

SHAWN FARROKHI, PT, DPT¹ • CHRISTINE D. POLLARD, PT, PhD² • RICHARD B. SOUZA, PT, MPT³
YU-JEN CHEN, PT, MS³ • STEPHEN REISCHL, PT, DPT, OCS⁴ • CHRISTOPHER M. POWERS, PT, PhD⁵

Trunk Position Influences the Kinematics, Kinetics, and Muscle Activity of the Lead Lower Extremity During the Forward Lunge Exercise

As an integral part of any rehabilitative, preventive, or maintenance program, proper exercise prescription can facilitate improvements in musculoskeletal function by addressing the specific needs of an individual. To this end, functional weight-bearing exercises have received a significant amount of attention as the preferred mode of exercise for lower extremity strengthening.^{8,10,28} The popularity of weight-bearing

exercises stems from the fact that they closely simulate activities of daily living, while simultaneously training multiple muscle groups. Furthermore, because skeletal muscle responds to the amount and type of demand (ie, tension) which is imposed upon it, proper selection of the activities that increase the mechanical and metabolic capacity of the muscle will dictate the nature and extent of the exercise adaptation.^{17,18} Although considerable attention has been given to the biomechanics of weight-bearing exercises such as squatting^{4,7,9,15,20,21,24,27} and stepping,^{3,12,16,22,23,26} little is known about the mechanical attributes (ie, kinematics and kinetics) and muscular responses associated with the lunge exercise and its variations.^{2,6,11,14}

It has been proposed that variations of the forward lunge exercise can alter the actions of the lower extremity muscle groups.¹¹ While the forward lunge can be modified by changing the step distance of the lead lower extremity from the starting position, this exercise also can be altered by changing the position of the trunk. It is conceivable that changing the trunk position during a weight-bearing exercise will affect the mechanics of the lower extremity by shifting the location

- **STUDY DESIGN:** Experimental laboratory study.
- **OBJECTIVES:** To examine how a change in trunk position influences the kinematics, kinetics, and muscle activity of the lead lower extremity during the forward lunge exercise.
- **BACKGROUND:** Altering the position of the trunk during the forward lunge exercise is thought to affect the muscular actions of the lead lower extremity. However, no studies have compared the biomechanical differences between the traditional forward lunge and its variations.
- **METHODS AND MEASURES:** Ten healthy adults (5 males, 5 females; mean age \pm SD, 26.7 \pm 3.2 years) participated. Lower extremity kinematics, kinetics, and surface electromyographic (EMG) data were obtained while subjects performed 3 lunge exercises: normal lunge with the trunk erect (NL), lunge with the trunk forward (LTF), and lunge with trunk extension (LTE). A 1-way analysis of variance with repeated measures was used to compare lower extremity kinematics, joint impulse (area under the moment-time curve), and normalized EMG (highest

1-second window of activity for selected lower extremity muscles) among the 3 lunge conditions.

- **RESULTS:** During the LTF condition, significant increases were noted in peak hip flexion angle, hip extensor and ankle plantar flexor impulse, as well as gluteus maximus and biceps femoris EMG ($P < .015$) when compared to the NL condition. During the LTE condition, a significant increase was noted in peak ankle dorsiflexion angle and a significant decrease was noted in peak hip flexion angle ($P < .015$) compared to the NL condition.

- **CONCLUSIONS:** Performing a lunge with the trunk forward increased the hip extensor impulse and the recruitment of the hip extensors. In contrast, performing a forward lunge with the trunk extended did not alter joint impulse or activation of the lower extremity musculature.

