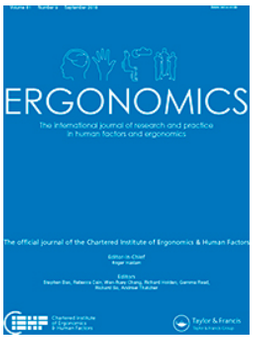

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Low back joint loading and kinematics during standing and unsupported sitting

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Keywords: Sitting position; posture; standing position; electromyography; lumbar spine; joint forces.

The aim was to examine lumbar spine kinematics, spinal joint loads and trunk muscle activation patterns during a prolonged (2 h) period of sitting. This information is necessary to assist the ergonomist in designing work where posture variation is possible — particularly between standing and various styles of sitting. Joint loads were predicted with a highly detailed anatomical biomechanical model (that incorporated 104 muscles, passive ligaments and intervertebral discs), which utilized biological signals of spine posture and muscle electromyograms (EMG) from each trial of each subject. Sitting resulted in significantly higher ($p < 0.001$) low back compressive loads (mean ± SD 1698 ± 467 N) than those experienced by the lumbar spine during standing (1076 ± 243 N). Subjects were equally divided into adopting one of two sitting strategies: a single 'static' or a 'dynamic' multiple posture approach. Within each individual, standing produced a distinctly different spine posture compared with sitting, and standing spine postures did not overlap with flexion postures adopted in sitting when spine postures were averaged across all eight subjects. A rest *component* (as noted in an amplitude probability distribution function from the EMG) was present for all muscles monitored in both sitting and standing tasks. The upper and lower erector spinae muscle groups exhibited a shifting to higher levels of activation during sitting. There were no clear muscle activation level differences in the individuals who adopted different sitting strategies. Standing appears to be a good rest from sitting given the reduction in passive tissue forces. However, the constant loading with little dynamic movement which characterizes both standing and sitting would provide little rest/change for muscular activation levels or low back loading.

1. Introduction

With increasing computer and deskwork associated with most jobs; sitting has become an integral part of most working environments. While prolonged sitting has often been identified as being associated with back pain (Frymoyer *et al.* 1980, Wilder *et al.* 1988, Chaffin and Andersson 1990) and specifically disc herniations (Kelsey 1975a, b), the link has been suggested to be due to the flexed curvature of the lumbar spine (Wilder and Pope 1996). To develop a scientific foundation to justify injury prevention strategies, this epidemiological evidence should be coupled with an understanding of the resultant tissue loading that leads to occupationally related disorders.

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While many studies have documented static spine postures, there have been a few that have examined the variation in spine postures assumed when seated for extended periods. Keegan (1953) noted that seated postures resulted in an increased flexion of the spine, with some postures approaching the same spinal flexion as a fully stooped position. However, most studies of spine postures appeared to have primarily examined postures assumed by subjects for very short periods, with representative measures of lumbar curvature recorded (Bendix and Hagberg 1984, Brunswic 1984, Frey and Tecklin 1986, Link *et al*. 1990, Eklund and Liew 1991, Bridger *et al*. 1992, Lord *et al*. 1997). Two studies have examined the variation in spine posture over a period of seated work (Bendix and Biering-Sorensen 1983, Bridger 1988). For example, Bridger (1988) examined seated work for 20 min with a measure of trunk angle taken from video once every minute, but there was no examination of the variation in trunk postures over the 20 min — only a comparison between average trunk angles adopted with different chairs and desks (sloped versus horizontal). Bendix and Biering-Sorensen (1983) also examined the effect of different chairs on trunk position. Six measures of lumbar spine angle were taken over 1 h with little or no change being found. When *in vivo* analysis of the relative range of flexion from upright standing to full flexion (Adams and Hutton 1982, Pearcy *et al*. 1984) is compared with lumbar spine motion during sitting, it appears possible that some motion segments could be strained close to the maximum of their *in vivo* limit. However, the seated postures examined were recorded for contrived positions assumed for the measurement of spine posture.

Intradiscal pressure in the lumbar spine, when seated, has been reported to increase when compared with standing (Nachemson 1966) although the link between disc pressure and spine loads has been brought into question due to the lack of incorporation of the load bearing capability of the apophyseal joints (Adams and Hutton 1981). In contrast the *in vivo* spine has been shown to increase in stature when seated after a period of standing (Althoff *et al*. 1992), leading to the conclusion that spinal loads are actually lower while seated. The comparison of muscle recruitment levels between standing and sitting postures has revealed little difference (Chaffin and Andersson 1990, Althoff *et al*. 1992). This lack of difference in muscle activation levels occludes any clarification in the difference in joint loading between seated and upright postures.

The primary purpose of this work was to determine low back loading and postural responses to a prolonged sitting posture. Secondary purposes were to assess the difference in joint loading between standing and a seated position and to determine whether spine postures are altered significantly to allow for unloading of the disc and/or load migration to any other tissues at risk. Additionally the range of motion of the lumbar spine was assessed before and following sitting to examine whether prolonged sitting increases the laxity of the spine. A further goal of this research was the examination of whether the physical demands of standing differ significantly from those of sitting so that this can provide a basis for resting the passive and active tissues in the seated worker.

