Influence of Prolonged Sitting and Psychosocial Stress on Lumbar Spine Kinematics, Kinetics, Discomfort, and Muscle Fatigue

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ABSTRACT

Low back pain (LBP) is a common occupational problem and continues to be the leading cause of occupational disability. Among diverse known risk factors, sitting is commonly considered as an important exposure related to LBP. Both modern living and contemporary work involve increased sedentary lifestyles, including more frequent and prolonged sitting. At present, however, the causal role of sitting on LBP development is controversial due to the contribution of several moderating factors (e.g., task demands, duration of exposures, and presence of muscle fatigue). A few studies have assessed low back loads in seated postures, but none has investigated the effects of prolonged sitting or time-dependent variations on spinal structure and spinal loading. Adverse effects of muscle fatigue on low back pain are well documented, yet the specific relationship between muscle fatigue and sitting-related low back pain are not fully established. In addition to these fundamental limitations in our understanding of the physical consequences of sitting, there is also little evidence regarding the effects of task requirements on muscle fatigue and spine loading.

Therefore, the main objectives of this work were, in the context of sitting, to: 1) develop and evaluate a method to assess paraspinal muscle fatigue using electrical stimulation; 2) develop and evaluate a method (model) to quantify biomechanical loads on the lumbar spine in a seated posture; and 3) quantify the effects of prolonged seated tasks on low back loads, body discomfort, and localized muscle fatigue (LMF). The primary hypothesis was that exposure to sitting-related LBP risks is influenced by task requirements and sitting duration.

A muscle stimulation protocol was developed to measure stimulation responses in the lumbar extensors. A stimulation protocol, which included one conditioning train along with three 16-second stimulation train at 2 Hz, was recommended as appropriate to measure those muscles potentially fatigued during prolonged seated tasks. A three-dimensional, sitting-specific, fatigue-sensitive, time-dependent, electromyography (EMG)-based

biomechanical model of the trunk was developed to investigate the effects of seated tasks and time-dependent variations on lumbosacral loading during sitting. Reasonable levels of correspondence were found between measured and predicted lumbosacral moments under a range of seated tasks. Lastly, the effects of prolonged sitting and psychosocial work stress on low back were quantitatively identified. Only prolonged sitting significantly increased trunk flexion angles and led to muscle fatigue. Relatively weak correlations were found between subjective and objective measures, though the two fatigue measurement methods (based on EMG and stimulated responses) showed a good level of correspondence.

Overall, this work provides a quantitative assessment of biomechanical exposures associated with seated tasks. The methods developed in this work make a contribution in terms of measurement/modeling approaches that can be used to assess LBP-relevant risks during prolonged sitting. The results of this work provide a better understanding of the effects of prolonged sitting on the risk of developing sitting-related LBP. Finally, results regarding the influences of prolonged sitting and psychosocial demands can be used to guide future job design.

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Chapter 1. Introduction

Low back pain (LBP) is one of the most common societal health problems, causing considerable disability, work absenteeism, and use of health services. LBP remains the leading cause of disability among those under 45 year of age (Guo et al., 1999; BLS, 2009). More than one-quarter of the working population is affected by LBP each year, with a total of 222,290 cases reported in 2008 (BLS, 2008a). Annually, 40.2% of those with LBP have persistent symptoms, 14.2% experience an aggravation of their symptoms, and 28.7% have a recurrence within 6 months (Cassidy et al., 2005). Furthermore, LBP comprises about 40% of all compensation claims in the United States (Lis et al., 2007) and 33% of total medical costs (Pope et al., 2002).

Among the identified occupational risk factors for LBP, sitting is commonly considered among these (Magora, 1972; Kelsey and White, 1980; Andersson, 1981; Wilder et al., 1988; Chaffin et al., 1999; Vingard et al., 2000; Pope et al., 2002). Sitting is a complex behavior that is affected by many moderating factors, such as sitting duration (Keyserling, 1991; Pitman and Ntuen, 1996; Stricker et al., 2003) or psychological task demands (van Dieën et al., 2001; McLean and Urquhart, 2002; Ellegast et al., 2007). Early work suggested that both too much and too little sitting are related to an increased incidence of LBP, and which can be described as a "U"-shaped relationship (Magora, 1972). Although studies have been conducted to quantify the relationship between these moderating factors and risks of sitting-related LBP, the specific mechanism(s) that underlie sitting-related LBP have not been well studied (Hartvigsen et al., 2000; Leboeuf-Yde, 2004; Womersley and May, 2006; Lis et al., 2007), and this lack has contributed to some controversy regarding the causal role of sitting on LBP (Cholewicki and McGill, 1996; Lis et al., 2007). Existing inconsistencies are likely due, at least in part, to the wide range of evaluation criteria used and the potential insensitivity of associated methods employed.

The effects of prolonged sitting have been measured by monitoring participants' muscle activities, postures, perceived workloads and discomfort (van Dieën et al., 2001; Gregory et al., 2006; Todd et al., 2007; Kingma and van Dieën, 2009), but spine kinetics during prolonged sitting have not been well quantified (Callaghan and McGill, 2001). In addition, an accurate assessment of spinal loading partitioning among passive and active components of the trunk during prolonged sitting, such as in biomechanical models, requires a realistic representation of time-dependent viscoelastic passive properties. The viscoelastic properties of the spine have been extensively studied, by directly or indirectly measuring the time-dependent creep and load-relaxation responses of the spine to applied loading (Burns et al., 1984; Holmes and Hukins, 1996; Kurutz, 2006). However, viscoelastic responses to prolonged sitting position have not been sufficiently described and it remains unclear how spine loading may change as a result. Therefore, an effective method is needed to assess the influence of prolonged sitting on spine kinetics.

Though low levels of muscle activity [<10% maximum voluntary contraction (MVC)] are required, postural muscles, such as the lumbar erector spinae, are continuously activated to stabilize the sitting posture (van Dieën et al., 2001). These continuous levels of muscles activation can cause muscle fatigue in the low back area, and the accompanying aches and cramping may further lead to low back pain (Kolich et al., 2001). However, measures of muscle fatigue using EMG-based methods have shown some conflicting results. Öberg et al. (1992; 1994) indicated that a muscle contraction at 15% MVC or higher is required to detect fatigue-related changes in EMG signal during different working levels. McLean and Goudy (2004) also indicated that during low level contractions, evidence of fatigue is inconsistent between different muscle groups and across individuals. However, Søgaard et al. (2003) and van Dieën et al. (2009) demonstrated the feasibility of using EMG to detect muscle fatigue during low-level contractions (as low as 2% MVC), though their results were derived based on 30-minute measurements and still showed a large inter-subject variation. Therefore, work is needed to test the sensitivity of current EMG-based methods and to find an efficient way (e.g., muscle stimulation) to detect muscle fatigue during prolonged low-level exposures (e.g., during sitting).

Many tasks, such as computer-based typing that requires prolonged sitting, also involve precision work, a high mental workload, a stressful working schedule, etc. The high prevalence of musculoskeletal disorders in psychologically stressful but light physical work has indicated that increased perceived workload plays an important role in the development of LBP (Jensen et al., 1998). Psychosocial task demands (e.g., time pressure and high mental workload) have been associated with the development of work-related musculoskeletal disorders in the forearms (Jensen et al., 1998; Blangsted et al., 2004; Hughes et al., 2007). Both physical and psychosocial task demands can substantially influence trunk kinematics and increase muscle activity (Pitman and Ntuen, 1996; van Dieën et al., 2001; Ellegast et al., 2007). Work duration, on the other hand, also has the potential to affect physical and mental task performance (Pitman and Ntuen, 1996; Liao and Drury, 2000). Psychosocial task demands seem to serve a mediator role with physical risk factors (such as prolonged duration) and potentially increase the risk for LBP development (Hughes et al., 2007). However, very few experimental studies have examined the contribution of such factors to LBP in a seated posture.

The main objective of this work was to develop sensitive methods to detect muscle fatigue and to quantify biomechanical loads in the spine during prolonged sitting. Additionally, the effects of different psychosocial task demands on spinal loading and muscle fatigue were also assessed. The primary hypothesis was that exposure to sitting-related LBP risks is influenced by psychological task demands and sitting duration. In this work, LBP-relevant risks were assessed using diverse approaches, including muscle activity and fatigue, trunk kinematics, body discomfort, and spine kinetics (lumbosacral forces and moments).

The material presented in this dissertation is organized in five chapters. In chapter 1, an introduction to the work was given, followed by a description of the context of the study, and its rationale. In chapter 2, a stimulation method is developed and evaluated with a goal of measuring muscle contractile status during seated postures. In chapter 3, the development and evaluation of an EMG-driven biomechanical model for internal loads estimation is presented. In chapter 4, an experimental study is described, to assess the effects of prolonged sitting and psychosocial stress on low back kinematics, muscle activities, body discomfort, and muscle fatigue. In chapter 5, research outcomes are summarized, highlighting the major results obtained and important limitations of the current work. A review of potential applications is also presented and areas described that should be considered for further research in this field.

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Chapter 2. Reliability of a stimulation method to assess the contractile status of the lumbar extensors in a seated posture

Abstract

The purpose of the present study was to develop and evaluate methods to assess stimulation responses of the lumbar extensors, as part of a longer-term goal of detecting fatigue during prolonged sitting. Three stimulation frequencies (2, 5, and 8 Hz) were tested in separate stages, which include three stimulation trains and four sampling blocks. Repeated Measures Analyses of Variance was used to determine if any significant differences in mean stimulation responses occurred with respect to stimulation frequency, sampling block, and stimulation train. Reliability of measured stimulation responses was assessed within and between sampling blocks using intraclass correlation coefficients (ICC). Stimulation frequencies significantly affected the stimulation responses and time-to-potentiation differed between the three stimulation frequencies, being highest for 2 Hz stimulation. All three stimulation frequencies resulted in excellent reliability within and between sampling blocks. Current protocol at 2 Hz is recommended as appropriate to measure the lumbar extensors status during prolonged sitting.

Keywords: Muscle stimulation; stimulation frequency; lumbar Extensors.

2.1. Introduction

Modern living increases the tendency to have a more sedentary lifestyle that involves prolonged sitting (Ford, et al., 2005; Varo, et al., 2003). In the US, for example, an earlier report indicated that more than 60% of adults are rarely physically active and 20% of all adults are completely sedentary (US Department of Health and Human Services, 1996). Current estimates are consistent in terms of leisure-time physical activity in the US (Pleis et al. 2009), and more generally that less than one-third of adults worldwide are physically active (Hallal et al., 2012). Recent evidence indicates that prolonged sitting postures compromises blood flow, increases intradiscal loads, and reduces muscle oxygenation (McLean and Urquhart, 2002; Vollestad, 1997). Such physiological changes can cause lumbar extensor fatigue that in turn may contribute to low back pain (Dankaerts, et al., 2006; Kolich, et al., 2001; NIOSH, 1997; O'Sullivan, et al., 2006; Womersley and May, 2006). Postural muscles are continuously activated to stabilize the sitting posture, though relatively low levels of muscle activity (< ~10% of maximal muscle capacity) are involved (van Dieën et al. 2001; Mork and Westgaard 2005; Mork and Westgaard 2009). Further, both objective and subjective signs of fatigue have been observed following 20 minutes or more of seated exposures (Blangsted, et al., 2005; Hakala, et al., 2006; McLean and Goudy, 2004; Sjøgaard, et al., 1986, 2003; van Dieën, et al., 2009).

Objective signs of muscle fatigue during sitting have typically been obtained from electromyographic (EMG) measures, though the use of EMG in this context has associated challenges and has provided some conflicting results. As noted, relatively low contraction levels are involved, and these can be lower than that needed for EMG to detect fatigue-related changes (Gamet, et al., 1993; Nagata, et al., 1990; Oberg, et al., 1994; Oberg, et al., 1992; Sood, et al., 2007). McLean and Goudy (2004) also indicated that during low contractions evidence of fatigue is inconsistent between different muscle groups and across individuals. While Søgaard et al. (2003) and van Dieën et al. (2009) demonstrated the feasibility of using EMG to detect muscle fatigue during low-level contractions (as low as 2% MVC), their results showed large inter-subject variability.

A common alternative means to identify muscle fatigue is through use of external stimulation to obtain stimulation responses. In this approach, fatigue is indicated when stimulation responses fall significantly below initial, pre-fatigue levels (Binder-Macleod and Snyder-Mackler, 1993; Edwards, et al., 1977; Johnson, 1998; Johnson, et al., 1995a). Muscle fatigue has been measured by comparing the decrement of response ratio between results to high frequency (50-100 Hz) stimulation and low frequency (1-20 Hz) stimulation (Byström and Kilbom, 1991; Davies and White, 1982; Edwards, et al., 1977). Others have measured muscle fatigue using a single stimulation frequency (Cooper, et al., 1988; Johnson, et al., 1995a). While a variety of stimulation protocols have been used, a greater loss in stimulation response can occur when evoked by low frequency stimulation (Cooper, et al., 1988; Fitch and McComas, 1985). Low frequency stimulation can also be advantageous since it is less likely to cause muscle fatigue or discomfort (Johnson, et al., 1995a). Low frequency muscle stimulation has been shown to be an effective method to assess muscle fatigue (Johnson, et al., 1995a) and has been successfully applied to the forearm muscles (Johnson, et al., 1995b; 1996; Mork and Westgaard, 2005). Yet, it remains unclear whether this approach would be similarly successful for assessing fatigue in the trunk musculature.

The current work is part of broader effort to use muscle stimulation (low frequency, specifically) to detect lumbar extensors fatigue during prolonged sitting. In this context, the present study was conducted to develop a reliable muscle stimulation protocol to measure stimulation responses in the lumbar extensors, specifically those muscles potentially fatigued during prolonged seated tasks. Several stimulation parameters and conditions can substantially affect stimulation response, thereby influencing the reliability of stimulation response measurements. During prolonged stimulation, stimulation responses undergo a gradual potentiation (Desmedt and Hainaut, 1968), which, if not accounted for, can mask or confound other effects. For example, an increase in stimulation response amplitude due to potentiation and a decrease due to fatigue can occur simultaneously (Rassier and MacIntosh, 2000) and has been observed during low-frequency fatigue caused by sustained low-level loading (Fowles and Green, 2003). Potentiated stimulation responses may also be a more sensitive index of contractile fatigue than un-potentiated stimulation responses (Kufel, et al., 2002). The relationship between stimulation response generation and excitation is also dependent on stimulation frequency, which can also influence the pattern of potentiation (Rassier and MacIntosh, 2000). In the current study, we evaluated the effects of several stimulation frequencies in terms of potentiation and the reliability of evoked stimulation responses. Results from this exploratory work were intended to facilitate the selection of stimulation protocols in future studies.

2.2. Methods

2.2.1. Overview

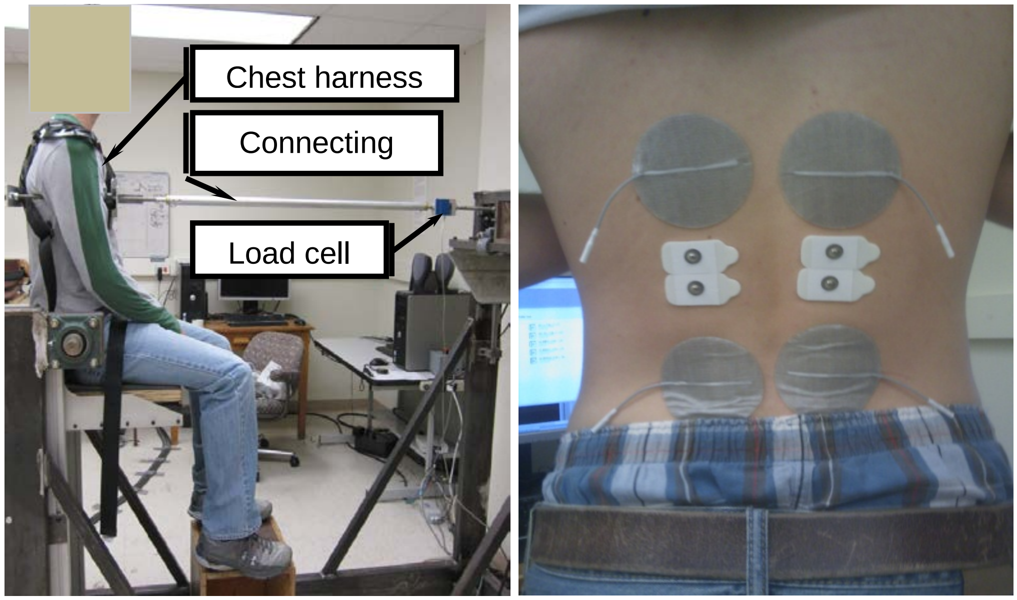
A stimulation protocol to assess lumbar extensor stimulation responses was developed. Effective stimulation sites and stimulus intensity were identified for each participant since these may vary between individuals due to anatomical differences. Successive muscle stimulation can lead to a gradual increase in measured muscle responses, a phenomenon known as muscle potentiation (Desmedt and Hainaut, 1968; Small and Stokes, 1992). Since potentiation could lead to decreased reliability in stimulation responses, and/or mask fatigue-related effects, it was considered important to control for these. To address this, a conditioning train was completed to potentiate target muscles to a stable status, and was done using each of three stimulation frequencies. Stimulation responses of the lumbar extensors, represented by the forces measured from resulting (evoked) trunk extension, were measured in response to each stimulation frequency using the same stimulation protocol and via a load cell and fixture. Responses to the three stimulation frequencies were compared to help identify an appropriate stimulation protocol.

