PUSHING/PULLING EXERTIONS DISTURB TRUNK POSTURAL STABILITY

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ABSTRACT

The stability of the spine can be estimated from kinematic variability and nonlinear analyses of seated balance tasks. However, processing methods require sufficient signal duration and test-retest experiments require that the assessment must be reliable. Our goal was to characterize the reliability and establish the trial duration for spine stability assessment. Stationarity, kinematic variability and nonlinear dynamic stability were quantified from kinetic and kinematic data collected during balance performance. Stationarity results showed that a minimum 30 seconds test duration is necessary. Intra-session reliability was excellent, however inter-session reliability needed more test trials to achieve excellent reliability.

Few studies have investigated the spinal stability during pushing and pulling exertions. Past studies suggest that the spine can be stabilized by paraspinal muscle stiffness as well as reflexes. We hypothesized that the stability of the spine decreases with exertion force and decreases during pushing more than during pulling exertion. Kinematic variability and nonlinear dynamic stability measurements were quantified from the balance performance during isometric pushing and pulling tasks. Results demonstrated that spinal stability decreased with exertion force and decreased a greater amount during pushing task than during pulling task. Stiffness alone may be insufficient to stabilize the trunk. Results may be able to be explained by slower

reflex delay. The results suggested that pushing and pulling exertions have a potential risk of low-back disorders.

ATTRIBUTIONS

The research described here was made possible through contributions from lab mates, family members, and friends.

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For teaching me, challenging me and having confidence in me. Thank you for being an excellent advisor.

Students of the Musculoskeletal Biomechanics Lab

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My parents and younger brother, for their support through this process and throughout my life. You inspired me to go on for higher education.

TABLE OF CONTENTS

List of Tables

List of Figures

Chapter 1 – Introduction

Repetitive lifting is a significant risk factor for occupational low-back disorders (Andersson, 1981; Marras et al., 1995). As a result, industry is rapidly changing the workplace to reduce lifting and frequently replaces it with pushing/pulling exertions. However, pushing/pulling exertions are also associated with low-back disorders (Damkot et al., 1984; Hoozemans et al., 1998). Thus, the injury rate is expected to increase with the trend toward a growing number of push and pull-related exertions in the workplace. To date, the biomechanics and neuromuscular control of pushing/pulling exertions remain poorly understood (Schibye et al., 2001; van der Beek et al., 1999).

The spine is stabilized, in part, by stiffness of the paraspinal muscles during active exertions (Granata & Wilson, 2001). It has often been assumed that spinal stability improves with exertion effort because stiffness increases with exertion force (Kearney & Hunter, 1990; Morgan, 1977; Moorhouse & Granata, 2005). However, it has recently been shown that stiffness alone may be insufficient to stabilize the torso in upright postures (Franklin & Granata, 2006; Moorhouse & Granata, 2005). Evidence suggests that muscle reflexes also play an important role in the stability of the spine. Reflex delay significantly impairs the control of spinal stability (Franklin & Granata, 2006) and reflex delay increases with exertion force (Granata et al, 2004). Therefore, considering the role of reflex delay on spinal stability, it is unclear whether spinal stability always improves with exertion effort. To investigate this, spinal stability changes under different force conditions will be studied.

Spinal instability is a significant risk factor of low-back disorders (Nachemson, 1985; Panjabi, 1992). Because trunk stability may differ between pushing/pulling exertions, one may have a greater injury risk than the other. Estimates of spinal stability from a biomechanical model suggest that the spine may be less stable during isometric pushing exertions than during pulling efforts (Granata & Bennett, 2005). However, we are aware of no existing empirical measurements to test whether stability is different during pushing versus pulling exertions. Thus, spinal stability differences between pushing/pulling exertions will be studied.

Current trends in clinical rehabilitation attempt to characterize torso stability and improve dynamic lumbar stabilization (Richardson et al. 2002; Saal & Saal, 1989). Therefore, reliable

measures of torso stability are required in both clinical and research environments. Reliability can be measured by intra-class correlation measures. This intra-class correlation reliability can only be achieved if individual trial duration achieves process stationarity. The trial duration required for process stationarity and reliable measures is unknown. This study investigates the necessary trial duration for reliable stability measurements.

SPECIFIC AIMS

1. Find the minimum test duration necessary for process stationarity of torso stability assessment.

2. Estimate the intra-session and inter-session reliability of stability measures recorded from seated postural torso movement.

3. Test empirically whether neuromuscular control of stability is affected by exertion force and exertion direction.

HYPOTHESES

1. Dynamic control of spinal stability will decrease with increased trunk flexion or exertion forces.

2. Spinal stability is lower in pushing exertions than in pulling exertions.

Chapter 2 - Background

2.1 The design of wobble chair

Cholewicki (2000) showed that spinal stability can be assessed empirically using an unstable chair design. In this study, subjects were asked to maintain a seated upright posture. During the balance performance, small biomechanical and/or neuromotor disturbances continuously disturbed the control system causing kinetic and kinematic variances. The lower extremity is fixed to a seat-pan. Thus, the neuromuscular control system maintains a stable posture by actively recruiting trunk muscles to return the disturbed posture toward the equilibrium state. The equilibrium state is defined as upright and zero-velocity posture regardless of lower extremity movements. Hence, spinal stability can be measured by his wobble chair test by recording the rate at which the kinematics are attracted toward the posture of static equilibrium.

Our own wobble chair was an adaptation of the design by Cholewicki (2000) to study postural control of the spine (Figure 2.1). It has four springs to support the body and the seat-pan weight. The springs were located on the four corners of an aluminum plate under the seat-pan. Their location was adjusted to change the system potential energy including seat-pan weight and subject's body weight and height. A ball bearing was fixed to the center of the aluminum plate. It fixed this center of the aluminum plate, but allowed it to pivot in any direction.

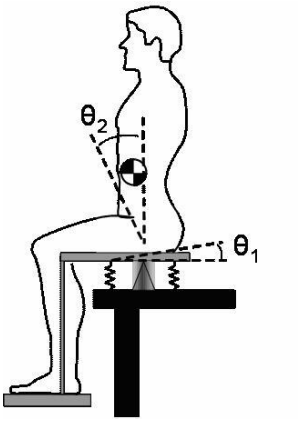


Figure 2.1. Wobble chair design. This wobble chair was modified Cholewicki's unstable chair design (2000). θ_1: seat-pan movement angle; θ_2: system center of mass movement angle.

The equations below were used to find the spring location, L, for balanced upright posture. The subject's trunk and the seat-pan move independently during the performance of a stability test. During the calculation of spring location, the trunk and spring angles were fixed and consider as a rigid body. The spring location was calculated from the free body diagram. The wobble chair system can be simplified as a single pendulum representing the trunk and seat-pan (Figure 2.2). The human control system is not accurate because the human can not maintain a constant force to correct feedback errors from sensory systems. Thus, subjects show dynamic movement necessary to balance posture. From this dynamic movement, we can derive a moment equation about pivot point O.

$$Mgh\sin(\theta) - KdL\sin(\theta) = I\alpha \qquad \text{Eq. 1}$$

where I: system inertia; α : system angular acceleration; M: system center of mass including a subject and a seat-pan; g: gravitational acceleration (9.8m/s^2); h: center of mass height from O; θ: center of mass and seat-pan movement angle from vertical and flat, respectively; K: spring stiffness (38 lbf/in); d: shorten spring length; L: spring location from the pivot point.

At the steady state condition, acceleration, α , is zero. Thus, equation 1 is represented as

$$Mgh\sin(\theta) = KL\cos(\theta)\sin(\theta) \qquad \text{Eq. 2}$$

where $d = L\cos(\theta)$. During balance performance on the seat-pan, movement is minimal, i.e. maximum movement range is -10° to 10°. Hence, we can apply a small angle assumption and equation 2 can be expressed as

$$Mgh = KL^2 \qquad \text{Eq. 3}$$

A spring location, L, can be calculated from equation 3.

$$L = \sqrt{\frac{Mgh}{K}} \qquad \text{Eq. 4}$$

In this equation, it is necessary to know Mgh. It can be calculated from the net difference moments between two postures, 10° forward tilt and 10° backward tilt. These two angles are

achieved by using wedges, angled 10°. In these two postures, trunk support helps maintaining upright trunk posture by keeping touch back to support (Figure 2.3). Therefore, the whole system, a body and the seat-pan, can be assumed to be a rigid body at, ±10°. At these two postures, the moment equations can be derived from equation 1;

$$\begin{aligned} Mgh\sin(10^\circ) &= M_1 \\ Mgh\sin(-10^\circ) &= M_2 \end{aligned} \qquad \text{Eq. 5}$$

where M_1 and M_2 are moments recorded from the force-plate at the forward and backward postures, respectively. System potential energy, Mgh, can be calculated from Eq. 5.

$$Mgh = \frac{M_1 - M_2}{\sin(10^\circ) - \sin(-10^\circ)} \qquad \text{Eq. 6}$$

A spring position can be calculated from Eq. 4 and Eq. 6. In this study, we call this spring position a 100% stability position. Various stability spring positions can be calculated from equation 3.

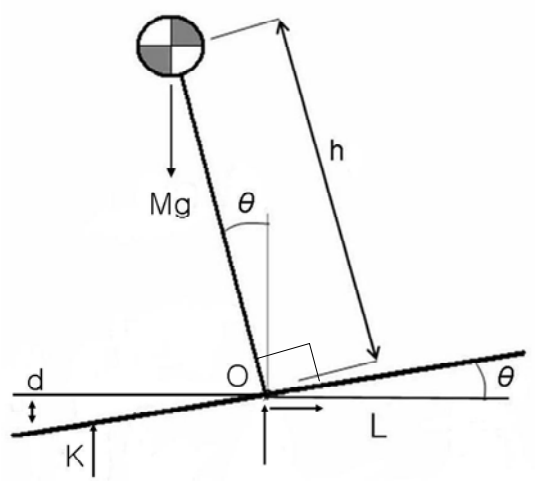


Figure 2.2. Free body diagram of wobble chair and subject. *O*: pivot point; *M*: system center of mass including a subject and a seat-pan; *g*: gravitational acceleration (9.8m/s^2), *h*: center of mass height from *O*; *θ*: center of mass and seat-pan movement angle from vertical and flat, respectively; *K*: spring stiffness (38 lbf/in); d: shorten spring length; *L*: distance from *O* to spring.

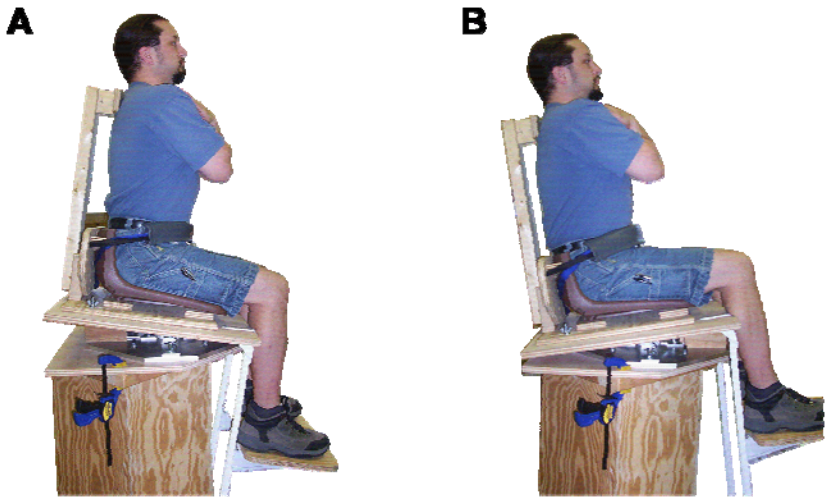


Figure 2.3. Two postures. (A) 10° forward tilt and (B) 10° backward tilt.

Ex.) 75% stability position L

$$0.75 \times Mgh = KL^2$$
$$L = \sqrt{\frac{0.75 \times Mgh}{K}}$$

2.2 Stationarity

Stationarity is defined as a process whose statistical properties do not change with time (Anderson 1971). In other words, the probability distributions of the process are time-invariant. In the case of postural sway if a process is non-stationary, there must be a difference between average initial orientation of the body and its average final orientation (Carroll 1993). This means that non-stationarity distorts the result of a time average. Thus, statistical inference regarding postural control behavior from empirical time series data requires that the dynamic process must achieve statistical stationarity.

Stationarity can be classified as first-order stationary and second-order stationary. A process is defined as first-order stationary if the mean, i.e. ensemble average, is a constant (no trend). In other words, the mean is independent of any time shift.

$$\overline{X}(t) = \overline{X}(t+\tau) = \mu \qquad \text{Eq. 7}$$

where $\overline{X}(t)$ is ensemble average at time t, τ is a time shift and μ is a constant. A signal is defined as second-order stationary if the mean is a constant and the variance is also constant (Chatfield 1984).

$$\begin{aligned} \overline{X}(t) &= \overline{X}(t+\tau) = \mu \\ Var(t) &= Var(t+\tau) = \rho \end{aligned} \qquad \text{Eq. 8}$$

where $Var(t)$ is ensemble variance at time t, τ is a time shift and ρ is a constant. This second-order stationarity implies that the process is first-order stationary, i.e. a constant mean. In this thesis, first-order stationary will be calculated.

2.3 Stability measures

2.3.1 Kinematic variability

Stability is defined as the capacity to maintain equilibrium posture of the spine and torso despite the presence of neuromuscular or kinematic disturbances (Granata & England, 2006). Many studies have measured kinematic variability of postural control to study postural stability. However, continuous biomechanical tasks such as postural control are dynamic movement wherein the neuromuscular system acts to control and attenuate input disturbances (neuromuscular errors, external disturbance forces, and kinematic perturbations) so as to recover movement errors by attraction toward the equilibrium state. Unfortunately, kinematic variability does not account for this neuro-dynamic mechanism.

Measurements of kinematic variability generally include a total center-of-pressure (CoP) path length traveled per second (PATH) (Eq. 10-a) and root mean square (RMS) (Eq. 10-b) and 95% ellipse area (EA) (Eq. 10-c) of CoP.

$$PATH = \sum_{i=1}^{N-1} \frac{\sqrt{\left[CoP_x(i+1) - CoP_x(i)\right]^2 + \left[CoP_y(i+1) - CoP_y(i)\right]^2}}{Duration} \qquad \text{Eq. 10-a}$$

$$RMS = \sqrt{\frac{\sum_{j=1}^{N} CoP_j(i)^2}{N}} \qquad \text{Eq. 10-b}$$

$$EA = 2\pi 3\sqrt{(S_x)^2(S_y)^2 - (S_{xy})^2}$$
$$S_{xy} = \frac{1}{N}\sum_{i=1}^{N} Cop_x(i)CoP_y(i) \qquad \text{Eq. 10-c}$$

where i = index of time sample, $j = x$ (antero-posterior), y (medio-lateral) and $r = \sqrt{x^2 + y^2}$, N is total data points, S_x and S_y are standard deviation of CoP_x and CoP_y time series and S_{xy} is covariance;

It is commonly assumed that these measurements represent stability. Investigation of low-back disorders is allowed by comparison to healthy controls with variability-base analyses (Radebold, 2001). However, these kinematic variability measurements are the record of the disturbance amplitude observed at the output of the human balancing performance. In other words, they represent average amplitude of output variance (RMS), a total disturbance distance (PATH), and the size of involuntary multi-dimensional movement (EA) (Prieto et al, 1996). Therefore, they ignore the time dependent attenuation of input disturbances. Thus, variability and stability should be considered independently (England & Granata, 2005, Li et al., 2005).

2.3.2 Measurements of Stability

Variability of kinematic measurements has been broadly used as a method of evaluating stability to study postural stability. However, they did not account for the time dependent

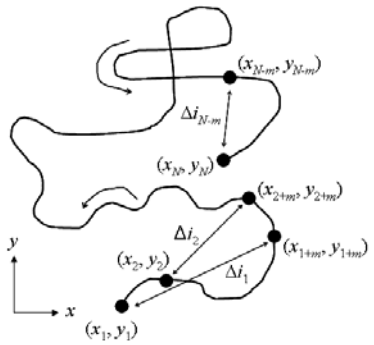
attenuation of disturbances. Therefore, new analysis methods, i.e. stabilogram diffusion analysis (SDA), Hurst rescaled range analysis (HRRA) and Lyapunov exponent, were proposed.

2.3.2.1 Stabilogram Diffusion Analysis (SDA)

Collins and De Luca (1993) proposed the stabilogram diffusion analysis (SDA) to study the characteristics of center-of-pressure (CoP) trajectories. The idea of the SDA started from Brownian motion which was studied by Einstein (Einstein 1905). Einstein showed that the mean square displacement $\langle \Delta i^2 \rangle$ was related to the time interval Δt by the expression;

$$\langle \Delta i^2 \rangle = 2D\Delta t \qquad \text{Eq. 11}$$

where $i = x, y, r$, $\langle \Delta r^2 \rangle = \langle \Delta x^2 \rangle + \langle \Delta y^2 \rangle$, parameter D is the diffusion coefficient and brackets $\langle\ \rangle$ indicate ensemble average over time or an ensemble average over a large number of samples. $\langle \Delta i^2 \rangle$ can be calculated by averaging the square of the displacements between all pairs of points separated in time by a specified time interval Δt (Eq. 13) (Figure 2.4).