2. Methods

2.1. *Participants*

Eight male volunteers were recruited from a university student population (age mean 22.4 years, SD 2.4, height mean 174.7 cm, SD 9.0, mass mean 74.4 kg, SD 7.0). All subjects were healthy and had not experienced any low-back pain for a minimum of 1

year. Informed consent was obtained from all subjects for the protocol, which had been reviewed and had received prior approval from the Human Research Ethics Committee of the University of Waterloo's Office of Human Research and Animal Care.

2.2. *Instrumentation*

Fourteen pairs of disposable Medi-Trace surface electromyogram (EMG) electrodes (Ag-AgCl) were applied to the skin bilaterally over the following muscles: rectus abdominis, 3 cm lateral to the umbilicus; external oblique, ~15 cm lateral to the umbilicus; internal oblique below the external oblique electrodes just superior to the inguinal ligament; latissimus dorsi, lateral to T9 over the muscle belly; thoracic erector spinae, 5 cm lateral to T9 spinous process; lumbar erector spinae, 3 cm lateral to L3 spinous process (McGill 1992); and multifidus, 3 cm lateral to L5 spinous process (MacIntosh *et al.* 1986). Before data collection, all subjects performed maximal isometric contractions for all monitored muscle groups to enable EMG normalization. Procedures for obtaining maximum myoelectric activity for normalization have been previously explained in McGill (1991). The raw EMG signal was prefiltered to produce a band width of 20 – 500 Hz and amplified with a differential amplifier (common-mode rejection ratio >90 dB at 60 Hz and input impedance >10 Mohms at >1 Hz) to produce peak to peak amplitudes of ~2 V. The amplified signal was A/D converted at 1024 samples s^{-1}.

Lumbar flexion/extension angle was monitored with a 3SPACE ISOTRAK (POLHEMUS, PO Box 560, Colchester, VT, USA) and A/D converted using customized software at 20.5 samples s^{-1}. The ISOTRAK source, which produces an electromagnetic field, was mounted on the sacrum using a custom-built harness and the sensor, which detects the rotational motion (three-directional cosines) with respect to the source, was mounted over the trunk midline at the T12/L1 spinal level. The zero position for lumbar flexion/extension angle was taken as normal relaxed upright standing.

Synchronization of the ISOTRAK and EMG signals was accomplished in the following way. The computer controlling the ISOTRAK at the beginning of the trial sent a pulse through the A/D converter of a second computer, which initiated collection of the EMG signals.

2.3. *Data collection and processing*

Collection protocol consisted of three tasks during which the level of muscle activation and spinal loading present were differentiated: sitting for 2 h, two trials of standing for 3 min each and four lumbar spine range of motion (RoM) tests (figure 1). The lumbar spine RoM tests were completed before and after each task. While some individuals often sit for periods exceeding this duration, 2 h was thought to represent the longest typical uninterrupted period of exposure for most office employees before receiving a coffee or lunch break. The subjects sat at a computer workstation while performing uncontrolled deskwork (reading, homework, computer work, etc.). The chair used by each subject was the seat pan of a normal stenographic type chair with the back support removed. A calibration posture (trunk flexed to 60° from the vertical with a lordotic spine, holding a 10 kg mass in the hands) was recorded before the sitting session (Neumann *et al.* 1995) to provide an EMG to moment relationship.

Both the EMG and the lumbar spine kinematics were sampled for 3 min out of every 15 min of sitting and for the full duration of the standing trials. The time-

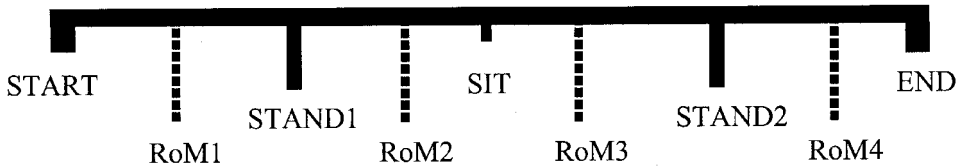

Figure 1. Time line of the testing protocol for the eight subjects.

varying demands on the lumbar spine and trunk musculature experienced by the subjects during sitting and standing were the focus of the study. Therefore, the amplitude probability distribution function (APDF) was used to examine the activation of the 14 muscles recorded and the flexion/extension spine postures. Briefly, digital processing of the raw EMG signals included full wave rectification followed by a Butterworth low-pass filter (2.5 Hz cut-off frequency) to produce a linear envelope. The filtered signals were then normalized to the maximum muscle activation that was elicited during the isometric contractions and synchronized to the ISOTRAK signal by reducing the EMG sample rate to 20.5 samples s^{-1}. Lumbar spine flexion was normalized to the first RoM test performed by the subject at the beginning of the test period. The normalized EMG and lumbar spine kinematics levels were then binned in 1% increments to generate an APDF for each individual in sitting and standing as well as an APDF function across all eight subjects for each of the 14 EMG channels and flexion/extension of the lumbar spine.

