

# The impact of stochastic resonance electrical stimulation and knee sleeve on impulsive loading and muscle co-contraction during gait in knee osteoarthritis

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## ABSTRACT

**Background:** Increased impulsive loading and muscle co-contraction during gait have been observed in individuals with knee osteoarthritis. Proprioceptive deficits in this population may contribute to these effects. Proprioception has been shown to improve with the combination of stochastic resonance electrical stimulation and a knee sleeve in knee osteoarthritis. Our goal was to determine whether stochastic resonance stimulation combined with a knee sleeve would decrease impulsive loading rates and muscle co-contraction during gait in knee osteoarthritis.

**Methods:** Gait kinetics, kinematics and muscle activity were assessed during walking in subjects with knee osteoarthritis during three different conditions: no stochastic resonance/no sleeve (control), stochastic resonance at 75% threshold/sleeve, and no stochastic resonance/sleeve. Loading rates were calculated from the ground reaction force. Muscle co-contraction was calculated from the ratio of vastus lateralis to lateral hamstring activity. Differences between conditions were assessed using a repeated measures analysis of variance ( $P < 0.05$ ).

**Findings:** The 75% threshold/sleeve and sleeve only conditions resulted in increased knee flexion at contact and reduced loading rates compared to the control condition ( $P < 0.05$ ). However, these measures did not significantly differ between the 75% threshold/sleeve and sleeve only conditions. Muscle co-contraction was found to decrease with the 75% threshold/sleeve condition compared to the other conditions.

**Interpretation:** Increased knee flexion and decreased loading rates may be a result of proprioceptive improvements resulting from the sleeve or sleeve/stimulation combination. The stochastic resonance stimulation did not demonstrate an ability to enhance the effects of the sleeve with the exception of reductions in muscle co-contraction.

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## 1. Introduction

Osteoarthritis (OA) is a common joint disorder affecting roughly 27 million people in the US (Helmick et al., 2008) and significantly contributes to increasing health care costs (Kotlarz et al., 2009). The disorder involves chronic breakdown of cartilage within a joint, which is associated with risk factors for development that include joint injury, obesity, and repetitive joint stress. More specifically, alterations in the mechanical environment of the knee joint due to the breakdown of cartilage can adversely affect load distribution, resulting in abnormal wear within the joint and further breakdown of cartilage. Reduced range of knee flexion and heightened muscular co-

contraction during the loading phase of gait are mechanical hallmarks of those with knee OA and together they represent what is known as the “stiffened knee response” (Childs et al., 2004; Schmitt and Rudolph, 2007).

Repetitive impulsive loading, a reflection of the peak force and the force immediately following ground contact, otherwise known as the heel strike transient (HST), is another mechanical factor that may play a role in the progression of knee OA (Collins and Whittle, 1989). Animal studies investigating the effects of repetitive loading have demonstrated that microfractures are present in the trabecular bone of rabbits when subjected to repetitive loading (Radin et al., 1973) and that greater cartilage fissuring results from the same magnitude impact loads applied at higher loading rates (Ewers et al., 2002). Subsequently, bone remodeling produces stiffening of the subchondral bone, thus minimizing its ability to absorb impact forces resulting in joint degeneration.

Deficits in proprioception, which is defined as the perception of limb position and movement within space (Lephart et al., 1998), could

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be the cause of ineffective muscle activation resulting in elevated impulsive loading. Also, these deficits may cause a sense of instability resulting in increased muscular co-contraction as a way to restabilize the joint, but at the expense of increasing compressive stresses across the joint. Correcting these proprioceptive deficits through a phenomenon known as stochastic resonance (SR) may help slow disease progression by decreasing impulsive loading and improper muscle activation. SR is a concept in which low-level noise improves a given system's sensitivity to weak stimuli. Somatosensory application of subsensory SR stimulation has demonstrated improvements in tactile sensation (Collins et al., 1996), muscle spindle output (Cordo et al., 1996), balance control (Gravelle et al., 2002; Priplata et al., 2005), and joint position sense (Collins et al., 2009). Previous work has shown that a knee sleeve/brace can improve proprioception (Birmingham et al., 1998; Birmingham et al., 2001; Herrington, 2005). Thus, by combining SR stimulation with a sleeve, greater improvements in proprioception may result. By enhancing the sensitivity of one's sensory system, proprioceptive improvements may alter gait, resulting in reduced joint loading, thus possibly delaying onset and/or slowing progression of OA.