2.2.2. Participants

Six participants (gender balanced) were recruited and completed an informed consent process approved by the Virginia Tech Institutional Review Board. Their mean (SD) age, height, body mass, and body mass index (BMI) were 28.2 (4.8) yrs, 170.3 (8.7) cm, 67.6 (13.2) kg, and 23.2 (3.2), respectively. Individuals with body mass index >30 were excluded due to the potential difficulty in evoking reliable muscle contractions. All participants were physically active and had no self-reported musculoskeletal or neurological diseases within the past year that restricted their daily activities.

2.2.3. Instrumentation

The lumbar extensors were stimulated bilaterally using a dual-channel, current-controlled muscle stimulator (Grass S88, AstroMed, Inc., West Warwick, RI) in series with a stimulus isolation unit (SIU5, AstroMed, Inc., West Warwick, RI) and a constant current unit (CCU1, AstroMed, Inc., West Warwick, RI). Two pairs of 7 cm diameter bipolar stimulating electrodes (PALS® Platinum Model 879300, Axelgaard Manufacturing Co.Lltd., Fallbrook, CA) were placed bilaterally over the lumbar extensors (see below, regarding stimulation site). During the experiment, participants sat in an experimental fixture (Figure 2-1A), in which motion of the pelvis was restrained using straps. An adjustable footrest was used to position the lower limb with right angles at the knee and ankle. Participants were positioned in a comfortable and relaxed upright sitting posture using a rigid rod connected to a chest harness at the T8 level, and which contained an in-line load cell (Interface SM2000, capacity = 2000 N, Scottsdale, AZ, USA). Stimulation responses were sampled at 1000Hz from the load cell. To minimize voluntary muscle activity during stimulation response measurements, the erector spinae were monitored using two pairs of surface EMG electrodes that were placed bilaterally at the L3 level (Figure 2-1 B). EMG signals were monitored visually throughout the experiments in real time. Muscle stimulation was started only after the observed level of muscle activation was (qualitatively) stabilized, specifically as reflected in no substantial increase/decrease in EMG levels.



(A) (B)

Figure 2-1. Illustration of (A) experimental fixture setup demonstrating a participant in an upright sitting posture, and (B) the locations of dual-channel muscle stimulation electrodes (grey) and bipolar EMG electrodes (white).

2.2.4. Experimental Procedures

Participants' skin in the dorsal lumbar region was prepared following procedures as described by Cram and Rommen (1989). As needed, the skin surface where the electrodes would be placed was first shaved, and then skin was cleaned with alcohol and dried. Based on earlier recommendations on initial electrodes locations and probing procedures (Baker, et al., 1993; Vanoncini, et al., 2006; Johnson et al. 1995a), subsequent preparatory procedures were completed to identify the most effective stimulation site for each individual. Initially, negative electrodes were placed at the level of the iliac crest and positive electrodes were placed at the level of the inferior margin of the rib cage (see Figure 1.1.B).

Stimulus intensity was then determined by increasing the stimulation current from 10 mA at each stimulation frequency (see below), until discomfort was reported or up to a max of 40 mA, and it was subsequently reduced by 15%. Discomfort/pain is related to the stimulation level, which is controlled by the pulse amplitude (current) and pulse duration. Measured stimulation responses are often a nonlinear function of either pulse duration or stimulation amplitude (Crago et al., 1980). During stimulation, it is important to minimize charge transfer to prevent damage to muscle tissue. At a given force level, a high stimulation amplitude requires less charge transfer per stimulation pulse, and charge transfer increases with increasing stimulation duration (Crago et al., 1974; Crago et al., 1980). As a compromise, we used a pulse duration of 0.3 msec. Current-controlled stimulation (at 150v) was also chosen, as it was easier to control and may induce less discomfort compared to voltage-controlled stimulation (Merletti, et al., 1992). After identifying the stimulus intensity, the test muscles were "warmed up" using a 4-minute stimulation block at the stimulation frequency. Finally, effective electrode sites were then determined by stimulating the muscle at each stimulation frequency and probing (i.e., adjusting the positive electrode locations) along superior-inferior direction until maximal evoked stimulation responses were measured using the load cell.

Each participant then completed three data collection stages (Figure 2-2 A), where each stage involved one of the three stimulation frequencies (2, 5, and 8 Hz), and a different presentation order was pre-defined for each participant using a Latin Square and a 10-min break was given after each stage. The same stimulation frequency and parameters were applied in the conditioning trains and in the subsequent measurement trains. Within each stage, stimulation responses were measured in four sampling blocks (Figure 2-2 B) with a 2-min rest in between. Within each sampling block, a conditioning train was applied (Figure 2-2 C) to first potentiate the muscles into a steady state (Desmedt and Hainaut, 1968; Edwards, et al., 1977; Kufel, et al., 2002; Rassier and MacIntosh, 2000). Each conditioning train was terminated when stimulation responses plateaued, determined in real-time by using a one-second moving window mean of peak responses measured. This Time-to-Potentiation (t_{pot}) was recorded along with the potentiated stimulation responses (F_{pot}). Immediately after the conditioning train, muscle stimulation responses were measured during three 16-second stimulation trains (Figure 2-2 C). Short breaks were provided between each train. To minimize potential effects of repositioning, however, participants were not allowed to leave the experimental fixture. After the second sampling block, participants were removed and repositioned in the experimental fixture. This was done to determine whether there were significant differences in stimulation responses.

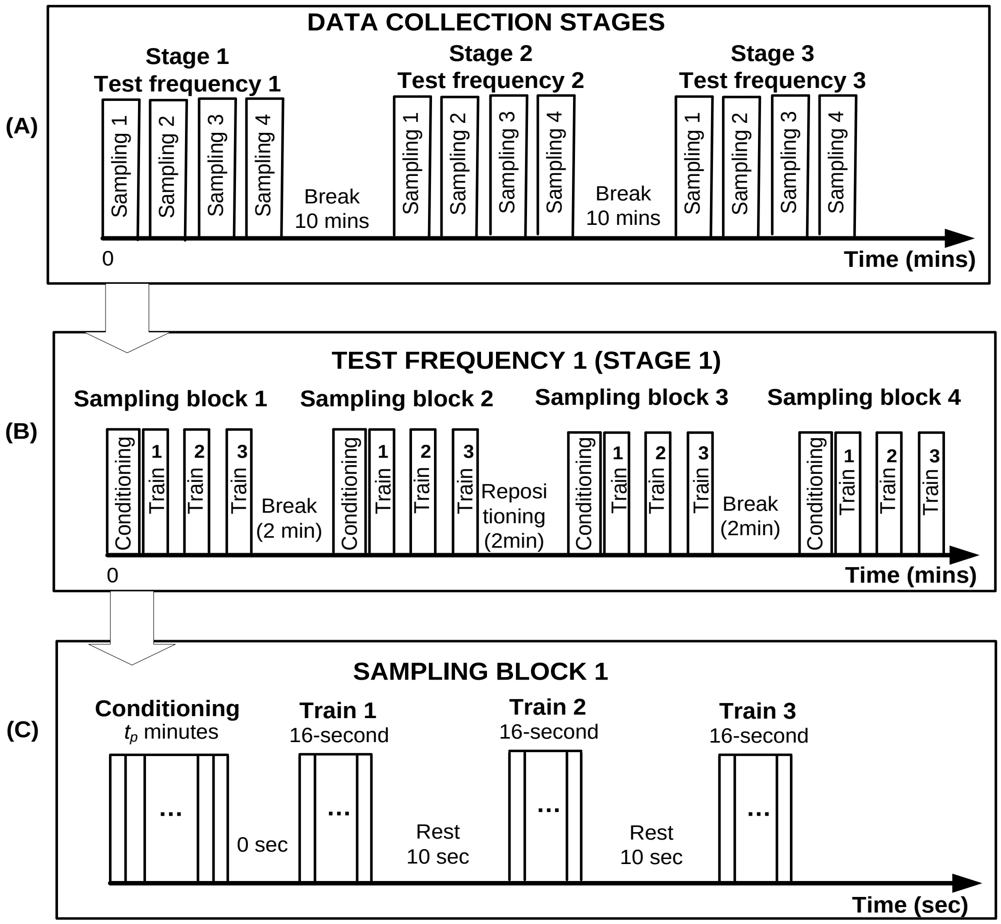


Figure 2-2. Schematic indication of stimulation procedures. (A) Three stimulation frequencies in separate test stages. (B) Muscle twitch protocol during one data collection stage. Conditioning: continuous stimulation at a given stimulation frequency to reach a plateau in twitch force. Train: stimulation at each stimulation frequency with rest in between. (C) Stimulation during one sampling block. In addition, participants were repositioned between 2nd and 3rd sampling blocks.

2.2.5. Data Processing and Analysis

Stimulation responses were band-pass filtered (1.5 - 10 Hz) to remove environmental and breathing artifacts (Beers, 2003). Peak responses from each stimulation pulse, within each train, were determined using Matlab software (MathWorks^{TM}, Inc, 2010a). Mean stimulation responses were the dependent measure in the Repeated Measures Analyses of Variance (RANOVAs). These were obtained as the means, across all peak stimulation responses recorded during the 16s block of each train, excluding the three largest and three smallest values. Such a truncated (trimmed) mean was used as an estimate of central tendency, as it is less sensitive to outliers (Rothenberg et al., 1966; Wilcox and Keselman, 2003).

RANOVAs were used to identify any significant differences in mean stimulation responses with respect to stimulation frequency, sampling block, and stimulation train. Where relevant, significant effects were further investigated using Tukey's honestly significant difference (HSD) post-hoc analyses. RANOVA was also used to identify any significant differences in t_{pot} related to stimulation frequency and sampling block. In all analyses, a p-value of <0.05 was considered statistically significant.

The reliability of stimulation responses was evaluated both within and between sampling blocks for each of the three stimulation frequencies. To allow for comparisons of twitch responses within and between participants, stimulation responses measured within each conditioning train and twitch train were normalized to stimulation responses collected after potentiation within the first sampling block. This approach was used, since F_{pot} was expected to be the most stable response measured. Reliability was evaluated using intraclass correlation coefficients (ICC), and was determined separately for each of the three stimulation frequencies. Within sampling blocks (Within-SP), reliability was determined by calculating ICC using data obtained from the three stimulation trains. Reliability between sampling blocks (*Between-SP*) was evaluated by calculating ICC from mean stimulation responses in each train and across all four sampling blocks. ICC was also calculated using mean stimulation responses across all three trains to evaluate reliability between sampling blocks (*Between-SP Mean*). Finally, ICC was used to assess the reliability of stimulation responses before and after repositioning the participant in the experimental fixture. ICCs were qualitatively interpreted (Cicchetti and Sparrow (1981) as poor (0.00–0.39), fair (0.40–0.59), good (0.60–0.74), or excellent (0.75–1.00).

2.3. Results

Stimulation responses were successfully evoked using all three stimulation frequencies (Figure 2-3). In general, trunk movements and stimulation responses in response to stimulation at 2 Hz were more noticeable visually, compared to the other two frequencies. Stimulation at 2 Hz resulted in the largest stimulation responses [mean (SD) = 30.8 (8.8) N], whereas stimulation at 5 Hz [mean (SD) = 19.6 (3.3) N] and 8 Hz [mean (SD) = 19.7 (4.5) N] yielded comparable but smaller responses. There was a significant effect of stimulation frequency ($F_{(2,175)}$ = 134.9; p<0.01) on stimulation responses (Figure 2-4). No significant main effects of stimulation train ($F_{(2,175)}$=0.4; p=0.65) or sampling block ($F_{(3,175)}$=0.9; p=0.34) were evident, or any interaction effects of stimulation frequency, stimulation train, or sampling block (p>0.91).

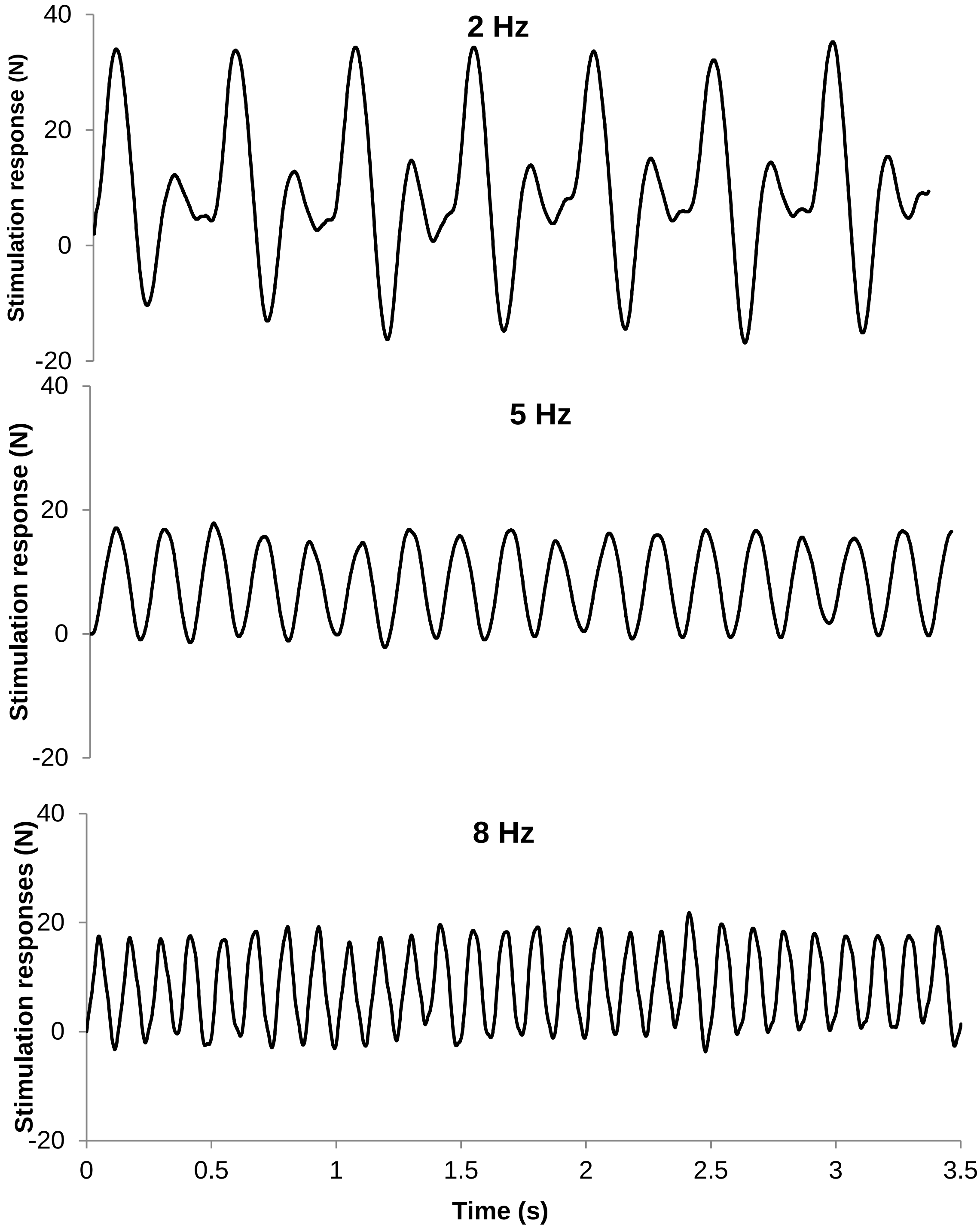


Figure 2- 3. Sample stimulation responses measured in response to three different stimulation frequencies. Positive (+) values represent forces recorded from the load cell during trunk extension caused by muscle stimulation. The secondary positive and negative (smaller) peaks, in response to 2Hz stimulation, were probably due to reflex of trunk flexors and extensors, similar to a damped pendulum response.