- **LEVEL OF EVIDENCE:** Therapy, level 5. *J Orthop Sports Phys Ther* 2008;38(7):403-409. doi:10.2519/jospt.2008.2634

- **KEY WORDS:** biomechanics, EMG, impulse, weight bearing

¹Doctoral Student, Division of Biokinesiology and Physical Therapy, University of Southern California, Los Angeles, CA. ²Assistant Professor of Research Physical Therapy, Division of Biokinesiology and Physical Therapy, University of Southern California, Los Angeles, CA. ³Doctoral Candidate, Division of Biokinesiology and Physical Therapy, University of Southern California, Los Angeles, CA. ⁴Adjunct Associate Professor of Clinical Physical Therapy, Division of Biokinesiology and Physical Therapy, University of Southern California, Los Angeles, CA. ⁵Associate Professor and Co-Director, Musculoskeletal Biomechanics Research Laboratory, Division of Biokinesiology and Physical Therapy, University of Southern California, Los Angeles, CA. This study was approved by the Health Sciences Institutional Review Board, University of Southern California, Los Angeles, CA. Address correspondence to Dr Christopher M. Powers, Division of Biokinesiology and Physical Therapy, University of Southern California, 1540 E. Alcazar St, CHP-155, Los Angeles, CA 90089-9006. E-mail: powers@usc.edu

of center of mass relative to the base of support.¹⁹ For instance, a common variation of the forward lunge that has the potential to increase the action of the hip extensors is the addition of trunk flexion. Conversely, performing the forward lunge with the trunk extended has the potential to increase the muscular response of the knee extensors. While these variations of the forward lunge are commonly used in clinical practice,¹³ it is not known if, and to what extent, trunk position influences lower extremity biomechanics during this activity.

The purpose of the current study was to examine how trunk positioning during a forward lunge influences the kinematics, kinetics, and muscle activity of the lead lower extremity. It was hypothesized that compared to a normal lunge with the trunk erect (NL), a lunge with the trunk forward (LTF) would result in kinematic and kinetic changes that would increase the muscular actions of the hip extensors and the ankle plantar flexors, while decreasing the muscular actions of the knee extensors. It also was hypothesized that a lunge with trunk extension (LTE) would result in kinematic and kinetic changes that would increase the muscle activity of the knee extensors and decrease the muscular response of the hip extensors and the ankle plantar flexors compared to the NL condition. Information obtained from this study will allow for a better understanding of the forward lunge exercise, which in turn will translate into a more appropriate selection and implementation of this exercise to recruit specific muscle groups.

METHODS

Subjects

TEN HEALTHY ADULTS (5 MALES AND 5 females) without a history of lower extremity pain or pathology participated in this study (mean \pm SD age, 26.7 ± 3.2 years). Subjects were excluded from participation if they reported having any of the following: (1) previous history of knee surgery, (2) history of traumatic

patellar dislocation, or (3) neurological involvement that would influence performing the required exercises. The average \pm SD height and mass of the participants were 1.73 ± 0.07 m and 62.5 ± 9.8 kg, respectively. The protocol for this study was approved by the Health Sciences Institutional Review Board of the University of Southern California. Subjects provided their informed written consent prior to participation.

Instrumentation

The dominant lower extremity (the lower extremity used to kick a ball) was instrumented for testing purposes. Lower extremity kinematics were collected using an 8-camera motion analysis system at 60 Hz (Vicon; Oxford Metrics LTD, Oxford, UK). Reflective markers (14-mm spheres) were placed over the following bony landmarks: the first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, the joint space between the fifth lumbar and the first sacral spinous processes and bilaterally over the greater trochanters and iliac crests. In addition, triads of rigid reflective tracking markers were placed on the lateral surfaces of the subject's thigh, lower leg, and heel counter of the shoe. Visual 3D software (C-Motion, Rockville, MD) was used to quantify motion of the hip, knee, and ankle, based on standard anatomical conventions (ie, relative motion between adjacent segments). Ground reaction forces were recorded at a rate of 1560 Hz, using an AMTI force plate (OR6-6-1; Advanced Mechanical Technology, Inc, Watertown, MA). The trajectory data from the reflective markers combined with the ground reaction forces were used to calculate the internal net joint moments using inverse dynamics equations. All moment data were normalized by body weight.

Electromyographic (EMG) activity of selected lower extremity muscles was recorded at 1560 Hz, using double-differential preamplified, bipolar, grounded surface electrodes (Motion Control, Salt Lake City, UT). Each electrode consisted

of 2 circular stainless contacts (12-mm diameter) separated by a distance of 17 mm. EMG unit specifications consisted of a differential input impedance greater than 100 000 M Ω , common-mode rejection ratio of greater than 100 dB, and a signal-noise ratio of 50 dB. Anti-alias filtering with a low-pass cutoff of 750 Hz was utilized.