The purpose of this study was to determine whether the application of SR electrical stimulation combined with a knee sleeve could decrease the HST and ground reaction force (GRF) loading rate as well as decrease the muscle co-contraction activity occurring during gait in subjects with OA of the knee. We hypothesized that the HST and GRF loading rates would be reduced at ground contact and muscular co-contraction between the hamstrings and quadriceps groups would decrease during the weight acceptance phase of gait with the application of a knee sleeve and further decrease with SR application.

## 2. Methods

### 2.1. Subjects

Following approval by the Institutional Review Board, 52 (30 females, 22 males) patients 40 years or older with doubtful to moderate (Kellgren–Lawrence KL grades 1–3) medial knee OA confirmed by physician's diagnosis volunteered to participate. Subjects who had a Body Mass Index (BMI) of 35 or greater, who had a previously diagnosed neurological condition, used a pacemaker or other implanted electronic device, had a diagnosis of musculoskeletal disease other than their knee OA, or had a lower extremity joint replacement were excluded. Subjects who were using an assistive walking device, or who had received a corticosteroid injection within 3 months prior to screening were also excluded. Subject demographics as well as self reported pain, stiffness, functionality, and instability measures are reported (Table 1). Of the 52 subjects, 28 had grade 3 OA, 11 had grade 2, and 13 had grade 1 OA. Twenty-four subjects indicated that they experienced no episodes of instability while only three subjects indicated that they were severely affected with daily episodes of instability.

**Table 1**

Mean (SD) demographics as well as subject reported pain, stiffness, functionality, and instability measures for all test subjects.

	Male (n = 22)	Female (n = 30)	Total (n = 52)
Age	58.6 (10.9)	63.0 (8.3)	61.2 (9.6)
Weight (kg.)	91.9 (12.4)	72.9 (12.7)	80.9 (15.7)
Height (cm.)	178.2 (8.1)	164.7 (6.4)	170.4 (9.8)
BMI	29.0 (4.1)	26.8 (4.2)	27.8 (4.3)
WOMAC Index (pain)	4.0 (4.1)	4.2 (3.0)	4.1 (3.4)
(Stiffness)	3.0 (2.1)	2.6 (1.6)	2.7 (1.8)
(Function)	12.7 (12.7)	12.2 (9.3)	12.4 (10.8)
WOMAC aggregate	19.6 (18.1)	19.0 (13.2)	19.2 (15.3)
Self reported instability (part A)	3.5 (1.3)	4.0 (1.3)	3.8 (1.3)
Self reported instability (part B)	6.2 (14.0)	10.6 (23.8)	8.7 (20.2)

Standing anterior–posterior radiographs taken with the knee in full extension were assessed by an orthopedist to determine knee OA severity based on a modified Kellgren–Lawrence (KL) grading system (Felson et al., 1995; Felson et al., 1997). Joint space narrowing was ensured to be greater on the medial side by visual inspection of standing radiographs. Each subject's more severely affected knee, excluding knees with grade 4 OA, was chosen for testing and in instances where both knees were equally affected, the subject's dominant knee was tested.