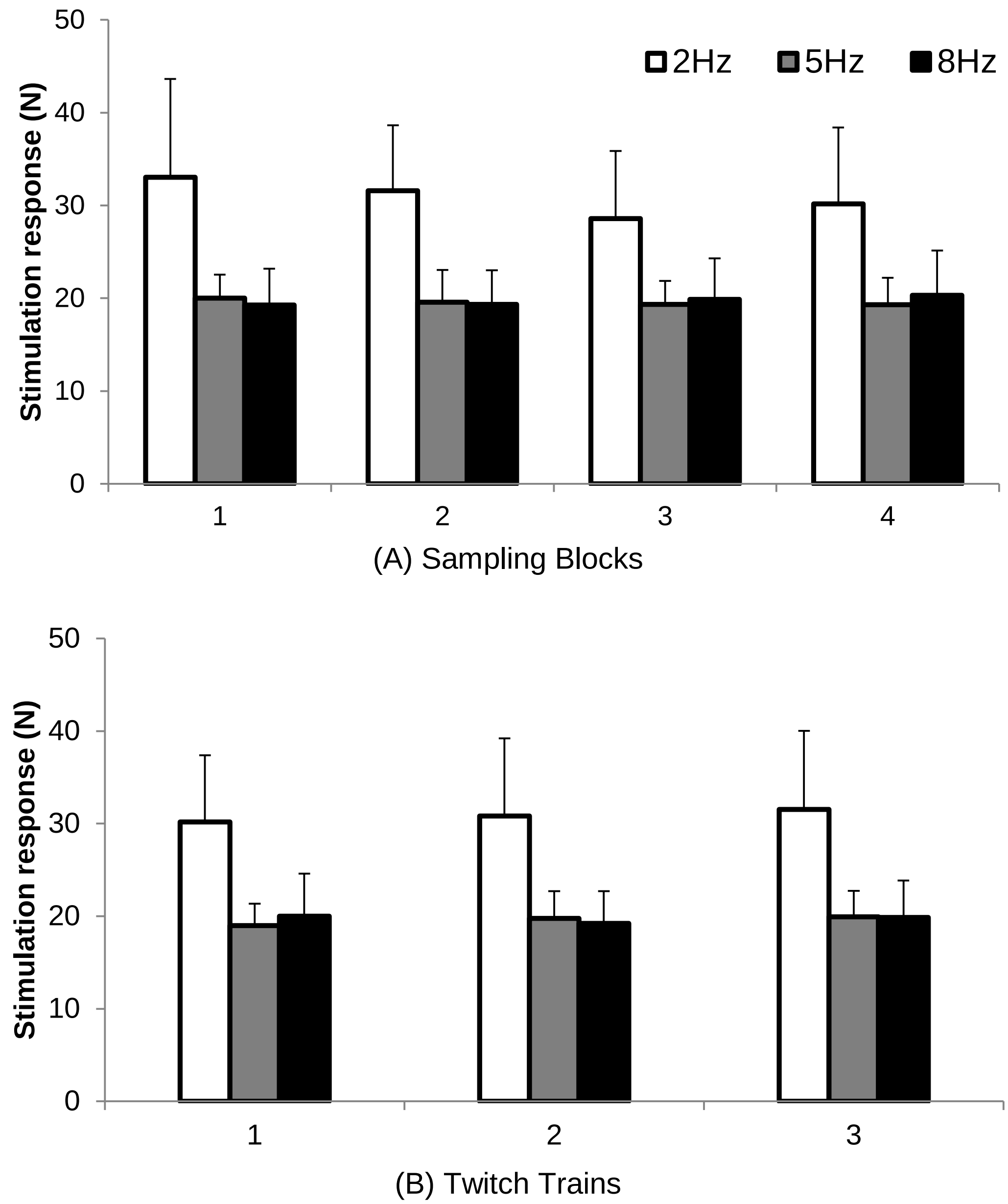


Figure 2- 4. Mean stimulation responses (error bars indicated SDs) to the three stimulation frequencies during four sampling blocks (A) and three trains (B). Stimulation responses (forces) measured at 2 Hz were significantly (p<0.05) larger than the other two stimulation frequencies in all cases.

Stimulation frequency also significantly ($F_{(2,60)}$=17.3; p<0.01) affected t_{pot}, which was longest with 2 Hz stimulation (mean = 261 sec), compared to 5 or 8 Hz (mean = 85 sec). Differences in t_{pot} between the four sampling blocks were not significant ($F_{(3,60)}$=1.2; p=0.32), and there was not a significant effect of the frequency x sampling block interaction ($F_{(6,60)}$ = 0.9, p = 0.45). Reliability of stimulation responses within each sampling block (Within-SP) was in the "excellent" range (>0.85) for all combinations of stimulation frequency and sampling block (Table 2-1). Qualitatively, reliability within sampling blocks was highest for stimulation at 2 Hz and consistent across the four sampling blocks. Excellent reliability was also evident between sampling blocks, with the highest levels found for stimulation at 8 Hz and the lowest at 2 Hz. This same pattern was evident for the reliability of mean stimulation responses between sampling blocks. Repositioning participants did not substantially influence reliability (Table 2- 2). Specifically, ICCs in each of the three trains were excellent, with one exception (2 Hz, train 3). These values were generally comparable with overall reliability between sampling blocks (Table 2-1), both in terms of magnitude and the influence of stimulation frequency.

Table 2-1. ICC values for three stimulation frequencies, determined within sampling blocks (Within-SB, for four sampling blocks), between sampling blocks (Between-SB, for three trains), and for mean twitch forces between four sampling blocks (Between-SB Mean).

	Within-SB ICC				Between-SB ICC			Between-SB Mean
	SB 1	SB 2	SB 3	SB 4	Train 1	Train 2	Train 3	
2 Hz	0.96	0.93	0.91	0.96	0.84	0.83	0.81	0.84
5 Hz	0.89	0.94	0.85	0.89	0.88	0.91	0.93	0.91
8 Hz	0.91	0.89	0.94	0.84	0.96	0.95	0.89	0.96

Table 2- 2 ICC values before and after repositioning a participant in the experimental fixture.

	Repositioning ICC		
	Train 1	Train 2	Train 3
2 Hz	0.87	0.84	0.76
5 Hz	0.94	0.82	0.94
8 Hz	0.97	0.88	0.97

2.4. Discussion

The objective of this study was to develop a method to measure stimulation responses of the lumbar trunk extensors in the context of seated postures. Muscle stimulation responses were successfully generated across all participants in response to all three stimulation frequencies. Stimulation responses collected at 2 Hz in particular resulted in a mean response of 32 N, comparable to a value of 35 N in an earlier study (Vanoncini, et al., 2006). Although the test condition was different than what was used here, the results from Vanoncini (2006) are considered to at least provide relevant information regarding the magnitude of stimulation responses in a seated posture. All three frequencies of stimulation produced measureable responses to muscle contractions, and no substantial pain or skin irritation was either noticed or reported. Rather, participants indicated only minor discomfort caused by the stimulation, and some minor skin reddening was evident under the stimulation electrodes, the latter considered a normal reaction to the long stimulation sessions involved (Baker, et al., 1993).

As an important aspect of stimulation, muscle potentiation was addressed. Potentiation is considered to be a result of phosphorylation of the myosin light chains, and depends on the fiber type (Hanson, 1974; Houston, et al., 1985). Specifically, Type II fibers are selectively and preferentially activated before Type I fibers by stimulation (Hanson, 1974; Sinacore, et al., 1990). The lumbar extensors (e.g., the erector spinae and multifidus) have a high proportion of Type I fibers. However, the percentage of Type II fibers can reach 35-50% (Mannion, et al., 1997; Thorstensson and Carlson, 1987), which emphasizes the importance of potentiating the muscle first while using stimulation to measure stimulation responses from these muscles. In the current study, twitch potentiation was observed in all participants within the first 85-261 seconds of the conditioning trails. Compared to the 2 Hz stimulation frequency, less time was required to reach a steady state with 5 and 8 Hz frequencies. This inverse relationship between time-to-potentiation and stimulation frequency is consistent with existing studies of the knee extensors (Binder-Macleod, et al., 2002; Eom, et al., 2002) and ulnar nerve (Griffin and Mettler, 2010). Eom et al., (2002) also indicated in their studies that the degree of force enhancement during potentiation decreased with the stimulation frequency, which was also observed in the current study. During potentiation, substantial enhancements of stimulation responses were observed using all three test frequencies; approximate increases were 50, 30, and 30% in response to 2, 5, and 8 Hz stimulation, respectively. This again indicates the need for potentiation, since measurements of stimulation responses without potentiation may introduce large variability to measured stimulation responses (Eom, et al., 2002; Johnson, et al., 1995a). As found here, muscle stimulation can lead to variable (increasing) magnitudes of stimulation responses (30%~50%) before reaching a steady state. No consistent pattern of potentiation in stimulated responses was found in the 2nd or 3rd trains within a sampling block. In a few cases, though, a fast (within first 2-3 stimulation pulses) and slight increase (≤ 5%) in stimulation responses was observed at the start of these trains. Any effects of such increases were minimized, however, during calculation of mean stimulation responses using truncated means. Since mean stimulation responses measured between several sampling blocks (Figure 2- 4) were consistent, the conditioning train appeared successful at achieving a steady state in terms of stimulation responses and it likely did not introduce any muscle fatigue.

The reliability of evoked stimulation responses was assessed in several ways. In general, all three stimulation frequencies exhibited excellent reliability in terms of stimulation responses within and between sampling blocks. Among the three frequencies, within-sampling block reliability for 2 Hz was somewhat higher than the other two, indicating that potentiated twitches at 2 Hz were more stable in the three stimulation trains. However, in comparison to 5 and 8 Hz stimulation, relatively larger trunk movements were observed in response to the 2 Hz stimulation. Since the seated posture was not precisely controlled between different sampling blocks, the sitting position at each sampling block may have varied. Different sitting positions may result in different trunk movement pattern and such differences may have contributed the somewhat lower reliability between different sampling blocks. Since 2 Hz stimulation provide more stable and larger stimulation responses, it is reasonable to expect more successful measurements in practical applications (e.g., to assess muscle fatigue). Stimulation at 5 Hz and 8 Hz, in contrast, though it resulted in higher ICC values between sampling blocks also yielded stimulation responses that were relatively lower in magnitude and less stable over the three stimulation trains.

Three potential difficulties need to be acknowledged. First, the most effective location of the stimulation electrode was achieved using protocols derived from earlier work (Baker, et al., 1993; Vanoncini, et al., 2006). However, isolation of different muscle groups may not be practically achievable. The lumbar extensors can be functionally divided into several different muscle groups, such as the erector spinae and multifidus, with the muscles structured in different layers making it difficult to identify that the exact target muscle group has been stimulated *in vivo*. However, stimulation response measurements here were done with participants sitting in the experimental fixture in a relatively controlled posture, which should ensure that the same muscles were stimulated at a similar electrode location within each sampling block. Therefore, the potential effects of changes in muscle 'cross-talk' and muscle geometry were assumed to be minimal. In addition, unlike voluntary muscle contraction, muscle contraction induced by stimulation does not follow the size principle, and muscle fibers are instead recruited without obvious sequencing related to fiber types (Gregory and Bickel, 2005; Kubiak, et al., 1987). Instead, both slow and fast muscle fibers are non-selectively activated during muscle stimulation regardless of force levels (Gregory and Bickel, 2005). As such, the potential effect of differing recruitment patterns can be minimized through stimulation. Therefore, for future applications, it is reasonable to assume that any measured changes in stimulation responses using these procedures are (predominantly) the result of changes in muscle contractile status (e.g., muscle fatigue) versus changes in muscle geometry and/or recruitment pattern. Further work will be completed to quantify the potential effects of trunk flexor and extensor activities on measured stimulation responses. In addition, it remains to determine whether stimulation responses remain reliable over longer blocks of measurement. For example, there may be important influences of diurnal, temperature, or electrode changes.

Second, the seated posture was not precisely controlled between sampling blocks, and the exact postures may have varied between different sampling blocks. Large postural changes could have introduced variability in measured stimulation responses. One reason to select a sitting position for muscle stimulation is to match the conditions (e.g. posture) during the actual tasks we plan in future studies (of seated work tasks). By using similar conditions, physiological properties of muscles (e.g., muscle length) and levels of voluntary contraction can be kept relatively consistent, and thereby help in distinguishing fatigue from potential confounding influences. Therefore, the effect of sitting postures could perhaps be minimized by controlling participant postures more precisely within and between different sampling blocks. In addition, monitoring trunk angles (e.g., using a motion tracking system), may facilitate methods to account for changes in measured stimulation responses caused by postural differences.

Third, voluntary contraction levels of the trunk flexors and extensors, as postural muscles, were controlled but not eliminated during muscle stimulation. Thus, the potential coexistence of muscle stimulation and voluntary contraction may have had a net effect on measured stimulation responses. Instead of pure twitch forces, the current methods provide a measure of net responses from both muscle stimulation and voluntary contraction of trunk flexors and extensors. Trunk extensors were monitored (real time EMG) throughout the current study, to minimize the levels of voluntary contraction. Qualitatively, this monitoring indicated relatively low and consistent levels of contraction during the sampling blocks. In addition, the sitting posture adopted during the sampling blocks placed the lumbar spine in flexion; this placed the lumbar motion segments beyond a neutral posture and activation of the trunk extensors was thus predominant. Thus, while potential effects of voluntary contraction were not eliminated, these were considered relatively small in the current study. Further work will be completed to identify the effects of trunk posture on measured stimulation responses and these results can be used to account for the effects of postural changes on measured stimulation responses.

Our long-term goal is to use muscle stimulation for detecting lumbar extensors fatigue during prolonged sitting. The developed method will be used to measure muscle fatigue in a future study, specifically to assess situations involving prolonged sitting. At present however, this approach may be limited to use in a laboratory environment, given the needs for a special seat and instrumentation. The current results suggest that stimulation at 2 Hz can provide suitable stimulation responses for this purpose. The higher responses captured using 2 Hz are considered beneficial, to allow for better separation of the muscle responses to stimulation from any background noise and/or potential voluntary muscle forces. Stimulation at this frequency required reasonable time blocks for potentiation (~4 min), and evoked stimulation responses had excellent reliability both within and between sampling blocks. The current results indicated high levels of reliability within and between sampling blocks and trains, While this suggests that only a single sampling train may be required, multiple sampling trains are recommended in future work to enhance the stability of obtained measures. .

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Chapter 3. Development and Evaluation of an EMG-based Model to Estimate Lumbosacral Loads during Seated Work

Abstract

Low back pain (LBP) is a common occupational problem and continues to be the leading cause of occupational disability. Among diverse known risk factors, sitting is commonly considered as an important exposure related to LBP risk, and modern living and contemporary work both involve increasing sedentary behaviors including more frequent and prolonged sitting. At present, however, the causal role of sitting on LBP development is controversial due to the variety of potential moderating factors. Specifically, a few studies have assessed lumbosacral loads in seated postures, but no sitting-specific model has been developed to investigate the effects of seated tasks and time-dependent variations on lumbosacral loading during sitting. Here, a three-dimensional, electromyography (EMG)-based biomechanical model of the trunk was developed to predict lumbosacral loads during a range of seated tasks. This model was a modification of an earlier approach, and specific modifications included a revised representation of lumbar muscle anatomy and viscoelastic soft-tissue properties, and a method to account for muscular fatigue during prolonged sitting. With these enhancements, the predictive ability of the model was assessed over a range of seated tasks that differed in terms of lumbar posture, time pressure, and mental workload. Predicted model parameters corresponded well with values reported earlier. Reasonable levels of correspondence were found between measured and predicted lumbosacral moments across all tested tasks. Physical exposures and injury risks related to seated work can potentially be estimated using this modeling approach, and which may facilitate future injury prevention strategies.

Key words: Sitting; Electromyography; Biomechanical model; Spine; Low back pain

3.1. Introduction

With increasing computer use and deskwork associated with most work and leisure activities, sedentary lifestyle has become a new trend in modern living, including substantial time in seated postures. Sitting has been often highlighted as a risk factor for low back pain (LBP) in the literature (Liira et al., 1996; Chaffin et al., 1999; Johanning, 2000; Vingard et al., 2000; Lee et al., 2001; Pope et al., 2002; Kopec et al., 2004; Corlett, 2006). Recent reviews, however, have suggested a less clear relationship between LBP and sitting at work (Hartvigsen et al., 2000; Hartvigsen et al., 2003; Leboeuf-Yde, 2004; Womersley and May, 2006; Lis et al., 2007; Bakker et al., 2009; Chen et al., 2009). One study (Hartvigsen et al., (2001) concluded that sedentary work was not a risk factor for LBP development, and might even have a protective effect. Sitting is a complex activity that can be affected by many moderating factors, such as sitting duration (Keyserling, 1991; Pitman and Ntuen, 1996; Stricker et al., 2003) and task requirements (van Dieën et al., 2001; McLean and Urquhart, 2002; Ellegast et al., 2007), and this complexity may underlie some of the controversy regarding the causal role of sitting on LBP (Cholewicki and McGill, 1996; Lis et al., 2007).

Several studies have been conducted to quantify the relationship between the noted moderating factors and the risk of sitting-related LBP. Yet, relatively little evidence exists regarding the specific mechanism(s) that might underlie this relationship. Several laboratory studies have focused on biomechanical mechanisms to help explain the association between sitting and LBP (Nachemson and Elfstrom, 1970; Magora, 1972; Wilke et al., 1999; Callaghan and McGill, 2001). While providing some support for biomechanical mechanisms, some of the results have been inconsistent. For example, *in vivo* measurements of intradiscal pressure (IDP) during standing have been reported to be lower than those experienced when sitting (Nachemson and Elfstrom, 1970; Andersson and Ortengren, 1974; Sato et al., 1999), yet recent studies indicated lower pressures during relaxed sitting compared to standing (Wilke et al., 1999; Rohlmann et al., 2001). Some of these inconsistencies may have been due, at least in part, to differences in measuring methods and technologies.

Disc compression forces have also been measured, both indirectly using stadiometric methods (Eklund and Corlett, 1987; Althoff et al., 1992) and more directly using instrumented implants (Rohlmann et al., 2001), to assess differences between a range of sitting and standing postures. Compression forces measured using both methods were lower in sitting postures than in upright standing. In contrast, Callaghan and McGill (2001) estimated compression forces using a biomechanical model during standing and prolonged sitting, and found substantially larger compression forces during prolonged sitting versus standing. These inconsistentces may be due to the fact that most studies comparing between sitting and standing required participants to perform standing/sitting tasks alternately. Spine length during sitting has been shown to increase after a period of standing, and which may lead to a decrease in lumbosacral loads while seated (Althoff et al., 1992). Further, both muscle fatigue and viscoelastic responses of spine passive structures (e.g., creep) could cause time-dependent changes in lumbosacral loads during relatively prolonged seated exposures (Bigland-Ritchie et al., 1995; McLean et al., 1997; Keller and Nathan, 1999; Beach et al., 2005). No existing study, to the author’s acknowledge, has assessed lumbosacral loads during seated tasks while considering the noted potential effects of muscle fatigue and viscoelastic behaviors.

