LEG MUSCLE ACTIVATION PATTERNS DURING SIT-TO-STAND ON VARIOUS COMPLIANCE SURFACES

A thesis submitted in partial fulfillment of the requirements
For the degree of Master of Science in
Kinesiology

By
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California State University, Northridge
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ABSTRACT

LEG MUSCLE ACTIVATION PATTERNS DURING SIT-TO-STAND ON VARIOUS COMPLIANCE SURFACES

By

Vadim Zaika
Master of Science in Kinesiology

BACKGROUND: Previous studies have compared electromyographic (EMG) activity patterns of leg muscles on stable versus unstable supporting surfaces. This research represents a novel approach because the surface stability was modified throughout eight different stiffness conditions ranging from very soft/unstable surface to very firm/hard surface.

OBJECTIVE: To analyze EMG activity patterns during sit-to-stand (STS) movement on various compliance surfaces.

HYPOTHESIS: The main hypothesis was to look if there were any differences in lower limb muscle activation patterns on various compliance surfaces during sit-to-stand task. A secondary hypothesis was to look if there were any significant bilateral differences in muscle activation between left and right legs on various compliance surfaces.

SETTING: Research was conducted at biomechanical lab at California State University, Northridge (CSUN). Supporting surface stiffness was modified with four eight-way
adjustable-stiffness shock absorbers mounted between two force plates. A custom built platform was also used in this experiment.

PARTICIPANTS: Seventeen CSUN student volunteers were recruited randomly by a pool of volunteers.

EXPERIMENTAL PROCEDURE: Each participant performed a self-paced STS under eight surface stiffness conditions in randomized order.

MEASURES: The mean EMG values of ten muscles during STS on various compliance surfaces were analyzed and compared according to eight distinct phases of force profile.

RESULTS: There were no statistically significant differences found in mean EMG of the muscles examined under different supporting surface stiffness conditions. Right and left leg muscle activities were highly correlated in all phases and not significantly different.

DISCUSSION: It was found that muscle activation patterns during sit-to-stand do not significantly change with variations of the surface compliance, suggesting that a sit-to-stand movement skill is preprogramed in a consistent way under all conditions. Regardless of the surface compliance left and right leg muscles were highly synchronized showing strong synergies. This study suggests that there were no neuromuscular adaptations to difference compliance surfaces in STS, but it is suggested to investigate whether long adaptation periods on a softer compliance surface environment will create any different results.
INTRODUCTION

Training on variable compliant surfaces has been increasingly popular in training of athletes, clinical rehabilitation, and injury prevention programs. A great variety of equipment, such as inflatable balls, foam mats, wobble boards and balance discs has been developed (Yaggie, & Campbell, 2006) for an unstable surface training (UST) (Cressey, West, Tiberio, Kraemer, & Maresh, 2007). It has been advocated that muscle activation patterns change upon changing of the supporting surface stiffness (Tung, Yang, Lee, & Wang, 2010). It is also suggested that UST promotes increased muscle activity (Imai, Kaneoka, Okubo, Shiina, Tatsumura, Izumi, & Shiraki, 2010), and develops balance and mobility at the same time by improving joint stability and rate of force development (Gollhofer, 2003).

As expected, training on various compliance surfaces is being commonly used to improve balance, coordination, and neuromuscular recruitment (Maior, Simão, Freitas de Salles, Miranda, & Costa, 2009), and is particularly valued in clinical rehabilitation settings (Cressey et al., 2007). Still, the research studies pertaining to the effects of UST on muscle activity have been producing mixed results (Lehman, 2007), causing a certain extent of disagreement regarding its application and benefits. Thus, the effects of an unstable surface training on muscle activity remain uncertain (Maior et al., 2009).

Previous studies have examined muscle activity patterns on various compliant surfaces (Dolbow, Gibson, Nguyen, Robertson, Sells, & Voight, 2008; Imai et al., 2010; Lehman, 2007; Maior et al., 2009; Tung et al., 2010), but not during sit-to-stand task. Notably, standing up from a seated position is one of the most common and essential
everyday tasks (Tung et al., 2010), and ability to stand up independently is a first step in retrieval of functional capacity for people with neuromuscular disorders (Mulroy et al., 2010). Cheng, Chen, Wang, and Hong (2004) observed EMG activity of the lower limb muscles during STS task, but not on various compliance surfaces. Findings of studies associated with UST were not consistent (Cressey et al., 2007; Dolbow et al., 2008; Lehman, 2007), suggesting that it is still unclear how the compliant support surface may alter muscle activity (Lehman, 2007; Maior et al., 2009). It should be acknowledged that previous studies have only examined and compared muscle activity patterns on stable versus unstable supporting surfaces. To our knowledge, there was no previous research done that have investigated muscle activation patterns surface that varies across a wide range of stiffness values by manually changing rigidity of the supporting surface.

The purpose of this research project was to analyze leg muscle activation patterns during sit-to-stand task on various compliant surfaces. A new apparatus with four variable stiffness shock-absorbers was constructed to create an experimental model where the researchers could establish a baseline of average responses that may be utilized in future studies to expand on the research topic. A secondary purpose was to examine if there is any bilateral muscle activation patterns.

It was hypothesized that executing sit-to-stand on various compliance surfaces would result in significantly different limb muscles activation patterns, specifically that more compliant surface would result in significantly greater leg muscle activation. For the secondary purpose it was hypothesized that there would be no significant differences in muscle activity between left and right legs, and both legs stiffness will be similarly adjusted from the central nervous system.
Dolbow et al. (2008) examined muscles activation timing of vastus lateralis (VL) and vastus medialis oblique (VMO) during free squats on stable and unstable surfaces. Nineteen female and six male participants were recruited from a pool of volunteers at Southeastern University. A medical history questionnaire was used to screen the applicants, and those with lower quarter injuries or surgeries within the past year were excluded from the study. Dolbow et al. (2008) identified two independent variables: the quadriceps muscle (VMO versus VL), and the type of surface (stable versus unstable). The relative time of muscle activation was defined as a dependent variable.