### 2.2. Study design

Kinetic, kinematic, and electromyographic (EMG) measures were recorded while subjects performed a 10-meter walk down a level platform located in the Sports Medicine Research Laboratory at UNC Chapel Hill. Each subject's threshold for detecting the SR stimulation was determined prior to gait analysis and a level of 75% of their threshold for detection was used as the strength of the SR stimulation in the experiment. Our level of 75% of threshold was chosen to be in line with previous studies that have demonstrated improvements in postural control with SR. Specifically, Reeves et al. applied SR in the general range of 25 to 90% of threshold (Reeves et al., 2009), while Priplata et al. identified 75% of threshold as an optimal level for improving postural control (Priplata et al., 2006). Additionally, our previous study investigating SR's effects on joint position sense in those with knee OA applied a level of 60% of threshold and no improvements were observed (Collins et al., 2010). During gait analysis, subjects were presented with four conditions in the following sequence: (1) no electrical stimulation and no sleeve (control1 NE:NS1); counterbalance design of two treatment conditions: (2,3) no stimulation/sleeve (NE:S), and 75% of threshold stimulation/sleeve (E75:S); (4) followed by a no stimulation/no sleeve condition (control2 NE:NS2). The two treatment conditions were presented in a counterbalanced sequence design in order to control for any lasting effects of the electrical stimulation. Additionally, any effects across the testing session were assessed by comparing measures in the two control conditions placed before and after the treatment conditions.

### 2.3. Testing protocol

After explanation of the study procedures and the associated risks, informed consent was obtained from each subject. Subjects then completed a self-reported measure of knee instability questionnaire adapted from the Knee Outcome Survey "Activities of Daily Living Scale" (Felson et al., 2007; Fitzgerald et al., 2004; Irrgang et al., 1998). The questionnaire asked each subject to rate his/her instability (0 to 5 scale) by answering the question, "To what degree does giving way, buckling, or shifting of the knee affect your level of daily activity?" and indicating how many times he/she had experienced instability within 3 months prior to testing. Each subject then completed the Knee and Osteoarthritis Outcome Score (KOOS) survey, which contains subscales of the WOMAC index, specifically knee stiffness, pain, and physical function, and also includes an overall quality of life measure (Roos et al., 1998). Five valid gait trials within each of the four testing conditions were collected with a valid trial defined as one in which the subject correctly landed on the force plate with no variation in stride length. Data collection commenced 3 s prior to ground contact and continued 2 s after contact.

### 2.4. Kinematic, kinetic, and EMG data collection

Subjects were instructed to walk at a self-selected "fast" pace with the foot of their test limb landing on a nonconductive force plate (model 4060nc, Bertec Corp., Columbus, OH, USA). Three electromagnetic position sensors (Flock of Birds, Ascension Technology Corp., Burlington, VT, USA) were placed on the sacrum, thigh and shank of

the test limb taking care to place them in areas of minimal subcutaneous tissue in order to minimize motion artifacts. The knee and ankle joint centers were defined as the midpoints between the digitized medial and lateral femoral condyles and medial and lateral malleoli, respectively. The hip joint center was determined using Leardini's method (Leardini et al., 1999). Knee joint angles were determined as the motion of the tibial reference frame relative to the femoral reference frame where flexion–extension was about the y-axis, valgus–varus was about the x-axis, and internal–external rotation about the z-axis. Walking speed was measured using an infrared timing system (Sparq XLR8 Digital Timing System, Nike, USA) to ensure walking speed did not vary by more than 10% between trials. In addition, mean forward velocity was calculated from the displacement of the sacral position sensor.

Surface electromyography (EMG) electrodes with preamplifiers (Delsys Inc., Boston, MA, USA) were placed on the vastus lateralis (VL), semitendinosus as a representative of medial hamstring (MH), and biceps femoral long head as a representative of lateral hamstring (LH) to determine electrical activity of these muscles. Electrodes were placed parallel to the muscle fibers over the longitudinal midline of the muscle belly. A common reference electrode was placed over the posterior aspect of the ipsilateral wrist.