2.4. *Estimation of passive tissue, muscle and joint forces*

The 3D moments and L4/L5 joint forces were calculated using an anatomically complex, dynamic EMG-driven low back model incorporating 104 muscles (McGill and Norman 1986, McGill 1992, Cholewicki and McGill 1996). The model utilizes calibrated and normalized EMG and low back kinematics to predict passive tissue and muscle forces (including forces on the intervertebral disc, ligaments, gut and lumbodorsal fascia). Briefly, first the passive tissue forces are predicted by assuming stress–strain or load–deformation relationships for the individual passive tissues. These passive forces are calibrated for the differences in flexibility of each subject by normalizing the stressstrain curves to the passive range of motion of the subject, detected by electromagnetic instrumentation that monitors the relative 3D lumbar angles. Then the remaining moment is partitioned among the many laminae of muscle based on their myoelectric profile and their physiological cross sectional area, and modulated with known relationships for instantaneous muscle length and either shortening or lengthening velocity (Sutarno and McGill 1995). This method of using biological signals to solve the indeterminacy of multiple load bearing tissues facilitates the assessment of the many individual ways in which loads are supported. In other words, using the EMG and spine posture signals, the model was sensitive to the subtle differences that occurred in each trial of every subject. Furthermore, the model was tailored to each individual using the calibration posture which allowed a relationship between EMG activity and reaction moment (while holding a 10 kg load) to be established. This provided a known flexion/extension reaction moment, calculated through a rigid link segment model. The resulting muscle activation levels and lumbar spine postures were used to drive the EMG model. The gain or error term within the EMG model, which essentially balances the reaction moment and the

moment produced by active and passive tissues, was then fixed for each individual. This approach has been used previously to assess low back loads during gait and yielded good agreement between measured and modelled moments (Callaghan *et al.* 1999).

2.5. *Statistical analyses*

One-way repeated measure analyses of variance (ANOVA) were performed on all levels of activation of the APDFs of the 14 EMG channels, lumbar compression in sitting versus standing, and lumbar spine RoM to determine whether there were significant differences between the sitting and standing activities. A 0.05 level of significance was used for rejection of the null hypotheses in all cases. *Post-hoc* multiple comparisons (Bonferroni) were used to examine matched data between the tasks when a significant difference was found.

3. Results

There were two clear patterns of sitting strategies adopted by the individuals in this study, or in other words not all subjects sat the same way. Generally, either a single 'static' posture was adopted for the duration of sitting or a more 'dynamic' multiple postures approach was taken (as shown by the APDFs in figures 2 and 3). The subjects were equally divided into the 'static' and 'dynamic' sitting behaviours. The APDFs function of the eight individual sitting and standing distributions (figure 4) show a distinctly different spine postures.

The peak flexions of the lumbar spine during the four RoM trials are shown in table 1. There was a trend towards an increased flexion following standing. The differences were not statistically significant.

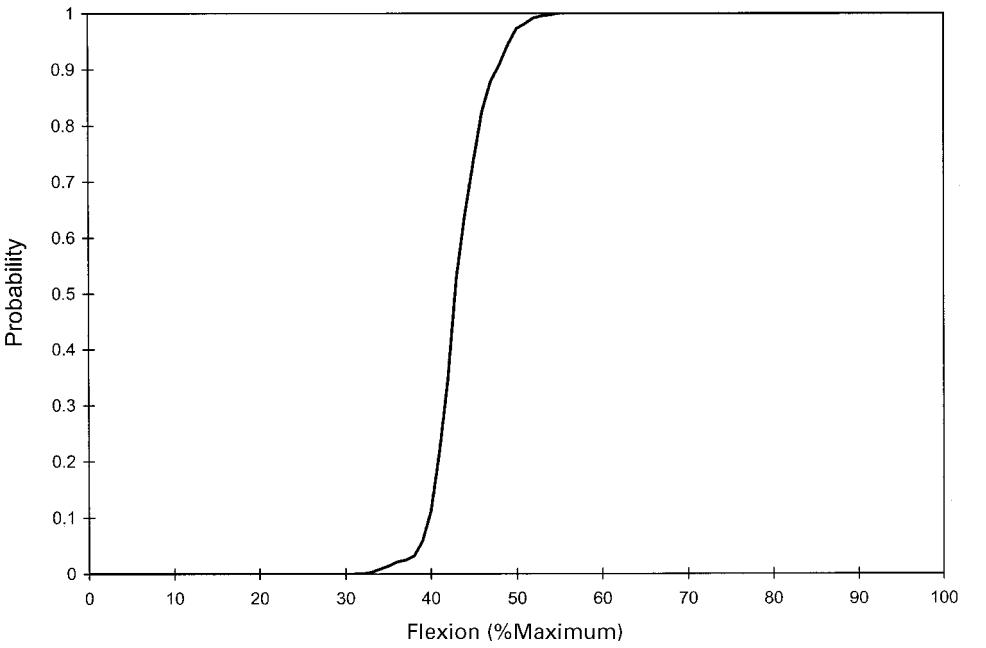

Figure 2. Amplitude probability distribution function of a static lumbar spine posture adopted by one subject for the duration of a 2 h sitting.