Prior to electrode placement, the skin was shaved, abraded with coarse gauze to reduce skin impedance, and cleaned with isopropyl alcohol. Electrodes were then placed over the gluteus maximus, biceps femoris, vastus lateralis, and the lateral head of the gastrocnemius muscles of the dominant lower extremity, in accordance with previously published literature.^{1,5,10} The gluteus maximus electrode was placed over the muscle belly midway between the second sacral vertebra and the greater trochanter. The biceps femoris electrode was placed over the muscle belly midway between the ischial tuberosity and the lateral epicondyle of the femur. The vastus lateralis electrode was placed over the muscle belly at the level of the mid thigh. The lateral gastrocnemius electrode was placed over the upper one third of the muscle belly.

Electrodes were connected to an EMG receiver unit, which was carried in a small pack on the subject's back. EMG signals were transmitted to an analog-to-digital converter using an 8-channel hardwired EMG unit. Differential amplifiers were used to reject the common noise and amplify the remaining signal (gain, 2000). EMG signals were then bandpass filtered (20-500 Hz) and a 60-Hz notch filter was applied. Data were full-wave rectified and a moving average smoothing algorithm (75-millisecond window) was used to generate a linear envelope. EMG processing and smoothing was performed using EMG Analysis software (Motion Lab Systems, Baton Rouge, LA).

All EMG signals were normalized to the maximum EMG signal recorded during a maximum voluntary isometric contraction (MVIC). The MVIC for the gluteus maximus muscle was performed



FIGURE. Different variations of the lunge exercise. (A) Starting position for all 3 conditions; (B) lunge with trunk extension (LTE); (C) normal lunge with the trunk erect (NL); (D) lunge with trunk forward (LTF).

with subjects prone on a treatment table with the knee flexed to 90°. Hip extension resistance was provided by a strap, positioned superior to the knee joint and secured around the table. The MVIC of the biceps femoris was performed with subjects supine with hip and knee flexed to 90°. This position was supported by a 45-cm stool placed under the lower leg. Subjects were instructed to maximally flex their knee into the stool. The pelvis was stabilized during this maneuver using a strap secured around the table. The vastus lateralis MVIC was performed with subjects seated and the knee flexed to 60°. Manual resistance was applied just above the ankle. Finally, the lateral

gastrocnemius MVIC was performed in standing. Subjects were instructed to perform a single-leg heel raise, with resistance supplied by a strap that formed a loop around the subject's shoulders and under the foot of the lower extremity of interest.

All MVICs were held for 5 seconds. The highest 1-second average of EMG signal during each MVIC was used for normalization purposes.

Lunge Protocol

Subjects were instructed to perform 5 repetitions of 3 different variations of the forward lunge that differed only in trunk and upper extremity position. Each sub-

ject performed the lunge variations in a random order. For all conditions, the subject started in a standing position with the trunk in an upright position and arms next to the body (FIGURE).

For the NL condition, subjects were instructed to step forward with their dominant lower extremity to a predetermined distance marked on the force platform, while maintaining a vertical position of their trunk. The length of the step was standardized for each subject and was equal to the distance from the greater trochanter to the floor as measured with the subject standing. This normalized distance was chosen based on pilot testing, in which a comfortable lunge step length was determined. Subjects were asked to lower their trunk by flexing their lead and trailing knees simultaneously to a point where the trail knee was approximately 2 to 3 cm short of contacting the ground (FIGURE). The lunge was completed when the subjects returned to the starting position.

The LTF condition was performed as described above; however, subjects were instructed to bring their arms beyond the knee of the lead lower extremity. This was accomplished by flexing the hip, trunk, and shoulders (FIGURE). For the LTE condition, subjects raised their arms overhead and backwards, which induced trunk extension. Subjects were instructed to reach up and back with their upper extremities as far as possible (FIGURE).