For a given Δt (spanning m data intervals):

$$\langle \Delta i^2 \rangle_{\Delta t} = \frac{\sum_{j=1}^{N-m} (\Delta i_j)^2}{(N-m)} \qquad \text{Eq. 13}$$

where i = x, y, r, N is total data points.

(Collins and De Luca, 1993)

Figure 2.4. CoP trajectory of two dimensions and equation for calculation of $\langle \Delta i^2 \rangle$

The Einstein's relation (Eq. 11) is generalized to the scaling law when statistical mechanics of time series data are fractal Brownian motion (Medelbrot & van Ness, 1968):

$$\langle \Delta i^2 \rangle \sim \Delta t^{2H} \qquad \text{Eq. 12}$$

where the scaling exponent H is a real number in the range $0 < H < 1$. This scaling exponent can be easily calculated from a log-log plot of the mean square displacement versus Δt curve. A log-log plot of postural sway $<\Delta i^2>$ versus Δt demonstrates bi-linear behavior, i.e. two values of H including a short-term (H_s) and long-term (H_l) scaling exponent (Figure 2.5). However, Delignieres (2003) pointed out that this bi-linear behavior exists because the signal is bounded, i.e. the CoP must remain within the extent of base support. In addition, Radebold (2001) reported that only the short term exponent discriminates between subject groups with and without low-back pain. Thus, in this study only the short-term region scaling exponent H_s will be used to quantify postural stability.

The interpretation of the scaling exponent is that the value $H = 0$ indicates the data are white noise which is a random signal (or process) with a flat power spectral density, $H = 0.5$ is observed for classical Brownian motion which is the random movement. A value of $H > 0.5$ is associated with a persistence phenomenon where a future movement direction tends to be positively correlated with a current movement direction. $H < 0.5$ is associated with anti-persistence where a future movement direction tends to be negatively correlated with a current movement direction. These two characteristics, persistence and anti-persistence, seem to be well matched to the stability definition. Thus, these parameters can be used to quantify postural stability.

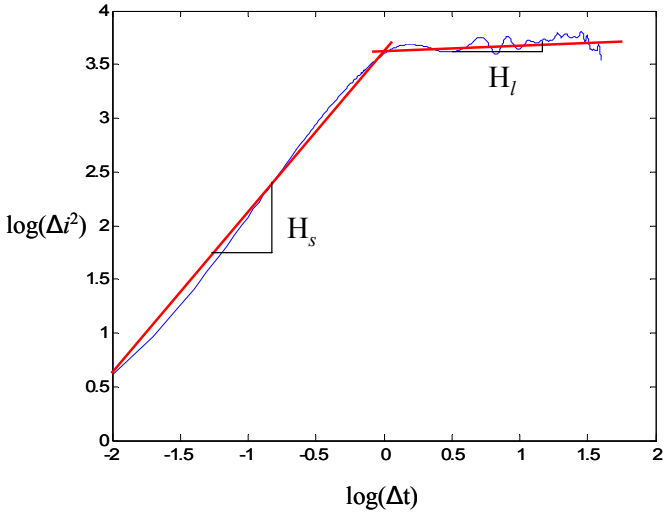


Figure 2.5. Example of a log-log stabilogram diffusion plot.

2.3.2.2 Hurst rescaled range analysis (HRRA)

Another method for measuring stability is Hurst rescaled range analysis (HRRA). HRRA is a very common method to find fractal property, widely used in econometry, geophysics, biology, and motor control (Delignieres, 2003). Unlike SDA this method does not have boundary limitation and it has the same characteristics as SDA: persistence and anti-persistence. These characteristics are found by calculating the scaling exponent ($H_{R/A}$) of HRRA.

The Hurst scaling exponent ($H_{R/A}$) can be calculated by assessment of the range of displacements of the locally integrated time series within each interval (Eq 14). Each interval is nonoverlapping intervals of length *n* divided from the time series of *N* numbers as *x(t)*.

$$X(t,n) = \sum_{k=1}^{t} \left\{ x(k) - \langle x \rangle_n \right\} \qquad \text{Eq. 14}$$

where $X(t,n)$ is an integrated series, $\langle x \rangle_n$ is the local average of the n data:

$$\langle x \rangle_n = \frac{1}{n} \sum_{t=1}^{n} x(t)$$

The difference between the maximum and minimum of the integrated data over a time interval Δt is defined as the range *R* for each interval Δt (Eq 15).

$$R = \max X(t,n) - \min X(t,n) \qquad \text{where } 1 \le t \le \text{n} \qquad \text{Eq. 15}$$

This range is then normalized by the local standard deviation of the original series *x(t)*. This calculation is repeated over all possible interval lengths, Δt. Delignieres(2003) recommended that in practice the shortest interval length is around 10 data points and the largest is *N*/2 . The R/S is related to interval length n by a power law (Eq 16).

$$R/S = (an)^{H_{R/S}} \quad \text{where } a \text{ is constant and } n \text{ is interval length} \qquad \text{Eq. 16}$$

Thus, the scaling exponent ($H_{R/S}$) of HRRA can be estimated as the slope of the log-log plot of *R/S* (Figure 2.6), as a function of interval length.

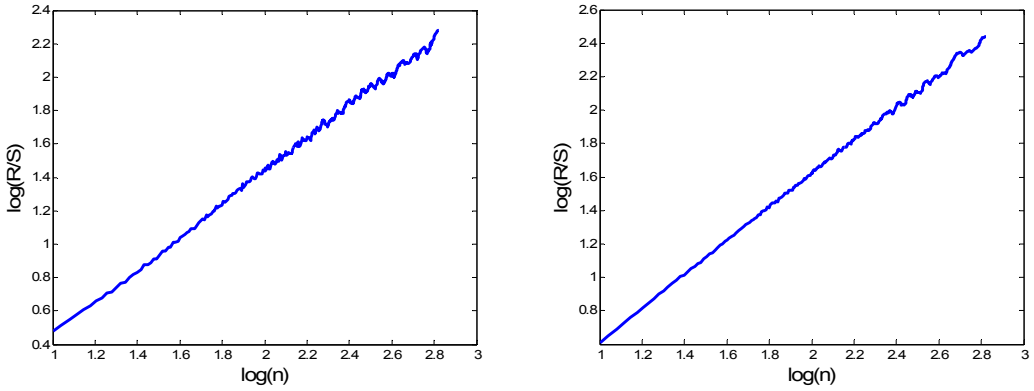


Figure 2.6. Hurst rescaled range analysis result plots; left plot: CoP; right plot: angle

The $H_{R/A}$ can vary between 0 and 1. A value $H_{R/S} = 0.5$ indicates the data are white noise. $H_{R/S} = 1$ represents Brownian motion. This means that the rescaled method shifts the H value by 0.5 compared to the exponent of SDA. Thus, the relationship between exponent H and $H_{R/S}$ is $H = H_{R/S} - 0.5$. The interpretation of modified rescaled range exponent H is same as exponent H of SDA with 0.5 subtracted. In other words, a value $H < 0.5$ means a persistent behavior and $H > 0.5$ means an anti-persistent behavior.

2.3.2.3 Lyapunov exponent

Lyapunov exponents quantify local dynamic stability determined by measuring the rate of exponential divergence of two nearest neighbors. The dynamic state at each instant in time was described by a four-dimensional vector including position and velocity in the anterior-posterior position and medio-lateral position, $\vec{q}(t) = [\theta_x, \theta_y \dot{\theta}_x, \dot{\theta}_y]$. A nearest neighbor is defined as the point which has the minimum Euclidean distance from the reference point and was found on separates orbits of this four dimensional state space similar to Figure 2.7 (A). The Euclidean distance is defined by the equation:

$$d(t) = \left\| q(t) - q(t^*) \right\|_2 \qquad \text{Eq. 17}$$

where *q(t)* is a four dimensional data point on the reference state-space trajectory, $q(t^*)$ is the nearest neighboring point on an adjacent trajectory and *d(t)* is the Euclidean distance between the each pair of points (Fig 2.7 (B)). The Euclidean distance, *d(t)*, represents a small disturbance from the equilibrium condition at time t. In order to reduce effects of random noise we computed the average distance to multiple nearest neighbors for each reference point (Kantz & Schreiber, 2004). Preliminary analyses demonstrated that the average of multiple nearest neighbors reduced the effects of stochastic noise in the signal. The average of more than three nearest neighbors failed to improve signal quality. Thus, in this thesis, *d(t)* represents the average of three initial nearest neighbors. The exponential divergence of these initial nearest neighbors is calculated for each data point through the time series (Eq. 18).

$$d(t+\Delta t) = d(t)e^{\lambda t} \qquad \text{Eq. 18}$$

where λ is the exponential divergences for each dimension at time t. The average system dynamics are quickly dominated by the maximum Lyapunov exponent, λ_{max} (Rosenstein et al., 1993). λ_{max} is determined from the mean divergence estimated by averaging the exponential divergence of all nearest neighbors. λ_{max} is defined by the equation:

$$d(t) = d_0 e^{\lambda_{\max} t} \qquad \text{Eq. 19}$$

where d_0 is the initial average distance. This λ_{max} is calculated as the slope (via the least squares "polyfit" command in Matlab) of *y(t)* in the linear region (Eq. 20). From the preliminary test, the linear region appeared in the range of 0.2 to 0.7 sec (Fig 2.8). Thus, λ_{max} is calculated at the region from 0.2 to 0.7 sec.

$$\lambda_{\max} = y(t) = \frac{1}{t}\ln\left(\frac{d(t)}{d_0}\right) \qquad \text{Eq. 20}$$

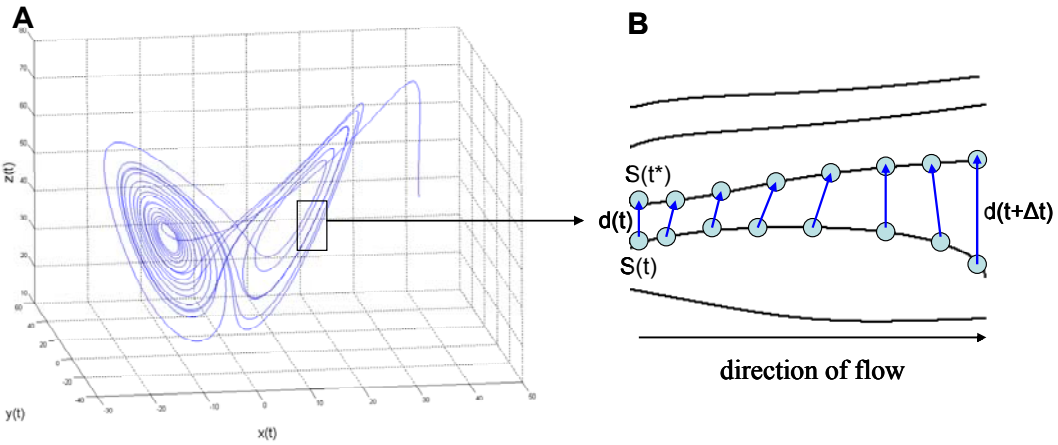


Figure 2.7. Schematic representation of state space construction. A traditional Lorenz attractor (A) and expanded view of a typical local region (B)

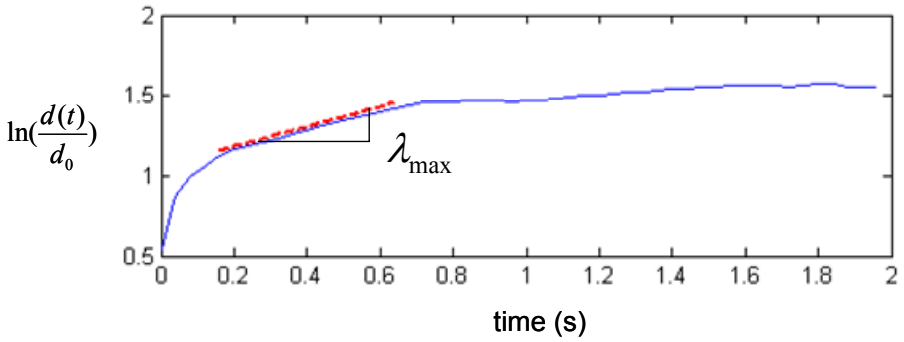


Fig 2.8. Example of maximum Lyapunov exponent for wobble chair test

Both SDA and HRRS have same interpretation of results and seem very similar. However, HRRA includes integration unlike SDA. Thus, HRRA avoid bi-linear characteristics of SDA and it shows linear plot. Unlike SDA and HRRA represent stability in one direction like anterior-posterior or medial-lateral direction, Lyapunov exponent indicates total system stability by being calculated from multi-dimension. Thus, in this thesis, these all three stability measurements are included.

2.4 Reference

Anderson T.W. (1971) The statistical analysis of time series. Wiley, New York.

Andersson G.B.J. Epidemiologic aspects on low back pain in industry. *Spine* 1981;**6:**53-60.

Baril-Gingras G., Lortie M. The handling of objects other than boxes: Univariate analysis of handling techniques in a large transport company. *Ergonomics* 1995;38:905-925.

Carroll J.P., Freedman W. Nonstationary properties of postural sway. *J. Biomech.* 1993;26:409-416.

Chatfield C. (1980) The analysis of time series: an introduction. London, New York.

Cholewicki J., Polzhofer G.K., Radebold A. Postural control of trunk during unstable sitting. *J. Biomech.* 2000;33:1733-1737.

Collins J.J., De Luca C.J. Open-loop and closed-loop control of posture: A random-walk analysis of center-of-pressure trajectories. *Exp Brain Res.* 1993;95:308-318.

Delignieres D., Deschamps T., Legros A., Caillou N. A methodological note on nonlinear time series analysis: Is the open- and closed-loop model of Collins and De Luca (1993) a statistical artifact? *J. Mot Behav.* 2003;35:86-96.

Damkot D.K., Pope M.H., Lord J., Frymoyer J.W. The relationship between work history, work environment and low-back pain in men. *Spine* 1984;9:395-399.

Duarte M., Zatsiorsky V.M. On the fractal properties of natural human standing. *Neuroscience Letters* 2000;283:173-176.

Einstein A. Über die von der molekularkinestisehen Theorie der Wärme geforderte Bewegung von in ruhenden Flüssigkeiten suspendieren Teilchen [On the movement of particles in suspension in resting liquids, postulated by the molecular-kinetic theory of warmth]. *Annalen der Physik*. 1905;322:549-560.

England S.A., Granata K.P. The influence of gait speed on local dynamic stability of walking. *Gait & Posture* 2005. In press.

Franklin T.C., Granata K.P. Role of reflex gain and reflex delay in spinal stability-A dynamic simulation. *J. Biomech*. 2006, Ref Type: In Press

Granata K.P., Wilson S.E. Trunk posture and spinal stability. *J. Biomech.* 2001;16: 650-659.

Granata K.P., Slota G.P., Bennett B.C. Paraspinal muscle reflex dynamics. *J. Biomech*. 2004;37:241-247.

Granata K.P., Bennett B.C. Low-back biomechanics and static stability during isometic pushing. *Human Factors* 2005;47:536-549.

Granata K.P., England S.A. Stability of dynamic trunk movement. *Spine* 2006, In press

Hollbrook T.L., Grazier K., Kelsey J.L., Stauffer R.N. (1994) The frequency of occurence, impact and cost of selected musculoskeletal conditions in the United States. Chicago, IL., *Am. Acad. Orthop. Surg*. Ref Type: Report 24-45.

Hoozemans M.J., van der Beek A.J., Frings-Dresen M.H., van Dijk F.J., van der Woude L.H. Pushing and pulling in relation to musculoskeletal disorders: A review of risk factors. *Ergonomics* 1998;41:757-781.

Kantz H., Schreiber T. (2004) Nonlinear time series analyses. 2 edn, Cambridge University Press, NY

Kelsey K.L., Githens P.B., White A.A. III et al. An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc. *J.Ortho.Res.* 1984;2**:**61-66.

Kearney R.E., Hunter I.W. System identification of human joint dynamics. *Crit. Rev. Biomed. Eng.* 1990;18:55-87.

Li L., Haddad J.M., Hamill J. Stability and variability may respond differently to changes in walking speed. *Hum Mov Sci.* 2005;24:257-267.

Marras W.S., Lavender S.A., Leurgans S., Fathallah F.A., Allread W.G, Ferguson S.A. et al. Biomechanical risk factor for occupationally related low back disorder risk. Ergonomics 1995;38:377-410.