In this experiment Dolbow et al. (2008) utilized a surface EMG machine and an analog-to-digital computer board to collect and process EMG signal. An electronic goniometer attached to the dominant leg was used to record beginning of the movement and a joint angle. Dyna-discs (round air inflated discs) were utilized as an unstable support surface. Participants warmed up for five minutes on a Biodex cycle ergometer before starting the experiment. Dolbow et al. (2008) had the participants flip a coin to determine the order of testing on different surfaces. EMG electrodes were attached to the skin over the VMO and VL of the participant's dominant leg, which was determined by kicking the ball.

Participants completed three free squats on each type of surface, bending the knees to a forty five degrees angle, and taking thirty second breaks between repetitions and 2-minute break when switching between the surfaces. As a safety precaution, spotters were used during the squats on Dyna-discs. EMG measurements were taken while the
participant was standing upright (baseline), and at the beginning of the muscle contraction upon initiation of a squat movement detected by goniometer. EMG data was averaged for all three measurements, and for each surface. A repeated two-way analysis of variance (ANOVA) with Bonferroni adjustment was used to test for the differences in activation timing between VMO and VL on both surfaces. A significance level was set at p < 0.05.

Dolbow et al. (2008) found that both muscles were firing significantly sooner during squats on an unstable support surface. However, there was no significant difference in activation timing between the VMO and VL observed, which did not correspond with previous reports. Dolbow et al. (2008) proposed that a delay in VMO activation on an unstable surface could occur due to the spotters’ assistance to the participants that were getting ready for squatting on an unstable surface. Dolbow et al. (2008) explained that because of spotter assistance, muscles could have continued to stay relaxed for some time, causing an activation delay.

Lehman (2007) reviewed six recent case studies, and, similar to Dolbow et al. (2008), also found unpredicted muscle responses to the exercising on unstable versus stable supporting surface. According to Lehman (2007), EMG increase upon replacing a stable support surface with an unstable one has not been observed in some studies. Lehman (2007) pointed out that during one of the experiments replacing a chair with an exercise ball did not produce increase in muscle activation. In the course of another experiment a reduction in the abdominal muscle activity during trunk extension on a stability ball versus trunk extension on a floor was observed. Lehman (2007) concluded that surface instability did not cause a neuromuscular system overload for some
individuals. Therefore, exercising on an unstable versus a stable surface did not always result into increased muscle activity.

Cressey et al. (2007) proposed that unstable surface training increases activity of the antagonist muscles, and suggested that UST may be detrimental to the force production and dynamic balance development. Cressey et al. (2007) conducted an intervention study to determine the effectiveness of a ten week lower-body UST on jumping, sprinting, and agility performance of nineteen Division 1 intercollegiate soccer players. Ten athletes participated in the intervention program, and nine soccer players formed a control group. Cressey et al. (2007) predicted significant changes in jumping test performance, sprinting speed and agility in participants who underwent lower-body unstable surface training in comparison to those who did not.

All athletes continued their regular training regimen, but the experimental group also underwent a ten-week supplemental lower-body UST program. An unstable surface workout consisted of squats, lunges, and dead lifts performed while standing on the inflatable rubber discs similar to the Dyna-discs used by Dolbow et al. (2008). To keep training load volume the same between both groups, Cressey et al. (2007) included a supplemental workout in the control group’s training regimen, but it was carried out on a stable supporting surface. Jumping, sprinting, and agility assessments were administered upon completion of the intervention group’s program.

Results were analyzed using IBM Statistical Product and Service Solutions (SPSS) 10.0 statistical software. The UST group demonstrated significant improvements in sprinting and agility, but did not make any significant development in jumping. On
contrary, the control group displayed significant progress in all tested areas, and showed
significantly greater improvements in jumping and sprinting in comparison to the
experimental group. Cressey et al. (2007) proposed that UST may not be useful and may
be even detrimental for strength training, and should only be applied in sport-specific
context. Nonetheless, Cressey et al. (2007) emphasized that rehabilitation value of the
unstable surface training should not be underestimated.

Imai et al. (2010) explained that the influence of surface stability on muscle
activation may vary, depending on the particular muscle or exercise. In a controlled
laboratory research, Imai et al. (2010) examined trunk muscle activity on various
compliance surfaces throughout five commonly used lumbar stabilization exercises. Nine
healthy male adults (mean age 24 years) without prior experience in trunk stabilization
training participated in this experiment. Candidates with lumbar spine or neurological
disorders were excluded from the study. Each participant performed elbow-toe, hand-
knee, curl-up, side-bridge, and back-bridge exercises on stable and unstable support
surfaces. Depending on the exercise, Bosu ball or a swiss ball were utilized to create an
unstable support surface.

One of the outstanding facts about this study is that Imai et al. (2010) used
intramuscular electrodes for recording of EMG signals. This is quite uncommon,
considering that most of the researchers use surface electrodes. To make intramuscular
electrodes, Imai et al. (2010) threaded the wire through a hypodermic needle, and bent
the tips into the hooks. EMG activity of erector spinae (ES), external obliques (EO),
lumbar multifidus (MF), rectus abdominis (RA), and transversus abdominis (TrA) was
collected during dynamic and isometric phases of each exercise. Participants executed a
dynamic phase of the exercise at their own pace, and maintained an isometric phase for three seconds. Imai et al. (2010) calculated the EMG amplitude root-mean-square for a 1-second window of the isometric phase and normalized EMG data as the percentage of the maximum voluntary contraction (MVC). Muscle activity during each exercise on a stable surface was compared to the muscle activity on an unstable surface.