The overall speed and duration of each lunge was controlled by using a metronome set at 60 beats per minute. Subjects were asked to adjust their speed so that the overall duration of each repetition of the lunge took 6 seconds (3 seconds for both the ascending and descending phases of the lunge). Errors and variability of performance were controlled by eliminating trials in which subjects demonstrated inconsistencies such as loss of balance or the inability to maintain the desired speed of movement. In addition, 1 of the investigators familiar with the lunging task provided simultaneous visual and auditory cues to subjects.

Data Analysis

The lunge cycle was identified as the time from initial contact of the lead lower extremity with the force platform to the time contact was terminated. The kinematic and kinetic variables of interest included peak sagittal plane hip, knee, and ankle angles, as well as sagittal plane hip, knee, and ankle joint impulse. We elected to evaluate sagittal variables as trunk flexion and extension would be expected to have the greatest influence on lower extremity mechanics in this plane. Joint impulse was calculated as the area under the moment-time curve during each lunge trial. Because impulse takes into consideration the magnitude and duration of the net joint moment, this variable gives a better indicator of the total torque experienced by a joint during a particular activity as opposed to the torque experienced at a single point in time (ie, peak moment).²⁹

The EMG variables of interest consisted of the highest 1-second average of normalized EMG signal of each muscle during the 3 lunge conditions. This was done to avoid averaging low levels of EMG during the lunge cycle (as was commonly observed at movement initiation and termination), with high levels of EMG (as would tend to occur at maximum hip and knee flexion). For all variables, data from the 5 trials were averaged for each subject for each of the 3 lunge conditions.

Statistical Analysis

Differences among the 3 lunge conditions were assessed using a 1-way analysis of variance (ANOVA) with repeated measures. This analysis was repeated for each dependent variable of interest. Post hoc testing, consisting of paired *t* tests with a Bonferroni correction, was performed when a significant ANOVA test was identified ($P < .015$). A Bonferroni adjustment was used to avoid potential type I errors associated with performing multiple *t* tests in the post hoc analysis. As a result, the critical threshold for significance was reduced to $P < .015$ (.05 divided by 3).²⁵

TABLE

SAGITTAL PLANE LOWER EXTREMITY PEAK JOINT ANGLES, AVERAGE JOINT IMPULSE, AND AVERAGE ELECTROMYOGRAPHIC (EMG) SIGNAL INTENSITY*

Measures	NL	LTF	LTE
Joint angles (deg)			
Ankle dorsiflexion	25.3 ± 5.7	24.3 ± 4.9	31.4 ± 3.5 ^{†‡}
Knee flexion	110.3 ± 5.9	104.3 ± 11.1	113.4 ± 7.4 [†]
Hip flexion	87.4 ± 11.8	107.9 ± 9.7 [†]	79.7 ± 11.5 ^{†‡}
Joint impulse (N·m·s/kg)			
Ankle plantar flexor	1.7 ± 0.4	2.5 ± 0.4 [†]	2.0 ± 0.5 [†]
Knee extensor	2.5 ± 0.6	2.1 ± 0.5	2.6 ± 0.5 [†]
Hip extensor	3.9 ± 0.7	5.2 ± 1.0 [†]	3.5 ± 0.9 [†]
EMG (% MVIC)			
Lateral gastrocnemius	19.5 ± 9.2	20.8 ± 5.2	22.6 ± 9.2
Vastus lateralis	45.6 ± 8.3	50.9 ± 13.2	50.1 ± 9.2
Gluteus maximus	18.5 ± 11.0	22.3 ± 12.0 [†]	19.3 ± 11.8
Biceps femoris	11.9 ± 6.4	17.9 ± 9.6 [†]	14.0 ± 9.3

Abbreviations: LTE, lunge with trunk extension; LTF, lunge with trunk forward; MVIC, maximum voluntary isometric contraction; NL, normal lunge with the trunk erect.

* All data are mean ± SD; n = 10 subjects.

[†] Statistically significantly different from NL ($P < .015$).

[‡] Statistically significantly different from LTF ($P < .015$).