### 2.5. SR stimulation and sleeve

Two pairs of surface SR electrodes designed to deliver the electrical stimulation via an electrical stimulator device (Afferent Corporation, Providence, RI, USA) were placed on the inferior and superior aspects of the knee joint line. Electrodes were placed approximately 2 cm above and below the joint line as measured from the joint line to electrode pad circumference. Each pair consisted of one electrode placed medial to the joint centroid and one lateral in order to create an alternating flow of current in the medial–lateral direction. Stimulation consisted of a Gaussian white noise signal (zero mean, 0–1000 Hz bandwidth) that was 75% of the subject's threshold level for detection determined prior to testing. SR threshold level for detection was determined by asking subjects to indicate at which amplitude they detected the presence of the electrical stimulation. SR electrodes remained in place during all testing conditions and subjects were blinded as to when the stimulation was applied. The electrodes were wired into an electrical stimulator device positioned in a fixed location. Wires connecting the electrodes were secured with straps along the subject's test leg and waist and exited the subject at a common location in order to minimize cable tension and interference with gait.

Subjects also wore a neoprene knee sleeve during the no electrical stimulation/sleeve (NE:S) and stimulation/sleeve (E75:S) treatment conditions. The sleeve was fit based on the girth of the test limb's thigh measured approximately 4 in. above the patella center per manufacturer's recommendations (Safe-T-Sport Model# 37–350, FLA Orthopaedics Inc., Miramar, FL, USA).

### 2.6. Data reduction

Kinematic, kinetic and EMG data acquisitions were synchronized using the Motion Monitor motion capture system (Innovative Sports Training, Chicago, IL, USA). All kinetic and SEMG data were collected at 1440 Hz while kinematic data were sampled at 144 Hz. Kinematic data were then filtered through a 4th order, zero lag Butterworth filter at a cutoff frequency of 6 Hz. Data were reported during the three phases of gait: preparatory phase (100 ms interval prior to initial ground contact through initial ground contact), weight acceptance phase (period from initial contact to peak knee flexion in first half of stance), and midstance/terminal (period from peak knee flexion to toe-off). Initial ground contact and toe-off were defined as the times at which the ground reaction force increased above 10 N and de-

creased below 10 N, respectively. Kinematic outcome measures include knee flexion angle at ground contact and forward velocity of subject's center of mass. Kinetic outcome measures include ground reaction forces in the anterior–posterior (x), medial–lateral (y), and superior–inferior (z) directions, which were acquired unfiltered and normalized to subject's body weight (N). Loading rate measures were calculated from the vertical ground reaction force (Fz) over increasing time domains in the following manner: 1.) Fz LR max (BW/s): the maximum slope from the 1st derivative of a 4th order polynomial fit between the point of initial ground contact and the peak heel strike transient (HST) 2.) Fz LR to HST (BW/s): the linear slope between the point of initial ground contact and the peak HST (Fz HST) and 3.) Fz LR to Peak (BW/s): the linear slope between the point of initial ground contact and the overall peak of the vertical ground reaction force (Fz Peak) (Fig. 1). The HST was determined by finding the first peak above 50 N of load within the ground reaction force waveform for each trial. Using the standard Waveform Peak Detection.vi within a customized Labview program (National Instruments), a quadratic least squares fit was applied along the ground reaction force waveform to identify the peak.

EMG data were bandpass filtered with a high pass cutoff frequency of 20 Hz and a low pass cutoff frequency of 450 Hz, full wave rectified, and filtered at 20 Hz using a zero lag 4th order Butterworth low pass filter to create a linear envelope EMG wave. The mean EMG amplitudes of the muscles (VL, LH) in each of the three phases of gait were calculated from the linear envelope EMG.

In order to account for any leakage of the SR stimulation to the linear envelope EMG levels, quiet trials of EMG were taken with the subject in a seated position with and without the stimulation applied. The difference in these mean linear envelope EMG levels for the quiet trials was subtracted from the filtered dynamic linear envelope EMG levels to compute corrected linear envelope EMG levels. The magnitudes of corrected EMG signals were no less than that in the quiet non-stimulated trial.

