QUANTIFICATION OF TRUNK MOTION IN RESPONSE TO COMPLEX PLATFORM PERTURBATIONS WHILE HOLDING WEIGHTS IN AN UPRIGHT POSTURE

MOHAMAD PARNIANPOUR*, KAUSHIK GHOSH**, KAMRAN BARIN**
WILLIAM S. MARRAS*

*Department of Industrial, Welding and Systems Engineering,
**Department of Otolaryngology, The Ohio State University, USA

ABSTRACT

This experiment investigated the role of the lumbar region in a simple load-holding situation for three asymmetric (0, 45 and 90 deg) platform perturbations for 18 normal, college age males. The platform perturbations included horizontal translation and rotation in addition to stable condition. The weight was 20% of the subjects' body weight. The Balance System (Chattecx Co.) was used to monitor balance performance while the Lumbar Motion Monitor was used to quantify low-back kinematics. The platform perturbation significantly affected both the lumbar motion parameters and the dispersion, an index of postural sway. The asymmetry of direction of perturbation also significantly affected both the postural sway and the lumbar motion involvement. Given the large mass of the spine, corrective/reactive fast movements to maintain the balance due to the unexpected perturbation to the base of support may lead to impulsive loading of the spine which is associated with the low back injuries. The study of postural stability, whilst an additional manual material handling task is being performed presents a more realistic and functional test than the traditional protocols used presently in Balance Clinics. This type of functional dual tasking is a rich paradigm that requires more research investigations.

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1. INTRODUCTION

The maintenance of upright posture is a very complex task. The center of mass is situated at a height equal to about two-thirds of stature. The large vertical distance of the center of gravity (CG) and the small base of support make upright standing an inher-

ently unstable task. The criteria for postural stability is that the CG must lie within the boundaries of the supporting base. Mechanically, internal muscle forces adjusting the spatial orientation of the body segments in such a way that the effect of the disturbance is offset compensate for deviations of the CG toward the edges of the supporting base. This finely tuned control system has a neural controller and a musculoskeletal actuator with over 200 degrees of freedom [4].

Physiological feedback to this neural control system is provided by the vestibular, proprioceptive and visual systems. Any disorder of these neural pathways makes the task of postural control even more difficult. The individual role of various sensory systems in balance restoration in normals and various neurological patient populations has been studied [3,29].

Epidemiology of Slip and Fall

The body of epidemiology literature that pertains to imbalance accidents seems to be divided into two major groups. One concerns the aged population, while the other investigates such accidents in the workplace.

Slips and falls are frequent causes of injury and death in the elderly population. Alfram [2] reports that more than 90 percent of all fractures of the hip are associated with trauma due to fall. Approximately 200,000 hip fractures occur in the United States every year and 64 percent of them are associated with the 65 and over age group. Associated health care costs for these fractures are estimated to be in the range of $2 billion [20]. Buck et al. [7] concluded that about 17 percent of all slipping and failing type accidents resulted in serious injury like fractures. Scheinder et al. [36] put the cost of such fractures at $1.6 billion in 1987 and predict that it will rise to $6 billion by the year 2040. Lotz et al. [19] found that fractures of the proximal part of the femur are an important public-health problem and a major source of mortality and morbidity among the elderly. Perhaps even more significant, from both a psychological and economic point of view is the severe loss of mobility and independence that are results of a fear of falling. Tinetti et al. [38] found from their study that 48 percent of the subjects, who fell at least one time over a one-year period, were afraid of falling while 26 percent reported that they had curtailed their daily activities due to this fear. The fear of falling is a major deterrent to daily mobility among the elderly [40]. This lack of mobility often leads to institutionalization with its associated high costs. Schneider et al. [36] estimated that the average cost of nursing home care for the elderly in 1985 was $23,600 per resident and totaled $31.1 billion. Falling is a common occurrence in the elderly and the most common cause of accidental death in this age group [5]. With the dramatic projected increase in the average age of the world population, some have proposed the possibility of a two to threefold rise in the total number of hip-fractures by the middle of the next century. Medical technology has substantially increased the life expectancy in the world. Scheinder et al. [36] predict that the number of U.S. residents above the age of 85 will have grown from 2.8 to 6.7 million. Due to the steady increase in the average life expectancy of the population in the U.S., the number of elderly that are institutionalized due to the lack of mobility (having actually experienced a falling-type accident or fear thereof) can be expected to rise dramatically. Femie et al. [10] predicts that in Canada, the increasing proportion of people in the higher age groups will greatly increase the incidence rates of falls. The incidence of serious injury (for example, fractures) increases exponentially with age. Consequently, nearly one out of every three women and one out of

every six men could sustain a fracture of the hip by age ninety.

Leamon et al. [18] report that falls accounted for about 11,700 deaths in the United States in 1989 and the number of in injuries in industry resulting from falls approaches about a third of a million. Buck et al. [7] report that in the U.K., 14 percent of all accidents reported to the Health and Safety Executive in manufacturing and construction industries were of a type classified as Slipping, Tripping, or Falling (STFL). Such accidents accounted for 40 percent of all reported injuries to railway staff over a two year period and 57 percent of all injuries to mining and quarrying staff over 5 year period. On an average, STFL accidents accounted for about 17.5 percent of all accidents. These authors estimate the cost of such accidents to be about 150 million pounds in the U.K.

In a study conducted on 13,958 employees of a car company, Manning et al. [22] found that "underfoot accidents" (moving surface, collapse of floor, slipping of the foot, tripping etc.) were associated with 80 of 122 accidents that resulted in back injuries. 57 of these were slips. In contrast, other unexpected events leading to spinal loading accounted for only 17 of the 122. However, a large number of patients were not able to attribute their low-back pain to a single cause, a probable indicator of a cumulative trauma problem. In the gearbox unit of the same plant, Manning [23] concluded that the lumbar spine was the part of the body most commonly injured in a lost-time, slipping accident. Baxter et al. [6] discusses the importance of the "underfoot factor" (slip, trip or fall) as the most common first unforeseen event leading to injury both at work and outside. Troup et al. [39] discovered that 36% of accidents causing back pain were falls and many more were slips. Falls were associated not only with increased length of absence from work during current attacks of LBP, but also with a higher recurrence rate of the affliction.