RESULTS

Ankle Joint

PEAK ANKLE DORSIFLEXION WAS significantly greater during the LTE condition (mean ± SD, 31.4° ± 3.5°) compared to the NL (mean ± SD, 25.3° ± 5.7°; $P = .008$) and the LTF (mean ± SD, 24.3° ± 4.9°; $P = .004$) conditions (TABLE). No difference in peak ankle dorsiflexion was found between the LTF and the NL conditions. Ankle plantar flexor impulse was significantly greater during the LTF condition (mean ± SD, 2.5 ± 0.4 N·m·s/kg) compared to the NL (mean ± SD, 1.7 ± 0.4 N·m·s/kg; $P < .001$) and the LTE conditions (mean ± SD, 2.0 ± 0.5 N·m·s/kg; $P = .005$) (TABLE). No difference in ankle plantar flexor impulse was found between the LTE and NL conditions. In addition, no differences in lateral gastrocnemius EMG were detected between the 3 lunge conditions (TABLE).

Knee Joint

The peak knee flexion angle during LTE condition was significantly greater than the LTF condition (mean ± SD, 113.4°

± 7.4° versus 104.3° ± 11.1°; $P = .003$); however, no differences were detected when the LTF and LTE conditions were compared to the NL condition (TABLE). Similarly, knee extensor impulse was significantly greater for the LTE condition compared to the LTF condition (mean ± SD, 2.6 ± 0.6 versus 2.1 ± 0.5 N·m·s/kg; $P = .008$); however, no differences were evident when the LTF and LTE conditions were compared to the NL condition (TABLE). No differences in vastus lateralis EMG were detected between the 3 lunge conditions (TABLE).

Hip Joint

The peak hip flexion angle was significantly greater during the LTF condition (mean ± SD, 107.9° ± 9.7°) compared to the NL condition (mean ± SD, 87.4° ± 11.8°; $P < .001$) and the LTE condition (mean ± SD, 79.7° ± 11.5°; $P < .001$) (TABLE). In addition, the peak hip flexion angle during LTE was significantly less when compared to the NL condition (mean ± SD, 79.7° ± 11.5° versus 87.4° ± 11.8°; $P = .01$) (TABLE). Hip extensor impulse for the LTF condition was sig-

nificantly greater (mean \pm SD, 5.2 ± 1.0 N·m·s/kg) compared to the NL (mean \pm SD, 3.9 ± 0.7 N·m·s/kg; $P = .001$) and LTE conditions (mean \pm SD, 3.5 ± 0.9 N·m·s/kg; $P < .001$) (TABLE). No difference in hip extensor impulse was detected between the LTE and the NL conditions.

The EMG of the gluteus maximus for the LTF was significantly greater (mean \pm SD, $22.3\% \pm 12.0\%$ MVIC) than the NL condition (mean \pm SD, $18.5\% \pm 11.0\%$ MVIC; $P = .009$) (TABLE). Similarly, the EMG of the biceps femoris for the LTF condition was significantly greater (mean \pm SD, $17.9\% \pm 9.6\%$ MVIC) when compared to the NL condition (mean \pm SD, $11.9\% \pm 6.4\%$ MVIC; $P = .005$) (TABLE). No differences in EMG were detected for either the gluteus maximus or the biceps femoris between the LTE and the NL conditions.

DISCUSSION

THE FORWARD LUNGE IS A COMMON rehabilitation exercise that simulates many activities of daily living. Clinically, variations of the forward lunge exercise, including the LTF and LTE, are implemented with the intent to target certain muscle groups, depending on an individual's needs.^{11,13} The current investigation supports this premise, in that the LTF and LTE affected the lead lower extremity biomechanics when compared to the NL condition. Based on the proposed hypotheses, our discussion will focus on how the LTF and LTE conditions differed from the NL condition.

Lunge With Trunk Forward

In agreement with our proposed hypothesis, the LTF was found to significantly increase the hip extensor impulse and hip extensor EMG when compared to the NL. These findings were consistent with the observation of greater peak hip flexion during the LTF compared to the NL condition. Although our findings would appear to suggest that the LTF better challenges the hip extensors compared to the NL, it should be

noted that the increases in hip extensor EMG with the LTF were relatively small (3.8% MVIC and 6.0% MVIC for the gluteus maximus and biceps femoris, respectively). Given as such, the clinical relevance of the observed EMG changes could be debated.