Imai et al. (2010) observed significantly higher muscle activity during the elbow-toe exercise on the unstable versus the stable surface for all muscles examined. Muscle activity during hand-knee technique on an unstable surface was also significantly greater for the RA, EO, and for the ES located on the opposite side to the arm lifted. With regard to the side bridge, Imai et al. (2010) recorded significantly higher activity of the RA on an unstable surface. Remarkably, in the course of performing the curl-up on an unstable surface, activity of the EO was significantly greater but activity of the TrA was significantly lower. On the other hand, there was no significant difference in muscle activity during the back-bridge exercise on an unstable versus stable support surface. Imai et al. (2010) concluded that trunk muscle activity (especially the global trunk muscles) was increased during rehabilitative lumbar stabilization exercises on an unstable support surface, with the exception of the back bridge exercise, which agreed with Lehman (2007).

Maior et al. (2009) suggested that incorporating unstable surface training into a joint rehabilitation and neuromuscular conditioning program is essential. Maior et al. (2009) conducted a study to compare activity of the quadriceps group muscles in male adults during a squat exercise on stable and unstable surfaces. However, unlike Imai et al. (2010) who used intramuscular electrodes, Maior et al. (2009) utilized a lot surface
EMG. During this research the EMG measurements of vastus lateralis (VL), vastus medialis (VM), and rectus femoris (RF) were recorded.

Twenty healthy males (more than twice that were used by Imai et al. (2010) that had a minimum of 12 months resistance training experience participated in the study. Maior et al. (2009) used a Physical Activity Readiness Questionnaire (PAR-Q) as an applicant screening device. In contrast to other studies described in this paper, Maior et al. (2009) organized six training sessions over a two week period so participants could get used to the experimental settings and procedures. Similar to Dolbow et al. (2008), Maior et al. (2009) established the dominant side of each participant by kicking leg preference. Maior et al. (2009) used a Smith Machine for the squat exercise, and Reebok Core Board Trainer was utilized as an unstable support surface. This was different from an experiment by Dolbow et al. (2008) where a free squat technique and the air-filled Dyna-discs were used. Maior et al. (2009) measured EMG of VL, RF, and VM of the dominant leg was during squats on unstable and stable surfaces.

Similar to the results of the previous study, increase in the muscle activity on an unstable surface was observed. Moreover, Maior et al. (2009) found significantly greater activation for all muscles during squats performed on an unstable versus a stable surface condition. Precisely, EMG of the RF was greater by eighteen percent, VL by 21 %, and VM by 16 %. Maior et al. (2009) proposed that increased muscle activity on unstable supporting surface was due to the bilateral contractions of the lower body, and similar to Gollhofer (2003) suggested utilizing an unstable platform during functional, coordination, and balance training.
Balance training has been successfully used in rehabilitation settings to improve neuromuscular conditioning and aid in recovery from an array of sensorimotor disorders including foot drop and ankle instability (McKeon, Ingersoll, Kerrigan, Saliba, Bennett, & Hertel, 2008). Yaggie et al. (2006) explained that balance training develops proprioceptive control, aiding in rehabilitation of unstable or injured joints. McKeon et al. (2008) conducted an intervention research study to investigate the effects of a four-week balance training program on static and dynamic postural control in people with chronic ankle instability (CAI). 31 physically active individuals of both genders with more than one instance of ankle sprain resulted in chronic ankle instability (unlike previously reviewed research studies) participated in the research. People with history of lower extremity injuries, surgeries, or neurological disorders affecting balance within last six months were excluded.

Similar to Cressey et al. (2007), all participants were randomly split into two groups. The group of six males and ten females underwent balance training, and six males and nine females formed a control group. Balance workout program consisted of 12 twenty-minute balance training sessions administered for the period of four weeks. Progressive balance training exercises included hop to stabilization, hop to stabilization and reach, hop to stabilization box drill, progressive single-limb stance balance activities with eyes open, and progressive single-limb stance activities with eyes closed. After four weeks, participants were tested for balance improvement. Foot and Ankle Disability Index (FADI) and the FADI Sport surveys (Martin, 2007) were used for self-report assessments; force plate measured center of pressure (COP) and time-to-boundary (TTB) analysis was used to examine participants’ static postural control, and a Star Excursion
Balance Test (SEBT) to assess their dynamic postural control before and after intervention program.

McKeon et al. (2008) found no significant differences in the pretest assessment between the groups. Conversely, upon completion of the training program the experimental group demonstrated significantly better balance in most measures compare to their pre-test scores as well as the post-test results of the control group. Specifically, the experimental group post-test results of the FADI and FADI Sport surveys, COP and TTB with eyes closed, and SEBT were significantly higher. Based on these results, McKeon et al. (2008) concluded that progressive balance training significantly improves sensorimotor capabilities, and is beneficial in recovery from neuromuscular disorders.

In people post-stroke, there tends to be altered responses of the neuromuscular system negatively affects muscle activation ability (Bradley, Hart, Mandana, Flowers, Riches, & Sanderson, 1998) and results in a number of sensory, motor and psychological dysfunctions (Hamzat & Peters, 2009; American Stroke Association (ASA)). Compromised muscle activation sequence following stroke often results into co-contraction of agonist and antagonist muscles (Kirker, Jenner, Simpson, & Wing, 2000). Hemiparesis is one of the most common post-stroke disabilities (National Stroke Association (NSA) associated with motor-sensory deficits (McEwen, Polatajko, Huijbregts, & Ryan, 2009) that negatively affect activities of daily living (ADL) (Campos, Barroso, & Menezes, 2010).