The magnitudes of corrected EMG signals were normalized to the average maximum activity of the specific muscle demonstrated during the control trials (conditions of NE:NS1 and NE:NS2). Each stance phase was normalized to 100 evenly distributed time points. Co-contraction indices between VL and LH were calculated from magnitude and time normalized EMG signals as described by Schmitt and Rudolph (Schmitt and Rudolph, 2007).

$$Index_{co-contraction} = \frac{\sum_{i=1}^{100} \left[ \frac{LowerEMG_i}{HigherEMG_i} \times (LowerEMG_i + HigherEMG_i) \right]}{100}$$

### 2.7. Statistical analysis

Paired *t*-tests were performed to compare control condition values (NE:NS1, NE:NS2) for each measure. Only the first control condition was used in subsequent statistical analysis to create three overall testing conditions (NE:NS1, NE:S, and E75:S). The second control condition values were not used in statistical analysis in order to avoid carryover effects of the stimulation and sleeve being incorporated into the control condition. Additionally, a HST was present in all but 3 trials for all subjects. Therefore, these trials were omitted during statistical analyses in order to assess differences between conditions when only a HST was present.

A one-way repeated measures analysis of variance (ANOVA) was performed to determine whether overall significant differences exist between conditions for each measure with further statistical differences between conditions assessed by the Student–Newman–Keuls posthoc method of multiple comparisons. Parametric analyses were performed and in instances where the data did not adhere to normality, non-parametric analyses were performed. Nonparametric analyses included

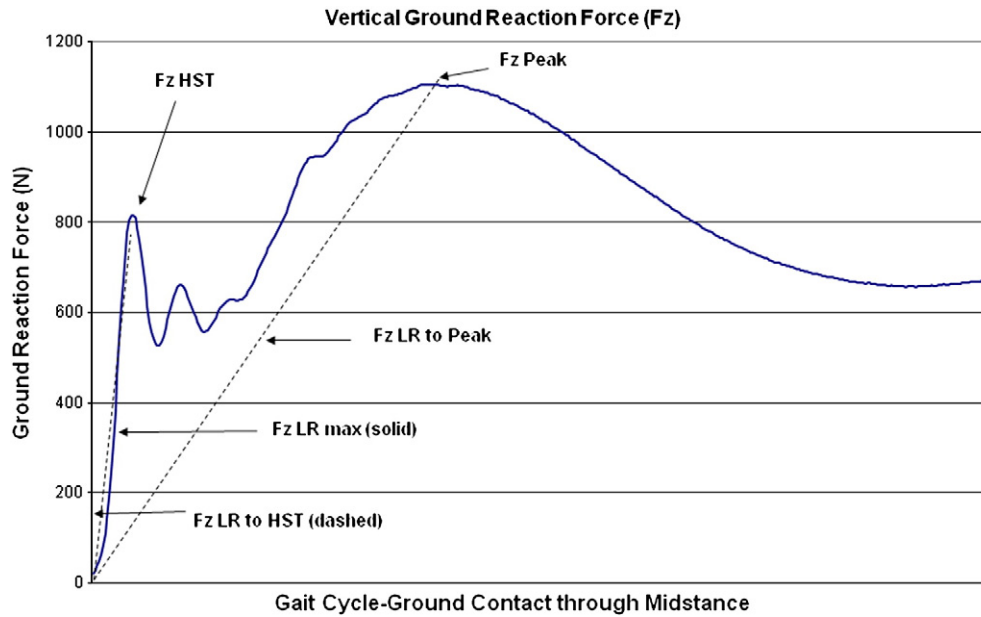


Fig. 1. Vertical component of ground reaction force demonstrating loading rate outcome measures with respect to the generated data plot from a sample test subject.

the Friedman Repeated Measures ANOVA on ranks as well as the Wilcoxon Signed Rank paired *t*-test. Statistical significance in the following results is reported from the appropriate test, either parametric or non-parametric.

A Type I error rate of 0.05 was chosen as an indication of statistical significance for each statistical test. All statistical tests were performed using SigmaPlot (Systat Software Inc., San Jose, CA, USA).