Involvement of the Low back in Balance Maintenance

The strategies used by the human body to maintain upright posture have been categorized into three major groups [30]. Movement of the whole body with the ankle as the fulcrum is referred to as an "ankle strategy". When the CG approaches the outside limit of the base of support, movement occurs about the hip and back. This is referred to as the "hip strategy". If both of these strategies prove insufficient, a "stepping strategy" is employed where the area of the supporting base is enlarged and, in essence, placed below the estimated position of the CG. This shows a hierarchy of balance maintenance strategies of which the hip and trunk is a part.

In the previous studies [3,4,17,30], the focus has been on the ankle joint and force plates have been used

to assess balance performance. The role of the ankle joint in compensating for support surface movement has been investigated [4,17]. These studies were concerned with sagittal plane motions only. Several studies have also been conducted that look at muscle activation patterns in challenging balance tasks. It seems to be widely accepted that perturbations of the support surface in the anterior-posterior (AP) direction are usually compensated for by the muscles attached to the ankle joint first [17]. There is a clear upwardly radiating pattern of Integrated Electromyographic (IEMG) activity from the lower extremity muscles (soleus, hamstring) to the lower back muscles (erector spinae, obliques) and then to the neck muscles in AP translations. Muscles above the lower extremity contribute only if the ankle compensation is not enough, a reflection of the hierarchical synergies [30]. However disturbances causing flexion extension of the ankle showed an almost inverse trend in the pattern of muscle activation. Keshner et al. [17] found that the muscles of the back are activated significantly (p<. 05) before the soleus and the hamstring in perturbations causing flexion-extension of the ankle joint.

The above-mentioned literature shows a clear case for lumbar involvement in the balance maintenance mechanism. Neural physiology reinforces this idea. Neurons originating in the lateral vestibular nucleus carry excitatory fibers from the semicircular canals to influence contraction of the extensor muscles of the trunk and extremities [8]. As, the semi-circular canals are the body's main balance sensors, it is logical to extrapolate that the trunk is involved in the postural maintenance mechanism.

The surveyed literature showed a paucity of work quantifying the motion of the back in the balance restoration function. Byl et al. [8] studied balance responses and body sways in healthy subjects and LBP patients. The LBP patient population demonstrated significantly greater postural sway, kept their center of pressure (COP) more posterior, and was less likely to be able to balance on one foot in the eyes closed test. The authors' subjective observation was that LBP patients balanced using their hip and back as a fulcrum in contrast to healthy subjects who used their ankle.

It is clear that the type of perturbation (flexion-extension or linear translation) affects the patterns of muscle recruitment and thus the usage of the various joints involved in the maintenance of upright posture. The ankle does not have significant degrees of freedom in the coronal and transverse planes. Iqbal [16] simulated a 4-link biped model and studied its postural stability in the frontal plane. He concluded that combinations of hip and trunk torques are needed to maintain uprightness. In the sagittal plane, the ankle torques carry out most of the stabilizing function, while in the frontal plane, the ankle torques are very small and the maximum contribution to stability comes

from the hip and trunk torques. It seems that the ankle is ill suited to compensate for medial-lateral disturbances. Hence, the inclusion of asymmetry in this experiment is considered important.

The directionality of a fall also has an effect on the seriousness of the potential injury. Lotz et al. [19] concluded that the energy available in a fall from standing height is about twenty times the strength of the femur: hence a fall could potentially cause a fracture every time were it not for the biomechanical factors (action of joints and muscles) that limited the peak forces at the point of impact. Lateral impacts to the greater trochanter are considered to have an elevated risk of serious injury. Experiments that deal with reducing the incidence and severity of fractures due to falls have simulated asymmetric falls. Robinovitch et al. [35] concluded from their experiment that reducing hip-fracture incidence requires more understanding of the biomechanics of the falling process. Their experiment dealt with a non-sagittally symmetric pelvis release experiment.

The purpose of this experiment was to quantify the kinematical parameters of low-back motion (ranges of motion, peak velocities and peak accelerations in all the three cardinal planes) and balance performance during the performance of a simple weight-holding task in three different asymmetric conditions. It investigates the effect of different kinds of support surface perturbation and the effect of the direction of perturbation on low-back motion and on balance performance.

2. METHODS

The focus of this study was to quantify low-back motion characteristics and balance performance during a simple manual material handling task (load-holding) in response to asymmetric platform perturbations. The experimental design counterbalanced and randomized these conditions to control order and learning effects. Eighteen healthy (with no prior history of back pain or of falling for the past six months) male volunteers between the ages of 20-33 participated in this study. The subjects' average (standard deviation) age, height and weight were 25 (3.2) years, 167.8 (11.5) cm and 66.8 (10.9) Kg respectively.

Apparatus

The Balance System (Chattecx Co., Chattanooga, TN) was used to perturb the upright posture and monitor the balance responses while the LNM was used to quantify the responses of the lumbar region. The Balance System consists of a programmable platform; a 286 based computer, two motors, signal conditioning circuitry, and two footplates each consisting of independent vertical force transducers for the ball and the

heel of each foot. The program calculates the position of the COP based on the measured forces. The platform can be programmed in one of three modes: a stationary mode and two movement modes. These modes are: 1) STABLE: platform remains stationary; 2) LINEAR: platform translates in the horizontal plane about the mean position; and 3) UP-DOWN (abbreviated from Toes Up-Down in the Balance System Manual): platform rotates about its axis, traveling in both directions beyond its mean position.

A separate motor controls each of the movement modes. The translation motor was replaced by a motor, which was more powerful than the one in the commercial version of the Balance System. The amplitudes and velocities of movement of the platform in both of these modes are found in Table 1. These are from the system specifications for the UP-DOMN case and from a simple measurement and timing experiment for the LINEAR case.