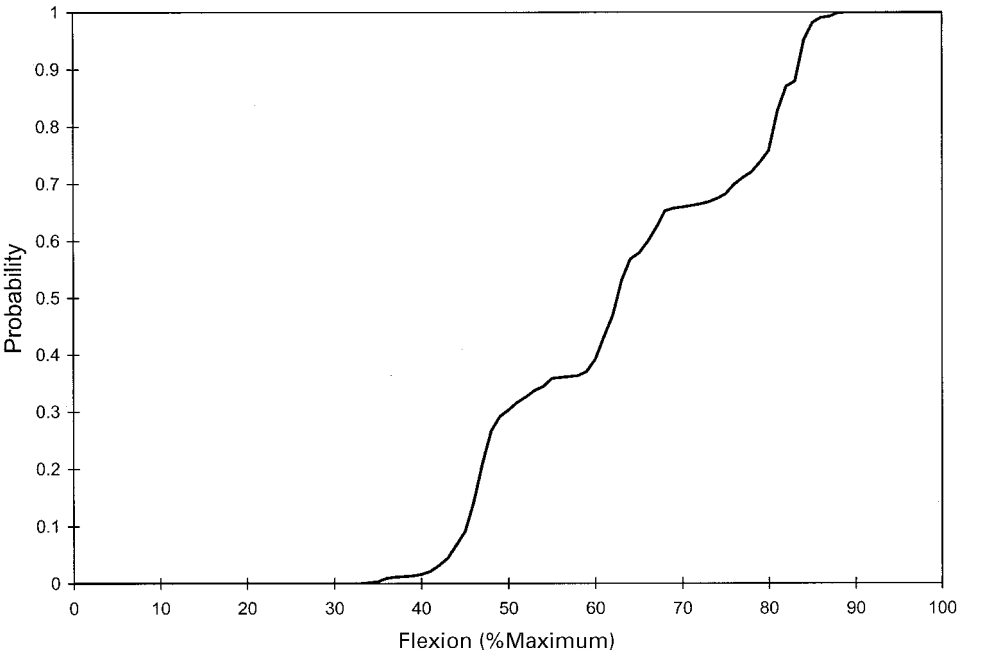

Figure 3. Amplitude probability distribution function of one subject who adopted 'dynamic' lumbar spine postures that varied over a wide range for the 2 h sitting. There are three clear postures in which the individual equally divided the sitting period.

The joint forces at L4/L5 calculated using the EMG driven model demonstrated higher low back compressive forces ($p < 0.001$) in the postures assumed when sitting compared with standing. In fact, on average compressive force was 500 N greater in sitting (figures 5 and 6). Individuals who exhibited 'static' sitting behaviour had one posture used to calculate joint compression in sitting. Individuals with 'dynamic' sitting behaviour had compression forces calculated for postures selected from their APDF of spine flexion to represent the range of postures to which they were exposed. The average EMG levels from all postures that matched with the selected postures were averaged to yield a representative muscle activation profile for that spine position.

The increase in joint compression when seated was due to increased force levels in the extensor musculature, as a result of slight increases in levels of activation, and increased passive force contribution (in both the ligaments and muscles) due to strains associated with a more flexed lumbar spine in sitting (shown for an individual in table 2). Both standing and sitting spine compressive forces were below the lifting biomechanical compressive tolerance of 3400 N proposed by NIOSH (1981).

There was no significant difference between the joint compressive forces calculated for the standing trial following sitting for 2 h compared with the standing trial before sitting (figure 6). Average anterior–posterior joint shear forces were higher ($p < 0.001$) during sitting than standing (135 ± 200 N and -13 ± 17 N respectively, where a positive shear indicates an anterior shearing of the trunk on the pelvis). However, both values are substantially below *in vitro* anteriorposterior

shear tolerances of 500 N for prolonged work (McGill *et al*. 1998). The medial lateral joint shears in both sitting and standing were insignificant (~30 N).

The upper and lower erector spinae muscle groups exhibited a shifting of the average APDF curves to the right during sitting, which indicates a lower probability of the same activation level occurring, thereby resulting in more time spent at higher activation levels. These shifts were statistically significant at the 1% MVC level for all four channels (minimum level of significance, $p < 0.05$). The upper erector spinae was also significantly shifted to the right at the 2% MVC level bilaterally (right, $p < 0.01$; left, $p < 0.05$) and at the 3% MVC level ($p < 0.05$) for the left upper erector spinae across subjects (figure 7). The right latissimus dorsi demonstrated a significant