Mendelbrot B.B., van Ness J.W. Fractal Brownian motions. Fractal noise and applications, *SIAM* 1968;10:422-437.

Moorhouse K.M., Granata K.P. Trunk stiffness and dynamics during active extension exertions. *J. Biomech*. 2005;38:2000-2007.

Moorhouse K.M., Granata K.P. Role of reflex dynamics in spinal stability: Intrinsic muscle stiffness alone is insufficient for stability. *J. Biomech* 2007, Ref Type: In Press.

Morgan D.L. Separation of active and passive components of short-range stiffness of muscle. *Am. J. Physiol*. 1977;232:c45-c49.

Nachemson A. Lumbar spine instability. A critical update and symposium summary. Spine 1985;10:290-291.

Newell K.M., Slobounov S.M., Slobounova B.S., Molenaar P.C.M. Short-term non-stationarity and the development of postural control. *Gait & Posture* 1997;6: 56-62.

Panjabi M.M. The stabilizing system of the spine. Part I Function, dysfunction, adaptation and enhancement. *J. Spinal Disorders* 1992;5:383-389.

Praemer A., Furner S., Rice D.P. (1992) Musculoskeletal conditions in the United States. 23-33. 1992. Park Ridge, IL., *Am. Acad. Orthop. Surg.* Ref Type: Report

Prieto T.E., Myklebust J.B., Hoffmann R.G., Lovett E.G., Myklebust B.M. Measures of postural steadiness: differences between healthy young and elderly adults. *IEEE Trans Biomed Eng.* 2006;43:956-966.

Radebold A., Cholewicki J., Polzhofer G.A., Green T.P. Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain. *Spine* 2001;26:724-730.

Richardson C.A., Snijders C.J., Hides J.A., Damen L., Pas M.S., Storm J. The relation between the transversus abdominis muscles, sacroiliac joint mechanics, and low back pain. *Spine* 2002;27:399-405.

Rosenstein M.T., Collins J.J., De Luca C.J. A practical method for calculating largest Lyapunov exponents from small data sets. *Physica D.* 1993;65:117-134.

Rowe M.L. Low back disability in industry: An updated position. *J.Occ.Med.* 1971;13**:**476-478.

Saal J.A.. Saal J.S. Nonoperative treatment of herniated lumbar intervertebral disc with radiculopathy. An outcome study. *Spine* 1989;14:431-437.

Schibye B., Sogaard K., Martinsen D., Klausen K. Mechanical load on the low back and shoulders during pushing and pulling of two-wheeled waste containers compared with lifting and carrying of bags and bins. *Clin. Biomech.* 2001;16: 549-559.

Van der Beek A.J., Hoozemans M.J.M., Frings-Dresen M.H., Burdorf A. Assessment of exposure to pushing and pulling in epidemiological field studies: An overview of methods, exposure measures, and measurement strategies. *Int. J. Ind. Ergon*. 1999;24:417-429.

Chapter 3

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Original Research Paper

Process Stationarity and Reliability of Trunk Postural Stability

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Abstract

Background: Empirical assessments of torso stability can be estimated from postural variability and nonlinear analyses of seated balance tasks. However, processing methods require sufficient signal duration and test-retest experiments require the assessment must be reliable. Our goal was to characterize the reliability and establish the trial duration for torso stability assessment.

Methods: Kinetic and kinematic data were recorded while subjects maintained a seated posture on a wobbly seat pan as per methods modified from Cholewicki (2000). Stability was evaluated from dynamic variability and nonlinear stability analyses. Process stationarity of the measured signals characterized the minimum necessary trial duration. Intra-class correlations measured within-session and between-session reliability.

Findings: Trial duration necessary to achieve process stationarity was 30.2 seconds. Shorter time to stationarity was observed with measures that included multi-dimensional movement behavior. Summary statistics of movement variability demonstrated good to excellent intra-session reliability, ICC = 0.64±0.13. Inter-session reliability for movement variance was good, ICC = 0.42±0.12. Nonlinear stability measures typically performed better than estimates of variability with inter-session reliability as high as ICC = 0.83. Process stationarity and reliability were improved in more difficult balance conditions.

Interpretation: To adequately capture torso dynamics during the stability assessment the trial duration should be at least 30 seconds. Excellent test-retest reliability can be achieved in intra-session analyses, but more repeated measurements are required for inter-session comparisons. Stability diffusion exponents, H_S, and the Lyapunov exponents provide excellent measures for intra-session analyses, while H_S provides excellent inter-session comparisons of torso stability.

3.1 Introduction

Spinal instability is a significant risk factor of low-back disorders (Nachemson A. 1985;Panjabi M.M. 1992). Current trends in clinical rehabilitation attempt to characterize torso stability and improve dynamic lumbar stabilization (Richardson et al. 2002;Saal J.A. & Saal J.S. 1989). Therefore, reliable measures of torso stability are required in both clinical and research environments. However, empirical assessments of stabilizing neuromuscular control of the torso have only recently become available. Radebold (Radebold et al. 2001) implemented an assessment wherein patients with low-back pain maintain seated balance on a wobbly chair by means of torso movement. Stability was quantified by summary statistics of movement variability and by nonlinear time-series analyses of movement response recorded during the task (Cholewicki, Polzhofer, & Radebold 2000). However, trial duration necessary to characterize the stability performance and reliability of the measures is unknown.

Trial duration can be established from the time necessary to achieve process stationarity. A process is considered stationary if its statistical properties do not change over time. Brief transient movements in postural control can influence the mean and variance of a signal if the signal duration is short. Therefore, measurements of standing postural control demonstrate that it is a non-stationary process in the short-term (Newell et al. 1997). This severely limits repeatability of stability measurements. Carroll and Freedman (Carroll & Freedman 1993) reported that stationarity occurred only for trial durations in excess of 15-20 seconds in standing postural control tests. We are aware of no reports to characterize the stationarity and minimum trial duration requirements for measurement of torso stability.

Reliability quantifies whether experimental results remain consistent over repeated tests under identical conditions (Carmines & Zeller 1979). Tracking therapeutic progress or research involving test-retest experimental designs require a reliable biomechanical assessment. Hence, intra-class correlation (ICC) measures have been reported for quiet standing postural control to document intra-session reliability (Corriveau et al. 2000;Lafond et al. 2004) and inter-session reliability (Benvenuti et al. 1999;Birmingham 2000). Note however, that inter-session reliability can only be achieved if individual trial duration achieves process stationarity and is influenced

by the number of trials collected during a session (Portney & Watkins 2000). To date, few studies have assessed the reliability of torso stability.

The specific aims of this study were to: 1) find minimum test duration necessary for process stationarity of torso stability assessment, and 2) estimate the intra-session and inter-session reliability of stability measures recorded from seated postural torso movement. To address these goals, kinetic and kinematic data were recorded while subjects maintained an upright seated posture without a seat-back support. The stabilizing control of the task was challenged by conditions wherein the seat pan of the chair was free to wobble using methods modified from Cholewicki (Cholewicki, Polzhofer, & Radebold 2000). Because the seat-pan was mechanically unstable, dynamic neuromuscular response from the torso musculature was necessary to maintain upright seated balance. Stability was characterized from signal variance as well as nonlinear stability analyses of the task.

3.2 Methods

3.2.1 *Protocol*

Twelve healthy volunteers (9 males, 3 females) with no previous history of low back pain participated after signing informed consent approved by institutional review board at Virginia Tech (Table 3.1). Each subject was instructed to maintain seated balance on a wobble chair with arms-crossed over their chest (Figure 3.1). The chair had no back rest and the seat pan pivoted freely in the sagittal and frontal directions. Foot support was attached to the seat-pan to avoid effects of lower limb movement. The torso was free to move independent of the seat pan so as to maintain stability on the wobbly structure. An electromagnetic motion sensor (Ascension Technology, Burlington, VT) recorded seat movement. Reaction forces from beneath the wobble chair were recorded by a force platform (Bertec, Columbus, OH).

Stability of the wobble chair was modulated by adjusting the rotational stiffness of the seat-pan. Potential energy of the system, including chair and subject, was represented as

$$V = \tfrac{1}{2} k_\theta \theta_S - G(\theta_S,\theta_T) \qquad (1)$$

where θ_S is a vector of the two-dimensional (anterior-posterior = x-axis, medial-lateral = y-axis) angle of the seat-pan with respect to horizontal and θ_T is the two-dimensional (2-D) vector of torso angle with respect to vertical. $G(\theta_S,\theta_T)$ is the gravitational potential energy of the body mass and is a function of the instantaneous chair and torso angles. Adjustable springs with rotational stiffness, k_θ, applied elastic restorative moment to the seat, i.e. return the seat pan to a horizontal configuration. Rotational stiffness of the seat pan, k_θ, was adjusted to achieve neutral stability about the base of support of the seat,

$$k_\theta + \nabla G = 0, \qquad (2)$$

The linearized gradient of gravitational moment $\nabla G = \partial / \partial \theta_S \, G(\theta_S, \theta_T)$ was recorded in static calibration measurements. Hence, the stability assessment accounted for differences in subject anthropometry intrinsically embedded in ∇G. Experimental conditions included three different stability conditions, i.e. $k_\theta = 100\% \nabla G$, $75\% \nabla G$, and $50\% \nabla G$. These represented neutral stability ($100\% \nabla G$) and two conditions wherein the biomechanical system is statically unstable ($75\% \nabla G$ and $50\% \nabla G$). To maintain an upright seated posture in these statically unstable

conditions, compensatory dynamic torso movement was necessary to correct innate disturbances and/or errors in seat angle.

Data trials of 60 seconds duration were recorded in each experimental condition. Prior to the test protocol, subjects performed three practice trials at k_θ=100%∇G and k_θ=75%∇G conditions and five practice trials at k_θ=50%∇G in order to become accustomed to the procedure. Pilot tests indicated these were sufficient to achieve consistent performance during the experimental tasks. Following acclimation, subjects performed a block of five repeated trials within each experimental condition. One minute of rest was provided between trials. Presentation order of experimental conditions was counter-balanced between subjects to remove statistical order effects. Identical test protocols were performed during two separate experimental visits with 1 week between visits.

3.2.2 *Analysis*

Chair angle data and reaction moments were recorded in the sagittal and frontal planes at 100 Hz and filtered using a 10 Hz, low-pass, 4th-order Butterworth filter. The filter cut off frequency was selected to assure sufficient bandwidth to represent torso dynamics; i.e. approximately ten times the natural frequency of the torso (Moorhouse & Granata 2005;Winter D.A. et al. 1998). Recorded reaction moments were normalized by subject body weight. Both angle and moment data were re-scaled so that the mean value of each trial was zero. Hence, measured data included nonparametric time-series of kinetic and kinematic performance.

Duration of experimental trials necessary for process stationarity was determined from the measured data. First-order stationarity is achieved when the mean is independent of the initial time value,

$$\left\langle \bar{x}(t) \right\rangle_{\Delta t} = \left\langle \bar{x}(t+\tau) \right\rangle_{\Delta t} \tag{3}$$

where $\bar{x}(t)$ is the average signal over a period from time t to t+Δt with Δt = 1, 2, 3, ··· 60 seconds. These were calculated from overlapping time epochs with initial time t = 0, 0.2, 0.4, ··· 60-Δt for each trial. Mean and standard deviation of the absolute value of each $\bar{x}(t)$ were computed across all subject trials and time for each independent value of Δt (Figure 3.2). Sample durations

greater than Δt were considered stationary when the mean value $\left\langle |\bar{x}(t)| \right\rangle_{\Delta t}$ was not statistically different than zero by student T-test.

Summary statistics of postural sway and associated reaction moments are often used as estimates of stability. Variance was characterized by: 1) root-mean-square (RMS) amplitude of the signals, 2) path-length (PATH), and 3) 95% ellipse area (EA). These were computed from the middle 40 seconds of each trial. This assured signal duration was sufficient to achieve process stationarity (see results) while avoiding potential transient artifact at the start and end of each data series. Path-length was estimated from the integral of the recorded signals divided by the signal duration. The 95% ellipse area represents the region of typical movement and was computed from

$$EA = 2\pi \cdot 3 \cdot \sqrt{S_y{}^2 \cdot S_x{}^2 - S_{yx}{}^2} \,. \tag{4}$$

Variables S_x and S_y were standard deviation of anterior-posterior and medial-lateral signal components, and S_{yx} was the covariance of the signals. The sway variance measures were recorded for kinetic and kinematic data in x-axis, y-axis, and radial directions, e.g. $\theta_r(t) = \sqrt{\theta_X(t)^2 + \theta_Y(t)^2}$.

Nonlinear stability analyses included the stabilogram diffusion analysis (SDA), the Hurst rescaled range analysis, H_R, and the maximum finite-time Lyapunov exponent, λ_{Max}. Statistical mechanics of fractal Brownian motion (Mendelbrot B.B & van Ness J.W. 1968) reveal that variance V(Δt) of time-series data change with time intervals Δt and can be expressed as

$$V(\Delta t) \sim \Delta t^{2H} \tag{5}$$

SDA analyses compute variance from the mean squared distances traveled $\left\langle \Delta d_i \right\rangle_{\Delta t}$ by the signal in time interval Δt (Collins J.J. & DeLuca C.J. 1993). H is operationally defined as the stability diffusion exponent and easily computed from the log-log slope of $\left\langle \Delta d_i{}^2 \right\rangle_{\Delta t}$ versus Δt = 0.01 to 2.0 seconds. Separate exponents were calculated for each axis component, *i* = [x-axis, y-axis data, radial distance]. Cholewicki (Cholewicki, Polzhofer, & Radebold 2000) demonstrates that this exponent demonstrates bi-linear behavior, i.e. short-term and long-term stability exponents.

Only the short term exponent discriminates between subject groups with and without low-back pain (Radebold, Cholewicki, Polzhofer, & Green 2001). Therefore, only the short-term exponent was retained for analyses, H_S.

The Hurst rescaled range analysis avoids this bilinear effect thereby presenting a single parameter to describe the dynamics of the task (Delignieres et al. 2003). Data series were divided into nonoverlapping intervals of length Δt. The difference between the maximum and minimum of each data segment was defined as the range, R, and normalized with respect to the standard deviation, R/S. Variance was represented by the mean value R/S across all time segments Δt, $\langle R/S \rangle_{\Delta t}$. The Hurst stability exponent, H_H, was computed from the log-log slope of variance versus Δt = 0.01 to 2 seconds (eqn 5).

Values of H_S and H_H closer to 0 indicates that perturbations are likely to be controlled and attracted toward the equilibrium state, whereas values greater than 0.5 indicate a disturbance will continue to grow toward infinity.

Lyapunov exponents quantify the mean rate of divergence with respect to an equilibrium state. The dynamic state at each instant in time was described by a four-dimensional vector including position and velocity in the x-axis and y-axis, $\vec{q}(t)=\left[\theta_x, \theta_y, \dot{\theta}_x, \dot{\theta}_y\right]$. Identical analyses were performed with reaction moments. For every data point $\vec{q}(t)$ in the time series we recorded the mean distance, $\overline{d}(t)$, to three data points that were nearest to it (Kantz & Schreiber 2004). The mean distance represents a small disturbance to the equilibrium condition at time t. Preliminary analyses demonstrated that the average of multiple nearest neighbors reduced the effects of stochastic noise whereas more than three failed to improve signal quality. Stability was quantified from the rate at which disturbances grow over a finite-time period Δt = 0.01 to 0.2 seconds, i.e. from d(t) to d(t+Δt) averaged over all time samples, t,

$$\left\langle d(t+\Delta t) \Big/ d(t) \right\rangle_t = e^{\lambda_{Max} \cdot \Delta t} \tag{6}$$

Large values of H_S, H_H, and λ_{max}, represent poor stability. Detailed description of these methods is available in Wolf (Wolf et al. 1985) and Rosenstein (Rosenstein, Collins, & Deluca 1993).

To quantify intra-session and inter-session reliability the intraclass correlation coefficient, ICC(2,1) was calculated for each dependent variable, EA, PATH, RMS, H_S H_H and λ_{max}. Statistically repeated measures ANOVA were used to test whether reliability was affected by independent variables of protocol difficulty ($k_\theta = 100\%\nabla G$, $75\%\nabla G$, $50\%\nabla G$), source data (kinetic, kinematic) and analysis method (sway variance, nonlinear stability). Pairwise comparisons between sway measures and stability measures were performed in post hoc analyses. Significance was determined at the level of $\alpha < 0.05$. Calculations were performed using SPSS (Chicago, IL).