### 3. Results

#### 3.1. Loading rate parameters

The heel strike transient peak (Fz HST), and loading rates calculated from ground contact to peak HST (Fz LR to HST) and the maximum loading rate to peak HST (Fz LR max) were significantly less in the NE:S and E75:S conditions compared to the control condition ( $P < 0.05$ ) (Fig. 2 and Table 2). The overall ground reaction force peak (Fz peak), the loading rate calculated from ground contact to the overall ground reaction force peak (Fz LR to peak), and the horizontal

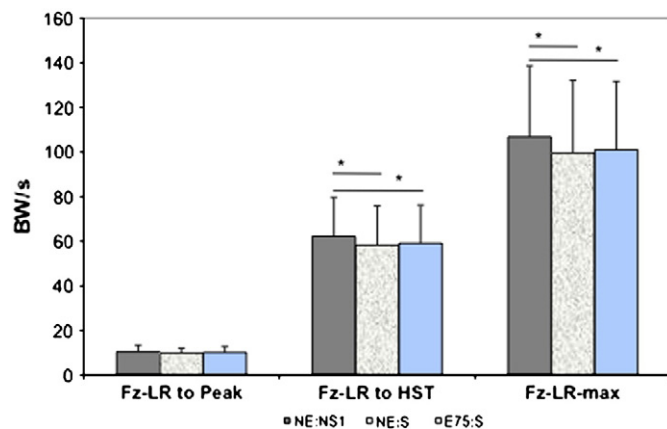


Fig. 2. Loading rate outcome measures within each of the three testing conditions. \* indicates significant differences between conditions at the end of the horizontal bars,  $P < 0.05$ .

component of the ground reaction force (Fx) were not significantly different among conditions.

#### 3.2. Kinematic parameters

The knee flexion angle at initial ground contact was significantly increased in the NE:S and E75:S conditions compared to the control condition ( $P < 0.05$ ), but not significantly different between the NE:S and E75:S conditions (Table 2). The mean forward velocity was not significantly different among conditions.

#### 3.3. VL/LH co-contraction

The index of co-contraction of the VL/LH muscles significantly decreased in the preparatory, weight acceptance, and midstance/terminal phases of gait in the NE:S and E75:S conditions compared to the control condition with further significant decreases from the NE:S to the E75:S condition ( $P < 0.05$ , Fig. 3).

#### 3.4. Control condition comparison

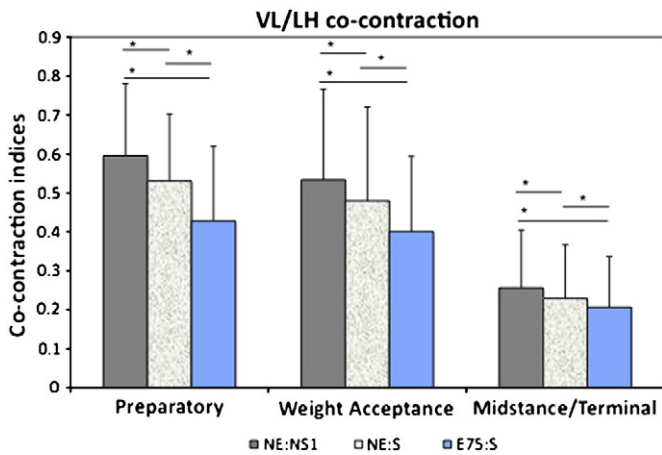
The index of co-contraction of the VL/LH muscles significantly decreased from control 1 (NE:NS1) to control 2 (NE:NS2) in the weight acceptance phase by 14.7% ( $P < 0.05$ ) and midstance/terminal phase by 9.4%. The preparatory VL/LH index was not significantly different between control 1 and control 2, decreasing by 3.7%.

Table 2

Mean (SD) kinematic and kinetic measures during gait within the three testing conditions.