Therefore, the purpose of this work was to develop a quantitative method to assess biomechanical exposures, specifically lumbosacral forces (compression and shear) and moments, during diverse seated tasks, and accounting specifically for time-dependent tissue responses and potential muscle fatigue. To this end, an EMG-based model was developed, by modifying an earlier model to be applicable to seated work under different working conditions. This model was subsequently used to predict lumbosacral loads during a range of experimental seated tasks. Including in this approach were distinct subject-specific calibration and model evaluation components. A longer-term goal of this effort was to facilitate estimates of physical exposures and potential LBP risks during seated work tasks, and to aid in the development and evaluation of injury prevention strategies for such tasks.

3.2. Method

3.2.1 Model and Procedural Overview

An EMG-based, three-dimensional biomechanical model of the trunk, suitable for application to seated tasks, was developed here as a modification of versions reported earlier (Nussbaum and Chaffin, 1998; Jia et al., 2011). As in these prior versions, the model was “driven” using surface EMG, trunk kinematics, and individual anthropometry, and was used to estimate lumbosacral (L5/S1) forces (compression and shear). In subsequent sections, an overview of and modifications (from Jia et al., 2011 version) to the major model components were presented. The three major model components are: 1) an anatomical model, representing lumbar trunk muscle geometry; 2) an EMG-based model, describing lumbar muscle force dynamics; and 3) a viscoelastic trunk stiffness model, characterizing the time-dependent behavior of trunk passive tissues. The latter two components were modified here, to address specific issues related to seated tasks. As in prior versions, when implementing the model a set of participant-invariant model parameters were first obtained from calibration procedures, and specific procedures were developed here, again to address seated tasks. Subsequently, a range of lab-based seated tasks were completed, the data from which were obtained to evaluate the current model performance both for different types of sitting work and for relatively prolonged sitting in which muscle fatigue occurs.

3.2.2 Structure of and Modifications to the Model

Anatomical model

The current anatomical model included six bilateral pairs of muscles, and was built using the AnyBody™ musculoskeletal modeling system (v5.0, repository 7, AnyBody Technology, Aalborg, Denmark). In this, a total of 92 fascicles were included. Muscles crossing the lower lumbar region were divided into two sets: 1) three flexors, including the internal oblique (IO), external oblique (EO), and rectus abdominis (RA); and 2) three extensors, including the erector spinae lumbar group (EL), erector spinae thoracis group (ET), and multifidus (MF). Initial insertions and origins were defined according to earlier data (de Zee et al., 2007), and were scaled using AnyBody™ based on individual stature and body mass.

EMG-based Muscle Model

Lumbosacral forces and moments were estimated as a function of normalized EMG, an EMG-to-force relationship, maximum muscle stress (for flexors and extensors separately), physiological cross-sectional area, electromechanical delay, and force-length relationships, as described earlier (Jia et al., 2011). Three modifications were made to account for seated conditions. First, the force-velocity relationship was excluded, since seated work is typically static or quasi-static, and given that existing studies have shown little to no improvement in the predictive ability of an EMG model in such conditions (McGill and Norman, 1986; Granata and Marras, 1995; Nussbaum and Chaffin, 1998; Nussbaum et al., 2000). Second, the time lag between muscle activation and force generation, also known as electromechanical delay (EMD), was determined by temporally shifting all EMG signals 130 milliseconds with respect to kinematic and kinetic data, as described in previous studies (van Dieën et al., 1991; Nussbaum and Chaffin, 1998). A fixed EMD was used, rather than the “global time delay” in Jia et al., (2011), to avoid having to assume linearity of the EMG-to-force relationship. Third, to account for the influence of muscle fatigue, a fatigue gain (G) was added, similar to an earlier approach (Sparto and Parnianpour, 1998). G was modified over time, based on assessing model prediction errors over time (see below for additional details).

Trunk Stiffness Model

Forces and moments caused by deformations of passive lumbar tissues (at L5/S1) were estimated by modeling trunk stiffness as a traditional 6x6 spine stiffness matrix as described earlier (Jia et al., 2011). Given the substantial trunk tissue deformations that occur in seated work over time, the model was updated to account for time-dependent viscoelastic behaviors of the passive lumbar tissues, specifically as related to trunk flexion (Toosizadeh et al., 2012). Viscoelastic behaviors of the lumbar were modeled as a three-parameter standard linear solid (SLS) subjected to prolonged axial loads (Keller et al., 1987; Taylor et al., 1990; Li et al., 1995; Keller and Nathan, 1999; Tsui et al., 2004; Brown et al., 2008; Groth and Granata, 2008; Watters et al., 2008; Toosizadeh et al., 2010).

Within the noted 6x6 stiffness matrix, the first three diagonal components (k_{11}-k_{33}) were modeled as a function of axial compressive loads, based on existing data (Stokes and Gardner-Morse, 2003; Gardner-Morse and Stokes, 2004), and as described earlier (Jia et al., 2011). The remaining three diagonal components (k_{44}-k_{66}) were represented as an SLS model, $\left[\frac{k_2\left(k_2 e^{-\left(\frac{k_1+k_2}{c}\right)t}+k_1\right)}{k_1+k_2}\right]$ using a Kelvin-Voigt body (k_1, c) and a spring (k_2), to account for viscoelastic behaviors (Poynting and Thomson, 1934; Keller et al., 1987; Toosizadeh et al., 2010). Using recent data (Toosizadeh et al., 2012), the coefficients k_1, k_2 and c were obtained as:

$$k_1 = -15.48 + 5.62 \times FR\% - 0.19 \times FR\%^2$$

$$k_2 = 23.57 + 0.42 \times FR\%$$

$$c = -2954263 + 326024.5 \times FR\% - 11247.6 \times FR\%^2 + 128.1 \times FR\%^3$$

where FR% represents the ratio of initial trunk flexion angle to the flexion-relaxation (FR) angle. FR angle is identified by a relative silence of lumbar extensor muscles activity when approaching the end range-of-motion in trunk flexion (Shirado et al., 1995; Callaghan and Dunk, 2002; Solomonow et al., 2003), and was determined here using procedures described earlier (Hendershot et al., 2011).

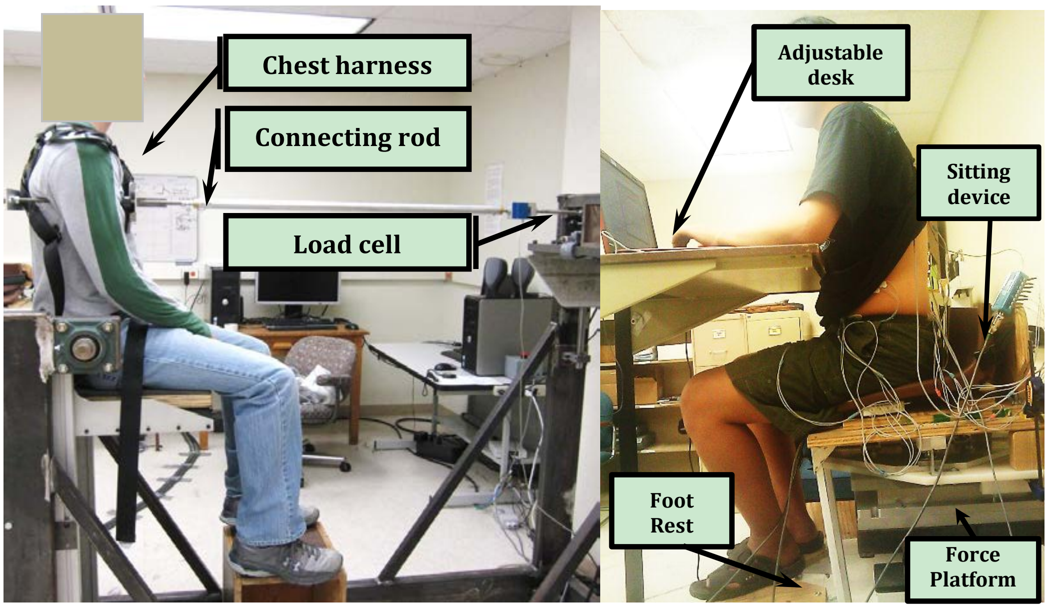
3.2.3. Participants and Experimental Procedure

Eight participants (gender balanced) completed the study, and were recruited from the university population and surrounding community. Mean (SD) age, stature, and body mass were 26.3 (3.7) yrs, 178.3 (4.7) cm, and 77.8 (12.4) kg, respectively, for males, and 26.5 (7.1) yrs, 163.9 (3.8) cm, and 61.4 (7.3) kg for females. Individuals with body mass index (BMI) >30 were excluded, due to associated challenges in measuring muscle activity using surface EMG. All participants reported being moderately physically active and having no illnesses or injuries that restricted their daily activities within the past year. Participants completed informed consent procedures approved by the Virginia Tech Institutional Review Board.

Each participant completed a series of calibration, evaluation, and fatiguing trials. During these, both trunk muscle activity (EMG) and kinematics were monitored. The bilateral activity of three extensor (MF, EL and ET) and three flexor (IO, EO and RA) muscle groups were monitored, using two telemetered EMG systems (TeleMyo 900, Noraxon, AZ). Pairs of disposable bipolar Ag/AgCl electrodes (AccuSensor, Lynn Medical, MI), with a 2.5 cm inter-electrode spacing, were placed bilaterally as described in existing studies (McGill, 1992; Mirka and Marras, 1993; Cholewicki and McGill, 1996; Potvin et al., 1996; Hides et al., 2008): IO, 10 – 12 cm lateral to umbilicus and superior to the inguinal ligament; EO, mid-point between 12th rib and anterior superior iliac spine; RA, 2 – 3 cm lateral to umbilicus; EL, 3 – 4 cm lateral to the spinous process at the L3 level, parallel to the muscle fibers; ET, 4-5 cm lateral to the spinous process at the T9/T10 level; MF, 1-2 cm lateral to the spinous process at the L3/L4 level. The skin was initially abraded and cleaned with a mild alcohol solution, and inter-electrode impedance was verified as < 10 KΩ. Trunk kinematics were monitored using two inertial measurement units (IMUs, Xsens Inc., Los Angeles, CA), placed at the T10 and L5/S1 levels.

<u>Calibration Trials: Maximum Voluntary Contractions</u>

Participants sat in a calibration fixture (Figure 3- 1 A), in which motion of the pelvis was securely but comfortably restrained by straps. An adjustable footrest was used to orient the lower limbs with right angles at the knees and ankles. Participants were set in an upright sitting posture, using a rigid rod connected to a chest harness at the T8 level, and which contained an in-line load cell (Interface SM2000, Scottsdale, AZ, USA). Participants completed four sets of maximum voluntary contractions (MVCs), involving static trunk flexion/extension and right/left lateral bending efforts. All MVCs consisted of ramp-up, hold, and ramp-down segments. At least three replications of each effort were completed, with at least one-minute of rest between each. Additional replications were completed if the third yielded the largest force. Lumbosacral moments were estimated from the recorded forces and measured moment arm, and peak EMG values for each muscle were obtained across all MVCs. After a 10-minute rest period, resting EMG activity was obtained for the ventral and dorsal muscles in relaxed supine and prone positions, respectively.



(A) (B)

Figure 3- 1. Illustration of calibration fixture setup (A), demonstrating a participant in an upright sitting posture, and the sitting device and adjustable desk setup (B) used for evaluation and fatiguing trials.

<u>Evaluation Trials: Simulated Seated Work</u>

Evaluation trials were completed with participants using a laptop while sitting on a custom sitting device (Figure 3- 1 B). This device included a combined seat pan and foot rest structure that was mounted on a force platform (AMTI OR6-7-1000, Watertown, Massachusetts). Footrest height was adjusted for each participant, so that both thighs were horizontal. A height-adjustable desk was also used, and the desk, laptop placement, and sitting device were adjusted according to workstation guidelines for each participant (HFES and ANSI, 2007). Evaluation trials involved a simple typing task, and which was done in several conditions described below. The specific task was adapted from a typing test program (Typing test TQ, Giletech e.K., Germany), and participants were asked to duplicate several paragraphs from this program using the laptop.

The typing task was done for two minutes in each of eight conditions, and which were intended to represent a range of seated work, including distinct postures and levels of muscle activity. The eight conditions included all combinations of two sitting postures, two levels of mental workload, and two levels of time pressure. The two sitting postures were lordosed (L) and kyphosed (K), and were used to represent relative extremes of common postures adopted during seated tasks (Delleman et al., 2004; Vieira and Kumar, 2004; Pynt et al., 2008). The lordosed posture was achieved by asking participants to sit in an unsupported upright posture, with the locations of the desk and laptop were adjusted to ensure the typing task could be completed without additional trunk movement. The kyphosed posture was achieved by asking participants to sit in an unsupported slumped posture, and with the same desk and laptop locations. The typing task alone was considered to have relatively low mental workload (M0), and participants were asked to type at a fast but comfortable pace with no additional time pressure (T0). A higher level of mental workload (M1) was introduced by asking participants to mentally calculate the product or summation of a pair of two-digit numbers displayed between paragraphs, similar to the approach in earlier studies (Kirschbaum et al., 1992; Hughes et al., 2007; Wang et al., 2011). Typing speed for each participant was measured after they completed an initial 1-min typing task. To introduce time pressure (T1), participants were offered a $10 bonus, as a stimulus, if they completed the experimental typing task at a rate ≥ 115% of their baseline speed. All the tasks were completed with the same requirements on typing accuracy. The order of presentation of the eight conditions was counterbalanced, using Latin Squares, and two minutes of rest were provided between each.

Fatiguing Trials: Unsupported Upright Sitting

A rest period (~5 min) was provided after the evaluation trials, and each participant then completed a single fatiguing trial involving unsupported sitting in the same device used in the evaluation trials (Figure 1B). Participants were instructed to maintain an active, upright sitting posture with lumbar lordosed (as in the "L" condition above), and this was held for 20 min or until a self-reported limit of endurance was reached.

3.2.4. Model Implementation and Evaluation

Calibration

Model parameters were specified for each participant, using data collected from the calibration trials. Raw EMG signals were amplified, band-pass filtered (20–400 Hz), and root-mean-squared converted (50 ms time constant) in hardware, then sampled at 1000 Hz. Processed EMG signals were then normalized using individual maximal (from MVCs) and resting EMG values (McGill, 1991; Mirka, 1991; Nussbaum and Chaffin, 1998; Marras et al., 1999). Trunk kinematics (IMUs) and exerted forces (in-line load cell) were both sampled at 100 Hz, and both were subsequently low-pass filtered (4 Hz cutoff; 4^{th} order Butterworth; bidirectional). Maximum muscle stresses (for flexors and extensors) and EMG-force relationships were then determined, using data from the MVCs, through an iterative procedure that minimized errors in model-based predictions of flexion/extension (sagittal plane) moments (Jia et al., 2011). This was done since both the MVC and evaluation tasks were completed in largely symmetrical postures, for which the major trunk rotations and largest lumbosacral loads were in the sagittal plane. Therefore, moments in sagittal plane were selected to calibrate model parameters and evaluate model performance. Two methods were used to assess the quality of model calibration, again focusing on sagittal plane, lumbosacral moments, and were similar to earlier approaches (McGill, 1992; Granata and Marras, 1995; Nussbaum and Chaffin, 1998; van Dieën and Kingma, 2005): 1) coefficient of determination (r^2), from a linear regression of measured vs. predicted moments in the sagittal plane; and (b) average absolute errors (AAE) between measured and predicted moments in the sagittal plane.

Evaluation

EMG and trunk kinematics were collected and processed as in calibration trials. Reaction forces and moments were recorded (at 100 Hz) from the force platform and subsequently low-pass filtered (4 Hz cutoff; 4^{th} order Butterworth; bidirectional). The model, calibrated for each participant, was then used to predict lumbosacral forces in each of the eight evaluation trials. Model predictions in each evaluation trial were assessed, as for the calibration trials, by comparing measured vs. predicted sagittal plane moments. The former were estimated based on force platform moments and inverse dynamics analyses. As described above, r^2 and AAE were used to quantify model performance for each trial. The cumulative distribution function (CDF) of r^2 was also derived to summarize results across all task combinations.

Separate analyses of variance (ANOVAs) were used to determine whether model performance (i.e., r^2 and AAE in the sagittal plane) differed depending on posture, mental workload, and time pressure. ANOVAs were also used to identify the effects of these same factors on predicted muscle EMG, sagittal plane lumbosacral moment, and lumbosacral forces (compression and shear). In these ANOVAs, within-trial mean values were the dependent measures. Significance was concluded when $p < 0.05$, and statistical analyses were performed with JMP™ (v9.0, SAS Institute Inc., Cary, NC). Summary results are presented as means (SD).

Fatiguing Trials

Data were processed using the same methods as in the evaluation trials, and the model was again evaluated by comparing measured and predicted lumbosacral moments (here, in the sagittal plane only, since other moments were relatively quite small). Associated errors in predicted lumbosacral moments were then assessed over the trial period (i.e., up to 20 min). If a temporal pattern in these errors was evident, it was presumed that this was substantially a result of muscle fatigue, given that postures were controlled and since such fatigue would alter the invariance of muscle stress and the nature of EMG-to-force relationship (Sparto et al., 1998). From qualitative examinations, these temporal patterns were mainly linear. Thus, linear regression was used to quantify the rate of change in prediction errors, and a linear adjustment to the fatigue gain (*G*) was implemented as: $G(t) = 1 + a(t)$. Here, the slope (a), was determined from regression analysis (minimizing total errors between measured and predicted lumbosacral moments in the sagittal plane), and *t* is time during the fatiguing trial. Coefficients of determination (r^2) and average absolute errors (AAE) were used to evaluate overall prediction errors (in the sagittal plane) with and without adjustments for fatigue gain. Differences in model prediction errors (between measured and predicted lumbosacral moments), with versus without adjustments to the fatigue gain, were evaluated using a paired *t* test.