3.3 Results

The trial duration necessary for process stationarity was computed from nonparametric kinetic and kinematic data (Table 3.2). Mean trial duration across all conditions necessary to achieve process stationarity was 30.2 ±11.8 seconds. For movement and reaction moments in the anterior-posterior direction the signal mean was not significantly different from zero for trial durations greater than 48 seconds at 100%∇G. Therefore, process stationarity is observed for trial durations greater than this time. During the difficult balance conditions the data were stationary at smaller durations, between 30 and 44 seconds. Kinetic and kinematic data showed similar stationary trial durations thresholds. Signals recorded from the radial direction, i.e. sum of squares of ML and AP components, achieved process stationarity in shorter time duration, 18.2 ±7.5 seconds.

Summary statistics of sway demonstrated good to excellent intra-session reliability in most conditions (Table 3.3). Mean intra-session reliability (±standard deviation) was ICC = 0.64 ±0.13. The easiest stability condition, 100%∇G, demonstrated the poorest reliability. Intra-session reliability performance was better in the more difficult conditions. Mean reliability in the 75%∇G and 50%∇G conditions were 0.70 ±0.09 and 0.71 ±0.06 respectively, whereas in the 100%∇G it was 0.51 ±0.14. The 75%∇G and 50%∇G conditions were significantly ($P<0.002$) more reliable than the 100%∇G (Table 3.4). The 75%∇G and 50%∇G conditions were not significantly different from each other. There were no significant differences between the kinetic and kinematic performance measures of sway variance. Most of the measures of sway variance measures performed equally well, i.e. the reliability of EA, PATH, and RMS summary statistics were similar. Only the kinetic PATH estimate of stability variance demonstrated excellent reliability performance at 100%∇G while all other metrics except medial-lateral RMS and anterior-posterior RMS demonstrated good reliability in the 100%∇G. In the 75%∇G and 50%∇G conditions several parameters showed excellent reliability including PATH computed from kinetic data as well as EA and RMS computed from kinematic data.

Inter-session reliability for sway variance demonstrated poor to good reliability, 0.42 ±0.12. Performance in the 75%∇G and 50%∇G conditions were 0.49 ±0.087 and 0.47 ±0.13

respectively but 100%∇G showed poorly repeatable results. Both 75%∇G and 50%∇G conditions were more reliable than the 100%∇G (P<0.002) but not different from each other. Kinematic measures, 0.55 ±0.05 were significantly (P<0.010) more reliable than kinetic data, but only for the 75%∇G and 50%∇G experimental conditions. Mean inter-session reliability during 75%∇G were 0.45 ±0.02 and 0.53 ±0.05 for the kinetic and kinematic data respectively. Mean reliability for the 50%∇G conditions were 0.37 ±0.10 and 0.57 ±0.04 for the kinetic and kinematic data respectively. Differences between dependent measures of EA, RMS, and PATH were small.

Nonlinear stability measures included SDA, Hurst exponents and Lyapunov analyses. The reliability of the stability results were not significantly different than the reliability of sway variance. However, intra-session reliability of nonlinear stability measures were significantly better than sway measures for the combined 75%∇G and 50%∇G experimental conditions. Reliability of mean intra-session nonlinear stability results were good to excellent, 0.60 ±0.21 when averaged across all stability metrics and experimental conditions (Table 3.5). The stability measures were significantly better when computing intra-session reliability from the kinetic data than the kinematic data, 0.68 ±0.20 and 0.51 ±0.19 respectively (P<0.003). There was a trend (P=0.078) wherein the most difficult experimental condition 50%∇G demonstrated greater intra-session reliability than the easy 100%∇G conditions, 0.65 ±0.20 and 0.53 ±0.23 respectively. The SDA short-term exponent, H_S, demonstrated better intra-session reliability than the Hurst exponent (P<0.002), 0.76 ±0.16 and 0.46 ±0.11 respectively. There were too few conditions to statistically compare with Lyapunov exponents with others, but the reliability of the Lyapunov method, 0.53 ±0.28, was qualitatively lower than Hurst stability analyses. However, when nonlinear stability measures were computed from the kinetic data the reliability of the SDA, 0.86 ±0.05 and Lyapunov exponents, 0.78 ±0.04, were excellent in all experimental conditions.

Inter-session reliability computed from the nonlinear stability measures was 0.47 ±0.18 when averaged across all stability metrics and experimental conditions. There was a trend (P=0.062) wherein inter-session reliability computed from the nonlinear stability measures were more reliable than sway variance. When limited to results from kinetic data this difference as statistically significant (P<0.002). The nonlinear stability measures were significantly better when computing intra-session reliability from the kinetic data than the kinematic data, 0.53

±0.17 and 0.41 ±0.17 respectively ($P<0.012$). There was no significant main effect of task difficulty. SDA short-term exponent, H_S, showed good inter-session reliability, 0.62 ±0.15; significantly ($P<0.001$) better than the Hurst exponent, 0.36 ±0.09 and qualitatively better than Lyapunov analyses.

3.4 Discussion

When the spine is stable under a specified spinal load then small neuromuscular or vertebral movement errors are automatically corrected without tissue damage (Bergmark A. 1989;McGill & Cholewicki 2001). Conversely, if the spine is unstable then biomechanical forces associated with small errors can cause sudden uncontrolled vertebral movement, i.e. spinal column buckling (Crisco & Panjabi 1992). These buckling movements impose acute strain on intervertebral tissues (Crisco et al. 1992). Repetitive strain of intervertebral discs and ligaments causes passive tissue laxity (Solomonow et al. 1999). Severe passive tissue laxity in the spine can be observed on clinical examination and is often described as a symptom of instability (Frymoyer J.W. & Selby D.K. 1985;Paris S.V. 1985) However, it is important to distinguish between instability and subsequent tissue laxity, i.e. tissue laxity is often misnamed clinical instability. Hence, stability is defined as the capacity to maintain equilibrium posture of the spine and torso despite the presence of neuromuscular or kinematic disturbances (Granata K.P. & England S.A. 2006).

Musculoskeletal stability is measured from the kinematic movement and control forces. The biomechanical system can be considered as a complex nonlinear input-output relation (Figure 3.3). Input disturbances include neuromuscular errors, external disturbance forces, and kinematic perturbations. These act on the musculoskeletal mass to cause output movements, i.e. dynamic states that are not equal to the equilibrium state. The neuromuscular system acts to control and attenuate these input disturbances so as to recover movement errors by attraction toward the equilibrium state. Hence, biomechanical measures of sway variability record the disturbance amplitude observed at the output of the system, i.e. RMS amplitude of output variance, PATH as a measure of total disturbance distance, and ellipse area to quantify the size of involuntary multi-dimensional movement (Prieto et al. 1996). However, variance represented by sway amplitude does not account for the neuron-dynamic mechanisms underlying the variability. Stability is achieved when small disturbances with respect to the equilibrium posture are attenuated and the disturbed state is attracted toward equilbirum. Nonlinear stability assessments including the SDA, Hurst exponent, and Lyapunov exponent record the attenuation of disturbances (Norris et al. 2005). Thus, variability and stability should be considered independently (England & Granata 2005;Li, Haddad, & Hamill 2005).

Duration of trials must be sufficient to fully characterize the movement and control dynamics of the seated balance task. If trial duration is brief then measurements may capture only part of the complex natural oscillations and thereby fail to achieve process stationarity. Results suggest that the seated stability protocol required trial duration of approximately 30 seconds to achieve process stationarity. Although the medial-lateral and anterior posterior data demonstrate similar trial length requirements, the radial direction data achieved stationarity in a shorter time duration, 18 seconds. This difference can be attributed to the coupled dynamics, i.e. movement dynamics in the ML direction can be transferred to AP movement. Whereas one-dimensional measures ignore this interaction, the radial measurements capture the full dynamics of the system (F.Takens 1981). Previously reported measurements of standing postural sway (Carroll & Freedman 1993) observed stationarity within 20 seconds, similar to our radial values. Conversely, power spectral density analyses of postural sway indicate that 120 seconds of data may be necessary to fully characterize the system dynamics (Powell & Dzendolet 1984). This discrepancy may be explained by analysis methods. The signals were determined to be stationary when the mean was not significantly different than zero, i.e. the mean was within 1.67 standard deviations of zero (P=.05 1-tailed t-value with 59 df). Therefore, the signal duration sufficient to achieve process stationarity was influenced by variability between means of individual time segments. Large variance of means will increase the range of mean values that fail to be statistically different than zero. Consequently, large variance of means will allow shorter trials to achieve process stationarity according to this statistical approach. An 18 second trial duration may under-estimates the time necessary to fully characterize the stabilizing dynamics.

The trial duration necessary to achieve process stationarity was shorter in more difficult balancing tasks. We attribute this to two factors. First, sway variance increased with task difficulty, i.e. larger values of RSM, PATH, and EA. As noted above, this variance may artificially reduce the time-duration necessary to achieve process stationarity. Seconds, more difficult conditions limited the region of stability. Zatsiorski (Zatsiorsky & Duarte 2000) describes two components in standing postural data including low-frequency rambling and faster trembling. These are also observed in seated postural movement of the torso (Bennett, Abel, & Granata 2004). This rambling behavior may be associated with the availability of a large biomechanically stable region that can be explored by the neuromuscular controller (Bottaro et al.

2005). However, as the rotational stiffness of the wobbly chair declined the available region of stability collapses. This prohibits large amplitude slow movements and may contribute to shorter time to process stationarity.

The intraclass correlation coefficient was recorded to document both within-day and between-day reliability of the torso stability assessment. ICC values greater than 0.75 are considered excellent while values less than 0.4 are considered poor (Fleiss 1986;Portney & Watkins 2000). Studies of standing postural sway report good to excellent reliability (Benvenuti, Mecacci, Gineprari, Bandinelli, Benvenuti, Ferrucci, Baroni, Rabuffetti, Hallett, Dambrosia, & Stanhope 1999;Birmingham 2000). Similarly, estimates of seated postural sway including signal path length (PATH) and RMS amplitude demonstrated excellent intra-session reliability. Likewise, the stability diffusion exponent, H_S and the Lyapunov exponent, λ_{Max}, calculated from kinetic data demonstrated excellent intra-session reliability. The remaining quantitative measures of sway variability and stability demonstrated good reliability, particularly in the $75\%\nabla G$ and $50\%\nabla G$ conditions. The only exception was the Lyapunov exponent computed from the kinematic data. This nonlinear method for calculating the dynamic stability requires much longer data sets (Rosenstein, Collins, & Deluca 1993). However, the Lyapunov analyses has the advantage of easier interpretation of state-space attraction toward the equilibrium state; the SDA and Hurst analyses are metrics by which the data are compared to Brownian random-walk, i.e. integral of random signal.

Inter-session reliability is typically poorer than intra-session reliability. Estimates of kinetic and kinematic sway variability demonstrated poor to good inter-session reliability, 0.42 ±0.12 averaged across dependent measures. Nonlinear stability estimates similarly demonstrated good reliability on average, 0.47 ±0.18. However, mean inter-session reliability was significantly improved in the difficult experimental conditions, $75\%\nabla G$ and $50\%\nabla G$. More difficult conditions required tighter neuromuscular control for success in the experimental task. Therefore, these challenging tasks permitted less day-to-day variability thereby providing results with greater repeatability. Reliability in difficult conditions may also be related to shorter duration to achieve stationarity in these conditions. If the stationarity duration is considered a sample of the dynamic system, then the difficult conditions were able to acquire the full set of dynamics, occasionally multiple dynamic sampled within a 40 second time-series signal whereas

the $100\% \nabla G$ recorded at most merely one time-sample of the full dynamic set. This suggests that longer trial durations may achieve improved test-retest reliability. There were few differences between dependent measures of seated sway variance. However, nonlinear stability analyses by methods of the SDA demonstrated significantly greater reliability than the Hurst Exponent. Collins (Collins J.J. & DeLuca C.J. 1993) attributes these short-term SDA exponents to open-loop biomechanical effects that may be dominated by involuntary response dynamics. The short-term exponents were separated from the remaining diffusion data by an inflection in the response curve (see Figure 3.2 in Cholewicki (Cholewicki, Polzhofer, & Radebold 2000)). In our data the critical time duration representing the upper bound of H_S was 0.67 ±.21 seconds. Therefore, voluntary movement influence the SDA and Lyapunov stability estimates less than the Hurst diffusion exponent. The long-term SDA and Hurst exponents are influence by voluntary effects that may contribute within-session and between-session variability in these parameters.

There were very few differences in sway results computed from kinetic and kinematic data. The reaction moments recorded under the wobbly seat were a mechanical consequence of seat rotations against the steel springs. Thus, the kinetic and kinematic data must be highly correlated and stability results should be similar. Unfortunately, these analyses overlook the control forces imparted by the torso, and the inertial load of the seat assembly. These effects behave as a mechanical filter, i.e. movement can be estimated from the convolution of the control forces and the mechanical filtering effects (Moorhouse & Granata 2005). This filtering behavior can limit the performance of nonlinear analyses as noted by poorer reliability of nonlinear stability measures computed from kinematic data. Therefore, the measured data are an adequate representation of the stabilizing control of this dynamic system, but stability estimates from kinematic data should take care to retain high frequency content in the signal.

Intraclass correlation analyses represent test-retest repeatability based upon the number of trials tested. We wanted to know how many experimental trials per condition should be collected to achieve excellent inter-session reliability. The Spearman-Brown formula (Shrout & Fleiss 1979) predicts the number of trials, k, to achieve a desired reliability ICC* based upon measured estimates of reliability from the current study,

$$k = \frac{ICC^*(1-ICC)}{ICC(1-ICC^*)}. \quad (8)$$

The target value ICC*=0.75 was used because it is defined as the threshold for excellent reliability (Fleiss 1986;Portney & Watkins 2000). Average of the values for the difficult conditions k_θ = 75%∇G and 50%∇G are reported because the easiest condition 100%∇G often demonstrated poor reliability and longer trial duration to stationarity. There were few differences between RMS values calculated from medial-lateral, anterior-posterior and radial components of the data, so the number of trials necessary to achieve ICC*=0.75 was computed from the average RMS reliability estimate. Similarly, results were computed for the mean value of the SDA exponent and the Hurst exponent. Results suggest that several measures of stability can be used successfully for intra-session reliability. The number of trials to achieve excellent inter-session reliability is typically large for most dependent measures, but eight trials is sufficient for the nonlinear SDA (Table 3.6). However, results are limited to interpretation of healthy subjects. Future studies should quantify repeatability using subjects with low-back pain.

In conclusion, trial duration and number of repeated trials necessary to evaluate torso stability were estimated from a dynamic seated task modified from a protocol by Cholewicki (Cholewicki, Polzhofer, & Radebold 2000). Process stationarity indicated that trial duration in excess of 30 seconds is recommended. If analyses include nonlinear stability estimates using the Lyapunov exponent, then longer trials are recommended. Excellent test-retest reliability can be achieved in intra-session analyses, but when studies require inter-session comparisons we recommend eight repeated trials per session. Longer trial duration may reduce the number of required trials. Using trials of less than one minute, the stability diffusion exponent H_S proposed by Collins (Collins J.J. & DeLuca C.J. 1993) and the Lyapunov exponent (Rosenstein, Collins, & Deluca 1993) are excellent measures for intra-session analyses and the stability diffusion exponent H_S is an excellent measure for inter-session comparisons of torso stability.

3.5 Acknowledgement

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3.6 Reference

Bennett, B. C., Abel, M. F., & Granata, K. P. 2004, "Seated postural control in adolescents with idiopathic scoliosis", *Spine*, vol. 29, no. 20, p. E449-E454.

Benvenuti, F., Mecacci, R., Gineprari, I., Bandinelli, S., Benvenuti, E., Ferrucci, L., Baroni, A., Rabuffetti, M., Hallett, M., Dambrosia, J. M., & Stanhope, S. J. 1999, "Kinematic characteristics of standing disequilibrium: reliability and validity of a posturographic protocol", *Arch.Phys.Med.Rehabil.*, vol. 80, no. 3, pp. 278-287.

Bergmark A. 1989, "Stability of the lumbar spine: A study in mechanical engineering", *Acta Orthop.Scand.Suppl.*, vol. 230, pp. 1-54.

Birmingham, T. B. 2000, "Test-retest reliability of lower extremity functional instability measures", *Clin.J Sport Med.*, vol. 10, no. 4, pp. 264-268.

Bottaro, A., Casadio, M., Morasso, P. G., & Sanguineti, V. 2005, "Body sway during quiet standing: is it the residual chattering of an intermittent stabilization process?", *Hum.Mov Sci.*, vol. 24, no. 4, pp. 588-615.

Carmines, E. G. & Zeller, R. A. 1979, *Reliability and Validity Assessment* Sage Publications, London.

Carroll, J. P. & Freedman, W. 1993, "Nonstationary properties of postural sway", *J Biomech.*, vol. 26, no. 4-5, pp. 409-416.

Cholewicki, J., Polzhofer, G. A., & Radebold, A. 2000, "Postural control of trunk during unstable sitting", *J.Biomechanics*, vol. 22, no. 12, pp. 1733-1737.