Measures	NE:NS1	NE:S	E75:S
Kinematic measures			
Mean forward velocity (m/s)	1.47 (0.19)	1.47 (0.22)	1.47 (0.20)
Knee flexion at contact (deg.)	12.40 (8.28)	14.44 (7.97) <sup>†</sup>	14.67 (8.03) <sup>†</sup>
Kinetic measures			
Fz-Peak (BW)	1.16 (0.14)	1.17 (0.14)	1.18 (0.13)
Fz-HST (BW)	0.73 (0.15)	0.70 (0.16) <sup>†</sup>	0.71 (0.15) <sup>†</sup>

<sup>†</sup> Indicates significant difference compared to the control (NE:NS1).



**Fig. 3.** Co-contraction indices of the VL/LH muscle groups. \* indicates significant differences (RMANOVA) between the conditions at the end of the horizontal bars,  $P < 0.05$ .

The maximum ground reaction force (Fz-peak) significantly increased from control 1 (NE:NS1) to control 2 (NE:NS2) ( $P < 0.05$ ) by approximately 2%. However, Fz-LR to peak, Fz-LR to HST, Fz-LR max and Fz-HST were not significantly different between control conditions. The knee flexion angle at contact and mean forward velocity were not significantly different between control conditions.

#### 4. Discussion

The results demonstrated that the HST and GRF loading rates during gait significantly decreased in the NE:S and E75:S conditions compared to the NE:NS1 condition, but were not significantly different between the NE:S and E75:S conditions. These results support our hypothesis that the application of the knee sleeve would decrease the HST and GRF loading rates during gait. However, these results do not support our hypothesis that the combination of SR and the knee sleeve would further decrease the HST and GRF loading rates compared to the sleeve alone.

The results of this study also demonstrated that the index of co-contraction of VL/LH significantly decreased in the NE:S and E75:S conditions during gait, but the E75:S condition decreased further than the NE:S condition. These results support our hypothesis that the application of the knee sleeve would decrease the VL/LH co-contraction during gait. These results generally do support the hypothesis that the application of SR would further decrease co-contractions compared to the application of the knee sleeve.

Overall, the results of the present study demonstrated that the current SR stimulation configuration was limited in its ability to enhance the effects observed with a sleeve alone. It is possible that a longer duration of SR application would produce greater effects; however, SR is a novel concept with its clinical application having limited coverage in the literature and no studies have investigated longitudinal effects of long-term SR use. As a result, it is difficult to know whether the short duration of SR application in the present study is the reason for the limited effects seen.

Several biomechanical and muscular activation abnormalities are present in those with knee OA and include increased loading rates and reduced knee flexion at contact (Mundermann et al., 2005) as well as increased co-contraction of the quadriceps–hamstring muscle groups (Hortobagyi et al., 2005). A neoprene knee sleeve has proven to be effective in improving proprioception via joint position sense relative to a no sleeve, control condition in those with knee OA (Collins et al., 2010). Previous studies investigating the effects of SR stimulation have found improvements in postural sway (Gravelle et al., 2002; Priplata

et al., 2005; Ross, 2007), tactile sensation (Collins et al., 1996), and proprioception (Reeves et al., 2009).