The relatively small increases in hip extensor muscle activity during the LTF may be due to multiple factors, thus a brief discussion is warranted. First, it is possible that the forward shift in the body center of mass during the LTF may not have been of sufficient magnitude to warrant meaningful increases in hip extensor muscle action. The second factor may be related to the fact that the subjects in our study were healthy young adults. It is conceivable that individuals with hip extensor weakness may exhibit greater amount of muscle recruitment when performing the various variations of the forward lunge exercise. For example, the observed changes in hip muscle EMG in our subjects during the LTF corresponded to a 20% and 50% increase in gluteus maximus and biceps femoris activity respectively compared to the NL condition. While this may not represent a meaningful increase in muscle activation for an individual who is performing the NL at a level of 20% MVIC, this relative increase could be substantial for a weaker individual who is performing the NL at 60% MVIC.

Inconsistent with our first hypothesis, we did not detect a decrease in knee extensor recruitment during the LTF condition. Based on the visual observation of multiple subjects, we speculated that the participants may have been unloading the trail lower extremity, while shifting more of their body weight onto the lead lower extremity, during the LTF condition. The increase in weight bearing on the lead lower extremity would be expected based on the nature of the LTF maneuver. To explore this hypothesis, a post hoc analysis of the vertical ground reaction forces during the LTF and NL conditions was performed. This analysis revealed that the individuals exhibited a

27% increase in vertical ground reaction force for the lead lower extremity during the LTF compared to the NL ($P < .001$). This increase in the vertical ground reaction force on the lead leg during the LTF condition would increase the impulse of all the lead lower extremity joints. Therefore, it may be the case that the expected decrease in the knee extensor impulse owing to a shift of the center of mass over the knee joint center (ie, reduced ground reaction force lever arm) was offset by an increase in the vertical ground reaction force under the lead lower extremity. Similarly, the increase in hip extensor impulse during the LTF condition was likely due to the combined effect of a more anterior position of the center of mass relative to hip joint center as well as an increase in the vertical ground reaction force experienced by the lead lower extremity.

While there were no differences in peak ankle dorsiflexion angles between the LTF and NL, subjects demonstrated a 47% increase in plantar flexor impulse during the LTF condition. Similar to the observed increase in hip extensor impulse, the increase in plantar flexor impulse was likely related to the forward shift of the center of mass relative to the base of support as well as the increased vertical ground reaction force during the LTF condition. The increase in plantar flexor impulse during the LTF condition was not, however, accompanied by an increase in the lateral gastrocnemius EMG.

Lunge With Trunk Extension

In contrast to the comparison between the NL and the LTF conditions, there were minimal differences between the LTE and NL. The primary differences were seen in the kinematics, which included a greater peak ankle dorsiflexion angle and a lesser peak hip flexion angle in the LTE condition. Despite the kinematic differences, however, there were no differences in joint impulse at the hip, knee, and ankle, or lower extremity EMG between the 2 conditions.

As suggested above, differences in lead lower extremity biomechanics during the 3 lunge conditions may have been the result of the center of mass excursion relative to the base of support, as well as the differences in the amount of weight bearing on the lead lower extremity. The more pronounced changes in lower extremity biomechanics exhibited during the LTF condition reflect the inherently greater range of hip and trunk flexion, thereby facilitating the forward movement of the body's center of mass and increased weight bearing on the lead lower extremity. Conversely, the lack of EMG and joint impulse changes between the NL and the LTE may be attributed to the fact there is less available motion of the trunk into extension as opposed to flexion. The reduced extension motion would limit the posterior displacement of the body's center of mass.