Collins J.J. & DeLuca C.J. 1993, "Open-loop and closed-loop control of posture: A random-walk analysis of center of pressure", *Exp.Brain Res.*, vol. 95, pp. 308-318.

Corriveau, H., Hebert, R., Prince, F., & Raiche, M. 2000, "Intrasession reliability of the "center of pressure minus center of mass" variable of postural control in the healthy elderly", *Archives of Physical Medicine and Rehabilitation*, vol. 81, no. 1, pp. 45-48.

Crisco, J. J. & Panjabi, M. M. 1992, "Euler stability of the human ligamentous lumbar spine: Part I Theory", *Clin.Biomech.*, vol. 7, pp. 19-26.

Crisco, J. J., Panjabi, M. M., Yamamoto, I., & Oxland, T. R. 1992, "Euler stability of the human ligamentous lumbar spine. Part II Experiment", *Clin.Biomech.*, vol. 7, pp. 27-32.

Delignieres, D., Deschamps, T., Legros, A., & Caillou, N. 2003, "A methodological note on nonlinear time series analysis: is the open- and closed-loop model of Collins and De Luca (1993) a statistical artifact?", *J Mot.Behav.*, vol. 35, no. 1, pp. 86-97.

England, S. & Granata, K. P. The Influence of Gait Speed on Local Dynamic Stability of Walking. Gait & Posture . 2005.
Ref Type: In Press

F.Takens 1981, "Detecting Strange Attractors in Turbulence," in *Dynamical Systems and Turbulence*, L.-S.Young, ed., Springer, New York, pp. 366-381.

Fleiss, J. L. 1986, *The Design and Analysis of Clinical Experiments* John Wiley & Sons, NY.

Frymoyer J.W. & Selby D.K. 1985, "Segmental Instability. Rationale for treatment", *Spine*, vol. 10, no. 3, pp. 280-286.

Granata K.P. & England S.A. Stability of dynamic trunk movement. Spine . 2006.
Ref Type: In Press

Kantz, H. & Schreiber, T. 2004, *Nonlinear Time Series Analyses*, 2 edn, Cambridge University Press, NY.

Lafond, D., Corriveau, H., Hebert, R., & Prince, F. 2004, "Intrasession reliability of center of pressure measures of postural steadiness in healthy elderly people", *Arch.Phys.Med.Rehabil.*, vol. 85, no. 6, pp. 896-901.

Li, L., Haddad, J. M., & Hamill, J. 2005, "Stability and variability may respond differently to changes in walking speed", *Hum.Mov Sci.*, vol. 24, no. 2, pp. 257-267.

McGill, S. M. & Cholewicki, J. 2001, "Biomechanical basis for stability: An explanation to enhance clinical utility.", *J.Ortho.Sports Phys.Ther.*, vol. 31, no. 2, pp. 96-100.

Mendelbrot B.B & van Ness J.W. 1968, "Fractal Brownian Motins. Fractal Noise and Aplications", *SIAM*, vol. 10, pp. 422-437.

Moorhouse, K. M. & Granata, K. P. 2005, "Trunk stiffness and dynamics during active extension exertions", *J.Biomech.*, vol. 38, no. 10, pp. 2000-2007.

Nachemson A. 1985, "Lumbar spine instability. A critical update and symposium summary", *Spine*, vol. 10, no. 3, pp. 290-291.

Newell, K. M., Slobounov, S. M., Slobounova, E. S., & Molenaar, P. C. 1997, "Stochastic processes in postural center-of-pressure profiles", *Exp.Brain Res.*, vol. 113, no. 1, pp. 158-164.

Norris, J. A., Marsh, A. P., Smith, I. J., Kohut, R. I., & Miller, M. E. 2005, "Ability of static and statistical mechanics posturographic measures to distinguish between age and fall risk", *J Biomech.*, vol. 38, no. 6, pp. 1263-1272.

Panjabi M.M. 1992, "The stabilizing system of the spine. Part I Function, dysfunction, adaptation and enhancement", *J.Spinal Disorders*, vol. 5, pp. 383-389.

Paris S.V. 1985, "Physical signs of instability", *Spine*, vol. 10, no. 3, pp. 277-279.

Portney, L. G. & Watkins, M. P. 2000, *Foundations of Clinical Research: Applications to Practice.*, 2 edn, Prentice-Hall, Upper Saddle River, N.J.

Powell, G. M. & Dzendolet, E. 1984, "Power spectral density analysis of lateral human standing sway", *J Mot.Behav.*, vol. 16, no. 4, pp. 424-441.

Prieto, T. E., Myklebust, J. B., Hoffmann, R. G., Lovett, E. G., & Myklebust, B. M. 1996, "Measures of postural steadiness: differences between healthy young and elderly adults", *IEEE Trans.Biomed.Eng*, vol. 43, no. 9, pp. 956-966.

Radebold, A., Cholewicki, J., Polzhofer, G. A., & Green, T. P. 2001, "Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain", *Spine*, vol. 26, no. 7, pp. 724-730.

Richardson, C. A., Snijders, C. J., Hides, J. A., Damen, L., Pas, M. S., & Storm, J. 2002, "The relation between the transversus abdominis muscles, sacroiliac joint mechanics, and low back pain", *Spine*, vol. 27, no. 4, pp. 399-405.

Rosenstein, M. T., Collins, J. J., & Deluca, C. J. 1993, "A Practical Method for Calculating Largest Lyapunov Exponents from Small Data Sets", *Physica D*, vol. 65, no. 117, p. 134.

Saal J.A. & Saal J.S. 1989, "Nonoperative treatment of herniated lumbar intervertebral disc with radiculopathy. An outcome study", *Spine*, vol. 14, no. 4, pp. 431-437.

Shrout, P. E. & Fleiss, J. L. 1979, "Intraclass Correlations - Uses in Assessing Rater Reliability",

Psychological Bulletin, vol. 86, no. 2, pp. 420-428.

Solomonow, M., Zhou, B., Baratta, R. V., Lu, Y., & Harras, M. 1999, "Biomechanics of Increased Exposure to Lumbar Injury Caused by Cyclic Loading: Part 1. Loss of Reflexive Muscular Stabilization", *Spine*, vol. 24, no. 23, pp. 2426-2434.

Winter D.A., Patla A.E., Prinve F., Ishac M., & Gielo-Perczak K. 1998, "Stiffness control of balance in quiet standing", *J.Neurophysiol.*, vol. 80, no. 3, pp. 1211-1221.

Wolf, A., Swift, J. B., Swinney, H. L., & Vastano, J. A. 1985, "Determining Lyapunov Exponents from A Time-Series", *Physica D*, vol. 16, no. 3, pp. 285-317.

Zatsiorsky, V. M. & Duarte, M. 2000, "Rambling and trembling in quiet standing", *Motor Control*, vol. 4, no. 2, pp. 185-200.

Table 3.1. Subject Anthropometry.

Gender	Age (year)	Height (cm)	Weight (kg)
Male	28.0 (± 3.1)	178.8 (± 9.5)	80.4 (± 14.8)
Female	25.7 (± 2.5)	165.7 (± 4.2)	60.4 (± 13.4)

Table 3.2. Stationarity. Results indicate the minimum trial duration (seconds) necessary to achieve process stationarity during the torso stability protocol. Kinetic data represent analyses of the normalized reaction moment time-series signals. Kinematic data represent analyses of chair seat-pan angle time-series.

		Task Difficulty		
		$100\%\nabla G$	$75\%\nabla G$	$50\%\nabla G$
Kinetic	AP	47	43	30
	ML	39	32	23
	r	29	15	11
Kinematic	AP	48	44	30
	ML	40	34	24
	r	26	16	12

AP anterior-posterior; ML medial-lateral direction; r radial direction $r = \sqrt{AP^2 + ML^2}$

Table 3.3. Reliability ICC(2,1) coefficients for seated sway summary statistics of variance from kinetic and kinematic data. Coefficients greater than 0.75 are considered excellent reproducibility of stability assessment.

		Intra-Session			Inter-Session		
		$100\%\nabla G$	$75\%\nabla G$	$50\%\nabla G$	$100\%\nabla G$	$75\%\nabla G$	$50\%\nabla G$
Kinetic	EA	0.51	0.69	0.68	0.35	0.42	0.36
	PATH	**0.87**	**0.87**	**0.76**	0.26	0.43	0.28
	RMS AP	0.42	0.68	0.65	0.25	0.47	0.48
	RMS ML	0.45	0.70	0.67	0.28	0.45	0.26
	RMS R	0.42	0.68	0.65	0.25	0.47	0.48
Kinematic	EA	0.52	**0.76**	**0.75**	0.32	0.59	0.55
	PATH	0.63	0.60	0.73	0.40	0.46	0.54
	RMS AP	0.38	**0.77**	0.65	0.22	0.47	0.55
	RMS ML	0.48	0.57	**0.79**	0.29	0.48	0.56
	RMS R	0.44	0.60	**0.80**	0.31	0.54	0.64

EA sway ellipse area; PATH sway path length; RMS sway amplitude

AP anterior-posterior; ML medial-lateral direction; r radial direction $r = \sqrt{AP^2 + ML^2}$

Values highlighted in bold indicate excellent reliability ICC > 0.75. Poor reliability is ICC < 0.40

Table 3.4. Statistical comparison of reliability performance. Bold values indicate statistically significant effects.

		Intra-Session	Inter-Session
	df	p-vlaue	p-value
Sway vs Stability (SS)	1	0.499	0.283
Kinetic vs Kinematics (KK)	1	0.142	0.824
Task Difficulty (Diff)	2	**0.001**	**0.001**
SS x KK	1	0.213	**0.036**
SS x Diff	2	0.164	**0.008**
KK x Diff	2	0.449	**0.003**

Table 3.5. Reliability ICC(2,1) coefficients for nonlinear stability measured from kinetic and kinematic data. Coefficients greater than 0.75 were considered excellent reproducibility of stability assessment.

		Intra-Session			Inter-Session		
		100%∇G	75%∇G	50%∇G	100%∇G	75%∇G	50%∇G
Kinetic	H_H AP	0.36	0.48	0.58	0.40	0.22	0.41
	H_H ML	0.42	0.62	0.67	0.47	0.45	0.53
	H_H R	0.33	0.32	0.49	0.43	0.27	0.37
	H_S AP	**0.82**	**0.80**	**0.85**	0.61	0.66	0.50
	H_S ML	**0.90**	**0.93**	**0.90**	**0.84**	**0.83**	**0.75**
	H_S R	**0.78**	**0.86**	**0.92**	0.65	0.72	0.65
	λ_{Max}	**0.83**	**0.76**	**0.75**	0.57	0.55	0.36
Kinematic	H_H AP	0.26	0.43	0.50	0.31	0.20	0.38
	H_H ML	0.40	0.46	0.53	0.34	0.34	0.46
	H_H R	0.56	0.36	0.42	0.32	0.33	0.30
	H_S AP	0.27	0.70	0.71	0.26	0.65	0.60
	H_S ML	0.66	**0.75**	**0.75**	0.37	0.70	0.63
	H_S R	0.51	**0.78**	**0.78**	0.45	0.62	0.70
	λ_{Max}	0.31	0.29	0.22	0.19	0.35	0.13

H_H Hurst exponent; H_S stability diffusion exponent; λ_{Max} max finite-time Lyapunov exponent

AP anterior-posterior; ML medial-lateral direction; r radial direction $r = \sqrt{AP^2 + ML^2}$

Values highlighted in bold indicate excellent reliability ICC > 0.75. Poor reliability is ICC < 0.40

Table 3.6. Number of recommended trials necessary per session. Results are based on the Spearman-Brown prophecy formula sufficient to achieve ICC = 0.75.

	EA	PATH	RMS	H_H	H_S	λ_{Max}
Intra-Session	6	6	7	16	4	15
Inter-Session	16	20	15	27	8	28

Figure 3.1. Subjects balanced on a wobble chair adapted from methods of Cholewicki (Cholewicki, Polzhofer, & Radebold 2000). Adjustable springs with equivalent rotational stiffness, k_θ, applied elastic restorative moment to the seat, i.e. return the seat pan to a horizontal configuration. Rotational stiffness of the seat pan, k_θ, was modulate the difficulty of the task.

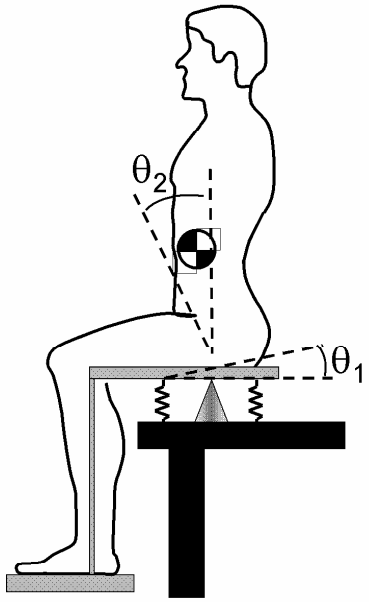

Figure 3.2. Stationarity test result. After the first 18 seconds the time averaged value of ensemble averaged data is not different from zero (P > 0.05).

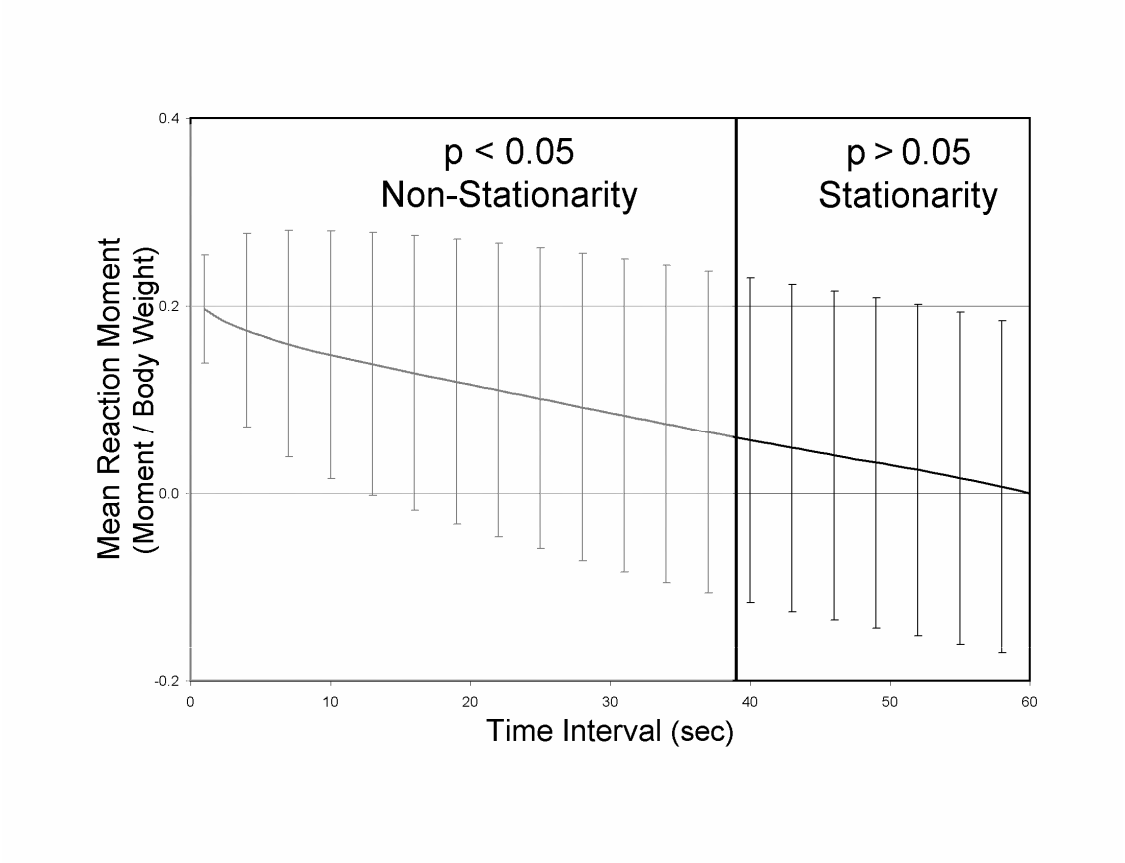

Figure 3.3. Interpretation of neuromuscular control stability can be considered a complex nonlinear input-output relation. Input disturbances include neuromuscular errors and external kinetic and kinematic perturbations.

Chapter 4

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Original Research Paper

Trunk Postural Stability during Voluntary Flexion and Extension Exertions

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Abstract

Background: Few studies have investigated the neuromuscular stabilizing control of the spine during pushing and pulling exertions even though they are one of the risk factors of low-back disorders. Our goal was to empirically test whether exertion force and direction affect the neuromuscular control of spine stability.

Method: Chair angle movements were recorded while subjects maintained a seated posture on a wobbly seat pan with different exertion forces (0N, 40N, and 80N) and in both flexion and extension directions. Stability was evaluated for dynamic variability and using nonlinear stability analyses.