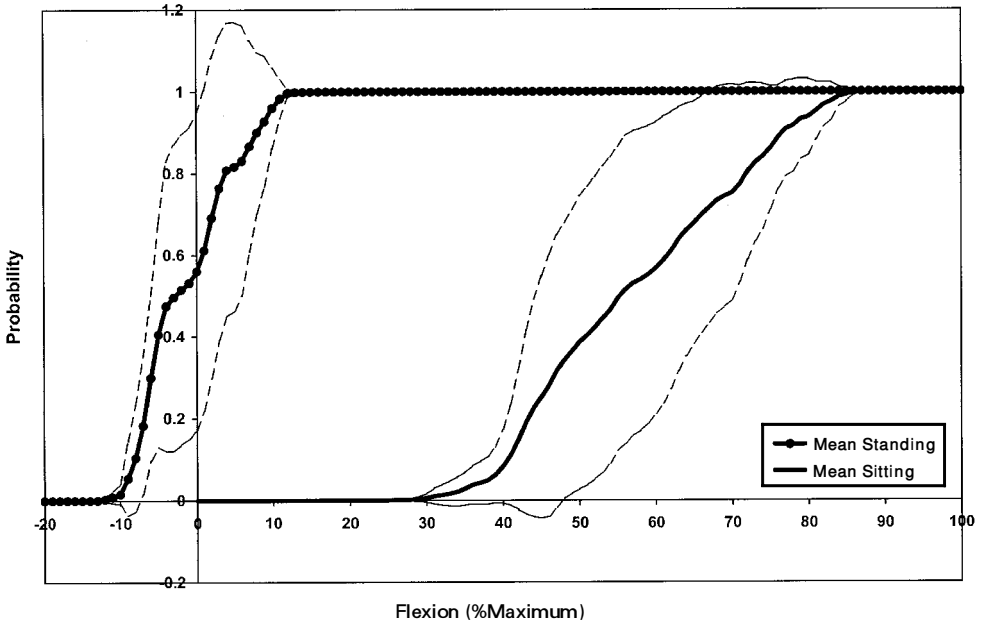

Figure 4. Average amplitude probability distribution function (±1 SD) of the postures adopted by the eight subjects while sitting and while standing.

Table 1. Peak lumbar spine flexion from the range of motion tests for each subject expressed as a percentage of the flexion in the initial test (RoM1).

Subject	RoM1	RoM2	RoM3	RoM4
1	100.0	102.0	101.8	101.9
2	100.0	105.3	102.9	110.1
3	100.0	101.0	99.8	102.3
4	100.0	98.8	94.5	97.0
5	100.0	97.0	96.4	98.2
6	100.0	100.7	95.9	96.1
7	100.0	105.3	109.2	113.9
8	100.0	98.8	100.4	103.0
Mean	100.0	101.1	100.1	102.8
SD	0.0	3.0	4.7	6.3

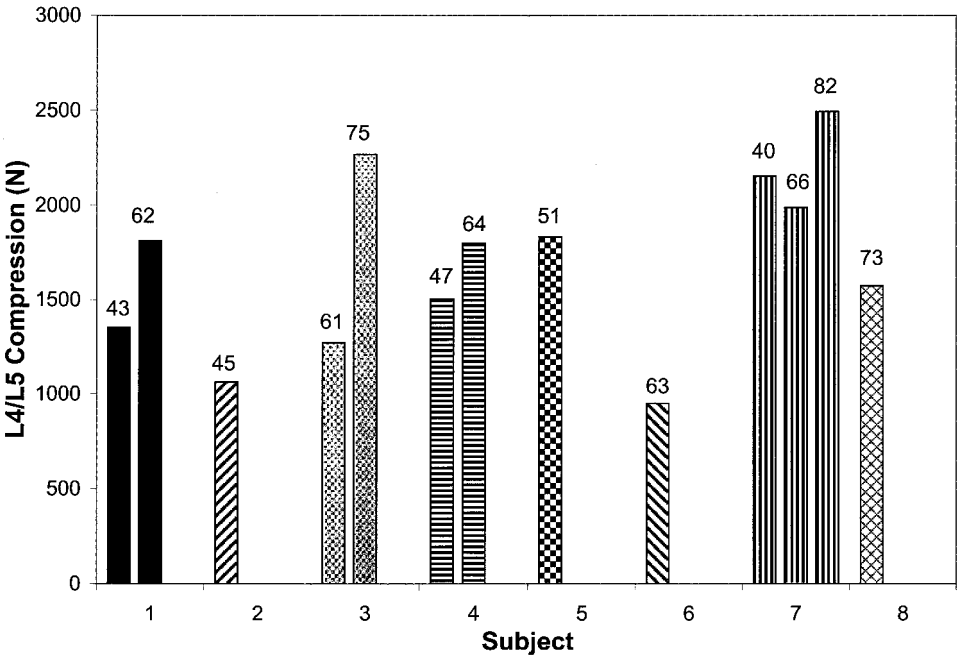

Figure 5. L4/L5 joint compression forces calculated for the sitting postures for each subject, selected based on the ADPF of their lumbar spine flexion to represent the range of postures to which they were exposed. Subjects with 'static' sitting behaviour had one posture whereas subject with 'dynamic' sitting behaviour had compression forces calculated for multiple postures. The lumbar spine flexion (% RoH) is indicated above each column.

shift ($p < 0.001$) to the right during sitting across subjects, but there was no corresponding change on the left side. The significant changes in the APDF curves represent a shift of the probability in the range 0.26–0.4 for a given muscle activation level. There were no changes in any of the abdominal muscle or multifidus activation levels when the sitting APDF was compared with the standing APDF. Individuals who adopted different sitting strategies did not demonstrate any clear differences in the corresponding muscle activation level APDFs. All muscle groups recorded demonstrated a probability greater than zero of achieving a 0% activation level across subjects indicating that the muscles were all shut off during both sitting and standing.