3.3. Results

<u>Calibration</u>

Calibration procedures resulted in predicted muscle stress values of 80 (25.10) N/cm^2 for the flexors and 86.7 (28.33) N/cm^2 for the extensors. Calibrated models yielded a relatively high correspondence between measured and predicted lumbosacral moments in the sagittal plane, with AAE = 3.30 (1.40) Nm and r^2 = 0.85 (0.06) across all MVC trials.

<u>Evaluation</u>

The CDF of r^2 values — which assessed the correspondence between measured and prediction sagittal plane lumbosacral moments — indicated that 45% of trials had $r^2 > 0.7$.

Overall, sagittal plane moments had r^2 values of 0.65 (0.30) and AAE of 5.90 (3.43) Nm. Time pressure resulted in significantly ($p = 0.03$) higher AAE values in the sagittal plane [6.80 (3.10) Nm] compared to when there was no time pressure [5.00 (3.60) Nm]. No significant effects of posture or mental workload, or any interaction effects among posture, mental workload, and time pressure, were found on model performance.

Within-trial mean muscle activation levels were generally lower than 10% MVC across the six task conditions. Higher mean muscle activations were found in the left and right MF ($p < 0.01$ for both) in kyphosed posture. No differences in mean muscle activation levels were found between the trials with different level of mental workloads. Higher time pressure, on the other hand, resulted in higher mean muscle activation levels in several muscles (Figure 3- 2).

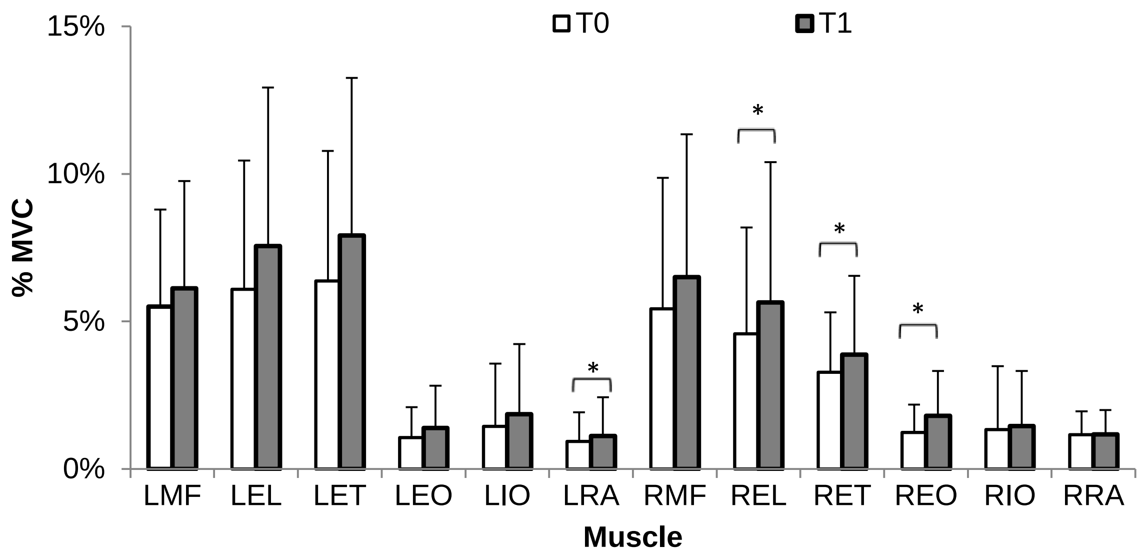


Figure 3- 2. Illustration of mean (SD) muscle activation levels for three trunk extensors (MF, EL, and ET) and flexors (IO, EO, and RA) collected bilaterally (L/R) under two level of time pressure. The * symbol indicates a significant pairwise difference.

All within-trial means of lumbosacral moments were flexion moments and all such means of antero-posterior shear were anteriorly directed; as such, results are presented for lumbosacral flexion moments and anterior shear. Across all trials mean anterior shear, compression, and lateral shear forces, were 116.90 (75.60), 683.1 (180.80), and 10.2 (10.40) N, respectively. Predicted lateral bending, axial rotation, and flexion moments were 2.47 (2.60), 1.56 (1.48), and 19.80 (9.70) Nm, respectively. The absolute values of mean lateral shear forces, lateral bending moments, and axial rotation moments, were presented (assuming there is little functional difference between bilateral and biaxial directions, respectively). There were significant main effects of posture and time pressure on flexion moments and predicted lumbosacral forces. The kyphosed posture resulted in lower anterior shear forces [34.90 (125.10) N vs. 128.80 (73.20) N] and higher flexion moments [21.80 (8.90) Nm vs. 17.70 (10.70) Nm]. Including time pressure resulted in higher anterior shear forces [104.60 (106.00) vs. 59.10 (115.10) N], larger compression forces [726.30 (173.80) vs. 639.90 (179.80) N], and higher flexion moments [21.4 (10.90) vs. 18.2 (8.00) Nm]. No significant main effects of mental workload, or interaction effects of posture, mental workload, and time pressure, were found on lumbosacral forces or flexion moments.

<u>Fatigue</u>

Adjusting the gain factor (over time) to account for fatigue generally improved the correspondence between measured and predicted sagittal plane moments (representative results shown in Figure 3- 3). Across all trials, the slope of the fatigue gain was -0.40 (0.30). There were significantly ($p < 0.01$) larger prediction errors without adjustment of fatigue gain. Using the model with adjustment yielded r^2 values of 0.56 (0.33), which were higher than values of 0.48 (0.38) without adjustment of fatigue gain. AAE was 5.40 (2.70) Nm vs. 0.60 (0.40) Nm with and without adjustment of fatigue gain, respectively.

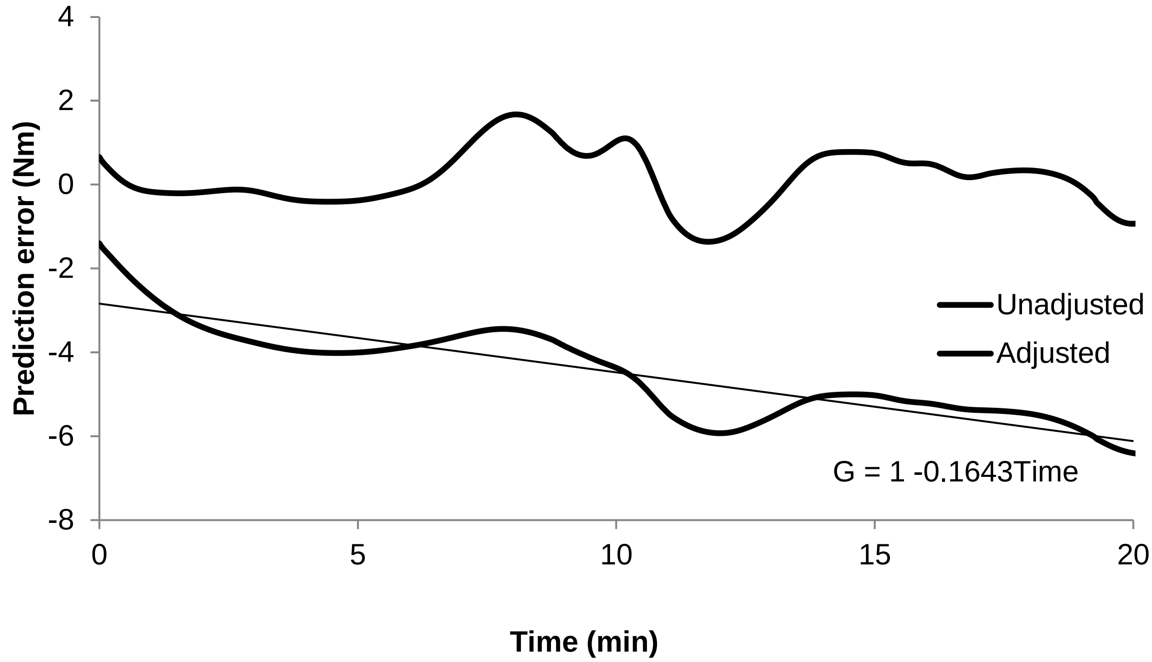


Figure 3- 3. Representative results showing prediction errors over time, between measured and model-predicted lumbosacral sagittal plane moments, both with (adjusted) and without (unadjusted) a time-variable gain.

3.4. Discussion

While some recent evidence indicates that workers who mainly work in sitting postures are more likely to experience LBP or exacerbate existing LBP (Dankaerts et al., 2006; O'Sullivan et al., 2006; Womersley and May, 2006), it remains unclear regarding the relationships between sitting and LBP and the specific mechanisms involved. Biomechanical analyses provide a quantitative method to examine potential mechanism(s) that might underlie this relationship. An accurate assessment of the partitioning of biomechanical loading among passive and active components of the trunk during sitting is needed, though, and requires representations of active and passive tissue properties and time-dependent viscoelastic properties. A model was presented here, to quantify lumbosacral loads during a range of seated tasks. Along with detailed representations of muscle anatomy and contraction dynamics, and participant-specific parameterization (calibration), the current model also incorporated new components to account for time-dependent viscoelastic behaviors of trunk tissues and muscle fatigue. The new model was evaluated in terms of predictive ability, and overall exhibited reasonable levels of correspondence between measured and predicted lumbosacral moments over the range of seated tasks examined.

Model parameters were determined for each participant using a set of maximal efforts, which was considered as a reasonable approach to estimate maximal muscle capacity during sitting and also account for intra- and inter-individual variability (McGill, 1991; Mirka, 1991). Muscle stresses estimated here (range: S_F=45-140 N/cm^2; S_E=35-115 N/cm^2) were comparable to ranges reported in earlier studies: S_F= 56-145 N/cm^2; S_E= 43-123 N/cm^2 in Bochen et al. (2011); S_F=40-140 N/cm^2 and S_E=37-105 N/cm^2 in Nussbaum and Chaffin (1998); S=37-93 N/cm^2 in Granata and Marras (1995); S=29-108 N/cm^2 in van Dieën and Kingma (2005). The wide range of estimated muscle stress values across participants likely indicates that the calibration procedures captured individual differences (e.g., PCSA and maximal stress), which would otherwise be difficult to measure directly.

In the current model, viscoelastic responses of passive trunk structure were determined using a time-dependent trunk stiffness model derived from recent empirical results

(Toosizadeh et al., 2012). Though not formally evaluated here, secondary analyses were completed to evaluate the model performance with versus without this viscoelastic component (i.e., comparing an elastic to a viscoelastic model). The results indicated a slight, but not substantial, improvement when including the viscoelastic component. A lack of more substantial differences is likely a result of the evaluation tasks lasting only 2 min. The current conditions also involved relatively low levels of passive tissue loading, versus previous reports of substantial viscoelastic responses during trunk flexion (Bazrgari et al., 2011; Hendershot et al., 2011).

Mean (SD) values of r^2 for sagittal plane moments during evaluation trials here [0.65 (0.30)] were comparable with those reported in earlier studies: 0.76 (0.15) in Nussbaum and Chaffin (1998) and 0.66 (0.20) in Kingma et al. (2001). Across all evaluation trials, the current model yielded good levels of correspondence between measured and predicted lumbosacral moments in the sagittal plane (45% of $r^2 > 0.7$), and there were no significant differences in this correspondence between tasks. Magnitudes of AAE here [5.9 (3.4) Nm] were lower than those in earlier reports – 14.1(7.4) Nm in Nussbaum and Chaffin (1998) and 56.5 (18.3) Nm in Kingma et al. (2001) – which may in part be due to the relatively low loads in the current study. No significant differences were found in terms of model prediction errors (AAE) between task conditions, except for the effects of time pressure. Although AAE magnitudes were slightly larger (1.7 Nm) in trials including time pressure, this difference was only ~ 8% of mean peak flexion moments.

In terms of predicted lumbosacral forces, the current model predicted relative smaller compression forces compared to those in previous studies (Sato et al., 1999; Wilke et al., 1999; Callaghan and McGill, 2001). Specifically, the mean compression forces predicted here was 683.1 (180.8) N in a lordosed posture, which is lower than values of 800-1000 N reported based on direct measurement in a upright sitting posture (Wilke et al., (1999); Sato et al., (1999), and substantially lower than values of ~ 1400 N reported by Callaghan and McGill (2001) using an EMG-based model. Two potential sources likely account for these differences. First, compression forces-reported earlier were at the L4/L5 joint, and these forces can be higher than at L5/S1 as assessed here (Stokes and Gardner-Morse, 1995). Second, a portion of upper body mass in the current study was supported by the desk, which likely resulted in lower compression forces vs. unsupported sitting. anterior shear forces predicted here [116.90 (75.60) N], however, were comparable to values of 135 (200) N reported by Callaghan and McGill (2001) .

Sitting posture and time pressure had significant effects on predicted lumbosacral loads. Mean anterior shear forces were higher in the lordosed posture, and which likely resulted from two sources. First, a larger portion of upper body weight, which can contribute to anterior shear, was probably supported by the desk in the kyphosed posture (though not directly measured, participants were observed to lean on the desk to a larger extent in this posture). Second, the lumbosacral joint is rotated forward in the lordosed vs. kyphosed posture. As such, gravitational loads make a larger contribution to anterior shear in a lordotic posture. Including time pressure resulted in higher anterior shear and compression forces, and higher flexion moments. In prior work, increased psychosocial stress, such as higher time pressure, has been shown to increase muscle activation levels, especially during situations involving low levels of physical demands (Marras et al., 2000; Hughes, 2004; Hughes et al., 2007). Consistent with this earlier evidence, a generalized increase in muscle activity was observed when time pressure was included (Figure 3- 2), and which likely accounts for the increased compression force. Specific causes of the larger anterior shear and flexion moments with time pressure are more difficult to discern, but may have involved changes in the patterns of muscle recruitment, postural adjustments, and/or task performance.

In most previous EMG-based models, invariance of muscle stress and a constant EMG-to-force relationship is assumed; either, however, may be violated when a muscle is fatigued. To address this, particularly during prolonged sitting, potential muscle fatigue was accounted for using a fatigue gain factor. Both linear and nonlinear EMG-to-force relationships have been reported for trunk muscles (Stokes et al., 1987; Potvin, 1992; Dolan and Adams, 1993; Potvin et al., 1996; Brown and McGill, 2008), and which especially influences model predictions for muscle activation levels between 20 and 80% of MVC (Potvin, 1992; Potvin et al., 1996). However, mean muscle activation levels here were typically < 10% of MVC (Figure 2), consistent with previous findings (Mork and Westgaard, 2005; Mork and Westgaard, 2009). Therefore, only the assumption of invariance of muscle stress was considered relevant in the current task conditions (prolonged sitting). Without an adjustment for fatigue, increasing prediction errors were found over time here, typically involving increasing overestimation of predicted moments, and consistent with existing studies (Sparto et al., 1997; Sparto et al., 1998), that muscle contraction “efficiency”, or force per unit EMG, declines with fatigue. Here, a simple linear correction was able to substantially address the overestimation of predicted moments with fatigue. Future work, though, may benefit from considering more complex time-dependent adjustments.

A few limitations in the present work should be acknowledged. First, all of the tasks evaluated here involve sagittally-symmetric conditions and as a result the model evaluation approach was only focused on the sagittal plane. Future work is needed to evaluate the model performance under more complex task conditions that include a wider range of trunk movements, especially those involve asymmetric postures such as use of a mouse or reaching to the side. Second, the current work did not include a chair backrest, due to the difficulties involved in tracking trunk posture (due to potential interference of a backrest) and capturing forces exerted by/on a backrest. Further studies are needed to overcome these limitations. Third, viscoelastic components were included in current model to account for the contribution of time-dependent changes in spinal soft tissues. However, the current task stress and sitting duration were both relatively low, which likely limited any observable effect of the viscoelastic component on model performance. Additional testing is needed to assess the potential benefits of the viscoelastic component, such as during more prolonged sitting especially in kyphosed postures without desk. Fourth, fatigue was assessed only in a limited aspect. Only a single gain factor was used for all muscles, which assumed that all muscles had equal fatigue levels during a fatiguing trial. In future research, as noted earlier, more complex fatiguing conditions should be used, and methods developed to account for more complex patterns of muscle fatigue/recovery.

In summary, an EMG-based biomechanical model has been developed to predict lumbosacral loads during various seated tasks. Through a calibration procedure, model parameters were specified on an individual basis and used to estimate lumbosacral forces. This model was assessed during a range of seated tasks, and demonstrated a reasonable level of predictive ability. Subsequently, model outputs (e.g., lumbosacral loads) may be of use as quantitative descriptions of physical demands and injury risks during more realistic task conditions. Further research is still needed, however, to evaluate predictive ability and to extend the model to more complex task conditions.

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Chapter 4. The Effects of Prolonged Sitting and Psychosocial Stress on Low Back Kinematics, Discomfort, and Localized Muscle Fatigue

Abstract

Changes in the nature of work have led to an increase in sedentary behavior. The effects of sitting duration and work stress on the low back, however, are not well documented, and quantifying these effects was the purpose of this study. Participants (n=14) completed prolonged (40-min) computer-based tasks, using a laptop without a desk. These were done under two levels of psychosocial stresses: low stress involved relaxed web browsing, whereas higher stress was induced through additional mental workload (mental calculation) and time pressure (fast typing speed) during a typing task. Both subjective (perceived discomfort) and objective measures obtained, the latter including lumbosacral kinematics, muscle activity, and measures of localized fatigue. Prolonged sitting significantly increased perceived discomfort and trunk flexion angles and led to muscle fatigue. No significant effects of psychosocial stress were found on any of the dependent measures. Relatively weak correlations were found between subjective and objective measures, though two fatigue measurement methods (based on EMG and stimulated responses) showed a good level of correspondence. This study provides a quantitative assessment of the effects of the prolonged sitting and psychosocial stress during a specific computer-based task, extending our understanding of sitting behaviors under different work conditions. Further research is needed to investigate more realistic occupational and non-occupational sitting conditions.