In people with post-stroke hemiparesis, the ability to stand up independently is often compromised by poor balance, postural instability (Britton, Harris, & Turton, 2008).
and impaired muscle activation patterns (Tung et al., 2010). Cheng et al. (2004) examined leg muscle activation patterns in people with post-stroke hemiparesis during sit-to-stand task. Cheng et al. (2004) were able to recruit 70 participants for this research, which is considerably more compared to the number of people that participated in the previously reviewed experiments. Medically stable individuals with post-stroke hemiparesis, including those with the history of falls, were engaged in this study. All experimental group participants were able to stand and walk independently, but had to wear ankle-foot orthoses for walking or carrying out daily activities. Applicants with previous traumatic brain injury, stroke, orthopedic or peripheral lower limb nerve injury, or other neurologic deficiency were excluded from this research. A control group consisted of 20 able-body healthy adults.

Cheng et al. (2004) utilized a backless, height-adjustable chair for a sit-to-stand task. An AMTI force platform, a multiple channel surface EMG and a personal computer was used for data collection and processing. Simultaneous collection of EMG and kinetic data was achieved by using a synchronization switch. During the experiment dressed in shorts, barefoot participants placed one foot on each force plate in parallel, and stood up from a chair adjusted to his/her knee height at a comfortable speed. Cheng et al. (2004) split the sit-to-stand movement into three segments according to participants’ body position and deviation of the exerted force: initial, executive, and standing. Kinetic measurements and EMG data of bilateral quadriceps (QUA), medial hamstring (HAM), tibialis anterior (TA), and soleus (SOL) were recorded throughout all three phases during three trials, and averaged for further investigation.
A statistical analysis included descriptive analysis to describe the types and characteristics of muscle activation patterns, analysis for categorical variables, and analysis of variance for continuous variables to compare results of the experimental and control groups. Cheng et al. (2004) also used post hoc comparison to determine the origin of the significant differences, and a paired t-test to compare the differences between affected and unaffected limbs of the post-stroke participants.

Cheng et al. (2004) found compromised muscle recruitment and great variability of muscle activation patterns in the affected leg of the post-stroke participants. Most of the post-stroke participants subjected to falling demonstrated significant reduction in motor output or interrupted motor-unit activation. In addition, activation of the TA in the affected limb of the post-stroke participants was significantly delayed, but SOL and HAM muscles on the contrary fired prematurely. Cheng et al. (2004) also noted that muscle activation patterns varied widely among the post-stroke participants. Cheng et al. (2004) proposed that people post-stroke who exhibit no or low activation of the TA and premature or excessive activation of the SOL, were predisposed to falling.
METHODS

Seventeen participants, 10 males and 7 females (see Table 1) were recruited randomly from a convenient sample of volunteer students attending classes at California State University, Northridge (CSUN). All participants completed an informed consent form approved by the human subjects review board at California State University, Northridge (CSUN). Individuals who had an injury within a previous year as well as those unable to stand up from a chair independently were excluded. It was confirmed that potential participants did not have any major injury within the last year.

All procedures were performed in the biomechanics lab at CSUN in accordance with a protocol approved by the human subjects review board. Participants were dressed in shorts and barefoot. A 16 channel Delsys Myomonitor system (Delsys, Boston, MA) was used to collect the data. EMG sensors were attached to skin over five leg muscles: rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA), and gastrocnemius (GA) on both right and left sides. A tri-axial accelerometer was affixed to the head. The EMG electrodes and the accelerometer were connected to a wireless transmitter mounted to the waist belt. The height of the chair was adjusted individually for each participant according to knee height. Two Kistler force plates (Kistler Group, Winterthur, Switzerland) were used to measure ground reaction forces. A seven camera MX Vicon system (Vicon-Peak, Oxford, UK) was used to collect the kinematic data. Two personal computers were utilized to acquire and process experimental data. A Custom-built platform was used for experimental setup (see Illustration 1).
Stiffness of the supporting surface was modified by changing the settings of four eight-way adjustable Enidine shock absorbers (ITT - Enidine Inc., Orchard Park, NY, USA) fitted between the force plates (see Illustration 2). Shocks were randomly adjusted between eight different settings, ranging from “1” (very soft - unstable surface) to “8” (very firm - hard surface). Each participant had to perform sit-to-stand under all eight different surface stiffness conditions in a randomized design order. Participants knew that they had to execute a sit-to-stand under different surface conditions ranging from soft to hard, but they did not know the exact condition during the data collection (one of eight different stiffness settings). The purpose for the randomization was to avoid the bias response from a participant that could alter his or her reaction to the changed supporting surface stiffness condition. This was explained to each participant prior to the experiment. After all adjustments were made, the tested participant placed his/her feet on the force plate in parallel. Then he or she was instructed to rise up from a chair at a self-paced comfortable speed upon looking into an LED light that were generated by the computer during the data collection process (see Illustration 3).

During the experiment muscle activity patterns in healthy adults were examined during standing up from a chair with feet planted on various stiffness surfaces. Surface stiffness was an independent variable, and the mean EMG amplitude was a dependent variable. Synchronized data collection made it possible to combine and analyze EMG and kinetic data. Mean EMG values of ten leg muscles (five on left and five on right) were collected twice from all participants under eight different surface stiffness conditions. Only one trial was used during the analysis. Each EMG signal was rectified and processed with a fourth order Butterworth linear envelope. For each muscle mean
Illustration 1. Custom-built platform.
Illustration 2. Adjustable shock absorbers mounted between the force plates.
Illustration 3. Participant is ready for the experiment.
EMG values were split and analyzed under eight distinct phases of the sit-to-stand force profile (see Figure 1).