Those subjects who adopted varying sitting strategies (figure 3) utilized up to 50% of the total lumbar spine flexion RoM while those adopting a single 'static' position (figure 2) only utilized $<10\%$ of the total flexion RoM of the lumbar. Standing presented a much narrower range of lumbar spine motion than sitting when averaged across all eight subjects (figure 4). Sitting resulted in lumbar spine postures that varied between ~ 30 and 80% of the lumbar spine RoM taking the zero reference posture as normal upright standing with 100% representing full forward flexion. Whereas standing resulted in a much narrower range of lumbar spine motion, varying by $\sim 10\%$ of the RoM around the normal upright standing posture

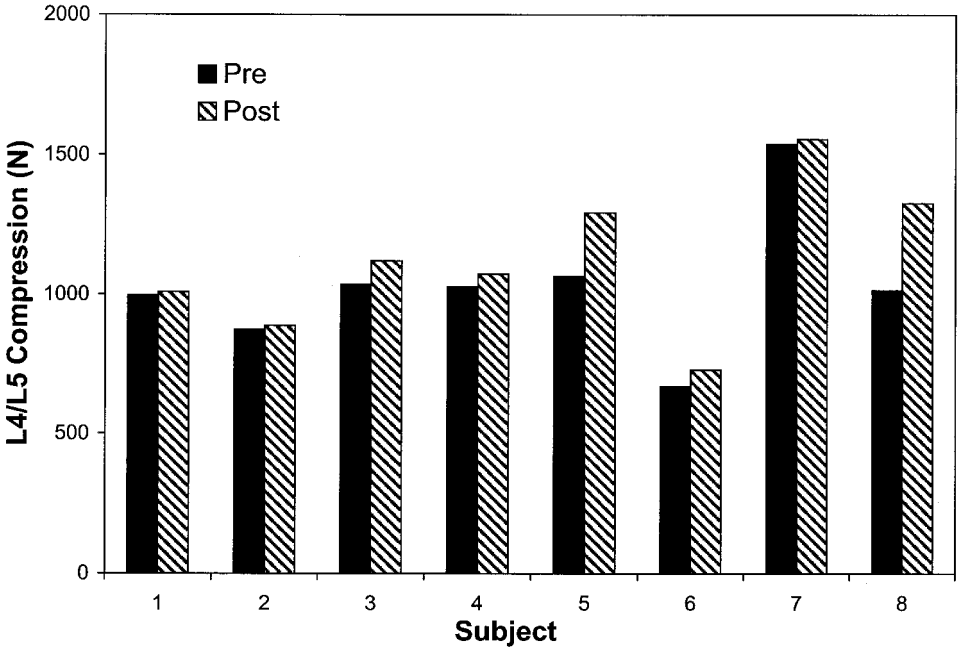

Figure 6. L4/L5 joint compression forces calculated during 3 min of standing, before and after sitting for 2 h.

of the lumbar spine. The APDF function of standing posture demonstrated a shift to postures completely outside the postures assumed during sitting.

4. Discussion

While spine loading during prolonged sitting and standing tasks was quite low, the two tasks present sufficiently different lumbar spine postures to constitute a rest break, at the tissue level, for workers who can alternate sitting with standing. However, static sitting or standing appears to be of concern via fatigue mechanisms given the prolonged loading caused by exposure to these tasks. The human body requires movement both to nourish structures, for example the nucleus pulposus and the intervertebral disc (Holm and Nachemson 1983), and provide periodic rest of muscles to prevent fatigue (Jonsson 1978, Veiersted *et al.* 1990) and occasional migration of loads between various passive tissues achieved from posture change. In sitting, individuals who adopted multiple postures and cycled between them across a wider band of lumbar spinal motion also created motion that appeared to prevent static loads on the spine. However, while passive tissues might be relieved in these individuals, there was little difference in the magnitude of the joint compressive forces present in the multiple positions adopted. Standing resulted in a very different lumbar spine posture to that adopted during sitting, which could provide relief/rest for the passive and active structures of the low back. Addition of such activities as fast walking would provide cyclic muscular activation and spine postures (Callaghan *et al.* 1999). This would suggest that walking could be a beneficial rest activity from prolonged sitting and/or standing.

Table 2. Muscle and ligament forces crossing L4/L5 from the EMG-driven model for subject 1 in standing trial 1 and sitting posture 2 (62% of lumbar spine flexion) and the contribution of the active and passive structures from each fascile to the total joint compression and shear at L4/L5.