Key words: Prolonged sitting; Psychosocial stress; Perceived discomfort; Muscle Fatigue; Lumbosacral kinematics/kinetics

4.1. Introduction

Sedentary lifestyle, which involves more prolonged sitting, becomes a new trend in modern living (Varo et al., 2003; Ford et al., 2005). Recent estimates indicate that more than half of all population do not maintain physical activity levels to achieve health benefit (Blair et al., 1999; Pleis et al., 2009), and more than one-third of adults worldwide are physically inactive (Hallal et al., 2012). Existing evidence suggests that workers who mainly work in sitting postures are more likely to experience or aggravate low back pain (LBP) (NIOSH, 1997; Dankaerts et al., 2006; O'Sullivan et al., 2006; Womersley and May, 2006). Sitting is a complex behavior that is affected by both the design of working environment and the tasks being performed (Magnusson and Pope, 1998; Delleman et al., 2004). Moderating factors, such as sitting duration (Keyserling, 1991; Pitman and Ntuen, 1996; Stricker et al., 2003), or psychological task stress (McLean and Urquhart, 2002; Ellegast et al., 2007), likely also play important roles in individual behaviors and responses (e.g., LBP) during sitting.

Prolonged sitting often results in discomfort and/or pain in the lower back (Pope, 1991; Fenety et al., 2000; Beach et al., 2005; Dunk and Callaghan, 2005; Gregory et al., 2006). Both muscle fatigue and viscoelastic properties of spine passive structures could cause time-dependent changes in lumbosacral loads during prolonged sitting (Bigland-Ritchie et al., 1995; McLean et al., 1997; Keller and Nathan, 1999; Beach et al., 2005). Muscle fatigue has been measured via muscle electromyography to assess the effects of prolonged sitting (McLean et al., 1997; Seghers et al., 2003).

Likely due to potential instability of EMG median power frequency caused by alterations in neuromuscular control strategies and geometric changes during low level muscle contraction, previous evidence indicates that fatigue-related changes in EMG signals can be detected consistently typically when muscle contraction level is greater than 15% of maximal muscle capacity (Oberg et al., 1992; Oberg et al., 1994). Typical activities generated by postural muscles during sitting, on the other hand, are usually at relatively low contraction levels of < 10% of maximal muscle capacity (van Dieën et al., 2001; Mork and Westgaard, 2005; Mork and Westgaard, 2009). Viscoelastic changes in spine structures (i.e., creep) during prolonged sitting have been investigated by measuring trunk flexion angles (Scannell and McGill, 2003; Beach et al., 2005). No existing study, to the author's knowledge, has quantified the time-dependent changes in lumbosacral kinetic during prolonged sitting with consideration of both potential effects of muscle fatigue and viscoelastic behaviors of trunk.

Modern jobs that require prolonged sitting also can involve precision work, a high mental workload, and a stressful working schedule. Psychosocial stress (e.g., time pressure and high mental workload) has been found to affect not only trunk kinematics but also to increase muscle activation levels (Pitman and Ntuen, 1996; van Dieën et al., 2001; Ellegast et al., 2007; Hughes et al., 2007). The high prevalence of musculoskeletal disorders in psychologically stressful, but light physical, work has indicated that increased perceived workload plays an important role in the development of LBP (Jensen et al., 1998; Davis and Heaney, 2000). More generally, psychosocial stress has been related to the development of work-related musculoskeletal disorders in upper extremities (Jensen et al., 1998; Hughes et al., 2007), shoulder (McLean and Urquhart, 2002; Blangsted et al., 2004), and low back (Marras et al., 2000). However, the effects of psychosocial stress on the low back during prolonged sitting have not been quantified.

Therefore, this study was designed to determine the effects of prolonged sitting and psychosocial stress associated with computer-based seated tasks on several outcomes including discomfort, lumbosacral kinematics, muscle activity, and localized muscle fatigue (LMF). Participants completed two computer-based seated tasks, intended to represent some common activities performed in a seated posture. We hypothesized that prolonged sitting and psychosocial stress would have significant effects on both objective and subjective measures, and that there would be a correspondence between these measurements. A longer-term goal of this effort was to expand our understanding of the potential adverse consequences of prolonged seated tasks and provide information to facilitate future practical interventions to reduce sitting-related risks.

4.2. Methods

4.2.1. Participants

Fourteen participants (gender balanced) completed the experiment and were recruited from the local student population and community. Mean (SD) age, body mass, and stature were 24.5 (4.1) yrs, 174.5 (9.1) cm, and 72.3 (11.1) kg, respectively. After a demonstration of the experimental protocols, all participants completed an informed consent process approved by the Virginia Tech Institutional Review Board. No participants had any self-reported musculoskeletal injuries or neurological diseases within the past year, and individuals with body mass index >30 kg/m^2 were excluded due to the potential difficulties in obtaining EMG measures and evoking reliable muscle stimulation response.

4.2.2. Experimental Design

Participants completed two tasks in separate trials that involved prolonged (40 min) use of a laptop in a seated posture without a desk, and which were done in two conditions involving different levels of psychosocial stress (low vs. higher). While some individuals often sit longer, a 40 min period was thought to represent a typical uninterrupted period of laptop usage under non-desk condition. The low psychosocial stress (LS) condition had participants engage in relaxed internet browsing, which involved participants performing random internet "surfing" (e.g., reading news, streaming video). A higher level of psychosocial stress (HS) was introduced by asking participants to complete a typing task in a situation involving both additional mental workload and time pressure (described in Chapter 3). During the typing task participants duplicated several paragraphs from a typing test program (Typing test TQ, Giletech e.K., Germany). To induce additional mental workload, pairs of two-digit numbers were displayed between each paragraph of text, and participants were asked to mentally calculate the product or summation of these paired numbers, similar to existing approaches (Kirschbaum et al., 1992; Hughes et al., 2007; Wang et al., 2011). To add time pressure, participants were asked to type as fast as they could and they were offered a $10 bonus, as an incentive, if they completed the typing task at a rate ≥ 115% of their baseline speed; this baseline was measured during a 1-min typing task completed at an individual's normal typing speed. The order of stress conditions was counterbalanced across participants, and at least four days were provided between each trial to minimize carry-over effects (Edwards et al., 1977; Jones et al., 1989).

4.2.3. Instrumentation and Data Collection

Trunk angles (relative rotations of the trunk vs. the pelvis) and muscle EMG were collected continuously during each trial, as described earlier (Chapter 3). Briefly, two telemetered EMG systems (TeleMyo 900, Noraxon, AZ) were used to collected EMG bilaterally from three lumbar extensors [erector spinae lumbar group (EL), erector spinae thoracis group (ET), and multifidus (MF)], and three lumbar flexors [internal oblique (IO), external oblique (EO), and rectus abdominis (RA)]. Raw EMG signals were sampled at 1000 Hz and then amplified, band-pass filtered (20 - 400 Hz), and root-mean-square (RMS) converted with a moving window of 50-millisecond duration in hardware. Trunk angles were monitored using two inertial measurement units (IMUs: Xsens Inc., Los Angeles, CA), fixed to the dorsal aspect of the trunk at the T10 and S1 levels. Trunk angles were collected at 100 Hz and subsequently low-pass filtered at 4 Hz (4^{th} order Butterworth; bidirectional). All data were collected simultaneously using LabVIEW software (version 8, National Instruments™, Austin, USA). RMS EMG signals were down sampled to 100 Hz to match the trunk angles.

Stimulated responses of the lumbar extensor muscles were measured (as an indicator of fatigue), following the protocols described in Chapter 2. In brief, a dual-channel system, consisting of a stimulator (Grass S88) in series with a stimulus isolation unit (SIU5) and a constant current unit (CCU1; all from AstroMed, Inc., West Warwick, RI) was used to stimulate the lumbar extensors bilaterally. After a conditioning train, the stimulation responses were collected (at 1000 Hz) during three 16-second stimulation trains at 2 Hz.

Perceived discomfort (PD), for the whole body and eight local body parts (neck, left/right shoulders, upper/lower back, buttocks and left/right thighs), was measured using a modified Borg CR-10 and Body Part Discomfort Scale (Corlett and Bishop, 1976; Borg, 1990; Kyung et al., 2008a; Figure 4-1).

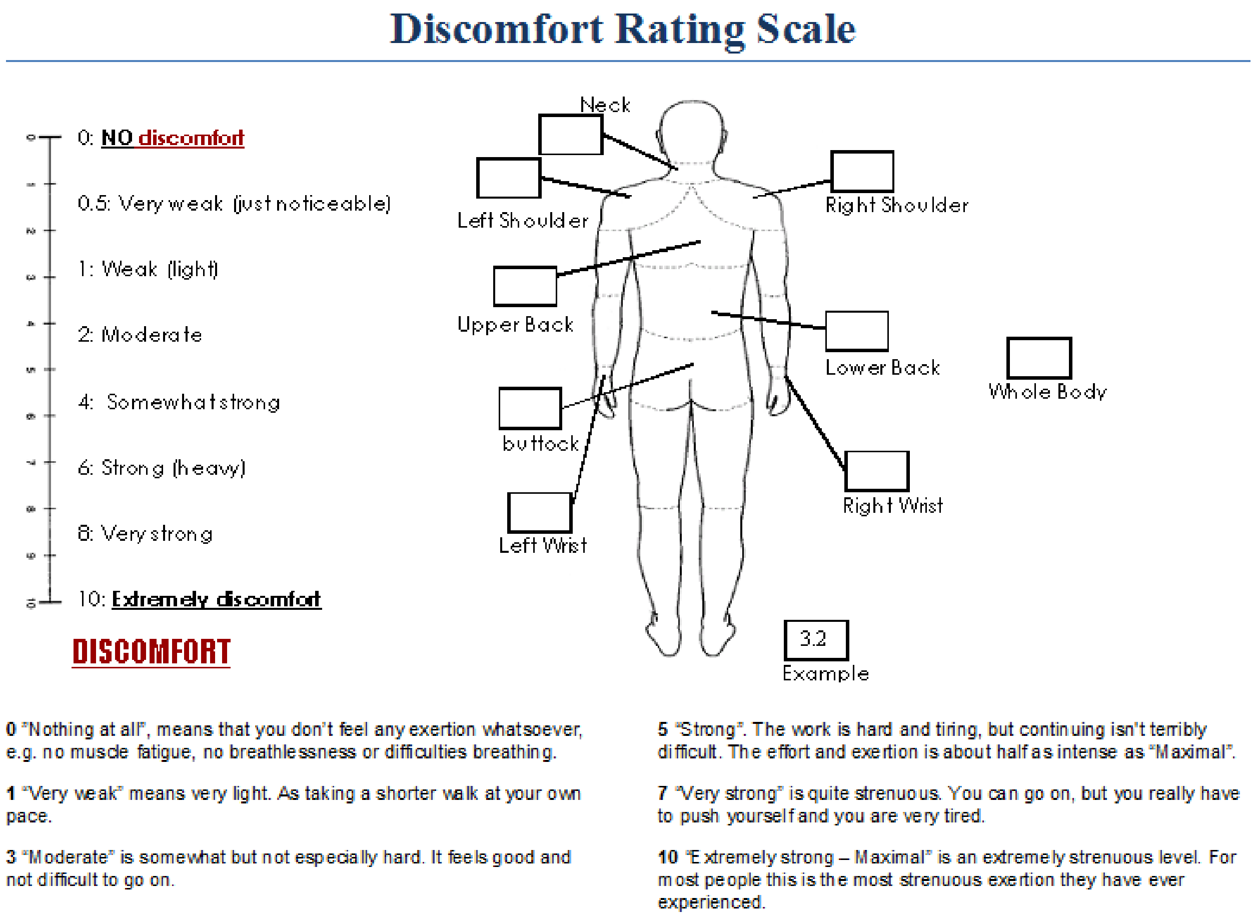

Figure 4- 1. Methods used to obtain subjective ratings of perceived discomfort and fatigue (modified from Kyung and Nussbaum, 2008b)

4.2.4. Experimental Procedures

Each trial consisted of initial calibration procedures, a 40 min seated task, and collection procedures of flexion-relaxation (FR) angle, stimulation responses, and perceived discomfort (PD) before and after seated task (Figure 4- 2). The FR angle of the trunk was collected. FR angle was measured by asking participants to bend forward to reach their end range-of-motion in trunk flexion (Shirado et al., 1995;

Callaghan and Dunk, 2002; Solomonow et al., 2003), and then the FR angle was determined by the relative silence of lumbar extensor muscle activity using procedures described earlier (Hendershot et al., 2011). Subsequently, muscle stimulation responses were collected after positioning the participant into a customized sitting fixture and completing a conditioning train (Chapter 2). Participants then completed a 40-min task in one of the two conditions. Each trial involved participants sitting on a stool with no back rest or back support. The height of the stool was adjusted for each participant, so that both thighs were horizontal and with a 90 degree included knee angle. The stool height, laptop placement, and participants' postures were monitored and the same setup was used for the second trial. Each participant performed the task with the laptop on their lap and in a kyphosed posture, and this configuration was used to represent one of the more common non-neutral sitting postures when using a laptop away from a desk (Moffet et al., 2002; Asundi et al., 2010; Gold et al., 2012). To avoid potential confounding from participants' movements and the ambient environment, participants were required to sit still but relaxed and were not allowed to communicate with the experimenters for the duration of the task. At the beginning of the task, participants were asked to rate their perceived discomfort. Right after the task trial, PD scores, FR angles, and stimulation responses were again collected, in a consistent sequence.

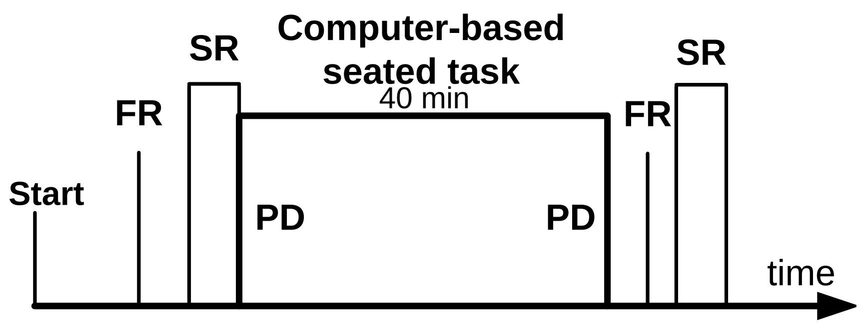

Figure 4- 2. Illustration of experiment procedure. FR: flexion-relaxation measurement; SR: stimulation-based fatigue measurement; PD: perceived discomfort via subjective rating.

4.2.5. Data Reduction and Processing

The first minute of each task was discarded since participants often needed this time to obtain a stable posture. Mean (SD) values of trunk angles (flexion), and EMG MPF, were obtained from the first five (i.e., 2 - 6 min) and the last five min of each task. Only trunk flexion angles were analyzed, since both trials were completed largely in symmetrical postures, for which the major trunk rotations were in the sagittal plane. Therefore, trunk flexion angles in sagittal plane were selected to evaluate the effects of prolonged sitting and psychosocial stress. FR angle, SR, and EMG MPFs were normalized to initial values collected before each task to facilitate comparisons between participants and different measures.

Localized muscle fatigue was assessed using two methods. The first method used raw EMG obtained during the trials, from which the median power frequency (MPF) of each EMG signal was obtained over 1-second windows. Mean EMG MPF obtained from the first five and the last five min of each task were calculated and compared. Changes between these mean values were used as a predictor of fatigue development in these muscles. The second method used mean stimulation responses to identify potential muscle fatigue. Muscle fatigue is identified when a drop was found in measured stimulation responses before and after each trial, as described in existing studies (Edwards et al., 1977; Binder-Macleod and Snyder-Mackler, 1993; Johnson et al., 1995a). Stimulation responses were obtained as the means, across all peak stimulation responses recorded during the 16s block of all three trains, and the three largest and three smallest values were excluded to minimize the influence of outliers on estimates of central tendency (Kogi and Hakamada, 1962; Rothenberg et al., 1964).

4.2.6. Statistical Analyses

Multivariate analysis of variance (MANOVA, using Wilks' Lambda) was used to determine the effects of prolonged sitting (specifically, the first five vs. last five minutes), psychosocial stress, and their interaction on the dependent measures as a whole. Specific dependent measures were, FR angle, trunk flexion angle, stimulation response, muscle EMG and MPF, and perceived discomforts. In the event of a significant MANOVA effect, univariate ANOVAs were performed for each dependent variable where relevant. Effects were considered significant when $p<0.05$. To assess whether there was any underlying relationship among perceived discomforts from the eight body segments, principal component analysis (PCA) was performed on the PD scores. For the same reason, PCA analysis was also performed on all EMG median frequencies. Correspondence between objective measures [trunk angle (flexion), muscle EMG MPF (PC component) and stimulation response] and subjective measures [perceived discomfort (PC component)] were evaluated using bivariate coefficients of correlation (ρ). To assess the consistency between the two current fatigue measurement methods, ρ values were also obtained between EMG median frequencies (PC component) and stimulation responses.