The significance of loading rate to the overall development and progression of knee OA has previously been demonstrated in animal studies (Ewers et al., 2002; Radin et al., 1973) with higher loading rates generating more surface fissuring of cartilage than lower loading rates (Ewers et al., 2002). Our results showed significant decreases in loading rates calculated over a shorter time domain with the application of a sleeve alone and in combination with SR whereas the loading rate calculated using the overall peak ground reaction force (Fz-LR to peak) was not affected. This is most likely the result of the significant decreases observed in the peak heel strike transient (Fz HST) between the sleeve alone and control condition as well as the stimulation/sleeve and control condition. The HST is an indirect measure of the amount of load experienced at the knee during the impact at ground contact. A decrease in this measure may suggest an overall reduction of harmful load experienced at the knee. Our observed improvements in loading rate are likely a result of the increased knee flexion at ground contact, as previous studies have found an attenuation of shock loading with greater knee flexion (Lafortune et al., 1996). In addition, Mundermann et al. demonstrated that subjects with knee OA landed in a more extended knee position with a greater loading rate than age-matched, healthy control subjects (Mundermann et al., 2005). This may suggest that the observed increase in knee flexion at contact with the sleeve condition is possibly a result of proprioceptive improvements, and these improvements may result in a return of the gait pattern of OA subjects to a more normal pattern. The use of knee braces and sleeves to improve loading rate is limited to only one study in which the authors were able to reduce the loading rate at initial ground contact through the use of a knee brace designed to provide feedback to the user (Riskowski et al., 2009). However, the use of a knee sleeve or simple, non-automated knee brace to reduce impact loading in knee OA has not been investigated. In the present study, our findings demonstrated significant increases in knee flexion at contact with the sleeve alone and in combination with SR stimulation though no differences were seen between the two treatment conditions.

Decreases in muscle co-contraction with the application of SR or sleeve alone may be the result of enhanced joint position sense (JPS), and this enhanced JPS may lead to a greater sense of joint stability. Previous studies have observed an improvement in JPS with a sleeve alone and a stimulation/sleeve condition compared to a control condition (Collins et al., 2010). Those with knee OA likely demonstrate increased co-contraction of the hamstrings and quadriceps muscle groups as a way to improve the stability of the knee. However, this strategy increases joint contact pressures, which exacerbates pain and degradation of the joint, thus highlighting the importance for treatment modalities targeting decreased muscle co-contraction.

While this study presents novel information to the existing field of knowledge, it is not without limitations. Increased knee flexion observed in our study could be due to passive restraint effects of the sleeve. However, this is unlikely due to the fact that the sleeve is made of a soft, pliable material contrary to rigid unloader braces, and was fit with the knee in full extension. Our results showed improvements in loading rate as well as co-contractions, but it is unknown whether these improvements are possibly due to proprioceptive enhancing effects of the sleeve or additional effects of the sleeve such as passive tension. It is also possible that the improvements seen in loading rate could be the result of reduced pain or improved confidence. Additionally, it is possible that the SR stimulation delivered was not at an optimal level. Some subjects became confused and were not sure if they were detecting the stimulation during the threshold for detection procedure, so it is possible that their detection amplitudes were incorrect. We also acknowledge the possibility of a washout effect. Specifically, almost a third of the subjects displayed changes in loading rate in the opposite sense than what was expected between the E75:S and NE:NS1 groups. We found a similar effect for the difference between NE:S and NE:NS1 groups, which

may suggest that the effect was more related to the presence of the sleeve. This possible washout effect may have dampened any differences between the groups. Lastly, we acknowledge the possibility of a minor fatigue effect during the testing with the maximum ground reaction force significantly increasing a mere 2% from the first control condition to the second control condition. Also, VL/LH co-contraction decreases in the weight acceptance and midstance phases from the first to second control conditions of 14.7% and 9.4% respectively, may be indicative of a learning effect, but also may suggest a carryover effect of the treatment conditions.

We postulate that over time, decreases in loading rates experienced during gait may translate into improvements in functionality, and reductions in pain and stiffness. While the differences in loading rate seen in this study are small, it is unknown what differences are considered clinically significant. In addition, these differences may grow with a more challenging task such as stair descent, with fatigue, or with prolonged use of a brace. Some studies have utilized a longer time domain when calculating loading rate, which diminishes the effect of initial impact (Liikavainio et al., 2007). By considering the time to the HST peak as the time period for the loading rate calculation, a more precise assessment can be made as to the loading rate experienced at contact.

## 5. Conclusions

Based on the results of this study, we can conclude that our reductions in loading rate were likely the result of increased knee flexion at contact and decreased muscular co-contraction, possibly due to a greater sense of stability provided by the proprioceptive enhancing effects of the sleeve or stimulation/sleeve combination.

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