Muscle	Force (N)		Compression (N)		Anterior/posterior shear (N)	
	Stand	Sit	Stand	Sit	Stand	Sit
R* rectus abdominis	17	9	17	8	2	2
L* rectus abdominis	22	19	21	18	2	6
R external oblique 1	4	5	4	5	2	1
L external oblique 1	12	8	11	8	4	4
R external oblique 2	8	3	6	2	4	2
L external oblique 2	17	7	13	4	7	5
R internal oblique 1	28	13	25	8	−15	−10
L internal oblique 1	27	16	24	14	−13	−8
R internal oblique 2	60	25	46	9	−11	−13
L internal oblique 2	53	58	41	27	−6	−28
R transverse abdominis	5	23	0	1	−3	−17
L transverse abdominis	10	21	1	1	−5	−16
R pars lumborum 1	12	52	11	51	1	14
L pars lumborum 1	11	67	11	65	1	17
R pars lumborum 2	13	61	13	60	1	16
L pars lumborum 2	12	78	12	76	1	20
R pars lumborum 3	16	70	14	66	−6	−12
L pars lumborum 3	15	88	13	84	−5	−12
R pars lumborum 4	19	79	15	72	−11	−23
L pars lumborum 4	17	102	14	94	−10	−26
R ileiocastalis lumborum	19	56	19	55	−2	11
L iliocastalis lumborum	25	66	24	65	−2	13
R longissimus thoracis	25	72	25	71	−2	14
L longissimus thoracis	30	85	30	84	−3	17
R quadratus lumborum	15	52	13	52	6	2
L quadratus lumborum	13	68	12	68	5	9
R latisimus dorsi 5	14	26	11	23	2	−2
L latisimus dorsi 5	8	23	7	22	1	2
R multifidus 1	8	31	7	29	0	5
L multifidus 1	7	39	6	36	0	7
R multifidus 2	8	31	7	30	2	8
L multifidus 2	6	39	6	38	2	10
R psoas 1	25	11	24	10	4	2
L psoas 1	17	30	16	29	3	5
R psoas 2	25	11	24	10	4	2
L psoas 2	17	30	16	29	3	5
R psoas 3	25	11	24	10	4	2
L psoas 3	17	30	16	29	3	5
R psoas 4	25	11	24	10	4	2
L psoas 4	17	30	16	29	3	5
Muscle subtotal	*724*	*1556*	*639*	*1402*	*−23*	*46*
Ligaments						
Anterior longitudinal	1	0	1	0	0	0
Posterior longitudinal	0	11	0	11	0	2
Ligamentum flavum	3	8	3	8	0	1
R inter transverse	0	1	0	1	0	1
L inter transverse	0	2	0	2	0	0

(continued)

Table 2. continued

Ligaments	Force (N) Stand	Force (N) Sit	Compression (N) Stand	Compression (N) Sit	Anterior/posterior shear (N) Stand	Anterior/posterior shear (N) Sit
R articular	0	1	0	1	0	0
L articular	0	3	0	2	0	2
R articular 2	0	0	0	0	0	0
L articular 2	0	3	0	2	0	0
Interspinous 1	0	2	0	2	0	1
Interspinous 2	0	16	0	10	0	14
Interspinous 3	0	14	0	8	0	12
Supraspinous	0	7	0	7	0	1
R lumbodarsal fascia	0	2	0	2	0	0
L lumbodarsal fascia	0	6	0	5	0	0
Ligament subtotal	*4*	*76*	*4*	*61*	*0*	*33*
Upper body weight (N)			350	350		
Total			993	1813	−23	79

*R, right; L, left.

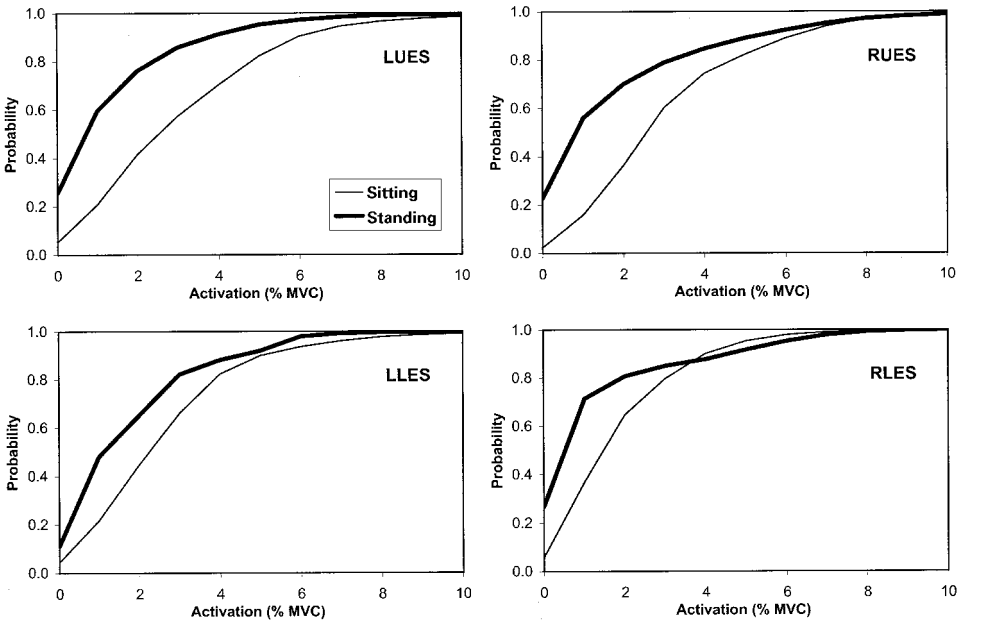

Figure 7. Average muscle activation amplitude probability distribution functions across eight subjects for the 2-h sitting and across the two standing periods. RUES, right upper erector spinae; RLES, right lower erector spinae; LUES, left upper erector spinae; LLES, left lower erector spinae.