The software accompanying the Balance System controls the movement of the platform through the motors. The protocol sequence was set-up as a single battery of tests for every subject. Each individual test was set-up for the type of platform movement: STABLE, LINEAR or UP-DOWN. The testing period was set to 10 seconds. As the equipment was originally designed to study the balance performance and postural stability in elderly and neurological patient populations, the apparatus also allows control of the speed of the test as a percentage of maximum speed. In this study, the speed was set to 100% for all subjects as no impaired subjects were tested.

The Lumbar Motion Monitor (LMM) was built and developed in the Biodynamics Lab at The Ohio State University to monitor motion parameters for people doing manual material handling (MMH) tasks [24]. Two pieces of orthoplast hold the LMM to the subject: one around the hip and the other around the shoulder. It is essentially like an exoskeleton, which follows the lumbar movement of the person in 3-D space. Potentiometers are used to measure the instantaneous changes in position for each plane. The power supply is a 9-volt battery that is attached to the waist piece. The signal from the LMM is sent via hard wire to the analog- to-digital converter board. The board interfaces with a Compaq 386 portable microcomputer for data storage. The LMM was calibrated using a reference frame to ± 30 degrees in the frontal and transverse planes and to ± 45 degrees in the sagittal plane. The calibration was found to be accurate to within ± 0.5 degrees [24]. The validation of the LMM data was done with video motion analysis system (Motion Analysis, Santa Rosa, CA), which used 60 Hz. as the sampling rate. This same sampling rate was used for this experiment because frequencies above 8 Hz. have negligible contribution (very little power) to kinematical signals [41]. Since the data acquisition for the LN4M was independent of the Balance System, a synchronization channel was used to monitor the progress of the test. A DC step voltage that goes high (about 3.5 V) when the test is in progress and low (about 0.2 V) when no test is being conducted was tapped off from the balance system and fed into the same A/D board as the LMM channels. The synchronization marker was used to extract the actual test data.

Experimental Design:

The experiment was designed to test the effect of asymmetry and platform movement on the lumbar response and balance performance. The experiment was randomized, with restrictions on the randomization for the independent variable asymmetry. The design counterbalanced the order of presentation of the asymmetry variable, while within a level of asymmetry; the presentation of the platform conditions was completely randomized. The design is therefore a completely crossed 3 x 3 design.

The dynamic nature of the balancing mechanism is such that the COP fluctuates incessantly, traversing large total excursions even during the quiescent act of two-legged stance [28], in this study, the sway index (dispersion) is essentially the amount of variation of the instantaneous COP values from the mean COP position. A detailed analysis for the calculation of dispersion with a discussion of validation studies and normative data can be found in Byl et al. [8]. The dependent variables in this experiment were the sway index dispersion (DISP) and the sagittal, coronal, and transverse ranges of motion (SROM LROM TROM), peak velocities (SPV, LPV, TPV) and peak accelerations (SPA, LPA, TPA).

Table 1. Parameters of the platform movement

Platform Movement	Cycles / 10 s	Amplitude	Velocity
LINEAR	3	2.5 cm	3 cm / s
UP-DOWN	1.5	4 deg	2.4 deg / s

There were two independent variables in this experiment: the kind of platform perturbation (PLAT) and the asymmetry of the direction of perturbation (ASYM). The three platform levels are STABLE, LINEAR and UP-DOWN. Figure 1 is a diagrammatic representation of the feet positioning of the subjects for the three-asymmetry conditions. Asymmetry can be described by imagining two independent coordinate systems. The first (fixed) coordinate system is defined by the platform's axes: XX' and YY' (Fig. 1). The horizontal translation of the platform takes place symmetrically along the YY' axis while the rotation occurs about the XX' axis. The other coordinate system is the subjects anatomical reference comprised of the anterior-posterior (AP) line and that medial-lateral (ML) line that passes through the ankle's axis of rotation (Fig. 1). The origins of both these coordinate systems are the center point of the ankle's axis of rotation. Hence, asymmetry refers to the relative rotation angle between the YY' axis of the platform and the subjects AP line and was simulated by positioning the footplates (and the subjects) on the platform in the three conditions described below, before the start of each test. The three levels to this independent variable: 0, 45 and 90 degrees can be described as: 1) 0 deg Asymmetry: the subjects' AP line is aligned with the platform's YY' line. Hence the medial-lateral axis is also aligned with the XX' axis of the platform (Fig. 1a); 2) 45 deg Asymmetry: the subjects AP axis makes 45 deg. with the platforms' YY' line (Fig. 1b); 3) 90

deg Asymmetry: the subjects AP line makes 90 deg. with the platform's YY' line (Fig. 1c).

Asymmetry in this experiment was achieved by rotating the subjects so that their mid-sagittal planes were at various angles to the direction of perturbation. In the context of this study, therefore, asymmetry did not refer to a "twisting" angle, but rather to the direction of perturbation with respect to the mid-sagittal plane of the subject.

Inclusion of a Weight-holding Task:

A pilot study had indicated that response of normal subjects to the perturbations of the Balance System was not significant in simple standing [13]. An amplification of the responses was needed because of the slow nature of the perturbations. Considerable debate exists in the literature regarding the protocol to evaluate postural stability. Geurts et al. [11] assert that it is important to record the kinematics of a balance task not only under basic conditions, but also to see how the same act is performed under more complex conditions (for example when the subject is performing some other task simultaneously). They suggest several tasks from the point of view of cognitive dual task interference. Byl et al. [8] created inherent instability in their study by having subjects perform balance tasks in positions with both feet aligned anteriorposteriorly (the two-feet tandem), the one-footed test and with their eyes closed. Alexander et al. [1] had "beam support" rather than a flat base for the sub-