Data were analyzed with two statistical methods using IBM SPSS Statistics 19 software. To test the first hypothesis, a repeated measures ANOVAs were used to analyze 80 dependent variables (eight phases times 10 muscles) across eight different surface stiffness conditions (independent variable). For the second hypothesis, paired t-tests compared left and right muscles under each condition. A total of 320 t-tests were used to analyze EMG activity of the five muscles throughout eight different supporting surface stiffness conditions during eight different phases of the STS force profile. Significance level was set at $p<0.05$. 
Figure 1. Sit-to-stand force phase diagram for top and bottom force plates. Force1 – top force plate, Force2 – bottom force plate. A is force onset, B is minimum reaction force, C is point where forces are positive again, D is point where positive impulse balances with negative impulse from A to C, E is when maximum force is generated, F is where local minimum Force is calculated during sway, G is the start of balance and steady state. Phase 1 is time span from A to B points, phase 2 is time span from B to C points, phase 3 is time span from A to C points, phase 4 is time span from C to D points, phase 5 is time span from D to E points, phase 6 is time span from E to F points, phase 7 is time span from F to G points, and phase 8 is time span from C to E.
RESULTS

The descriptive data of the 17 participants that included ten males and seven females (mean age 23.18 years old, mean height 170.18 cm, mean weight 73.03 kg) are given in Table 1.

The 80 (eight phases by ten muscles) repeated measures ANOVA revealed no statistically significant difference in mean EMG activity of the muscles in the eight support surface stiffness conditions (p > .05). Upon examination of Wilks’ lambda multivariate tests it was determined that F values within subjects were not significantly different (for example see Table 2). Occasionally, there were some statistical significant differences among the eight stiffness conditions when the SPSS procedures were at the level of Repeated Simple Comparisons (condition 8 with condition 1, condition 8 with condition 2 etc.). However, never those comparisons were coupled with an overall multivariate F statistic significant result. Thus, those comparisons were considered statistically random, and it was expected that at an alpha level of 0.05 with eighty statistical procedures to have four procedures being statistically significant by chance.

Further data analysis consisted of 320 paired t-tests between five left and five right leg muscles under eight different surface stiffness conditions within each of the eight phases of individual force profiles. The results of the paired t-tests showed that 242 out of the 320 Pearson’s r values were statistically significant correlated (see an example at Table 3-a, b, c, d and e, and Figure 2). Furthermore, there were no statistically significant differences found between the right and left leg muscle activation throughout all phases with the exception of some instances (see Table 4 and Figure 2).
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<td>172.72</td>
<td>106.59</td>
<td>35.56</td>
<td>19.05</td>
</tr>
<tr>
<td>11</td>
<td>m</td>
<td>21</td>
<td>185.42</td>
<td>81.65</td>
<td>27.94</td>
<td>19.05</td>
</tr>
<tr>
<td>12</td>
<td>f</td>
<td>21</td>
<td>170.18</td>
<td>58.06</td>
<td>30.48</td>
<td>17.78</td>
</tr>
<tr>
<td>13</td>
<td>f</td>
<td>32</td>
<td>160.02</td>
<td>58.51</td>
<td>27.94</td>
<td>17.78</td>
</tr>
<tr>
<td>14</td>
<td>m</td>
<td>22</td>
<td>182.88</td>
<td>74.84</td>
<td>27.94</td>
<td>17.78</td>
</tr>
<tr>
<td>15</td>
<td>f</td>
<td>34</td>
<td>160.02</td>
<td>72.57</td>
<td>30.48</td>
<td>15.24</td>
</tr>
<tr>
<td>16</td>
<td>f</td>
<td>19</td>
<td>167.64</td>
<td>59.87</td>
<td>27.94</td>
<td>16.51</td>
</tr>
<tr>
<td>17</td>
<td>f</td>
<td>21</td>
<td>157.48</td>
<td>58.97</td>
<td>25.40</td>
<td>16.51</td>
</tr>
<tr>
<td>Mean</td>
<td></td>
<td>23.18</td>
<td>170.18</td>
<td>73.03</td>
<td>28.35</td>
<td>18.03</td>
</tr>
</tbody>
</table>
Table 2. Example of F values corresponding with Wilks’ lambda multivariate tests results for the left tibialis anterior, showing that none of the dependent variables on each force phase were significant different because of the eight different stiffness conditions.

<table>
<thead>
<tr>
<th>Force Phase</th>
<th>F</th>
<th>Probability</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.484</td>
<td>.093</td>
</tr>
<tr>
<td>2</td>
<td>2.322</td>
<td>.110</td>
</tr>
<tr>
<td>3</td>
<td>.382</td>
<td>.893</td>
</tr>
<tr>
<td>4</td>
<td>1.734</td>
<td>.207</td>
</tr>
<tr>
<td>5</td>
<td>.548</td>
<td>.781</td>
</tr>
<tr>
<td>6</td>
<td>1.772</td>
<td>.199</td>
</tr>
<tr>
<td>7</td>
<td>.464</td>
<td>.840</td>
</tr>
<tr>
<td>8</td>
<td>2.772</td>
<td>.070</td>
</tr>
</tbody>
</table>
Table 3-a. Paired samples Pearson’s correlation between right and left legs for rectus femoris under surface stiffness condition eight (8) for force phases one through eight.

<table>
<thead>
<tr>
<th>Paired Correlations</th>
<th>Pearson’s $r$</th>
<th>Probability ($p$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus Femoris Left-Right phase 1</td>
<td>0.671</td>
<td>0.003</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 2</td>
<td>0.705</td>
<td>0.002</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 3</td>
<td>0.682</td>
<td>0.003</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 4</td>
<td>0.581</td>
<td>0.014</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 5</td>
<td>0.761</td>
<td>0</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 6</td>
<td>0.677</td>
<td>0.003</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 7</td>
<td>0.862</td>
<td>0</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 8</td>
<td>0.774</td>
<td>0</td>
</tr>
</tbody>
</table>
Table 3-b. Paired samples Pearson’s correlation between right and left legs for tibialis anterior under surface stiffness condition eight (8) for force phases one through eight.