4.3. Results

MANOVA and ANOVAs results are summarized in Table 4- 1. The former indicated that both prolonged sitting ($p<0.01$) and psychosocial stress ($p=0.04$) had significant effects on the set of dependent measures, but no significant interaction effect ($p = 0.90$). Subsequent univariate ANOVAs indicated that prolonged sitting significantly affected, trunk flexion angles, stimulation responses, MPF from most muscles, and perceived discomfort in all body regions evaluated. Psychosocial stress did not have significant effects on any of the dependent measures.

Table 4- 1. Summary of ANOVA results for effects of prolonged sitting and psychosocial stress.

		Prolonged Sitting		Psychosocial Stress	
		$F_{(2,1057)}$	p	$F_{(1,1057)}$	p
Flexion-relaxation angles		0.49	0.49	2.83	0.10
Trunk angle		8.32	**<0.01**	1.09	0.30
Stimulation Response		4.01	**0.05**	0.11	0.74
EMG Median Frequency	LMF	2.16	0.15	0.07	0.79
	LEL	7.65	**<0.01**	0.15	0.71
	LET	5.36	**0.02**	1.02	0.32
	LEO	5.46	**0.03**	0.11	0.75
	LIO	6.51	**0.02**	0.25	0.62
	LRA	2.71	0.12	2.59	0.12
	RMF	10.58	**<0.01**	0.17	0.68
	REL	6.25	**0.02**	3.11	0.09
	RET	4.34	**0.04**	3.02	0.09
	REO	0.03	0.86	0.13	0.73
	RIO	6.38	**0.02**	0.94	0.34
	RRA	5.10	**0.03**	1.23	0.74
Discomfort	Neck	35.31	**<0.01**	1.01	0.32
	L shoulder	22.59	**<0.01**	0.01	0.94
	R shoulder	32.45	**<0.01**	0.09	0.77
	Upper back	106.53	**<0.01**	1.94	0.17
	Lower back	39.64	**<0.01**	0.03	0.86
	Buttock	27.86	**<0.01**	0.32	0.57
	L wrist	5.68	**0.02**	1.71	0.20
	R wrist	11.85	**<0.01**	0.48	0.49
	Whole body	104.07	**<0.01**	1.02	0.32

Across trials, mean levels of muscle activity were typically lower than 10% of MVC and with the highest levels of activity found for the bilateral erector spinae thoracis. No significant (p>0.10) differences on mean muscle activity levels were found between two psychosocial stress conditions. For the extensor muscles, however, activation levels were typically higher in the HS condition (Figure 4- 3). Activation levels of trunk flexors, on the other hand, were generally lower in the HS condition exception the left IO.

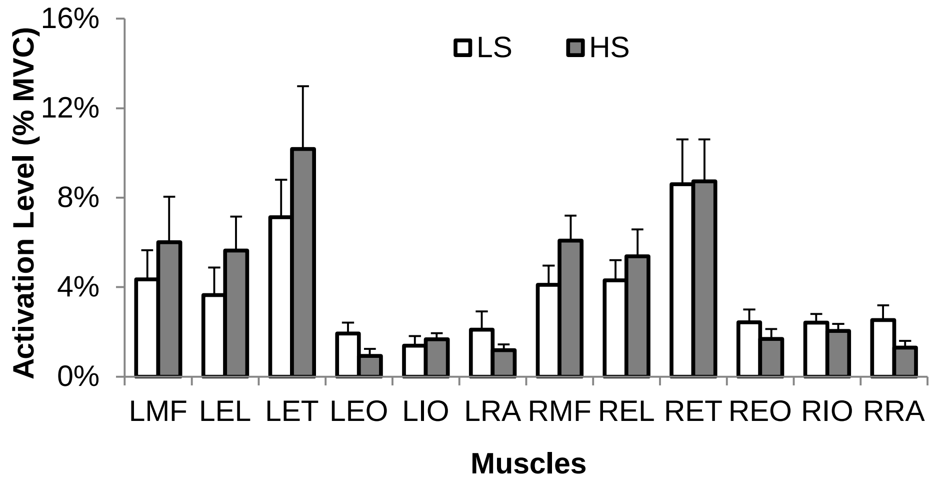

Figure 4- 3. Summary of muscle activation levels (% of MVC) between the two task conditions (LS = low psychosocial stress, HS = higher psychosocial stress).

FR angle, muscle stimulation responses, and EMG MPF, and are summarized in Figure 4- 4. Although not significant (*p*=0.44), FR angle generally increased at the end of each task. Signs of muscle fatigue were found in both stimulation response and EMG MPF measures. Prolonged sitting resulted in a significant (*p*=0.05) decrease in stimulation responses and significant declines of MPF from all muscles except the REO. No main or interaction effects of psychosocial stress effects were found for stimulation response, EMG MPF, or FR angle. Though not significant, both fatigue measurements indicated more fatigue in the HS condition.

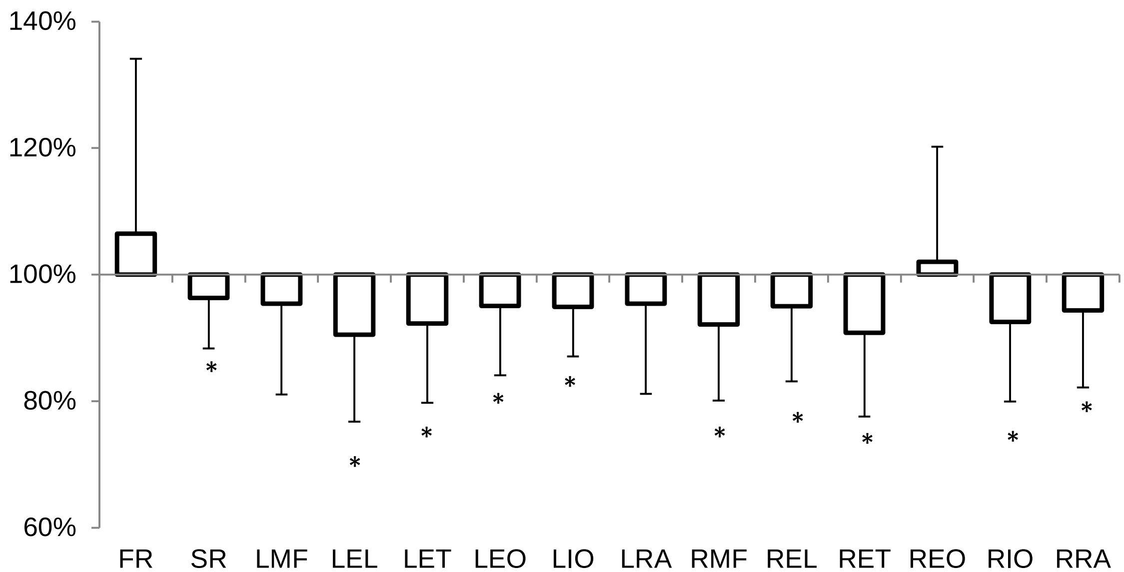

Figure 4- 4. Illustration of mean changes (after vs. before task) in normalized flexion-relaxation angles (FR), stimulation responses (SR), and median frequencies. All measures were normalized to initial values (i.e., 100% = no change over time). The * symbol indicates a significant effect of prolonged sitting, and error bars indicate standard deviations.

Perceived discomfort results are summarized in Figure 4- 5. Perceived discomfort of the whole body and each of the eight local body parts significantly (p<0.01) increased by the end of the tasks. Largest mean perceived discomfort was found in upper back, followed by lower back and whole body. No main or interaction effects of psychosocial stress were found on perceived discomfort.

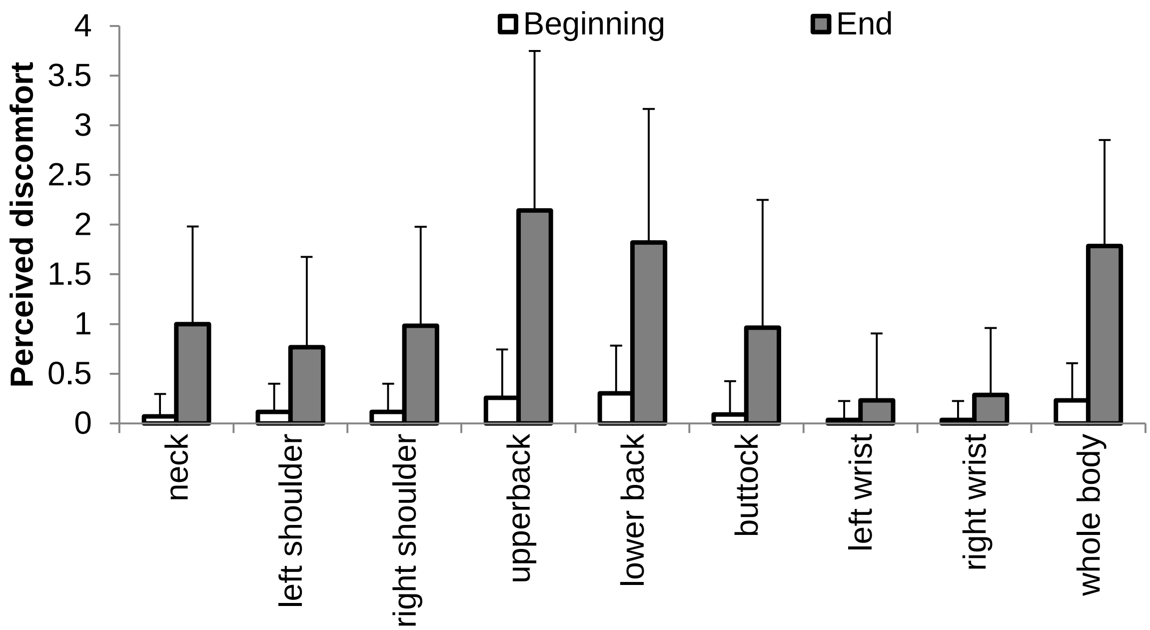

Figure 4- 5. Mean (SD) values of perceived discomfort for each body part. All PD scores significantly (p<0.01) increased at the end of the task.

For both EMG and PD scores, one dominant PC was found for each, and which accounted for > 38% of the variance in EMG and > 67% of the variance in PD scores (Table 4- 2). PC1 for both EMG MPF (EPC1) and PD scores (PPC1) can be roughly considered as representing a mean across individual MPF and PD measures, in that most of the coefficients have comparable magnitudes with the same sign. Therefore, these two PCs were denoted as the components that represent overall measures of EMG and PD.

Table 4- 2. Principal component (PC) coefficients for EMG and PD scores.

PD	Neck	Left Shoulder	Right Shoulder	Upper Back	Lower Back	Buttock	Left Wrist	Right Wrist	Whole Body
PPC1	0.3	0.36	0.35	0.32	0.32	0.33	0.34	0.35	0.33

EMG	LMF	LEL	LET	LEO	LIO	LRA	RMF	REL	RET	REO	RIO	RRA
EPC1	0.15	0.37	0.12	0.37	0.19	0.40	0.27	0.35	0.17	0.20	0.31	0.38

As shown in Figure 4- 6, no strong or significant relationships were found between objective and subjective measures. Although not significant, PPC1 was positively related to trunk angle (flexion), compression forces, and flexion moments, and was negatively related to stimulation responses and EPC1. A positive, though not significant (ρ=0.23, p=0.08) relationship was found between EPC1 and stimulation responses.

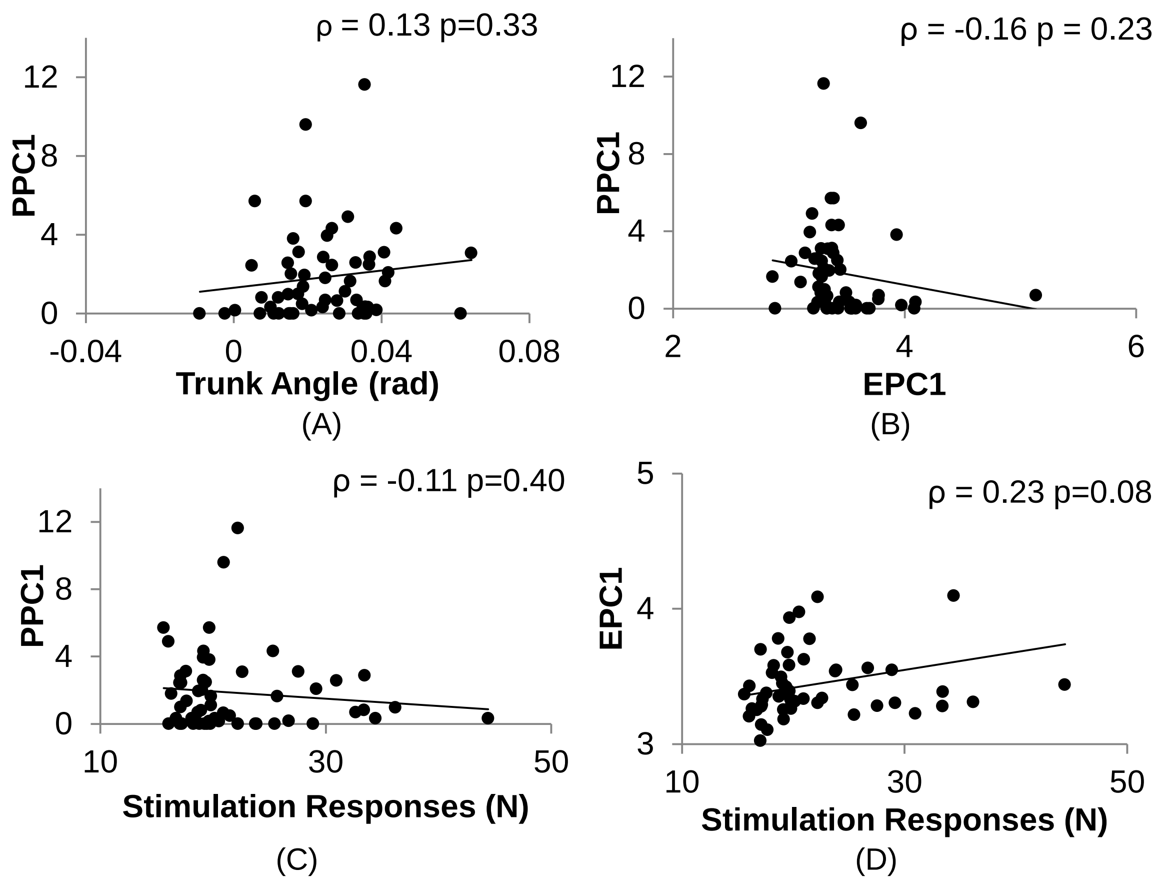

Figure 4- 6. Relationships between subjective (PPC1) and objective measures [trunk angle (A), EPC1 (B), and stimulation responses (C)], as well as between stimulation responses and EPC1 (D).

4.4. Discussion

While some moderating factors, such as prolonged sitting and psychosocial stress could have a direct impact in individual behavior, these may also potentially increase the risk for LBP development though few studies have investigated the effects of such factors. The objective of this study was to determine the effects of prolonged sitting and psychosocial stress associated with computer-based seated tasks on trunk kinematics, perceived discomfort, and localized muscle fatigue.

Only prolonged sitting had significant main effects on several of the dependent measures (Table 4- 1), with no significant main or interactive effect of psychosocial stress found. The latter suggests only additive effects of prolonged sitting, or that the effects of prolonged sitting exposure are consistent across different levels of stress. It may be that the levels of both factors examined here were relatively narrow, and thus not able to capture potential interactive effects. It is also unclear if this lack of an interaction effect will generalize to a larger range of exposure conditions. Previous work indicated that increased psychosocial stress could contribute to increased muscle activation, especially during conditions involving relatively low physical demands (Marras et al., 2000; Hughes, 2004; Hughes et al., 2007), and such increased muscle activity was found here (Figure 4- 4), though such difference were not significant.

There were significant increases in trunk flexion angles at the end of both tasks. Participants were asked to maintain a static, kyphosed sitting posture, with minimum posture adjustments or in-chair movement. Both creep and fatigue in the low back can develop with static trunk flexion (Kazarian, 1975; Keller et al., 1987; Adams and Dolan, 1996; Kanlayanaphotporn et al., 2002), and creep may occur after only five minutes of static flexion (Adams and Dolan, 1996; Kanlayanaphotporn et al., 2002).

FR angle is often used as an indicator of creep deformation in spinal structure under flexion exposures, with results from such studies indicating that prolonged trunk flexion increases FR angle (Olson et al., 2004; Olson et al., 2006; Shin et al., 2009). Here, prolonged sitting resulted in an increase in FR angles in both stress condition (Figure 4- 5), indicating likely creep deformation of the spine, and supporting that observed increases in trunk angles were likely due to creep responses during prolonged sitting.

Muscle fatigue was measured using both muscle stimulation and EMG MPF. Prolonged sitting resulted in significant drops in stimulated responses and declines in EMG MPF at the end of both task conditions, but no effects of psychosocial stress were evident. Stimulation responses have been used to measure muscle fatigue during sustained, low level muscle contractions (Johnson et al., 1995a; Johnson, 1998; Mork and Westgaard, 2005), with fatigue identified by a drop in measured stimulation response (Kazarian, 1975; Edwards et al., 1977). Using this criterion, muscle fatigue in trunk extensors was apparent as a result of 40 min seated task in a kyphosed sitting posture. Muscle fatigue was also apparent from a significant decline of EMG MPF, which was found for several flexor and extensor muscles. Some of the inconsistencies between muscles may be attributed to functional differences between the trunk flexors and extensors. During prolonged sitting, trunk extensors are usually considered as postural muscles, and which are required to be continuously activated to stabilize the sitting posture (van Dieën et al., 2001; Mork and Westgaard, 2005; Mork and Westgaard, 2009). Therefore, the observed decline of EMG MPF among three trunk extensors (EL, ET and MF) could be used as a good representation of muscle fatigue caused by prolonged sitting. The decline of EMG MPF in the bilateral both IO suggests that this muscle also plays an important role in stabilizing the trunk posture during sitting, and this is consistent with existing evidence (Snijders et al., 1995). Instead of testing the relationship between stimulation responses and each individual EMG MPF, the correlation between the two muscle fatigue measurement methods was instead completed using the first principal component for all EMG MPF. Though only approaching significance EMG MPF was positively correlated with stimulation responses (Figure 4- 6 F), and both measurements declined over time. Consistency between these different measures supports that fatigue indeed developed during the tasks, and that both approaches have potential value in assessing muscle fatigue during prolonged sitting.