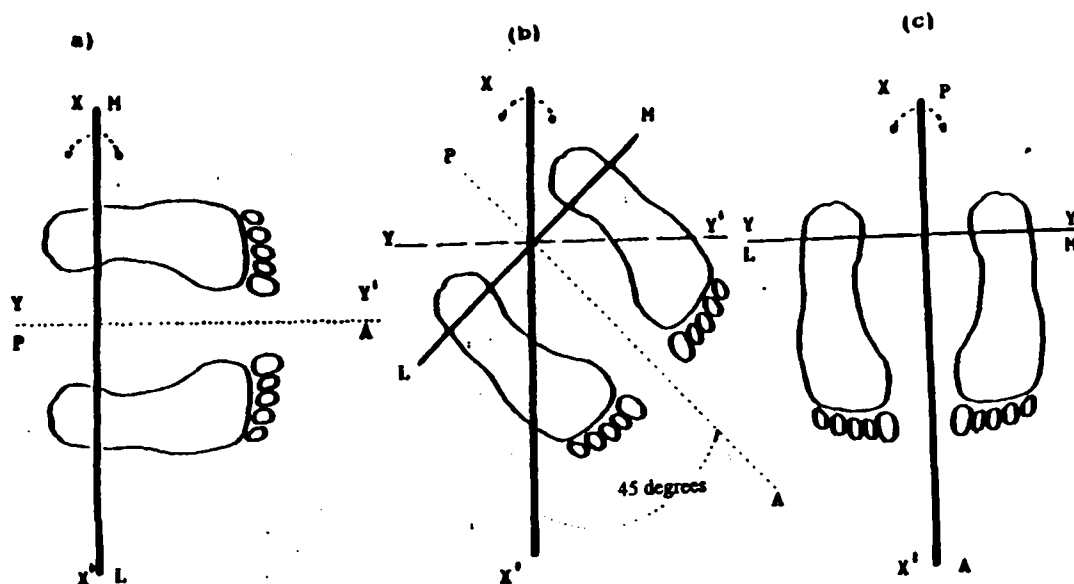Fig. 1. Feet positioning of the subjects for the three asymmetric positions. The platform coordinate system XX', YY' is fixed while the subjects coordinate system AP, ML is rotated with respect to the platform's frame of reference. Asymmetry refers to the angle between the YY' line of the platform and the AP line of the subject: a) 0 degree condition, b) 45 degree condition, c) 90 degree condition

jects to stand on in one of their conditions. Winter et al. [40] suggests that responses to postural perturbations are so task specific that a single assessment technique cannot serve as a true predictor of balance performance.

For this study, a motor task seemed more appropriate. A weight-holding task was chosen because of the ergonomic applicability to MMH tasks in industry. The results of a second pilot study had indicated that holding a weight of 20 percent of body mass did indeed increase the instability of the normal subject for platform perturbations in symmetric conditions [13]. It was important to select a posture and weight-level combination that was challenging without unduly fa-

tiguing any subject. Further, the posture had to be one that would be naturally employed to carry moderately heavy, compact loads. Subjects were instructed to maintain their elbow angle at 90 deg through the duration of the test and keep the load from touching their bodies. Hence the weight was held at elbow level. The weight was presented in a compact box, with rotating handles, which enabled subjects to keep their wrists in the neutral position. The weight level was set at 20 percent of body mass. A percentage rather than a fixed weight was chosen so that the effect would be comparable across all subjects.

Procedure

Table 2. Means and Standard Deviations of the 10 dependent variables grouped by Asymmetry (ASYM) and platform condition (PLAT).

PLAT		VARIABLES*	ASYM=O Mean	ASYM=O SD	ASYM=90 Mean	ASYM=90 SD	ASYM=45 Mean	ASYM=45 SD
P L A T F O R M	S T A B L E	DISP	7.0	2.3	6.0	2.9	6.2	1.4
		LROM	6.2	2.6	5.9	2.7	4.6	2.2
		SROM	7.1	3.4	6.5	1.0	6.7	1.1
		TROM	2.3	1.8	3.8	5.3	3.2	3.3
		LPV	13.8	5.3	15.9	7.8	13.8	5.5
		SPV	18.6	7.0	19.6	5.6	19.0	7.6
		TPV	13.9	9.8	12.4	5.2	11.5	4.0
		LPA	25.8	14.2	26.6	14.2	29.4	11.4
		SPA	40.6	11.1	32.7	11.9	41.1	9.8
		TPA	27.9	18.8	20.8	15.8	29.8	10.2
	L I N E A R	DISP	17.4	1.0	24.9	4.5	27.4	8.1
		LROM	5.5	2.2	6.4	2.1	4.9	2.6
		SROM	9.7	2.5	6.6	3.5	8.5	2.8
		TROM	5.4		5.0	2.4	3.7	2.2
		LPV	3.1		23.9	7.9	16.6	6.4
		SPV	25.9	9.3	29.0	9.4	33.2	6.0
		TPV	25.4	7.8	17.7	7.5	19.3	8.5
		LPA	15.3	7.3	33.5	16.7	33.6	13.8
		SPA	33.4	17.0	51.8	17.4	56.0	14.9
		TPA	48.3	12.9	30.4	15.1	33.9	11.5
			26.4	11.4				
	U P D O W N	DISP	12.5	1.9	25.7	2.3	23.6	2.5
		LROM	5.6	2.1	6.6	1.6	5.6	2.2
		SROM	7.6	4.2	6.4	4.6	8.2	2.1
		TROM	5.7	2.0	3.3	3.4	4.1	3.6
		LPV	22.4	10.5	26.0	16.5	19.5	11.6
		SPV	27.2	10.2	28.3	13.5	28.8	12.3
		TPV	18.1	10.6	20.3	12.7	17.7	9.0
		LPA	33.4	14.6	28.9	13.8	41.6	19.6
		SPA	40.4	17.0	43.7	10.1	46.2	23.2
		TPA	23.9		24.8		28.1	
			16.5		23.1		11.8	

* For the sake clarity of the table units of variables are provided here: DISP is in mm; SROM, LROM, TROM are in degrees ( ); SPV, LPV, TPV are in /s ; and SPA, LPA, TPA are in /s^2

The experiment was conducted in one two-hour session. The subjects were first familiarized with the equipment and the experimental setup and questioned about any history of low back and balance disorders. Their heights and weights were then measured and the weighted box was prepared. The importance of holding this box consistently at a 90 deg elbow angle with the wrists in a neutral posture and with the box not touching the body was impressed on them. They were then fitted with the LMM and asked to simply stand erect in their natural posture. The data from the LMM was checked at this stage to insure that all the channels are recording accurately. Subjects then performed the sequence of tests as determined by the experimental design. They stood in three different positions to simulate the three directions of force plate perturbation while holding the loaded box. Their lumbar and sway index response to the three levels of platform perturbations were recorded.