<table>
<thead>
<tr>
<th>Paired Correlations</th>
<th>Pearson’s $r$</th>
<th>Probability ($p$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibialis Anterior Left-Right phase 1</td>
<td>.697</td>
<td>.002</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 2</td>
<td>.730</td>
<td>.001</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 3</td>
<td>.597</td>
<td>.011</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 4</td>
<td>.856</td>
<td>.000</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 5</td>
<td>.916</td>
<td>.000</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 6</td>
<td>.948</td>
<td>.000</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 7</td>
<td>.951</td>
<td>.000</td>
</tr>
<tr>
<td>Tibialis Anterior Left-Right phase 8</td>
<td>.970</td>
<td>.000</td>
</tr>
</tbody>
</table>
**Table 3-c.** Paired samples Pearson’s correlation between right and left legs for gastrocnemius under surface stiffness condition eight (8) for force phases one through eight.

<table>
<thead>
<tr>
<th>Paired Correlations</th>
<th>Pearson’s $r$</th>
<th>Probability ($p$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gastrocnemius Left-Right phase 1</td>
<td>.323</td>
<td>.206</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 2</td>
<td>.310</td>
<td>.225</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 3</td>
<td>.225</td>
<td>.386</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 4</td>
<td>.277</td>
<td>.282</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 5</td>
<td>.225</td>
<td>.385</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 6</td>
<td>.416</td>
<td>.097</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 7</td>
<td>.818</td>
<td>.000</td>
</tr>
<tr>
<td>Gastrocnemius Left-Right phase 8</td>
<td>.561</td>
<td>.019</td>
</tr>
</tbody>
</table>
Table 3-d. Paired samples Pearson’s correlation between right and left legs for vastus lateralis under surface stiffness condition eight (8) for force phases one through eight.

<table>
<thead>
<tr>
<th>Paired Correlations</th>
<th>Pearson’s $r$</th>
<th>Probability ($p$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vastus Lateralis Left-Right phase 1</td>
<td>.976</td>
<td>.000</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 2</td>
<td>.984</td>
<td>.000</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 3</td>
<td>.958</td>
<td>.000</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 4</td>
<td>.173</td>
<td>.508</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 5</td>
<td>.291</td>
<td>.258</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 6</td>
<td>.396</td>
<td>.115</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 7</td>
<td>.720</td>
<td>.001</td>
</tr>
<tr>
<td>Vastus Lateralis Left-Right phase 8</td>
<td>.635</td>
<td>.006</td>
</tr>
</tbody>
</table>
Table 3-e. Paired samples Pearson’s correlation between right and left legs for biceps femoris under surface stiffness condition eight (8) for force phases one through eight.

<table>
<thead>
<tr>
<th>Paired Correlations</th>
<th>Pearson’s $r$</th>
<th>Probability ($p$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps Femoris Left-Right phase 1</td>
<td>.728</td>
<td>.001</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 2</td>
<td>.604</td>
<td>.010</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 3</td>
<td>.709</td>
<td>.001</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 4</td>
<td>.760</td>
<td>.000</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 5</td>
<td>.689</td>
<td>.002</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 6</td>
<td>.699</td>
<td>.002</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 7</td>
<td>.753</td>
<td>.000</td>
</tr>
<tr>
<td>Biceps Femoris Left-Right phase 8</td>
<td>.328</td>
<td>.199</td>
</tr>
</tbody>
</table>
Figure 2. Average right and left rectus femoris activation patterns, across the eight phases of the STS force profile when socks are set at stiffness condition eight, showing statistically significant correlations between left and right sides, and no statistically significant differences. Abbreviations: LRF is left rectus femoris, and RRF is right rectus femoris.
Table 4. Paired samples t-test between right and left legs for rectus femoris under surface stiffness condition eight (8) for force phases one through eight.

<table>
<thead>
<tr>
<th>Paired-Sample t-test</th>
<th>t</th>
<th>Probability</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus Femoris Left-Right phase 1</td>
<td>-1.04</td>
<td>.314</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 2</td>
<td>-1.097</td>
<td>.289</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 3</td>
<td>-0.86</td>
<td>.402</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 4</td>
<td>-0.973</td>
<td>.345</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 5</td>
<td>-0.173</td>
<td>.865</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 6</td>
<td>0.503</td>
<td>.622</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 7</td>
<td>-0.101</td>
<td>.921</td>
</tr>
<tr>
<td>Rectus Femoris Left-Right phase 8</td>
<td>0.914</td>
<td>.374</td>
</tr>
</tbody>
</table>
There were 313 not statistically significant differences and 7 statistically significant differences between left and right muscle comparisons. Statistically there were expected 16 comparisons (5 percent of 320 comparisons) to show significance. The 7 statistical differences did not have any pattern under any condition or any force curve phase, so there were considered statistically random.

The same statistical results were replicated when the signals were normalized with Maximum Volunteer Contractions (MVCs) that were collected during data acquisition. There were no statistically significant differences when either raw or normalized integrated EMGs were analyzed during the eight force phases. A preliminary statistical analysis of the right and left leg muscles onset firing has shown that most muscles was firing in synchronicity, and close proximity of the EMG onset across legs was typically observed (see Figure 3).

The segmental movement sequence was observed quantitively and qualitatively and it had some predictable patterns for all participants. Head and shoulder markers were the first to move and then the tibialis anterior muscles were activated. The next benchmark was the force onset as registered from the force plate followed by EMG onset of the rest of the muscles. A typical firing order is illustrated in Figure 4.
Figure 3. Synchronization of the onset firing for the left and right gastrocnemius.

Point A - stimulus on LED light mean activation (muscle tone) before the stimulus was calculated as a baseline. Point B - EMG onset calculated as a point where the signal exceeds the mean EMG signal by three standard deviations.
Figure 4. Typically observed muscle activity pattern, where tibialis anterior (LTiBANt) EMG onset is ahead of force production (FP2Z), and bicep femoris (RBicFem) and gastrocnemius (RGastro) muscles EMG onsets are following FP2Z.
DISCUSSION

The purpose of this research was to analyze muscle activation patterns during sit-to-stand task on surfaces with various stiffness values, and to establish a baseline of average healthy-adult responses for future studies with people with neuromuscular problems (e.g. stroke). It was hypothesized that executing a sit-to-stand on surfaces with increased stiffness would significantly increase lower limb muscles activation patterns. A secondary hypothesis was that there would be significant bilateral muscle activity correlations. During the study muscle activation patterns of biceps femoris, rectus femoris, vastus lateralis, tibialis anterior, and gastrocnemius were examined in a wide range of the surface compliance. The surface compliance was modified by changing the settings of four eight-way adjustable shock absorbers fitted between the force plates.