Interpretation of the findings of this investigation should be made with an appreciation for the study's limitations. First, our subject sample consisted of a cohort of young healthy individuals. Given as such, care should be taken in extrapolating these findings to various patient populations that may exhibit weakness, range-of-motion deficits, or other impairments. Second, it could be argued that our sample size was inadequate to find significant differences in impulse and EMG variables between the NL and LTE. However, relatively small differences in joint kinematics were detected between these 2 conditions (6°-7°), suggesting that our study was powered appropriately. Third, lower extremity kinematics were not strictly controlled between subjects (subjects were permitted to vary their joint kinematics to accomplish each lunge task according to the specific performance criteria). Therefore, the degree to which the observed differences in lower extremity biomechanics and muscle activation patterns between the 3 lunge conditions can be attributed solely to differences in trunk positioning is unknown.

CONCLUSION

OUR DATA SUGGEST THAT TRUNK AND upper extremity position during the lunge exercise can significantly affect the biomechanics of the lower extremities. In general, the LTF was characterized by an increase in hip extensor impulse and a concomitant increase in the muscular actions of the hip extensors when compared to NL. In contrast, performing a forward lunge with the trunk extended did not alter joint impulse or the EMG signals of the lower extremity musculature. These findings should be considered when prescribing variations of the lunge as part of a lower extremity rehabilitation program. ●

KEY POINTS

FINDINGS: Performing a forward lunge with trunk forward increases the recruitment of the hip extensors. Performing a forward lunge with trunk extension does not alter the activation of the lower extremity musculature.

IMPLICATION: Performing a forward lunge with the trunk forward may be desirable when the goal is to increase recruitment of the hip extensors.

CAUTION: The subjects utilized in this study consisted of healthy young adults. Care should be taken in extrapolating these findings to various patient populations.

REFERENCES

1. Basmajian JV, De Luca CJ. *Muscles Alive: Their Functions Revealed by Electromyography*. Baltimore, MD: Williams & Wilkins; 1985.
2. Bennell KL, Talbot RC, Wajswelner H, Techovanich W, Kelly DH, Hall AJ. Intra-rater and inter-rater reliability of a weight-bearing lunge measure of ankle dorsiflexion. *Aust J Physiother*. 1998;44:175-180.
3. Beutler AI, Cooper LW, Kirkendall DT, Garrett WE, Jr. Electromyographic Analysis of single-leg, closed chain exercises: implications for rehabilitation after anterior cruciate ligament reconstruction. *J Athl Train*. 2002;37:13-18.
4. Boyden G, Kingman J, Dyson R. A comparison of quadriceps electromyographic activity with the position of the foot during the parallel squat. *J Strength Cond Res*. 2000;14:379-382.