Statistical Analysis

All statistical analyses were performed by the Statistical Analysis Software (SAS, Cary NC). The first step was to calculate the descriptive statistics of all the dependent variables. The next step in the analysis was to perform a Multiple Analysis of Variance (MANOVA) to determine the significance of the independent variables on the entire set of dependent variables. Analysis of Variance (ANOVA) was then conducted. Post-hoc (Tukey) tests were executed for

those variables, which proved significant from the ANOVA to separate the levels of the independent variables.

3. RESULTS

Descriptive statistics were calculated for all the dependent variables (Table 2). The MANOVA indicated significant effects of asymmetry, platform condition and their interaction on the collective set of dependent measures at the level of $p<0.05$ (Table 3). Hence further univariate analysis was performed for all the effects.

Effect of Platform Movement

Dispersion responded significantly to the platform movement (Table 4). Six lumbar variables also responded significantly to changes in the platform condition. They were the sagittal range of motion (SROM), the peak velocities (SPV, LPV, TPV) in all three cardinal planes and the lateral and sagittal peak accelerations (LPA, SPA).

Post-hoc Tests for the Effect of Platform Movement

The Tukey test for platform movement was performed to distinguish the three levels of platform perturbation (STABLE, UP-DOWN or LINEAR) that were significantly different from one another for each

Table 3. MANOVA and ANOVA p-values of the main and interaction effects for the balance and back motion parameters

		Main Effects		Interaction Effect
	Variables	**ASYM** '	**PLAT**	**ASYM X PLAT**
MANOVA	**All**	0.0001*	0.0001*	0.0001*
ANOVA	**DISP**	0.0001 *	0.0001*	0.0001*
	LROM	0.0713	0.7388	0.4922
	SROM	0.0008*	0.0038*	0.0854
	TROM	0.3827	0.0631	0.1038
	LPV	0.0877	0.0001 *	0.2482
	SPV	0.0529	0.0001 *	0.4944
	TPV	0.7508	0.0101*	0.3866
	LPA	0.0333*	0.0080*	0.0596
	SPA	0.0927	0.0020*	0.0640
	TPA	0.0159*	0.0601	0.1421

Note: * indicates significance at the 0.05 level
All main effects had 2 degrees of freedom and interaction effects had 4 degrees of freedom.

SWAY INDEX (BALANCE EXPERIMENT)
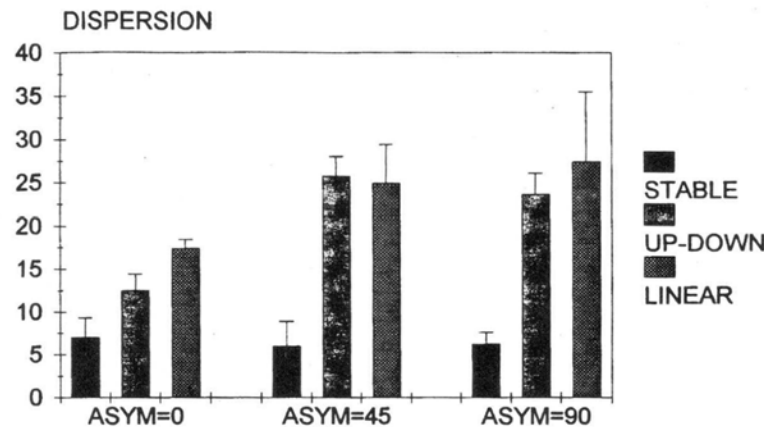## INTERACTION OF ASYMMETRY AND PLATFORM

Fig. 2. Mean and standard deviations of SROM for the three levels of ASYM and PLAT used in this experiment showing the interaction between these two variables. DISP increased almost linearly from the stable, up-down to the linear condition in the 0 degrees condition, but increased dramatically from the stable condition to both the perturbed conditions in the 45 and 90 degrees conditions.

of the 7 significant dependent measures. Table 4 summarizes this post-hoc analysis.

Sway Index

The DISP means were significantly different at all three levels. The STABLE condition of platform had the smallest mean. Dispersion increased significantly in the UP-DOWN condition and further in the LINEAR condition. (Tables 4 and 5)

Lumbar Motion Parameters

All the sagittal variables (SROM, SPV and SPA) were significant (Tables 4 and 5). The largest response was from the LINEAR condition of platform movement, followed by the UPDOWN condition with the smallest mean in the STABLE condition for all these variables. The Tukey test separated the three levels of platform into overlapping groups for SROM: significant increase over the STABLE condition was seen only in the LINEAR condition of perturbation. The sagittal peak velocity (SPV) was separated into two different groups (STABLE and LINEAR/UP-DOVIN in Table 5). The acceleration variable significantly increased in the LINEAR condition of platform movement: the other two conditions were grouped together by the Tukey test.

The significant lateral variables, LPV and LPA, were dynamic parameters had similar responses (Tables 3 and 4). LPV and LPA were highest in the UP-DOWN condition of platform movement followed by

the LINEAR condition. Tukey analysis separated the levels of platform condition into two groups (Table 4). Only the response in the STABLE platform condition was significantly lower than the other two for both these variables: hence the source of the significance lay in the movement of the platform.

In the transverse plane, only peak velocity (TPV) was significant. The highest mean was in the UP-DOWN condition of platform movement, followed by the LINEAR condition, though the Tukey test could not distinguish these two levels. The STABLE condition of platform perturbation had a significantly lower response than both movement conditions.