Results of this research show that executing STS on more compliant (less stable) surfaces does not induce significantly greater leg muscles activation and that the muscle activation patterns during STS do not significantly change with variation of the surface compliance. There were no differences between right and left muscles but muscle activation was highly correlated showing a robust synchronicity between homologous muscles. Regardless of the value of the stiffness setting of the support surface, the muscle of activity of the left and right leg muscles were strongly correlated, even in conditions where the body’s momentum may have created instabilities between left and right kinematics.

These findings contradict with results of the previous research conducted by Maior et al. (2009) who found significantly greater leg muscle activation during squats
performed on an unstable versus a stable surface. This was most likely because Maior et al. (2009) examined the muscle activity during repetitive squats, thus neuromuscular system had some time to adapt. Additionally the current study used shock absorbers that change the stiffness in the vertical dimension of movement and not in three-dimensions as Maior et al. (2009).

Furthermore, to create an unstable surface Maior et al. (2009) used a Reebok Core Board Trainer, so the participants were continuously under reduced stability conditions. Contrary, during the current experiment the surface stiffness was randomly adjusted between eight different conditions unknown to the participants. Hence, the participants could not alter their response to the supporting surface stiffness condition from the start, but it had to be a reactionary effect, which was not statistically significant.

It seems that STS is a gross body movement and neuromuscular activation was predetermined from the start of the movement. Another postulation was that neuromuscular reactions may have considerable delay in relation to the perceived stiffness changes.

On the contrary, results of the current study support the outcome of the research conducted by Dolbow et al. (2008) who observed no significant changes in firing pattern of VMO and VL on stable versus unstable surface (Dyna-discs). The outcome also corresponds with conclusions of Lehman (2007), who reviewed six recent case studies and suggested that training on an unstable surface does not always result into increased muscle activity. Findings also agree with findings of Imai et al. (2010), supporting the
proposition that influence of the surface stability on muscle activation may vary, depending on the particular muscle or exercise.

Although McKeon et al. (2008) concluded that progressive balance training on unstable surface significantly improves sensorimotor capabilities, and is beneficial in recovery from neuromuscular disorders; the current study could find a direct connection between changes in vertical surface compliance and participants’ motor response. Thus, the no statistically significant effects maybe related in the single dimensionality of the experiment that was restricted only in the vertical dimension.

On another issue Cheng et al. (2004) also noted that muscle activation patterns varied widely among post-stroke participants and the increased variability may influence the statistical results. During the current study the participants were healthy college students and the bilateral correlations were statistically significant pointing to consistent firing of the muscle and not a random effect. If there were no correlation between the left and right muscles the results may the result of random muscle firing however this is not the case in the current study.

The study represents a continuation of a research paradigm that builds in a significant aspect of biomechanical research of muscle responses under different surface compliance conditions. The custom made apparatus with force plates and adjustable-stiffness shock absorbers that was used in this investigation is unique because it enables a researcher to manually modify surface stiffness throughout eight different settings. This is a novel approach and allows to measure muscle responses under multiple conditions.
The differences on the current study with studies that observed some differences (Maior et. all, 2009) may rely on the fact that the other studies looked only two conditions.

The findings of this study show when gross movement such as sit-to-stand is executed on different compliance surfaces muscle activation patterns do not change. The results suggest that an average person should not have a specific response under variable compliance conditions if the person does not know the stiffness condition of the surface. If a difference in the persons’ responses is observed the alternative hypothesis will be that the specific person does not fit the normal curve responses.

Still, it is important to acknowledge that the study had a limited number of participants and the results are delimited to physically active college students. Hence, further research involving much larger number of participants and more diverse population is necessary. Additionally to clarify the issue more study is warranted in the area of adaptation, and whether participants were acclimatized over a longer period of time with multiple repetitions the results maybe different and the muscle activity may be different. Another interesting study could be to measure the muscular activity of participants during drop jumps, with the delimitation of one-dimensional stiffness change.

The current results converge on the conclusion that for gross everyday movements when compliance conditions are unknown participants will not alter their muscle activation. There is a strong bilateral correlation between homologous muscles and this association is not altered under different compliance conditions in any of the phases of the measured ground reaction forces.
REFERENCES


APPENDIX A

Bibliography


You are being asked to participate in a research study. Participation in this study is completely voluntary. Please read the information below and ask questions about anything that you do not understand before deciding if you want to participate. A researcher listed below will be available to answer your questions.

**RESEARCH TEAM**

**Researcher:**
Vadim Zaika  
Department of Kinesiology  
Telephone Number (818) 677-3205

**Faculty Advisor:**
Professor Konstantinos Vrongistinos, Ph.D.  
Department of Kinesiology  
18111 Nordhoff St.  
Northridge, CA 91330-8287

**PURPOSE OF STUDY**

The purpose of this research study is to analyze electromyographic activity patterns during sit-to-stand task on various compliance surfaces.

**SUBJECTS**

**Inclusion Requirements**

You are eligible to participate in this study if you are between 18 and 69 years old.

**Exclusion Requirements**

You are not eligible to participate in this study if you are unable to stand up from a chair independently or if you have any major injury within the last year.