5. Cram JR, Kasman GS. *Introduction to Surface Electromyography*. New York, NY: Aspen Publishers, Inc; 1998.
6. Cronin J, McNair PJ, Marshall RN. Lunge performance and its determinants. *J Sports Sci*. 2003;21:49-57.
7. Dionisio VC, Almeida GL, Duarte M, Hirata RP. Kinematic, kinetic and EMG patterns during downward squatting. *J Electromyogr Kinesiol*. 2008;18:134-143. <http://dx.doi.org/10.1016/j.jelekin.2006.07.010>
8. Escamilla RF, Fleisig GS, Lowry TM, Barrentine SW, Andrews JR. A three-dimensional biomechanical analysis of the squat during varying stance widths. *Med Sci Sports Exerc*. 2001;33:984-998.
9. Escamilla RF, Fleisig GS, Zheng N, Barrentine SW, Wilk KE, Andrews JR. Biomechanics of the knee during closed kinetic chain and open kinetic chain exercises. *Med Sci Sports Exerc*. 1998;30:556-569.
10. Flanagan S, Salem GJ, Wang MY, Sanker SE, Greendale GA. Squatting exercises in older adults: kinematic and kinetic comparisons. *Med Sci Sports Exerc*. 2003;35:635-643. <http://dx.doi.org/10.1249/01.MSS.0000058364.47973.06>
11. Flanagan SP, Wang MY, Greendale GA, Azen SP, Salem GJ. Biomechanical attributes of lunging activities for older adults. *J Strength Cond Res*. 2004;18:599-605. [http://dx.doi.org/10.1519/1533-4287\(2004\)18<599:BAOLAF>2.0.CO;2](http://dx.doi.org/10.1519/1533-4287(2004)18<599:BAOLAF>2.0.CO;2)
12. Graham VL, Gehlsen GM, Edwards JA. Electromyographic Evaluation of Closed and Open Kinetic Chain Knee Rehabilitation Exercises. *J Athl Train*. 1993;28:23-30.
13. Gray G. *Total Body Functional Profile*. Adrian, MI: Wynn Marketing, Inc; 2001.
14. Hefzy MS, al Khazim M, Harrison L. Co-activation of the hamstrings and quadriceps during the lunge exercise. *Biomed Sci Instrum*. 1997;33:360-365.
15. Isear JA, Jr., Erickson JC, Worrell TW. EMG analysis of lower extremity muscle recruitment patterns during an unloaded squat. *Med Sci Sports Exerc*. 1997;29:532-539.
16. Jonsson H, Karrholm J. Three-dimensional knee joint movements during a step-up: evaluation after anterior cruciate ligament rupture. *J Orthop Res*. 1994;12:769-779. <http://dx.doi.org/10.1002/jor.1100120604>
17. Kisner C, Colby LA. *Therapeutic Exercise: Foundations and Techniques*. Philadelphia, PA: F.A. Davis; 1996.
18. Lieber RL, Lieber RL. *Skeletal Muscle Structure, Function and Plasticity: The Physiological Basis of Rehabilitation*. Philadelphia, PA: Lippincott Williams & Wilkins; 2002.
19. Mathiyakom W, McNitt-Gray JL, Requejo P, Costa K. Modifying center of mass trajectory during sit-to-stand tasks redistributes the mechanical demand across the lower extremity joints. *Clin Biomech (Bristol, Avon)*. 2005;20:105-111. <http://dx.doi.org/10.1016/j.clinbiomech.2005.02.001>

clinbiomech.2004.08.005

20. McLaughlin TM, Dillman CJ, Lardner TJ. A kinematic model of performance in the parallel squat by champion powerlifters. *Med Sci Sports*. 1977;9:128-133.
21. McLaughlin TM, Lardner TJ, Dillman CJ. Kinetics of the parallel squat. *Res Q*. 1978;49:175-189.
22. Mian OS, Thom JM, Narici MV, Baltzopoulos V. Kinematics of stair descent in young and older adults and the impact of exercise training. *Gait Posture*. 2007;25:9-17. <http://dx.doi.org/10.1016/j.gaitpost.2005.12.014>
23. Miller JP, Sedory D, Croce RV. Leg Rotation and Vastus Medialis Oblique/Vastus Lateralis Electromyogram Activity Ratio During Closed Chain Kinetic Exercises Prescribed for Patellofemoral

Pain. *J Athl Train*. 1997;32:216-220.

24. Ninos JC, Irrgang JJ, Burdett R, Weiss JR. Electromyographic analysis of the squat performed in self-selected lower extremity neutral rotation and 30 degrees of lower extremity turn-out from the self-selected neutral position. *J Orthop Sports Phys Ther*. 1997;25:307-315.
25. Portney LG, Watkins MP. *Foundations of Clinical Research: Applications to Practice*. Upper Saddle River, NJ: Prentice Hall Health; 2000.
26. Protopapadaki A, Drechsler WI, Cramp MC, Coutts FJ, Scott OM. Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals. *Clin Biomech (Bristol, Avon)*. 2007;22:203-210. <http://dx.doi.org/10.1016/j.clinbiomech.2006.09.010>

org/10.1016/j.clinbiomech.2006.09.010

27. Salem GJ, Powers CM. Patellofemoral joint kinetics during squatting in collegiate women athletes. *Clin Biomech (Bristol, Avon)*. 2001;16:424-430.
28. Salem GJ, Salinas R, Harding FV. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. *Arch Phys Med Rehabil*. 2003;84:1211-1216.
29. Zatsiorsky VM. *Kinetics of Human Motion*. Champaign, IL: Human Kinetics; 2002.



MORE INFORMATION
WWW.JOSPT.ORG