Effect of Asymmetry

Dispersion increased significantly with increasing asymmetry. Changes in asymmetry had significant effects on lumbar parameters in all three planes (Table 3). The sagittal range of motion (SROM), the lateral peak acceleration (LPA) and the transverse peak acceleration (TPA) were all significant (Table 3). Post-hoc Tukey analyses distinguished those of the three levels of asymmetry (0, 45 and 90) that were significantly different from one another. Table 5 is a summary of the results of this analysis.

Sway Index

The DISP was significantly affected by the Asymmetry. The 90 deg asymmetric condition had the highest sway index followed by the 45 deg condi-

tion. The zero deg condition had the lowest response and was significantly lower than the other two groups (Table 5).

Lumbar Motion Parameters

All the significant lumbar parameters showed an interesting U-shaped trend in response to increasing asymmetry. SROM was the only sagittal variable significantly affected by asymmetry. The highest mean was in the 0 deg asymmetric condition, followed by the 90 deg and then by the 45 deg asymmetric condition. Only the 45 deg condition (smallest mean) was significantly different than the other two.

The LPA was significant between asymmetry conditions. The lateral peak acceleration had the highest mean in the 90 deg asymmetric condition followed by the 0 deg condition. The smallest mean was in the 45 deg asymmetric conditions, but the Tukey test could not identify the differences to be significant.

The TPA was significantly affected by the Asymmetry (Table 3). The order of the means from highest to lowest was 90, 0 and 45. Again, the Tukey test could not identify the separate groupings (Table 5).

Interaction (Plat x Asym) Effects

The only dependent variable that was significant was the dispersion (Table 3). Figure 2 shows the response of the sway index to the asymmetry and platform. Platform movement elicits larger responses than the STABLE condition, but the increase is more dramatic when asymmetry is 45 and 90 deg. There does not seem to be any discernable difference between the dispersion index in the 45 and 90 deg asymmetry conditions.

4. DISCUSSION

In this study, subjects performed a simple load-holding task in addition to the task of balance maintenance for different support surface movement conditions. Subjects simply held a 20% of body weight load with the elbow angle at 90 deg. The reasons for choosing this task, level of weight and posture were guided by considerations of risk of injury, onset of fatigue and a moderate level of challenge. The maximum and minimum load that any subject was asked to hold was between the AL and the MPL [31]. This

Table 4. Post-hoc (Tukey) test for the effect of platform movement on the significant variables (from ANOVA). Means with the same letter are not significantly different (N=54 and α >.01).

Variable	Tukey Grouping	Mean	PLAT
DISP	A	23.29	Translation
	B	20.63	Rotation
	C	6.42	Stable
SROM	A	8.31	Translation
	B A	7.46	Rotation
	B	6.84	Stable
SPV	A	29.26	Translation
	A	28.12	Rotation
	B	19.50	Stable
SPA	A	52.08	Translation
	B	43.45	Rotation
	B	38.17	Stable
LPV	A	22.60	Rotation
	A	22.19	Translation
	B	14.56	Stable
LPA	A	34.68	Rotation
	A	33.55	Translation
	B	27.32	Stable
TPV	A	18.70	Translation
	A	17.40	Rotation
	B	12.66	Stable

task could relate to an assembly operation task in industry where the operator is holding a part in his hand for the next job to reach him on the conveyer belt for subsequent assembly. Extrapolated to carrying, the posture and weight (between 10 and 18 kg) is seen in a large number of load- carrying tasks in industry and in daily activities.

Subjects held this weight at elbow level. This large vertical distance might have been expected to substantially increase the required moment of force [15,40] needed to be produced by the ankle musculature in postural compensation. This was supported by the large increases in dispersion due to load holding versus the no load condition seen in the second pilot study [13]. This agrees with Davis [9] who quantified stability as antero-posterior and lateral ankle-angle subtended by the Center of Gravity and found that "stability decreases when laden".

Effect of Platform Movement

The sway index (DISP) responded significantly to the platform movement (Table 3). All three conditions of movement were distinguishable by the Tukey test (Table 4). The translational perturbation caused more sway than the rotational perturbation of the platform. The dispersion (sway index) found in this experiment had a range of values that were, in general smaller than the data in Byl et al. [8] but well within the range of the normative data for dispersion quoted by these authors. The variability in the data was also much less compared to the normal population tested by them. A possible explanation may lie in the tested subject population and the differing protocols used. Six of the lumbar motion parameters responded significantly to platform movement (Tables 3,4). The translational perturbation also affected the sagittal

lumbar motion variables (SROM and SPA) more than the rotational perturbation. Keshner et al. [17] found that muscles of the lumbar region were activated before the soleus and the hamstring in flexion-extension perturbations suggesting a heavier use of the "hip-strategy" of balance maintenance while for AP translations, the lower extremity muscles were recruited before or simultaneously, suggesting a heavier use of the "ankle strategy" in such perturbations. In this study, neither type of perturbation (horizontal translation or angular rotation) could be singled out as inducing more low-back response. The significant difference (post-hoc Tukey) lay not between the two perturbation conditions, but between the stable condition and both perturbation (horizontal translation and rotational) conditions (Table 4).

This study however differs from Nashner et al. [29,30] and Keshner et al. [17] in that only the asymmetric condition of 0 degrees corresponded to "pure" disturbances. The UP- DOWN condition of platform movement provides a pure flexion-extension of the ankle joint only in the asymmetry condition of zero because the ankle axis of rotation is aligned with the axis of rotation of the platform (Fig. 1). In the 45 deg asymmetry condition, the platform rotation forces a mixture of plantar flexion and extension and lateral inversion-eversion. Platform rotation in the 90 deg asymmetric condition provides a pure inversion-eversion pattern because there is no sagittal component to the rotation. Similarly, horizontal movement of the platform is an AP translation only in the asymmetric condition of 0 deg- the 90 deg condition corresponds to a pure medial lateral disturbance while the 45 deg asymmetric angle was a combination of both AP and medial lateral translation. Hence the type of movement induced was a function of the asymmetry.

Table 5. Post-hoc (Tukey) test for the effect of asymmetry on the significant variables (from ANOVA). Means with the same letter are not significantly different (N=54 and α >.01).