Perceived discomfort also significantly increased with prolonged sitting. A prolonged static sitting posture, such as that imposed by the computer-based tasks in this study, has been known to lead to discomfort of the neck, shoulders, and back (Grandjean et al., 1983; Stokes et al., 1987; Gay et al., 2006), and to increase with time during uninterrupted tasks over the working day (Marras et al., 1995; Ostensvik et al., 2009). To minimize the potential confounding effects of in-chair movement, participants here maintained a kyphosed sitting posture, and which can increase intradiscal pressures (Wilcox and Keselman, 2003), stretch the posterior spine ligaments (Owen et al., 2010), and restrict nutrient supplies to nerves (Hallal et al., 2012). These factors may have contributed to the observed increases in discomfort.

While subjective measures (e.g., perceived discomfort) provided direct assessment of sitting discomfort, objective measures can potentially provide unbiased estimations that are related to sitting discomfort (Lee et al., 1993; Kyung et al., 2008a). Support for objective measures, as useful additions to subjective measurements of sitting discomfort, would be provided if a good correlation existed between these two types of measures. Several studies have been completed to identify the relationship between subjective measures (e.g., perceived discomfort) and a range of objective measures (e.g., trunk posture, muscle EMG and pressure distribution; Eklund and Corlett, 1987; Lee et al., 1993; Salewytsch and Callaghan, 1999; Vergara and Page, 2000). From these, pressure distribution was found to be most clearly related to subjective measures (Thakurta et al., 1995; Vergara and Page, 2000; Kyung and Nussbaum, 2008b).

In this study, the first principal component (PC1) of PD scores was used as a representation of overall perceived discomfort across all several measured body parts and the whole body, and correlations were obtained between this and several objective measures (trunk angle, muscle EMG MPF and stimulation responses). No significant correlations between subjective and objective measures were found. Trunk angle was positively related to perceived discomfort, specifically with higher trunk flexion angles related to higher perceived discomfort, and which was consistent with previous evidence (Eklund and Corlett, 1987). Although not significant, declines in both fatigue measures (EPC1 and stimulation response) were related to an increase in perceived discomfort. This suggests that participants' perceived higher discomfort at least in part as a result of muscular fatigue, as has been described earlier (Rohmert and Luczak, 1978; Zhang et al., 1996). Although no strong association between subjective and objective measures was found in this study, the objective measures used may still provide useful information, such as in future work to predict discomfort without potential biases involved with subjective measures.

Two potential limitations of this work need to be acknowledged. First, the effects of prolonged sitting and psychosocial stress were evaluated under a constrained sitting condition. Each experimental task was completed with participants sitting in a single, fixed posture, and with minimized in-chair moments. In a realistic sitting situation, participants usually make several posture adjustments or take microbreaks to reduce body discomfort (Salewytsch and Callaghan, 1999; McLean et al., 2001). Therefore, the results from the current study may have limited ability to generalize to realistic use of a laptop in a non-desk condition. Future work is needed to test the effects of prolonged sitting and psychosocial stress under more complex task conditions that include a wider range of trunk movements and sitting postures. Second, the duration of the prolonged task was determined based on the results from our pilot studies, where muscle fatigue was observed after 30 min of quiet sitting. There is a lack of evidence, to the author's knowledge, regarding typical exposure durations for laptop use without a desk, so the applicability of the current work, in terms of the duration of actual use, is unknown. Additional study is needed to identify the common exposure durations of sitting under different conditions.

In summary, prolonged sitting was found to have important impacts on lumbosacral kinematics, and to lead to the development of muscle fatigue and perceived discomfort. The current findings suggest that individuals who sit for prolonged periods can be at increased risk of injury and that this risk may also be higher with increased psychosocial stress. These outcomes were evident after 40 min of sitting under experimental conditions involving laptop use without a desk, which could be of particular concern for some conditions of laptop use (e.g., in an airport). This study highlighted the potential influence of sitting-related moderating factors on individual behaviors and provided evidence to facilitate better understanding of sitting-related risks. Further research is still needed, however, to assess the effects of prolonged sitting and psychosocial stress in more complex occupational and non-occupational sitting conditions.

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Chapter 5. Conclusions and Recommendations

Sitting is a complex behavior, though outcomes of which are likely affected by a number of potential moderating factors such as sitting duration or psychological task stress (McLean and Urquhart, 2002; Ellegast et al., 2007). The relationship between these moderating factors and risks of sitting-related LBP has been studied previously, yet there remains little and inconsistent evidence regarding the specific mechanism(s) that underlie sitting-related low back pain (Hartvigsen et al., 2000; Leboeuf-Yde, 2004; Womersley and May, 2006; Lis et al., 2007). There is also a lack of effective methods to assess the influence of prolonged seated tasks on spine loading or muscle fatigue, two outcomes that are likely to reflect or influence low back pain. To address these needs, three laboratory studies were completed to: 1) develop methods that can be used to detect muscle contractile status responses in the lumbar extensors, as part of a longer-term goal of detecting muscle fatigue during prolonged sitting; 2) develop and evaluate a biomechanical model to assess lumbosacral loads in a seated posture; and 3) quantify the effects of prolonged sitting and psychosocial task demands on spinal loading and muscle fatigue.

Muscle contractile status

This work (Chapter 2) developed and assessed a new method to evaluate the status of the lumbar extensor muscles during prolonged sitting conditions. A muscle stimulation protocol was developed to measure stimulation responses in the lumbar extensors, specifically those muscles potentially fatigued during prolonged seated tasks. Results indicated that the protocol gives measures of muscle status with high levels of reliability in a sitting posture. Further, a specific stimulation protocol was recommended that includes one conditioning train along with three 16-second stimulation train at 2 Hz.

For the torso muscles, EMG-based muscle fatigue measurements are typically used. As there are debates and inconsistencies regarding the ability of traditional EMG-based fatigue method at detecting muscle fatigue during low-level sustained muscle activity, the developed stimulation-based fatigue method has the potential to resolve these issues by detecting low-level sustained muscle activity and related muscle fatigue during long-term exposures. This study successfully applied muscle stimulation method to detect muscle contractile status in trunk extensors, which could be further used as a method to detect muscle fatigue during prolonged sitting.

Assessing lumbosacral loads during sitting

To facilitate assessments of seated tasks and time-dependent variations in lumbosacral loading during sitting, a new electromyography (EMG)-based biomechanical model of the trunk was developed. Through calibration procedures, several subject-invariant model parameters were identified, and which had a good correspondence with values reported in earlier studies. Reasonable levels of correspondence were found between measured and predicted lumbosacral moments under a range of seated tasks, supporting the predictive ability of the model. Model outputs may be of use as quantitative descriptions of physical demands and injury risks during more realistic task conditions.

This study provided an innovative way to assess internal loads during prolonged seated tasks. Due to the fact that the capability of the spinal structure to resist prolonged loading is influenced by muscle fatigue and the viscoelastic properties of the disc, ligaments, and muscle tissues (Li et al., 1995), it is important to model potential time-dependent changes of spine structures such as during prolonged seated tasks. Therefore, a 3D EMG-based spine model was developed, by incorporating a trunk stiffness component to account for the viscoelastic changes of the spine, with a goal of provide a more precise estimation of spine kinetics as well as time-dependent spinal changes during prolonged seated tasks.

Effects of prolonged sitting and psychosocial stress

Prolonged sitting was found to increase trunk flexion angles, likely due to viscoelastic deformation, and to lead to muscle fatigue, even though at low level muscle contractions (<10% MVC), likely as a result of prolonged low-level muscle contractions. Psychosocial stress, though, did not modify the influence of prolonged sitting. Relatively weak correlations were found between subjective and objective measures, though muscle fatigue was positively associated with perceived body discomfort, suggesting the former is at least a component of the latter. A good level of consistency was been found between two fatigue measurement methods, supporting the value of both approaches.

This study provided a quantitative assessment on the potential role of prolonged sitting and psychosocial task demands on the pathway from exposures to the biomechanical risk factors to the development of LBP. While the main effects of prolonged sitting and psychosocial task demands have been investigated independently, the combined effects have received little attention. This study quantitatively examined the pathway between psychosocial task stress and physical/physiological implications during prolonged sitting. Furthermore, this study also provided a more detailed assessment of the correspondence between biomechanics (i.e., trunk kinematics and LMF) and perceptual responses (i.e., perceived discomfort and fatigue).

Future research directions

First, there are several limitations of this work that should be addressed in the future research. During the development of the muscle stimulation method, there was difficulty in identifying the exact target muscle group that has been stimulated. Future work is needed to quantify the potential effects of various trunk muscle activities on measured stimulation responses, and such results can help in identification the most effective location of the stimulation electrodes. In addition, during muscle stimulation, the potential coexistence of muscle stimulation and voluntary contraction may have a net effect on measured stimulation responses. Further work should be completed to identify the effects of trunk posture on measured stimulation responses, and these results can be used to account for the effects of postural changes on measured stimulation responses. Regarding EMG model development, model prediction capability may be limited based on collection of data only from a set of symmetric tasks completed in an unsupported sitting condition. Future work is needed to evaluate the model performance under more complex task conditions, that include a wider range of trunk movements, and a better representation of interaction between the human body and a sitting device. In addition, the contributions of viscoelastic components and muscle fatigue gain on model performance were not fully evaluated. Further testing is needed to assess the potential benefits of the viscoelastic component and muscle fatigue gain, such as during longer sitting periods and including more complex fatiguing task conditions. In terms of evaluating the effects of prolonged sitting and psychosocial stress on the low back, this work involved a relatively constrained condition, including a restricted sitting position and single, relatively short exposure duration. Future work could be completed to test the effects of the prolonged sitting and psychosocial stress under more realistic task condition with several exposure durations.

Second, to bring the current research to practice, a few application-oriented research directions should be considered. The current work provided a quantitative assessment of biomechanical exposures associated with seated tasks. By incorporating more direct measures of time-dependent spinal loads, the results from this work could contribute to refining existing ergonomic sitting or chair design guidelines, to account for the effects of time-dependent changes in muscle and spine structure on sitting behaviors. In addition, existing evidence has shown conflicting results regarding the causal role of sitting on LBP development (Cholewicki and McGill, 1996; Lis et al., 2007; Ramond et al., 2011). For example, some studies have reported that people working in a sitting posture may have a lower risk of LBP due to relative low physical requirement in sitting (Levangie, 1999; Vingard et al., 2000), while other studies have indicated that exposure to sitting, especially along with other moderating factors, such as, long duration, work stress, may lead to more severe LBP (Lee et al., 2001; Lis et al., 2007). Therefore, by estimating time-dependent spinal load change in a seated posture, the results from this work could further help identify more precisely the effects of sitting on the development of LBP.

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Appendix

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants

in Research Projects Involving Human Subjects

Title of Project: Influence of Prolonged Sitting and Psychosocial Stress on Lumbar Spine Loads, Discomfort, and Muscle Fatigue

Investigator:

Bochen Jia – (540) 239-7581 – Department of Industrial and Systems Engineering
Maury A. Nussbaum, Ph.D. – (540) 231-6053 – Department of Industrial and Systems Engineering

Purpose

The purposes of this proposed project are, in the context of sitting, to quantify low back loads and localized muscle fatigue development, and to assess the effects of different seated tasks and a novel seating intervention.

Procedures

It is important for you to understand that we are not evaluating you or your performance in any way. You are helping us to collect data that will be used to estimate the physical demands along the presented tasks and improve the design of a seating intervention. Any tasks you perform, or opinions you have will only help us do a better job in this evaluation and design. Therefore, we ask that you perform normally and be as honest as possible. The information and feedback that you provide is very important to this project. The total experiment time will be approximately 3.5 hours.

During the course of this experiment you will be asked to perform the following tasks:

1) Read and sign an Informed Consent Form (this form).

2) Allow several measures to be obtained from you (e.g. height, weight, leg length, etc.).

3) Allow experimenters to place electromyography (EMG) and neuromuscular electrical stimulation (NMES) electrodes over low back region and place constrain harness about 10th thoracic vertebra (T10) level. Depending on your preference, a same- or opposite-gender experimenter will be applying the electrodes. The purpose and procedures involved with these are explained below.

4) Be instructed in seated tasks and techniques.

5) Complete a set of seated tasks, with and without using the prototype seating intervention device.

6) Complete customized questionnaire regarding body comfort and discomfort.

This experiment will take place within the Safety Engineering Laboratory, in the Department of Industrial and Systems Engineering. We may apply low frequency (<5Hz) artificial neuromuscular electrical stimulation to your low back muscles. This involves relatively brief and controlled electrical stimulation, but note that the applied stimulations are at relatively low levels. In addition, we may also apply quick but small perturbations to record the stiffness of your spine. Our experiment will quantify the physical demands at the lower back during prolonged seated tasks, with and without an intervention.

In order to estimation the physical demands at the lower back, we need to measure several things while you perform the tasks. We will monitor muscle activities through electromyography (EMG). This procedure involves measuring the electrical activity given off by muscles when they contract (similar to an EKG for your heart). While you do the seated tasks, we will also measure the position and orientation of you torso, using markers that we attach to your skin or clothing (using double sided tape). In addition, we will have you sit on a pressure mapping system that will record interface pressure at your buttocks position. Using these data, and the measures we take of your height and weight, we will use a biomechanical model to estimate the loads on your lower back and the presence of muscle fatigue.

Risks and Benefits

There are minimal risks to you as a participant in this study as follows.

1) You may experience minor muscle strain as a result of performing the experimental tasks.
2) You may experience some muscle soreness, 1-2 days after the experiment.
3) You may experience some discomfort as a results of the electrical stimulation.

Participants in a study are considered volunteers, regardless of whether they receive payment for their participation. Under Commonwealth of Virginia law, workers compensation does not apply to volunteers. Appropriate health insurance is strongly recommended to cover these types of expenses.

This research project will help quantify the effects of prolonged seated work and a new chair design, which if effective may benefit people who sit for long periods (e.g., at work). While this research may yield such benefits, no promise or guarantee of benefits will be made to participants. Participants may contact the investigators listed at the end of Consent Form to inquire about the results and conclusions of this research.

Extent of Anonymity and Confidentiality

Your personal information and identity will be kept in the strictest of confidence. No names will appear on questionnaires or surveys, and a coding system will be used to associate your identity with questionnaire answers and data. The list associating names with answers will be destroyed one month after completion of data collection. Photographing might occur for assisting in the assessment of postures. However, any images used in documentation will have your face blacked out to maintain confidentiality. All information will be collected in a file and locked when not being used, and only the investigators have access to the data. It is possible that the Institutional Review Board (IRB) may view this study's collected data for auditing purposes. The IRB is responsible for the oversight of the protection of human subjects involved in research.

Informed Consent

You will receive two informed consent forms to be signed before beginning the experiment; one for your record and one for the experimenter's record.

Compensation

You will be compensated for your participation at a rate of $10 per hour. Compensation will be limited to time spent in the experimental session (e.g., you will not be compensated for your travel to or from the study). Your total payment will vary, depending on the duration required, but the total compensation will be approximately $35.

Freedom to Withdraw

You are free to withdraw from this study at any time without penalty or reason stated, and no penalty or withholding of compensation will occur for doing so. If you choose to withdraw, you will be compensated for the portion of time of the study for which you participated. Furthermore, you are free not to answer any question or respond to experimental situations without penalty. There may be circumstances under which the investigator may determine that the experiment should not be continued. In this case, you will be compensated for the portion of the project completed.

Approval of Research

The Department of Industrial and Systems Engineering has approved this research, as well as the Institutional Review Board (IRB) for Research Involving Human Participants at Virginia Tech.

Participant's Acknowledgments

Check in the box if the statement is true:

☐ I have U.S citizenship.
☐ I am not under the influence of alcohol or drugs.

☐ I have no current or recent (past year) musculoskeletal problems (the experimenter will discuss this with you).

Participant's Responsibilities

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

1. To read and understand the aforementioned instructions
2. To answer questions, surveys, etc. honestly and to the best of my ability
3. Be aware that I am free to ask questions at any point time

Participant's Permission

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

If I participate, I reserve the right to withdraw at any time without penalty. I agree to abide by the responsibilities noted above, to the best of my ability, or to inform the investigators if I am unable to comply with these.

Participant’s Signature Date

Experimenter’s Signature Date

Signature Page

I have read the description of this study and understand the nature of the research and my rights as a participant. I hereby consent to participate with the understanding that I may discontinue participation at any time if I choose to do so.

Participant's Signature Date

Printed Name

The research team for this experiment is led by Dr. Maury Nussbaum. He may be contacted at the following address and phone number:

Dr. Maury A. Nussbaum

Professor

Department of Industrial and Systems Engineering

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In addition, if you have any detailed questions regarding your rights as participant in University Research, you may contact the following individual:

Dr. David Moore

Chair, Virginia Tech Institutional Review Board

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