Findings: Stability declined with exertion force. Both kinematic variance and nonlinear dynamics stability estimates increased significantly with exertion ($p < 0.041$). Flexion exertions were less stable than extension exertions. RMSr, EA, and Lyapunov exponent were greater during flexion exertions than during extension exertions. However, there was no direction effect for SDAr estimates of stability.

Interpretation: Voluntary exertion force and direction affect neuromuscular control of stability significantly. Although the results are not from direct invasive measures of spinal stability, they provide insight into the control of spinal stability. The results demonstrate that pushing and pulling exertions can be risk factors of low-back disorder by increasing the spinal instability.

Keywords: Stability, Pushing and Pulling exertions, reflex

4.1 Introduction

Repetitive lifting has been reported as a significant risk factor of low-back disorders in the occupational setting (Andersson, 1981; Marras et al., 1995). As such, industry is rapidly modifying the workspace to reduce lifting, and frequently replacing them with pushing and pulling exertions. However, pushing and pulling exertions are also associated with low-back disorders. In the United States, Canada, and the U.K., 20% of industrial low-back injury have been attributed to pushing or pulling activities (Damkot et al., 1984; Hoozemans et al., 1998). This injury rate is expected to increase with the trend toward a growing number of push and pull-related exertions in the workplace. Despite the fact that 50% of industrial manual materials handling include pushing and pulling exertions (Baril-Gingras & Lortie, 1995), the biomechanics and neuromuscular control of pushing and pulling exertions remain poorly understood (Schibye et al., 2001; van der Beek et al., 1999).

The spine is stabilized, in part, by stiffness of the paraspinal muscles during active exertions (Granata & Wilson, 2001; Frankling & Granata, in press; Moorhouse, 2005). This stiffness increases with exertion force (Kearney & Hunter, 1990; Morgan, 1977; Moorhouse & Granata, 2005). Therefore, it is often been assumed that spinal stability improves with exertion effort. However, evidence suggests that reflexes also play an important role in the stability of the spine. Reflex gain can help to stabilize the spine, but reflex delay limits this benefit of this gain (Franklin & Granata, in press). Reflex gain decreases and reflex delay increases with exertion force (Granata et al, 2004). Considering the role of reflexes on spinal stability, it is unclear whether spinal stability always improves with exertion effort. We are unaware of any studies investigating the effects of exertion force on stability of the spine.

Estimates of spinal stability from a biomechanical model suggest that the spine may be less stable during isometric flexion (pushing) exertions than during extension (pulling) efforts (Granata & Bennett, 2005). Antagonistic co-contraction of the lumbar paraspinal muscles may be recruited to help maintain spinal stability during pushing exertions. However, it is unclear whether this compensatory co-contraction response is sufficient to maintain stability. We are unaware of any studies investigating the effects of exertion direction (trunk flexion or extension) on stability of the spine.

The goal of this study was to empirically test whether stability of the spine is affected by exertion force and exertion direction. We hypothesized 1) stability decreases with increased trunk flexion or extension forces, and 2) stability is less in pushing exertions than pulling exertions.

4.2 Methods

A total of 12 healthy adult volunteers participated including 7 males (aged 25.7 ± 6.9 years, height 178.7 ± 6.9 cm, mass 79.9 ± 9.0 kg), and 5 females (aged 21.4 ± 1.7 years, height 161.3 ± 8.2 cm, mass 59.6 ± 8.3 kg). Participants reported no history of low back pain, and provided informed consent approved by the Institutional Review Board at Virginia Tech prior to participation.

Spinal stability was estimated during seated balance on a wobbly chair (Cholewicki, Polzhofer, & Radebold 2000). The seat pan of the chair pivoted freely in the sagittal and frontal directions (Figure 4.1). Destabilizing moments arise from gravitational effects associated with small angle deviations from neutral, $M_{Grav}(\theta_S,\theta_T)$, where θ_S is a vector of the anterior-posterior and medial-lateral angles of the seat-pan with respect to horizontal, and θ_T is the vector of torso angles with respect to vertical. Adjustable springs with an effective rotational stiffness, k_θ, applied elastic restorative moment to the seat, i.e. return the seat pan to a horizontal configuration. This stiffness was set at $k_\theta = 50\% \nabla M_{Grav}$ where $\nabla M_{Grav} = \partial / \partial \theta_S \, M_{Grav}(\theta_S,\theta_T)$ is the linearized gradient of gravitational moment recorded in static calibration measurements. Hence, the stability assessment accounted for differences in subject anthropometry intrinsically embedded in ∇M_{Grav}. Since the stabilizing restorative stiffness was less than the destabilizing gravitational gradient, the biomechanical system was inherently unstable. Therefore, dynamic torso movement was necessary to maintain an upright seated posture, and to correct disturbances and/or errors in seat angle. A foot support attached to the seat pan supported the lower limbs so that the upright seated posture was maintained predominantly by movement at the lumbar spine.

External horizontal isotonic forces were applied at the T8 level of the torso. This required voluntary trunk extension and flexion moments to maintain an upright posture while balancing on the wobbly chair. Forces 0, 40 and 80N were generated by tensioning elastic bands between the subject's chest harness and a distant wall of the laboratory. The length of the elastic bands was 15 m, thereby assuring that change in length of the band from torso movements was small relative to the band length, i.e. forces were nearly isotonic. The externally applied isotonic forces caused a tipping moment about the pivot point under the seat pan. To offset this tipping

moment, the seat pan was adjusted anteriorly and posteriorly to achieve steady-state equilibrium under the applied load, i.e. static gravitational moment balanced the externally applied forces. Note, however, that these equilibrium conditions were statically unstable as described above.

Data were recorded during trials of 60 seconds duration to achieve process stationarity (Granata and Lee, in press). Experimental conditions included three force levels (0, 40, 80 N) and two force directions (flexion, extension). The 0N force condition was always tested first. The presentation order of the 40N and 80N conditions was counter-balanced across directions to remove potential order effects except for the 0N condition. Subjects completed a block of five repeated trials in each experimental condition with one minute of rest between trials. Prior to the test protocol, subjects performed five practice trials at each condition to become accustomed to the procedure. Pilot tests indicated that these were sufficient to achieve consistent performance during the experimental exertions.

Stability Analysis

Seat angle in both the anterior-posterior and medial-lateral planes was sampled at 100 Hz using an electromagnetic motion sensor (Ascension Technology, Burlington, VT). Elastic band force was sampled at 100 Hz using a uniaxial load cell (Interface Inc., Scottsdale, AZ). Seat angle and elastic band force data were filtered using a 6 Hz, low-pass, 4^{th}-order Butterworth filter. The filter cut off frequency for seat angle data was selected to assure sufficient bandwidth to represent torso dynamics; i.e. natural dynamics of the torso is approximately 1 Hz (Moorehouse & Granata, 2005; Winter et al. 1998). Seat angle data were re-scaled so that the mean value of each trial was zero.

Stability of each experimental condition was recorded from summary statistics of seat angle variance and by estimating the strength of the state-attractor, i.e. nonlinear dynamic stability analyses. Summary statistics of seat angle included: root-mean-square (RMS) amplitude of the seat angle, and 95% ellipse area (EA). The RMS amplitude was estimated from the Euclidean norm of the 2-D seat angle data, $\theta(t) = \sqrt{\theta_{Sx}(t)^2 + \theta_{Sy}(t)^2}$ because this includes the dynamic interaction between multi-dimensional movements, and because previous analyses

demonstrate it is a more reliable measurement than single-axis data (Granata & Lee, in press). The 95% ellipse area represents the region of typical movement and was computed from

$$EA = 2\pi \cdot 3 \cdot \sqrt{S_y^{\,2} \cdot S_x^{\,2} - S_{yx}^{\,2}}\,. \qquad (1)$$

where S_x and S_y were standard deviation of anterior-posterior and medial-lateral movements, and S_{yx} was the covariance of the signals.

Nonlinear stability analyses included the stabilogram diffusion analysis (SDA) and the maximum finite-time Lyapunov exponent, λ_{max}. Statistical mechanics of fractal Brownian motion (Mendelbrot & van Ness, 1968) reveal that the variance $V(\Delta t)$ of the time-series seat angle data change with time intervals Δt and can be expressed as:

$$V(\Delta t) \sim \Delta t^{2H} \qquad (2)$$

SDA analyses compute variance from the mean squared distances traveled $\left\langle \Delta d_i^{\,2} \right\rangle_{\Delta t}$ by the signal in time interval Δt (Collins & De Luca, 1993). H is operationally defined as the stability diffusion exponent and can be computed from the log-log slope of $\left\langle \Delta d_i^{\,2} \right\rangle_{\Delta t}$ versus $\Delta t = 0.01$ to 2.0 seconds. Cholewicki (Cholewicki et al., 2000) showed that this exponent demonstrates bi-linear behavior, i.e. short-term and long-term stability exponents. Only the short term exponent discriminates between subject groups with and without low-back pain (Radebold et al., 2001). Therefore, only the short-term exponent, H, was retained for analyses. Values of H closer to 0 indicate that perturbations are likely to be controlled and attracted toward the equilibrium state, whereas values greater than 0.5 indicate a disturbance will continue to grow toward infinity.

Lyapunov exponents quantify the mean rate of divergence with respect to an equilibrium state. The dynamic state at each instant in time was described by a four-dimensional vector including angle and velocity in the sagittal and frontal planes, $\vec{q}(t) = \left[\theta_{Sx}, \theta_{Sy}, \dot{\theta}_{Sx}, \dot{\theta}_{Sy}\right]$. For every data point $\vec{q}(t)$ in the time series we recorded the mean distance, $\overline{d}(t)$, to three data points that were nearest to it (Kantz & Schreiber, 2004). In an asymptotically stable system, the distance between these points must grow smaller with time, i.e. attractor. Preliminary analyses

demonstrated that the average of multiple nearest neighbors reduced the effects of stochastic noise whereas more than three failed to improve signal quality. Stability was quantified from the rate at this which $\overline{d}(t)$ changes over a finite-time period $\Delta t = 0.01$ to 0.2 seconds, i.e. from d(t) to d(t+Δt) averaged over all time samples, t,

$$\left\langle d(t+\Delta t) \Big/ d(t) \right\rangle_t = e^{\lambda_{Max} \cdot \Delta t} \tag{3}$$

Large values of H and λ_{max} represent poor stability. Detailed description of these methods is available in Wolf (Wolf et al., 1985) and Rosenstein (Rosenstein et al., 1993).

A repeated-measures ANOVA was used to investigate the effects of force level (0, 40, 80N), and force direction (flexion, extension) on sway variance and stability. Interaction between force and exertion conditions was performed in post hoc analyses. Significance was determined at the level of $p<0.05$.

4.3 Results

Stability declined with exertion force. Both kinematic variance and nonlinear estimates of stability increased significantly with exertion (Table 4.1). RMS and EA values were greater in 40 N exertions than during unloaded, 0 N trials, and greater at 80 N than at 40 N (Table 4.2). The SDA during 80 N tasks was greater than 0 N and 40 N conditions. Maximum Lyapunov exponents , λ_{max}, during 40 N and 80 N exertions were greater than unloaded conditions.

Flexion exertions were less stable than extension exertions. Kinematic variance recorded in terms of RMS and EA was greater during flexion tasks then during extension exertions (Table 4.1). Flexion tasks represent conditions wherein the external isotonic force was directed posteriorly, thereby requiring voluntary torso flexion moment to maintain upright posture. During 0 N force exertions there is no difference between flexion and extension exertions, and thereby no difference in RMS or EA variance (Table 4.2). However, post-hoc analyses of a significant force-by-direction interaction revealed RMS variance was greater during the flexion exertions than during extension in both the 40 N and 80 N tasks. The main effect of direction significantly affected EA but there was no significant 2-way interaction. Lyapunov exponents were significantly greater during flexion exertions, i.e. less stable, than during extension efforts. Analyses of the interaction revealed that this difference was observed only during the 40 N force conditions. There were no direction effects for SDA estimates of stability.

4.4 Discussion

Lifting tasks are rapidly being replaced by pushing or pulling tasks in the industrial workplace because repetitive lifting is a significant risk factor of low-back disorders. However, pushing and pulling exertions also have risk associations. The goal of this study was to empirically test whether neuromuscular control of stability is affected by exertion force and exertion direction. Results matched our hypotheses that trunk postural stability is poorer during voluntary pushing exertions than during pulling exertions, and that trunk postural stability decreased as exertion force increased.

Biomechanical analysis of spinal stability suggests that paraspinal muscle co-contraction is necessary during pushing tasks despite the fact that these muscles are antagonistic to the flexion exertion (Granata & Bennett, 2004). However, co-contraction is considered open-loop control and may be insufficient for reactive feedback (closed-loop) control in dynamic environments. In this study, the experimental task was specifically designed to be statically unstable, i.e. $k_\theta = \tfrac{1}{2}\nabla G$. Therefore, reactive control was necessary to maintain stability. This approach was selected because previous studies demonstrated that reactive neuromuscular control contributes approximately 40% to the total stabilizing behavior of the torso (Moorhouse & Granata, 2006).

Contrary to published computational estimates of spinal stability (Cholewicki & McGill, 1996), empirical results demonstrated poorer stabilizing control with increasing exertion force. Co-contraction increases with exertion (Kearney & Hunter, 1990; Morgan, 1977; Moorhouse & Granata, 2005), but the stabilizing task could not be achieved with co-contraction alone, i.e. neuromuscular response to small movements was necessary for successful balance (Franklin & Granata, in press; Moorhouse, 2005). Researches (Granata et al., 2004) reported that reflex gain and reflex delay increased with the trunk flexion preload. Thus, reflex gain and reflex delay may increase with exertion effort in this study. Although increased reflex gain helps to stabilize the spine with increasing co-contraction, reflex delay significantly impairs control of spinal stability (Franklin & Granata, in press). Thus, in dynamic environments, risk of tissue strain injury from instability may be greater in heavy exertion tasks than in light manual materials handling.

Results supported the hypothesis that neuromuscular control of stability is poorer during voluntary flexion exertions than during extension exertions. Co-contraction during trunk flexion exertions was approximately twice the level of co-contraction during equivalent extension tasks (Granata et al., 2005). Thus, reflex time during flexion exertion might be slower than reflex time during extension exertion. Results suggest that manual materials handling requiring pushing tasks may be less stable than lifting or pulling tasks. It is traditionally assumed that workplace design may benefit by replacing lifting tasks with pushing or pulling. This is based on evidence that spinal compression is less in pushing tasks than in lifting (Granata & Bennett, 2005). However, results from the present study showed that both pushing and pulling tasks may also impair spinal stability. This may be due to increased spinal shear force during pushing and pulling exertions. A risk of spinal compression during heavy lifting is well established. This study demonstrates a risk caused by spinal shear force.

This study has limitations that must be addressed through further study. First, exertion forces were fixed (0N, 40N, and 80N). These exertion conditions did not consider the subject's anthropometry. Thus, there is a potential possibility that subjects experienced different levels of task difficulty. Future studies should consider the subject's anthropometry when choosing exertion force conditions. Second, EMG data analysis was not included in this study. Changes in reflex and co-contraction could be better explained with EMG data. Future studies should include EMG data collection and analysis should be included.

In conclusion, nonlinear dynamical systems analyses and kinematic stability estimates demonstrate significant effects of voluntary exertion force and exertion direction on neuromuscular control of stability. These results are limited by the fact that the measurements record stability of the torso rather than direct invasive measures of spinal stability. However, the measured data represent the first buckling mode of the spine, and therefore provide insight into the control of spinal stability. This study may aid in understanding whether pushing and pulling exertions are also risk factors of low-back disorders caused by spinal instability. Further analyses are necessary to investigate the occupational risks associated with spinal instability during tasks involving pushing and pulling.

4.5 Acknowledgements

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4.6 References

Andersson, G. B. J., 1981. Epidemiologic aspects on low back pain in industry. Spine 6, 53-60.

Baril-Gingras, G., Lortie, M., 1995. The handling of objects other than boxes: Univariate analysis of handling techniques in a large transport company. Ergonomics 38, 905-925.

Cholewicki, J., Polzhofer, G. A., Radebold, A., 2000. Postural control of trunk during unstable sitting. J. Biomechanics 22, 1733-1737.

Collins, J. J., DeLuca, C. J., 1993. Open-loop and closed-loop control of posture: A random-walk analysis of center of pressure. Exp. Brain Res. 96, 308-318.

Damkot, D. K., Pope, M. H., Lord, J., Frymoyer, J. W., 1984. The relationship between work history, work environment and low-back pain in men. Spine 9, 395-399.

Franklin, T. C., Granata, K. P., 2006. Role of reflex gain and reflex delay in spinal stability-A dynamic simulation. J. Biomechanics Ref Type: In Press

Granata, K. P., Marras, W. S., 2000. Cost-benefit of muscle co-contraction in protecting against spinal instability. Spine 25, 1398-1404.