Variable	Tukey Grouping	Mean	ASYM
DISP	A	19.08	90
	A	18.92	45
	B	12.34	0
SROM	A	8.18	0
	A	7.88	90
	B	6.55	45
LPA	A	34.91	90
	A	30.91	0
	A	29.73	45
TPA	A	30.68	90
	A	26.11	0
	A	25.38	45

Effects of Asymmetry

The investigation of the effects of asymmetry on the lumbar involvement in postural balance maintenance was the major theme of this experiment. The underlying idea is that because of the relative "inability" of the ankle joint in compensating for asymmetric perturbations, the "hip strategy" would need to be used more in asymmetric perturbations. This was expected to show up in increasing values of the dispersion and kinematic variables with more asymmetry: viz. range of motion, peak velocity and peak acceleration.

The dispersion was significantly higher in the asymmetric conditions (90 and 45 degree) than the 0 deg condition (Table 5). However, the response of the low-back parameters to asymmetry showed a trend inconsistent with these hypotheses. The 45-degree condition seemed to elicit the least low-back response in all the cardinal planes of motion. The sagittal range of motion (SROM) had the highest response in the zero degree condition. This value dropped significantly (Table 5) in the 45 deg conditions and then increased in the 90 deg condition. LPA was the highest in the 90 deg condition and decreased somewhat in the zero degree condition and further in the 45 deg condition. TPA also showed a similar response. All kinematic variables showed a U-shaped kind of response. An experiment with more levels to the asymmetry variable, that also measures the dynamic involvement of ankle motion, might provide a clearer picture of this trend.

Interaction effects

Dispersion changed significantly between the levels of the platform variable. However the increase was more dramatic with increasing asymmetry: i.e. the interaction effect was very significant as well (Table 3). Dispersion increased only slightly in the 0 deg condition from the stable condition to the movement conditions of the platform. In contrast, this increase was much more dramatic in both asymmetric conditions (Figure 2).

The dispersion is a sensitive index of whole body stability and quantifies two-dimensional distances of the COP's excursion from the time-averaged mean. This result indicates that the asymmetry of the perturbations caused larger COP deviations from the mean position. Simulations of biped postural stability [16] have shown that the ankle (which is the body's first choice in compensating underfoot perturbations) does not contribute significantly to stability in the frontal plane and other, perhaps less suitable strategies need to be used when ankle compensation is not enough [8,30]. This decreased overall stability bears out the initial supposition that the body's balance mechanism is not so effective in compensating for asymmetric perturbations as it is for symmetric movements. This relation-

ship of decreased stability with increasing asymmetry has implications to safety in the workplace. In this experiment, these instabilities were caused by underfoot perturbations. However, activities like lateral bending, twisting, one-handed manipulation of weights and laterally shifting loads also cause instability in non-sagitally symmetric directions [15]. Lateral shear forces at the shoe-floor interface are inevitable in such MMH tasks. Grieve [14] discuss from their slip- chart model that a slip is almost inevitable in some manual materials handling tasks. Modeling of the biomechanics of hip fractures by Robinovitch et al. [35] has shown that lateral impacts pose a greater risk of fracture. Hence such disturbances are not only more difficult for the human body to compensate, but entail a higher risk of serious injury if they result in impacts.

The Role of Unexpectedness

Close inspection of the data indicated that the peak dynamic (velocity and acceleration) parameters occurred immediately (less than 0.5 seconds) after initial platform perturbation. The next cycles of platform perturbation did not seem to affect lumbar response as significantly [13]. The initial activity is perhaps attributable to the degree of unexpectedness associated with the experiment. As the subjects were not instructed to close their eyes, and the test was initiated by hitting keys on the computer keyboard, they got visual and audio feedback about the timing of the start of the test: however, the type of platform condition was unknown to them. Marras et al. [25] found that mean trunk muscle forces for unexpected loading situations exceeded those for expected situations by a factor of 2.5 and peak muscle forces were 70% higher. The onset rate of all the muscle forces, were also higher when the load was unexpected. Hence, unexpectedness plays a major role in the muscular response of the lumbar region to load handling. In other words, a sudden unanticipated loading serves as a "trigger event"; causing the muscular system to over-react and thus damaging already weakened tissues containing noiceceptors. He speculates that handling of loads within the recommended AL [31] could lead to peak compressions exceeding these values in situations of sudden spinal loading. The cumulative trauma pathway of low-back injury has been the focus of ergonomic research of late and the documentation of these underfoot, unexpected events immediately preceding lumbar injury in industrial settings, are poor. Further research is needed quantifying lumbar involvement in the biomechanics of unexpected events that cause lumbar injury. The present study tried to address the issue of the kinematic lumbar involvement in the postural control mechanism. Inclusion of such kinematic parameters as lumbar velocity and acceleration in these models might lead to better risk of falling prediction.

Results obtained by those authors who have tried to build predictive models of risk of falling accidents based on COP parameters (like median frequency, preferred directionality of the movement of the locus etc.) have shown that more factors may be involved. Gehlsen et al. [12] found only a moderate (R squared=0.37) relationship in their model (which included traditional strength and range of motion predictors in addition to performance in balance tests). They attempted to classify elderly patients as falters and non-fallers. Fallers demonstrated significantly smaller hip and ankle ranges of motion and lesser leg strength. Bartlett et al. [5] constructed a model based on the speed of sway of the movement of the COP to classify falters and non-fallers. They found that their model had a large probability of misclassification (95% confidence interval of the probability was 0.38 to 0.49). These authors discuss that one of the reasons for this is that the model failed to account for other factors contributing significantly to falling liability. Perhaps one of the missing elements is the kinematic involvement of the higher joints, which these models have not included.

