and time delay, T_d, were observed in each trial. However, the analyses implemented $n = 6$ and $T_d = 200$ ms to ensure consistent interpretation of every trial. Treating these parameters as trial-dependent co-factors may provide further information content when assessing dynamic stability using λ_{max}. Evidence suggests that fatigue influences muscle recruitment. However, the present study did not investigate empirically the changes in muscle recruitment patterns due to lumbar fatigue.

In conclusion, fatigue of the trunk extensor muscles has been identified as a risk factor for low-back injury (Biering-Sorensen 1984, Luoto *et al.* 1995). Theoretical analyses suggest that the mechanism of this risk may be related to stability (Granata *et al.* 2004). Spinal instability is a common clinical condition associated with low-back pain (Pope and Panjabi 1985), wherein an unstable spine is susceptible to intersegmental hypermobility and tissue strain injury (Panjabi 1992b). Results demonstrate that fatigue of the trunk extensor muscles impairs the neuromuscular stabilising control of dynamic torso movement. Further studies must be conducted to examine whether torso and spinal stability can be used as a risk factor for discrimination and prediction of low-back pain.

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