Granata, K. P., Wilson, S. E., 2001. Trunk posture and spinal stability. J. Biomechanics 16, 650-659.

Granata, K. P., Slota, G. P., Bennett, B. C., 2004. Paraspinal muscle reflex dynamics. J. Biomechanics 37, 241-247.

Granata, K. P., Bennett, B. C., 2005. Low-back biomechanics and static stability during isometic pushing. Human Factors 47, 536-549.

Granata, K. P., Lee, P. E., Franklin, T. C., 2005. Co-contraction recruitment and spinal load during isometric trunk flexion and extension. Clin. Biomech. 20, 1029-1037.

Granata, K. P., Lee, H., Process stationarity and reliability of trunk postural stability. Clin. Biomech. Ref Type: In Ppress.

Hoozemans, M. J., van der Beek, A. J., Frings-Dresen, M. H., van Dijk, F. J., van der Woude, L. H., 1998. Pushing and pulling in relation to musculoskeletal disorders: A review of risk factors. Ergonomics 41, 757-781.

Kantz, H., Schreiber, T., 2004. Nonlinear Time Series Analyses, Second ed. Cambridge University Press, New York.

Kearney, R. E., Hunter, I. W., 1990. System identification of human joint dynamics. Crit. Rev. Biomed. Eng. 18, 55-87.

Marras, W. S., Lavender, S. A., Leurgans, S., Fathallah, F. A., Allread, W. G., Ferguson, S. A., Ferguson S. A., Allread W. G., Rajulu, S. L., 1995. Biomechanical risk factors for occupationally related low back disorder risk. Ergonomics 38, 377-410.

Matthews, P. B., 1991. The human stretch reflex and the motor cortex. Trends Neurosci. 14, 87-91.

Mendelbrot, B. B., van Ness, J. W., 1968. Fractal Brownian Motions. Fractal Noise and Applications. SIAM 10, 422-437.

Morgan, D. L., 1977. Separation of active and passive components of short-range stiffness of muscle. Am. J. Physiol. 232, c45-c49.

Moorhouse, K. M., Granata, K. P., 2005. Trunk stiffness and dynamics during active extension exertions. J. Biomechanics 38, 2000-2007.

Moorhouse, K. M., Granata, K. P., 2007. Role of reflex dynamics in spinal stability: Intrinsic muscle stiffness alone is insufficient for stability. J. Biomechanics, Ref Type: In Press.

Radebold, A., Cholewicki, J., Polzhofer, G. A., Green, T. P., 2001. Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain. Spine 26, 724-730.

Rosenstein, M. T., Collins, J. J., DeLuca, C. J., 1993. A practical method for calculating largest Lyapunov exponents from small data sets. Physica D 65, 134.

Schibye, B., Sogaard, K., Martinsen, D., Klausen, K., 2001. Mechanical load on the low back and shoulders during pushing and pulling of two-wheeled waste containers compared with lifting and carrying of bags and bins. Clin. Biomech. 16, 549-559.

Viitasalo, J., Komi, P. V., 1981. Interrelationships between electromyographic, mechanical, muscle structure and reflex time measurements in man. Acta Physiol. Scand. 111, 97-103.

Winter, D. A., Patla, A. E., Prinve, F., Ishac, M., Gielo-Perczak, K., 1998. Stiffness control of balance in quiet standing. J. Neurophysiol. 80, 1211-1221.

Wolf, A., Swift, J. B., Swinney, H. L., Vastano, J. A., 1985. Determining Lyapunov exponents from a time-series. Physica D 16, 285-317.

Van der Beek, A. J., Hoozemans, M. J. M., Frings-Dresen, M. H., Burdorf, A., 1999. Assessment of exposure to pushing and pulling in epidemiological field studies: An overview of methods, exposure measures, and measurement strategies. Int. J. Ind. Ergon. 24, 417-429.

Table 4.1. ANOVA Table

	RMS_r	EA	SDA_r	λ_{max}
Force	$P < 0.001$	$P < 0.001$	$P < 0.041$	$P < 0.001$
Direction	$P < 0.001$	$P < 0.046$	$P < 0.597$	$P < 0.002$
Force x Direction	$P < 0.001$	$P < 0.367$	$P < 0.171$	$P < 0.047$

Table 4.2. Results of different force conditions

Force	RMSr	EA	SDAr	λ_{max}
0 N	0.83 ± 0.26	21.26 ± 11.25	0.762 ± 0.017	0.552 ± 0.126
40 N	0.94 ± 0.27*	25.04 ± 9.88*	0.763 ± 0.016	0.594 ± 0.108*
80 N	1.10 ± 0.36*,†	31.06 ± 13.20*, †	0.766 ± 0.016*	0.613 ± 0.105*

* indicate significant ($p<0.05$) difference from 0N force

† indicate significant ($p<0.05$) difference from 40N force.

Figure 4.1. Postural dynamic stability measured during active pushing and pulling exertions.

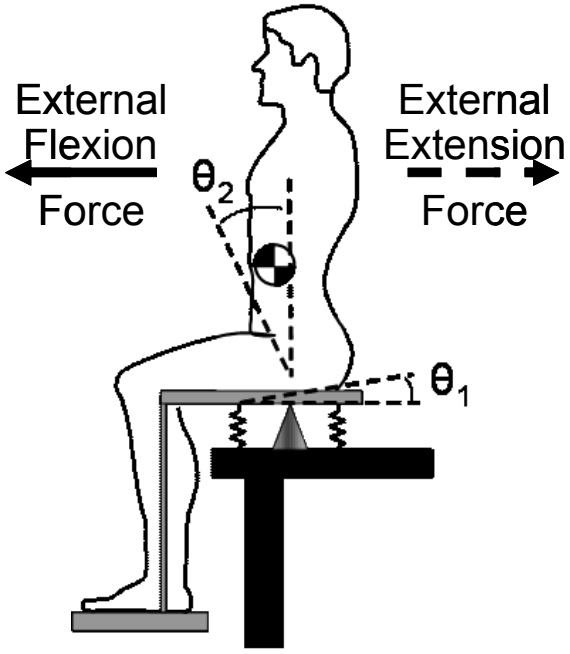
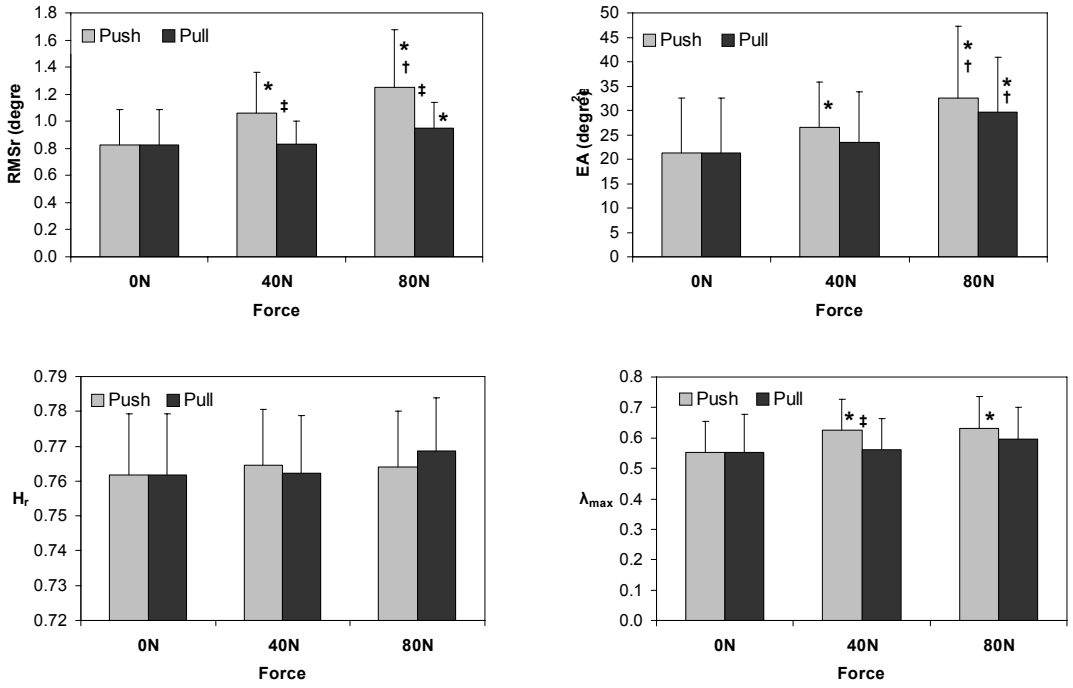


Figure 4.2. Kinematic variance (RMS, EA) and the rate of state-space expansion (H, λ_{max}) increase monotonically with force which indicates that stability is less with greater exertion effort. RMS variance was greater during flexion exertions (40, 80 N), EA was greater during flexion exertions at 80 N, and λ_{max} was greater during flexion exertions at 40 N. These each suggest that poorer stability was less during flexion exertions than during extension efforts.



* indicates a significant ($p<0.05$) difference from 0N force
† indicates a significant ($p<0.05$) difference from 40N force.
‡ indicates a significant ($p<0.05$) difference between flexion and extension

CONCLUSIONS

Test duration and reliability were examined in the first study, and trunk stability during pushing/pulling tasks with exertion forces was examined in second study. Results from the first study demonstrated that the test duration should be at least 30 seconds for the kinematic variability and stability measurements to be reliable. The second study showed that trunk stability was disturbed by both exertion force and exertion direction.

Process stationarity indicated that a trial duration greater than 30 seconds is recommended. However, this result is limited to college students. If patients with low-back pain or elderly persons are recruited to test trunk stability, the test duration should be examined again. If the nonlinear stability measures include the analysis of the Lyapunov exponent, longer trials are recommended. Five trials are enough to achieve excellent test-retest reliability in intra-session analyses, but for studies that require inter-session comparisons we recommend at least eight repeated trials per session. Using trials of between 30 seconds and one minute, the stability diffusion exponent H_S proposed by Collins (Collins J.J. & DeLuca C.J. 1993) and the Lyapunov exponent (Rosenstein, Collins, & Deluca 1993) are excellent measures for intra-session analyses. The stability diffusion exponent H_S is an excellent measure for inter-session comparisons of torso stability.

Nonlinear dynamical stability analyses and kinematic variability estimates demonstrate significant effects of voluntary exertion force and exertion direction on neuromuscular control of stability. Thus, both exertion force and pushing/pulling exertions can be significant risk factors of low-back disorders. These results are limited by the fact that the measurements record stability of the torso rather than direct invasive measures of spinal stability. However, the measured data represents the first buckling mode of the spine, and therefore provide insight into control of spinal stability. Further analyses are necessary to investigate the risks associated with spinal instability during occupational tasks involving pushing and pulling.

APPENDICES

Appendix A – ICC test results

Reliability ICC(2,1) coefficients for seated sway summary statistics of variance from kinetic and kinematic data. Coefficients greater than 0.75 are considered excellent reproducibility of stability assessment. TD test duration; EA sway ellipse area; PATH sway path length; RMS sway amplitude. AP anterior-posterior; ML medial-lateral direction; r radial direction $r = \sqrt{AP^2 + ML^2}$. Values highlighted in bold indicate excellent reliability ICC > 0.75. Poor reliability is ICC < 0.40.

a) Kinematic Variance

Data	TD	Var	Intra-Session			Inter-Session		
			100% ∇G	75% ∇G	50% ∇G	100% ∇G	75% ∇G	50% ∇G
Kinetic	40s	EA	0.51	0.69	0.68	0.35	0.42	0.36
		PATH	**0.87**	**0.87**	**0.76**	0.26	0.43	0.28
		RMS AP	0.42	0.68	0.65	0.25	0.47	0.48
		RMS ML	0.45	0.70	0.67	0.28	0.45	0.26
		RMS R	0.42	0.68	0.65	0.25	0.47	0.48
	50s	EA	0.49	0.69	0.74	0.33	0.39	0.39
		PATH	**0.88**	**0.90**	**0.79**	0.27	0.44	0.29
		RMS AP	0.39	0.71	0.69	0.22	0.46	0.52
		RMS ML	0.48	0.74	0.66	0.31	0.45	0.28
		RMS R	0.46	0.66	0.68	0.34	0.48	0.42
	60s	EA	0.46	0.68	**0.75**	0.30	0.38	0.41
		PATH	**0.90**	**0.91**	**0.79**	0.28	0.44	0.29
		RMS AP	0.38	0.71	0.70	0.21	0.47	0.52
		RMS ML	0.45	0.69	0.71	0.30	0.38	0.33
		RMS R	0.41	0.69	0.69	0.27	0.46	0.43
Kinematic	40s	EA	0.52	**0.76**	**0.75**	0.32	0.59	0.55
		PATH	0.63	0.60	0.73	0.40	0.46	0.54
		RMS AP	0.38	**0.77**	0.65	0.22	0.57	0.55
		RMS ML	0.48	0.57	**0.79**	0.29	0.48	0.56
		RMS R	0.44	0.60	**0.80**	0.31	0.54	0.64
	50s	EA	0.51	**0.77**	**0.81**	0.33	0.57	0.58
		PATH	0.66	0.69	**0.77**	0.42	0.48	0.54
		RMS AP	0.37	**0.78**	0.70	0.18	0.58	0.61
		RMS ML	0.52	0.61	**0.79**	0.37	0.47	0.57
		RMS R	0.52	0.64	**0.82**	0.45	0.55	0.66
	60s	EA	0.54	0.72	**0.81**	0.34	0.55	0.62
		PATH	0.69	0.69	**0.76**	0.40	0.49	0.56
		RMS AP	0.40	**0.79**	0.71	0.21	0.58	0.64
		RMS ML	0.57	0.55	**0.82**	0.40	0.42	0.63
		RMS R	0.57	0.63	**0.84**	0.48	0.54	0.73

b) Stability measurements

Data	TD	Var	Intra-Session			Inter-Session		
			100%∇G	75%∇G	50%∇G	100%∇G	75%∇G	50%∇G
Kinetic	40s	H_H AP	0.36	0.48	0.58	0.40	0.22	0.41
		H_H ML	0.42	0.62	0.67	0.47	0.45	0.53
		H_H R	0.33	0.32	0.49	0.43	0.27	0.37
		H_S AP	**0.82**	**0.80**	**0.85**	0.61	0.66	0.50
		H_S ML	**0.90**	**0.93**	**0.90**	**0.84**	**0.83**	**0.75**
		H_S R	**0.78**	**0.86**	**0.92**	0.65	0.72	0.65
		λ_{Max}	**0.83**	**0.76**	**0.75**	0.57	0.55	0.36
	50s	H_H AP	0.38	0.49	0.56	0.39	0.24	0.42
		H_H ML	0.42	0.62	0.69	0.50	0.42	0.56
		H_H R	0.36	0.49	0.46	0.41	0.29	0.38
		H_S AP	**0.84**	**0.87**	**0.84**	0.60	0.70	0.50
		H_S ML	**0.89**	**0.93**	**0.89**	**0.85**	**0.83**	0.74
		H_S R	**0.81**	**0.85**	**0.94**	0.70	0.70	0.66
		λ_{Max}	**0.85**	**0.79**	**0.84**	0.58	0.52	0.49
	60s	H_H AP	0.44	0.42	0.52	0.39	0.25	0.43
		H_H ML	0.44	0.62	0.66	0.54	0.41	0.60
		H_H R	0.45	0.43	0.39	0.45	0.22	0.41
		H_S AP	**0.84**	**0.87**	**0.85**	0.60	0.69	0.49
		H_S ML	**0.89**	**0.94**	**0.89**	**0.85**	**0.82**	0.72
		H_S R	**0.83**	**0.86**	**0.94**	0.70	0.66	0.62
		λ_{Max}	**0.88**	**0.86**	**0.83**	0.61	0.53	0.50
Kinematic	40s	H_H AP	0.26	0.43	0.50	0.31	0.20	0.38
		H_H ML	0.40	0.46	0.53	0.34	0.34	0.46
		H_H R	0.56	0.36	0.42	0.32	0.33	0.30
		H_S AP	0.27	0.70	0.71	0.26	0.65	0.60
		H_S ML	0.66	**0.75**	**0.75**	0.37	0.70	0.63
		H_S R	0.51	**0.78**	**0.78**	0.45	0.61	0.70
		λ_{Max}	0.31	0.29	0.22	0.19	0.35	0.13
	50s	H_H AP	0.19	0.33	0.45	0.25	0.18	0.41
		H_H ML	0.29	0.46	0.58	0.25	0.34	0.52
		H_H R	0.55	0.40	0.39	0.25	0.26	0.32
		H_S AP	0.35	0.73	0.72	0.31	0.58	0.68
		H_S ML	0.68	**0.83**	**0.79**	0.51	0.73	0.63
		H_S R	0.54	**0.82**	**0.81**	0.50	0.59	0.67
		λ_{Max}	0.48	0.34	0.25	0.19	0.35	0.20
	60s	H_H AP	0.17	0.26	0.38	0.16	0.16	0.42
		H_H ML	0.29	0.43	0.60	0.18	0.33	0.58
		H_H R	0.62	0.38	0.37	0.30	0.24	0.34
		H_S AP	0.33	0.71	**0.77**	0.33	0.67	0.70
		H_S ML	0.73	**0.85**	**0.81**	0.50	0.74	0.69
		H_S R	0.55	**0.84**	0.73	0.47	0.60	0.66
		λ_{Max}	0.63	0.27	0.21	0.27	0.34	0.23