The compressive forces on the low back were increased when a sitting posture was assumed. The difference between the standing and sitting joint loading was due to the increased flexed posture of the lumbar spine assumed when seated, which

resulted in increased passive tissue forces which were also due to slight increases in exterior EMG levels. Low levels of EMG were recorded for all muscles in both sitting and standing activities. There seemed to be a trend (non-significant) of decreased flexion RoM after sitting, which would appear to support the finding that the lumbar spine increases in height while exposed to sitting (Althoff *et al.* 1992) (equating increased disc height with reduced joint laxity). However, one potential confounder that could explain the findings of this study (of an increased compressive force during sitting and a decreased RoM following sitting) is the finding that motion increases nutrition to the intervertebral disc (Holm and Nachemson 1983). Subjects used a larger range of spine flexion, which could present less static loading on the spine than standing. During sustained flexion tests, the nucleus has been shown to redistribute its volume posteriorly, resulting in increased stress concentrations in the posterior aspect of the intervertebral disc (Adams and Hutton 1985, Adams *et al.* 1994). This shifting of stress could also be another factor that could alter the flexion RoM test following sitting.

Of the few studies that have examined the variation in spine procedures adopted when seated for extended periods, most have only reported small variations in lumbar or trunk angle during the period examined (Bendix and Biening-Sorensen 1983, Bridger 1988). In contrast, only half of the subjects in the present study adopted a single 'static' posture that remained relatively constant throughout the 2 h of sitting. By examining an extended period of seated work, it was also discovered that 50% of the individuals studied chose to alter their lumbar spine posture frequently. A consideration for chair design/use driven by the examination of individual sitting postures suggests that a chair should allow individuals to vary sitting posture easily, which allows the spine to move, rather than attempting to constrain individuals to an 'ideal' position.

An examination of the muscle activation patterns reported in the literature does not provide any clarification of the difference in spinal joint loading between seated and standing postures. The comparison of muscle recruitment between standing and sitting has revealed little difference in activation profiles (Chaffin and Andersson 1990, Althoff *et al.* 1992). These findings were supported by the current study for the abdominal musculature, latissimus dorsi and multifidus. However, both the upper and lower erector spinae exhibited a decreased amount of time spent at lower levels of contraction during sitting. Levels of activation over 1 or 2% without complete rest have been shown to be problematic (Aaras 1994, Veiersted 1994). While all EMG channels demonstrated a rest component when averaged across the eight subjects in this study, the increased probability of higher levels of activation while seated could be a potential source of the back pain reported by many individuals when exposed to prolonged bouts of sitting (Frymoyer *et al.* 1980, Wilder and Pope 1996). Each of the individual APDFs for the eight subjects and all 14 EMG channels recorded had activation levels that fell below the static levels proposed by Jonsson (1978, 1988b) ($p = 0.1$ with activation levels not exceeding 2 – 5% MVC) to avoid muscle overload.

While electromyographic analysis of the trunk musculature has shown that both lumbar supports and increased seat back inclination angle reduce activation levels (Knutsson *et al.* 1966, Andersson *et al.* 1979, Chaffin and Andersson 1990), the current study only examined sitting with no back support. This was primarily due to instrumentation requirements but also provided insight into the unconstrained lumbar spine kinematics present during sitting. Additionally, when people work at a desk, they often lean forward and do not rest against the back support of a chair.

Description of spine posture change using an APDF provides insight into the total time spent in a posture together with an illustration of the range of postures adopted, but the APDF does not indicate the number or distribution of posture changes. Future work must develop thresholds of the frequency and magnitude of postural shifts that are beneficial to the involved tissues. These thresholds will then allow the number and magnitude postural changes to be assessed during prolonged sitting. The model used to predict both passive tissue and muscle forces (McGill and Norman 1986, McGill 1992) has been used to assess a wide variety of occupational and athletic tasks for years. Individual tissue loads cannot be individually validated. However, the anatomical detail, incorporation of known passive stress–strain data, known modulators of muscle force predicted from EMG and physiological cross-sectional area, provide some content validity. At least the 3D moments about the low back can be directly measured and compared with the predicted moments. For sagittal plane tasks, like those evaluated in this study, the moments typically match well.

5. Conclusions

The forces experienced by the joints of the lumbar spine during sitting and standing activities were found to fall well below any traditional single exposure tissue tolerance value. However, prolonged static loads could still present a fatigue injury mechanism either due to low but prolonged muscle contraction and/or prolonged flexed postures of the spine leading to accumulated damage to the posterior component of the annulus. Standing appears to be a good rest from sitting, for the passive tissues, given the change in lumbar spine posture and shift in loading of the posture dependent passive tissues. Therefore, standing, used alternatively as a rest from sitting, could form a basis for injury prevention when designing work. However, the constant loading with little dynamic movement present in both standing and sitting would provide little rest/change for muscular activation levels and the resultant low back loads.

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