According to Nashner et al. [29,30], support surface disturbances create moments of force not only about the ankle, but also about the higher joints like the knee and the trunk. The large mass of the upper body and the CG of the whole human body make for an inherently unstable system. The moment of inertia of the head, arms and trunk (HAT) about the ankle joint is about 55 $kg.m^2$ while that about the hip is 7 $kg.m^2$ [40,41]. "Because the moment of inertia about the ankle is about 8 times that about the hip, the ankle muscles would need about eight times the moment-of-force to accomplish the same angular acceleration of the HAT as would the hip muscles." Hence it would seem that the involvement of the lumbar region is perhaps a more attractive strategy for compensatory accelerations of the HAT. Byl et al. [8] showed that there are differences in the ways in which normals and LBP patients use their backs in compensatory mechanisms, an indication that impairment in the low-back leads to different, if not less effective pathways, of neuro-muscular postural control. Biomechanically, this means that kinematic involvement of the lumbar region might be a more important factor in the maintenance of posture than has yet been realized.

Limitations of the Study

It would seem that most of the low-back involvement was the result of unexpectedness in this study as well. There was no method of introducing an unexpected asymmetry of perturbation with the Balance System as horizontal or rotational displacements of the platform were possible only in one direction and the subject had to be turned in asymmetric directions. Temporal unexpectedness could be incorpo-rated into a similar experiment in many ways. The subjects could be deprived of both audio and visual feedback by the test being initiated remotely, that is the platform perturbations could be started by the experimenter at random times outside the visual or audio range of the subjects, for example from outside the room. Alternatively the Balance System software could start platform perturbations so that the platform moves after a random amount of time without the need of the experimenter to actually hit any keys.

Suggestions for Future Research.

Traditionally, clinical evaluations of balance performance have used postures, which are simple variations of upright standing [8,17,20,21,27,33] without any weight handling being required. Load-holding was incorporated into this experiment because of the ergonomic applicability in providing insight into the postural control mechanisms employed while manipulating an external load. Few other studies [9,15] in the surveyed literature used load-holding. However, most MMH tasks in industry require the manipulation of substantial external loads. In addition, a large number of these industrial tasks also necessitate trunk flexion/extension, twisting and lateral bending. Oddsson [32] showed that even without load-handling, such tasks are inherently more unstable than normal standing because of displacement of the body segments. He conducted an experiment on stability and control in voluntary flexion and extensions of the trunk and found that the COP oscillates, on the average, about six times faster in such tasks than in pure standing. Handling loads further destabilizes this system: underfoot accidents like slips and trips are even more difficult to recover from. He speculates that if a person carrying or lifting a load loses balance, then the first concern of the CNS is to regain equilibrium. The resulting response would then perhaps be "beneficial for posture, but be potentially harmful for load carrying tissue, such as muscles, tendons and ligaments". Such choice of one strategy at the cost of another can help explain why a large number of slips that do not actually result in impacts cause the onset of LBP. Future experiments could be designed to evaluate the lumbar involvement in the balance responses to underfoot perturbations during more ergonomically relevant tasks: for example load-carrying (walking with the load) or lifting tasks.

Schultz [37] discusses the causes of mobility impairments and its relationship to balance. According to him, the kinematic and kinetic parameters of activities (like ROM, muscle strength and joint torques requirements) are critical elements of balance maintenance and need to be identified and quantitatively documented so that they can be compared against maximum capacities. If the requirements are well within these capacities, there is a greater degree of neuromuscular control and a smaller

neuromuscular control and a smaller risk of accident. Kinetic data from experiments of underfoot perturbations when common MMH tasks (like lifting, carrying etc.) are being performed provide the basis for biomechanical analysis of required torques at the higher joints for these activities. This knowledge could be further used to establish thresholds of postural stability and identify MMH tasks, which are potentially at risk of causing imbalance accidents.

Most experiments that have investigated the postural control issue have looked at the temporal aspects of muscle EMG patterns [17,30,34]. The focus was on the latencies in muscle activation (from the time of perturbation) to identify motor strategies in balance maintenance. No attempt was made in these studies to quantify the muscular forces involved or to model the loading of the joints in these mechanisms. However, the quantification of the impulsive loading on the spine that occurs during slips, trips and falls require an estimation of these internal muscle forces. EMG analysis could be incorporated into future experiments to be used in the 3-D biomechanical modeling of spinal loading. It might be particularly interesting to see if asymmetric perturbations, as used in this experiment, lead to asymmetric patterns of lumbar muscular activity. Spinal shear would result from unbalanced medial-lateral, anterior-posterior forces and torsion. These shear forces are thought to increase the propensity for injury. This might shed more light on the processes that lead to the large incidence of low-back injury during the recovery- response from slips.

5. CONCLUSIONS

This experiment investigated the role of the lumbar region in a simple load-holding situation for three asymmetric (0, 45 and 90 deg) platform perturbations for 18 normal, college age males. The weight was 20% of the subjects' body weight. The platform movement consisted of a stable condition, a linear translation condition and an angular rotation condition. Significant reactions to the platform movement were seen in the sway index (DISP) and six other kinematic parameters. They were the sagittal range of motion (SROM), the peak velocities in all three cardinal planes and the lateral and sagittal peak accelerations (LPA & SPA). Significant reactions to asymmetry were seen in the sway index and lumbar parameters in all three planes (SROM, LPA and TPA). Unique combinations of asymmetry and platform movement significantly affected Dispersion.

The results indicate that the lumbar region is definitely involved in the corrective mechanisms of balance maintenance. In general, most of the significant low-back parameters were dynamic (velocity and acceleration) parameters rather than positional (ROM)

parameters. This shows that biomechanically, the task of maintaining uprightness under different conditions of instability involves small changes in the range of motion, but requires relatively quick reactions to keep the CG within the bounds of the supporting base. The kind of platform movement (rotation or linear translation) was shown to be less important in determining the extent of lumbar involvement.

The issue of kinematic lumbar involvement in postural control mechanisms is worthy of further investigation and is directly applicable to the safe, biomechanical design of the workplace. Future studies should consider the effects of stronger, unexpected perturbations on normal subjects to simulate realistic slips, trips and falls, whilst more ergonomically pertinent tasks (i.e. lifting and carrying, asymmetric load-handling) are being performed. Subsequent kinematic and kinetic modeling of joint torque requirements and reactions necessary to maintain balance may shed more light on lumbar loading and on the processes that lead to the high incidence of low back injuries following slips and trips.

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