Appendix B - Number of recommended trials

Number of recommended trials necessary per session. Results are based on the Spearman-Brown prophecy formula sufficient to achieve ICC = 0.75. TD test duration; H_H Hurst exponent; H_S stability diffusion exponent; λ_{Max} max finite-time Lyapunov exponent

AP anterior-posterior; ML medial-lateral direction; r radial direction $r = \sqrt{AP^2 + ML^2}$

a) Kinematic Variance

Data	TD	Var	Intra-Session			Inter-Session		
			100% ∇G	75% ∇G	50% ∇G	100% ∇G	75% ∇G	50% ∇G
Kinetic	40s	EA	14	7	7	28	21	27
		PATH	2	2	5	43	20	39
		RMS AP	21	7	8	45	17	16
		RMS ML	18	6	7	39	18	43
		RMS R	21	7	8	45	17	16
	50s	EA	16	7	5	30	23	23
		PATH	2	2	4	41	19	37
		RMS AP	23	6	7	53	18	14
		RMS ML	16	5	8	33	18	39
		RMS R	18	8	7	29	16	21
	60s	EA	18	7	5	35	24	22
		PATH	2	1	4	39	19	37
		RMS AP	24	6	6	56	17	14
		RMS ML	18	7	6	35	24	30
		RMS R	22	7	7	41	18	20
Kinematic	40s	EA	14	5	5	32	10	12
		PATH	9	10	6	23	18	13
		RMS AP	24	4	8	53	11	12
		RMS ML	16	11	4	37	16	12
		RMS R	19	10	4	33	13	8
	50s	EA	14	4	4	30	11	11
		PATH	8	7	4	21	16	13
		RMS AP	26	4	6	68	11	10
		RMS ML	14	10	4	26	17	11
		RMS R	14	8	3	18	12	8
	60s	EA	13	6	4	29	12	9
		PATH	7	7	5	23	16	12
		RMS AP	23	4	6	56	11	8
		RMS ML	11	12	3	23	21	9
		RMS R	11	9	3	16	13	6

b) Stability measurements

Data	TD	Var	Intra-Session			Inter-Session		
			100%∇G	75%∇G	50%∇G	100%∇G	75%∇G	50%∇G
Kinetic	40s	H_H AP	27	16	11	23	53	22
		H_H ML	21	9	7	17	18	13
		H_H R	30	32	16	20	41	26
		H_S AP	3	4	3	10	8	15
		H_S ML	2	1	2	3	3	5
		H_S R	4	2	1	8	6	8
		λ_{Max}	3	5	5	11	12	27
	50s	H_H AP	24	16	12	23	48	21
		H_H ML	21	9	7	15	21	12
		H_H R	27	16	18	22	37	24
		H_S AP	3	2	3	10	6	15
		H_S ML	2	1	2	3	3	5
		H_S R	4	3	1	6	6	8
		λ_{Max}	3	4	3	11	14	16
	60s	H_H AP	19	21	14	23	45	20
		H_H ML	19	9	8	13	22	10
		H_H R	18	20	23	18	53	22
		H_S AP	3	2	3	10	7	16
		H_S ML	2	1	2	3	3	6
		H_S R	3	2	1	6	8	9
		λ_{Max}	2	2	3	10	13	15
Kinematic	40s	H_H AP	43	20	15	33	60	24
		H_H ML	23	18	13	29	29	18
		H_H R	12	27	21	32	30	35
		H_S AP	41	6	6	43	8	10
		H_S ML	8	5	5	26	6	9
		H_S R	14	4	4	18	10	6
		λ_{Max}	33	37	53	64	28	100
	50s	H_H AP	64	30	18	45	68	22
		H_H ML	37	18	11	45	29	14
		H_H R	12	23	23	45	43	32
		H_S AP	28	6	6	33	11	7
		H_S ML	7	3	4	14	6	9
		H_S R	13	3	4	15	10	7
		λ_{Max}	16	29	45	64	28	60
	60s	H_H AP	73	43	24	79	79	21
		H_H ML	37	20	10	68	30	11
		H_H R	9	24	26	35	48	29
		H_S AP	30	6	4	30	7	6
		H_S ML	6	3	4	15	5	7
		H_S R	12	3	6	17	10	8
		λ_{Max}	9	41	56	41	29	50

Appendix C – Critical Point

Critical point (sec)

Data	TD	Dir	100%∇G	75%∇G	50%∇G
Kinetic	40s	AP	0.38 (± 0.33)	0.70 (± 0.35)	0.75 (± 0.32)
		ML	0.40 (± 0.29)	0.51 (± 0.33)	0.64 (± 0.37)
		R	0.31 (± 0.22)	0.47 (± 0.24)	0.49 (± 0.20)
	50s	AP	0.38 (± 0.33)	0.69 (± 0.38)	0.75 (± 0.32)
		ML	0.41 (± 0.30)	0.52 (± 0.34)	0.64 (± 0.36)
		R	0.33 (± 0.26)	0.50 (± 0.23)	0.49 (± 0.19)
	60s	AP	0.40 (± 0.35)	0.72 (± 0.37)	0.76 (± 0.31)
		ML	0.41 (± 0.31)	0.54 (± 0.34)	0.61 (± 0.35)
		R	0.34 (± 0.27)	0.51 (± 0.24)	0.49 (± 0.19)
Kinematic	40s	AP	1.02 (± 0.33)	0.97 (± 0.33)	0.91 (± 0.29)
		ML	0.87 (± 0.25)	0.87 (± 0.24)	0.87 (± 0.26)
		R	0.72 (± 0.26)	0.66 (± 0.22)	0.58 (± 0.19)
	50s	AP	1.06 (± 0.33)	0.98 (± 0.30)	0.90 (± 0.28)
		ML	0.87 (± 0.26)	0.89 (± 0.26)	0.86 (± 0.25)
		R	0.73 (± 0.24)	0.68 (± 0.21)	0.59 (± 0.19)
	60s	AP	1.05 (± 0.30)	0.99 (± 0.30)	0.90 (± 0.27)
		ML	0.87 (± 0.25)	0.88 (± 0.23)	0.86 (± 0.24)
		R	0.75 (± 0.25)	0.70 (± 0.20)	0.59 (± 0.18)

Appendix D – IRB Approval Form

VirginiaTech

Office of Research Compliance
Institutional Review Board
1880 Pratt Drive (0497)
Blacksburg, Virginia 24061
540/231-4991 Fax: 540/231-0959
E-mail: moored@vt.edu
www.irb.vt.edu

FWA00000572(expires 7/20/07)
IRB # is IRB00000667.

DATE: January 5, 2007

MEMORANDUM

TO: Kevin P. Granata

Approval date: 1/21/2007
Continuing Review Due Date:1/6/2008
Expiration Date: 1/20/2008

FROM: David M. Moore

SUBJECT: **IRB Expedited Continuation 5:** "Musculoskeletal Biomechanics of Movement and Control ", IRB # 05-769

This memo is regarding the above referenced protocol which was previously granted expedited approval by the IRB. The proposed research is eligible for expedited review according to the specifications authorized by 45 CFR 46.110 and 21 CFR 56.110. Pursuant to your request, as Chair of the Virginia Tech Institutional Review Board, I have granted approval for extension of the study for a period of 12 months, effective as of January 21, 2007.

Approval of your research by the IRB provides the appropriate review as required by federal and state laws regarding human subject research. As an investigator of human subjects, your responsibilities include the following:

1. Report promptly proposed changes in previously approved human subject research activities to the IRB, including changes to your study forms, procedures and investigators, regardless of how minor. The proposed changes must not be initiated without IRB review and approval, except where necessary to eliminate apparent immediate hazards to the subjects.
2. Report promptly to the IRB any injuries or other unanticipated or adverse events involving risks or harms to human research subjects or others.
3. Report promptly to the IRB of the study's closing (i.e., data collecting and data analysis complete at Virginia Tech). If the study is to continue past the expiration date (listed above), investigators must submit a request for continuing review prior to the continuing review due date (listed above). It is the researcher's responsibility to obtained re-approval from the IRB before the study's expiration date.
4. If re-approval is not obtained (unless the study has been reported to the IRB as closed) prior to the expiration date, all activities involving human subjects and data analysis must cease immediately, except where necessary to eliminate apparent immediate hazards to the subjects.

cc: File

Invent the Future

Appendix E – Consent Forms

Title of Project: Musculoskeletal Biomechanics of Movement and Control
Investigators: K.P. Granata

Purpose of this Research

To understand musculoskeletal injury and improve clinical diagnoses of injury it is necessary to understand how muscles control force and movement. The purpose of this study is to measure the relation between human movement, force generation and muscle activity. We are also interested in observing how gender, fatigue and physical conditioning influence these parameters. Throughout the course of this project more than 450 subject volunteers will participate including healthy individuals from the age of 18 to 55.

Procedures

We will tape adhesive markers and sensors on your skin around your trunk, legs and arms. These sensors are EMG electrodes that measure the activity of your muscles and position sensors to measure how you move. After some preliminary warm up stretches, we may ask you to push and/or pull as hard as you can against a resistance. We may then ask you to hold or lift a weight or weighted-box and to bend forward and back. We may also ask you to do some fatiguing exertions such as holding or lifting a heavy weight or pushing/pulling against a bar or cable for several minutes. We may also apply a quick but small force to record reflexes. You may be requested to return for repeated testing. Between test sessions you may be asked to participate in specified physical conditioning as per the American College of Sports Medicine recommended guidelines.

Risks

The risks of this study are minor. They include a potential skin irritation to the adhesives used in the tape and electrode markers. You may also feel some temporary muscle soreness such as might occur after exercising. Subjects participating in physical conditioning may experience muscle soreness and/or musculoskeletal injury associated with inherent risks of cardiovascular, strength training and therapeutic exercise. To minimize these risks you will be asked to warm-up before the tasks and tell us if you are aware of any history of skin-reaction to tape, history of musculeoskeletal injury, cardiovascular limitations.

Benefits

By participating in this study, you will help to increase our understanding musculeoskeletal control of movement and musculoskeletal injury mechanisms. We hope to make this research experience interesting and enjoyable for you where you may learn experimental procedures in biomechanical sciences. We do not guarantee or promise that you will receive any of these benefits and no promise of benefits has been made to encourage your participation.

Anonymity and Confidentiality

Experimental data collected from your participation will be coded and matched to this consent form so only members of the research team can determine your identity. Your identity will not be divulged to unauthorized people or agencies. Digital video recorded during the experimental trials will be used to track the movement of the sensors by means of computer analyses and is insufficient video quality to observe individual participant identifying characteristics. Secondary VHS-style video may be recorded to validate the digital motion data. This camera angle is placed to avoid facial or other identifying characteristics. Sometimes it is necessary for an investigator to break confidentiality if a significant health or safety concern is perceived or the participant is believed to be a threat to himself/herself or others.

Compensation

Participants required to return for multiple test sessions or participate in physical conditioning for this protocol will receive payment per the number of test sessions as well as a bonus for full completion of the multi-session research protocol. Subjects participating in experiments as part of course or laboratory procedures will receive appropriate credit for analysis of specified data as described in the course syllabus

but not for personal performance during the experimental session. If course credit is involved and the subject chooses not to participate alternative means for earning equivalent credit will be established with the course instructor.

Freedom to Withdraw

You are free to withdraw from a study at any time without penalty. If you choose to withdraw, you will be compensated for the portion of the time of the study (if financial compensation is involved). If you choose to withdraw, you will not be penalized by reduction in points or grade in a course (if course credit is involved). You are free not to answer any questions or respond to experimental situations that they choose without penalty.

There may be circumstances under which the investigator may determine that you should not continue as a subject. You will be compensated for the portion of the project completed

Approval of Research

This research project has been approved, as required, by the Institutional Review Board for Research Involving Human Subjects at Virginia Polytechnic Institute and State University, by the Department of Engineering science and Mechanics.

21 January 2007	20 January 2008
IRB Approval Date	Approval Expiration Date

Subject's Responsibilities

I voluntarily agree to participate in this study. I have the following responsibilities:

- Inform the investigators of all medical conditions that may influence performance or risk
- Comply to the best of my ability with the experimental and safety instructions
- Inform the investigator of any physical and mental discomfort resulting from the experimental protocol

Subject's Permission

I have read and understand the Informed Consent and conditions of this project. I have had all my questions answered. I hereby acknowledge the above and give my voluntary consent:

Subject Name (Print):_____

Subject signature: Date

_____ Date_____

Witness (Optional except for certain classes of subjects)

Should I have any pertinent questions about this research or its conduct, and research subjects' rights, and whom to contact in the event of a research-related injury to the subject, I may contact:

Investigator(s): _____E-mail: _____ Phone 231-2022

Faculty Advisor: K.P. Granata E-mail: Granata@vt.edu Phone 231-7039

Ishwar Puri 231-3243

Departmental Reviewer/Department Head Telephone/e-mail

David M. Moore
Chair, IRB
Office of Research Compliance
Research & Graduate Studies
540-231-4991 / moored@vt.edu

This Informed Consent is valid from 21January 2007 to 20 January 2008.

Subjects must be given a complete copy (or duplicate original) of the signed Informed Consent

Appendix F – Data Collection Form

Data Collection Form for study 1

Reliability test

Sub # _____ Data _____

	Forward	Backward		
Moment	_____	_____	Gender _____	Weight _____
Angle	_____	_____	Age _____	Hwight _____

MGH _____

sp 100 ( )	sp 75 ( )	sp 50 ( )	File Name
Practice 1	Practice 1	Practice 1	
Rest 1 min	Rest 1 min	Rest 1 min	
Practice 2	Practice 2	Practice 2	
Rest 1 min	Rest 1 min	Rest 1 min	
Practice 3	Practice 3	Practice 3	
Rest 1 min	Rest 1 min	Rest 1 min	
		Practice 4	
		Rest 1 min	
		Practice 5	
		Rest 1 min	
Trial 1	Trial 1	Trial 1	s#sp#trial1
Rest 1 min	Rest 1 min	Rest 1 min	
Trial 2	Trial 2	Trial 2	s#sp#trial2
Rest 1 min	Rest 1 min	Rest 1 min	
Trial 3	Trial 3	Trial 3	s#sp#trial3
Rest 1 min	Rest 1 min	Rest 1 min	
Trial 4	Trial 4	Trial 4	s#sp#trial4
Rest 1 min	Rest 1 min	Rest 1 min	
Trial 5	Trial 5	Trial 5	s#sp#trial5

Data Collection Form for study 2

Pushing/puling test

Sub # _____ Data _____

	Forward	Backward		
Moment	_____	_____	Gender _____	Weight _____
Angle	_____	_____	Age _____	Hwight _____

MGH _____ L – 50 _____

0N	Flexion (Push)		Extension (Pull)		File Name
	40N	80N	40N	80N	
Practice 1	Practice 1	Practice 1	Practice 1	Practice 1	
Practice 2	Practice 2	Practice 2	Practice 2	Practice 2	
Practice 3	Practice 3	Practice 3	Practice 3	Practice 3	
Practice 4	Practice 4	Practice 4	Practice 4	Practice 4	
Practice 5	Practice 5	Practice 5	Practice 5	Practice 5	
Test 1	Test 1	Test 1	Test 1	Test 1	s#force#trial1
Test 2	Test 2	Test 2	Test 2	Test 2	s#force#trial2
Test 3	Test 3	Test 3	Test 3	Test 3	s#force#trial3
Test 4	Test 4	Test 4	Test 4	Test 4	s#force#trial4
Test 5	Test 5	Test 5	Test 5	Test 5	s#force#trial5

VITA

HyunWook Lee

HyunWook Lee was born in Jinhae, Korea on March 10, 1976. He attended Kangseo High School in Mokdong, Korea. He attended Inha University in Inchon, Korea, where he graduated Bachelors of Engineering degree in Mechanical, Aerospace & Automation in February, 2003. He transferred to Virginia Polytechnic Institute & State University (Virginia Tech) for graduate school and obtained a Master of Science degree in Biomedical Engineering in May, 2007. Working as a graduate research assistant in the Musculoskeletal Biomechanics Laboratory, His research focused on spinal stability. His research was published in Clinical Biomechanics. During his free time, HyunWook enjoys movies, snowboarding, and playing basketball.