**Time Commitment**

This study will involve approximately 2 hours of your time.
PROCEDURES
The following procedures will occur:

All procedures are going to be performed in accordance with a protocol approved by the Human Subject Committee of California State University, Northridge. During this experiment all participants undergo the same protocol.
- You will be dressed in shorts and barefoot.
- Electromyography (EMG) sensors will be affixed to your skin to monitor five muscles. It will take about 30 to 45 minutes to prepare and connect all sensors. Data from sensors will be transmitted to the data collection computer with a wireless transmitter.
- You will sit on the chair on the platform. There is a force plate that monitors the forces you put while you will move from sit-to-stand position. Under the force plate there are some shock absorbers that sometimes will be in a soft position and sometimes in a hard position
- You will execute sit-to-stand eight times under different conditions. Sometimes the surface will be softer and sometimes it will be harder. You may need to repeat some trial if the data collection is not correct.
- The height of the chair will be adjusted to make it more comfortable to execute the sit to stand.
- Your feet should be placed in a comfortable position but they should be on top of the force plate.
- The researcher will instruct to be ready for the data collection and you should execute the sit to stand position as soon as you see the light in front of you is on at a self-paced comfortable speed.
- It will take about 30 to 45 minutes to collect all data.

RISKS AND DISCOMFORTS

This study involves no more than minimal risk. There are no known harms or discomforts associated with this study beyond those encountered in normal daily life. However, you may have discomfort while attaching the EMG electrodes to your skin. You may have a minimal irritation or discomfort when the electrodes are removed but this usually disappears after a short period of time, unless you are allergic to some of the gel of the EMG electrodes

A sit-to-stand is an activity of daily living. Nonetheless, if it is not executed correctly it can result into a fall or you may feel dizzy while executing it.

BENEFITS
Subject Benefits

You will not directly benefit from participation in this study.

Benefits to Others or Society

The proposed research findings will help us understand how people move in soft and in hard surfaces. Some of these data will be used to identify differences in people with stroke.

ALTERNATIVES TO PARTICIPATION

This is a descriptive study and no benefits are expected for any participant.

COMPENSATION, COSTS AND REIMBURSEMENT

Compensation for Participation
You will not be paid or in general compensated in any form for your participation in this research study.

Costs

There is no cost to you for participation in this study.

You will be responsible for any health costs if you are injured during the experiment.

Reimbursement

You will not be reimbursed for any out of pocket expenses, such as parking or transportation fees.

WITHDRAWAL OR TERMINATION FROM THE STUDY AND CONSEQUENCES

You are free to withdraw from this study at any time. Withdrawing from the study will not jeopardize any relation you may have with California State University, Northridge. If you decide to withdraw from this study after you participated you should notify the research team immediately, so your data can be excluded from the study. The research team may also end your participation in this study if you do not follow instructions, miss scheduled visits, or if your safety and welfare are at risk.

CONFIDENTIALITY

Subject Identifiable Data:

All identifiable information that will be collected about you will be removed at the end of data collection and will be replaced with a code.

Data Storage

- All research data will be stored on a computer that is password protected and without identifying personal information.
- All research data will be stored in a locked file cabinet in RE174. There will be no data identifying your personal information
- Initially the [video recordings] will also be stored in a locked file cabinet in RE174; then analyzed and erased as soon as possible.
- The video recordings will be converted into electronic files that have only the position of your segments while you are moving.
- All research data will be destroyed three years (or the time required from the research journal) after the study is published.

Data Access

Only the researcher and faculty advisor named on the first page of this form will have access to your study records. Any information derived from this research project that personally identifies you will not be voluntarily released or disclosed without your separate consent, except as specifically required by law. Publications and/or presentations that result from this study will not include identifiable information about you.

Data Retention
• All research data will be destroyed three years (or the time required from the research journal) after the study is published. Archived data will not include any identifiable information about you.

IF YOU HAVE QUESTIONS
If you have any comments, concerns, or questions regarding the conduct of this research please contact the research team listed on the first page of this form.

If you are unable to reach a member of the research team listed on the first page of the form and have general questions, or you have concerns or complaints about the research study, research team, or questions about your rights as a research subject, please contact Research and Sponsored Projects, 18111 Nordhoff Street, California State University, Northridge, Northridge, CA 91330-8232, or phone 818-677-2901.

VOLUNTARY PARTICIPATION STATEMENT
You should not sign this form unless you have read it and been given a copy of it to keep. Participation in this study is voluntary. You may refuse to answer any question or discontinue your involvement at any time without penalty or loss of benefits to which you might otherwise be entitled. Your decision will not affect your future relationship with California State University, Northridge. Your signature below indicates that you have read the information in this consent form and have had a chance to ask any questions that you have about the study.

I agree to participate in the study.

Subject Signature _______________________________ Date __________________

Printed Name of Subject ________________________________

Researcher Signature _______________________________ Date __________________

Printed Name of Researcher ________________________________
EXPERIMENTAL SUBJECTS
BILL OF RIGHTS

The rights below are the rights of every person who is asked to be in a research study. As an experimental subject I have the following rights:

1) To be told what the study is trying to find out,

2) To be told what will happen to me and whether any of the procedures, drugs, or devices is different from what would be used in standard practice,

3) To be told about the frequent and/or important risks, side effects or discomforts of the things that will happen to me for research purposes,

4) To be told if I can expect any benefit from participating, and, if so, what the benefit might be,

5) To be told the other choices I have and how they may be better or worse than being in the study,

6) To be allowed to ask any questions concerning the study both before agreeing to be involved and during the course of the study,

7) To be told what sort of medical treatment (if needed) is available if any complications arise,

8) To refuse to participate at all or to change my mind about participation after the study is started. This decision will not affect my right to receive the care I would receive if I were not in the study.

9) To receive a copy of the signed and dated consent form.

10) To be free of pressure when considering whether I wish to agree to be in the study.

If I have other questions I should ask the researcher or the research assistant, or contact Research and Sponsored Projects, California State University, Northridge, 18111 Nordhoff Street, Northridge, CA 91330-8232, or phone (818) 677-2901.

X ____________________________
Signature of Subject            Date

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