Kinematics and Kinetics of 2 Styles of Partial Forward Lunge

Daniel J. Wilson, Kyle Gibson, and Gerald L. Masterson

Objective: To evaluate the anterior shift of the body’s center of gravity (CG) and the associated inertial forces produced by 2 styles of a partial forward lunge. Setting: Gait-analysis laboratory of a research institution. Participants: 10 healthy volunteers. Intervention: 3 trials of each lunge. Main Outcome Measures: Kinematic data were collected, and inertial reaction forces were resolved into net compressive and shear forces using an inverse dynamic model. Results: Significantly (P < .001) greater anterior translation of the CG was found with an arms-in-front v arms-across-chest lunge style. No significant differences were found between the average peak inertial compressive and shear forces of the 2 styles (427 ± 184 N v 426 ± 187 N, −536 ± 113 N v −538 ± 127 N). Conclusion: Anterior translation of the CG was larger with the arms-forward partial-lunge position, creating increased balance demands. Both styles produced clinically safe (posteriorly directed) inertial shear forces, with greater anterior CG shift with the arms-forward style. Keywords: balance, center of gravity, inertial forces, physical therapy

Clinically, the standard 1-leg forward lunge (arms at chest or side) has been reported to be a safe exercise to include in rehabilitation programs because of the similarity between this exercise and the squat, an exercise popular in postinjury rehabilitation programs. The lunge can be safely used to strengthen the biarticular quadriceps and hamstrings muscles critical to the proper rehabilitation of gait and activities of daily living, with the added advantage of imposing balance demands beyond those of the more common squat. Biomechanically, the forward body position of the lunge results in a larger anterior translation of the center of gravity (CG) than the lunge and a change in the base of support created by the forward step. The effects of perturbations to balance created by the interaction of these 2 variables have not been reported.

Rehabilitation patients often have balance deficits resulting in a variety of compensatory mechanisms that have the potential to affect the effectiveness of the exercises prescribed. We have seen many rehabilitation patients adopt an excessive backward (posterior) lean of the upper extremities despite the increased anteroposterior position of the base of support. In addition, we have seen many patients use this excessive posterior lean during the forward lunge in combination

Wilson and Masterson are with the Dept of Health, Physical Education and Recreation, Missouri State University, Springfield, MO 65897. Gibson is with the Dept of Physical Therapy, University of Missouri–Columbia, Columbia, MO 65211.
with hip and ankle strategies designed to shift the body’s CG posteriorly to alleviate perceived balance difficulties. Ankle strategies are somatosensory-invoked muscle contractions at the lower extremity level in response to a mild balance threat. Hip strategies are similar muscle contractions in the upper and lower limbs and movement in the hips in response to a greater-intensity balance threat. This can negate the potential benefits of the balance training associated with the lunge.

As an alternative to the full lunge, a partial lunge can be used to reduce some of the muscle-strength and -coordination demands of the full lunge. In contrast to the full lunge, a partial lunge is performed through a more limited range of motion. Rather than flexing the forward-leg knee until the forward thigh reaches horizontal, a partial flexion of the forward knee allows for a limited downward movement. Theoretically, the reduced balance and strength demands of the partial lunge will result in anterior translation of the CG, rather than the use of postural strategies that prevent the desired outcome. As an added benefit, reducing posteriorly directed body movements during the performance of the partial lunge might reduce forces imposed on the anterior knee. To further shift the CG forward, clinical manuals have suggested allowing patients to hold their arms horizontally forward during the exercise as a natural progression.

The biomechanical effects of a partial squat on the movement of the CG have not been reported. Clinical observations of patients using these styles of partial lunge have led therapists to question the effects of these exercises on knee safety. Ohkoshi et al have suggested that shifting the CG anterior to the knee during a rehabilitation exercise (squat) can enhance safety, because of the increasing net posterior shear force (posterior drawer force) imposed on the knee as the CG translates anteriorly. Posterior shear unloads the anterior cruciate ligament (ACL), whose function is to prevent posterior translation of the femoral condyles across the tibia (anterior translation relative to the tibia) during knee flexion. Thus, a forward lunge has the benefit of increased safety against potential knee injury while strengthening the large postural muscles. The objective of our study was to evaluate differences in kinematics and kinetics of body position (trunk lean) produced by 2 styles of partial forward lunge commonly used in rehabilitation. As an ancillary to the primary objective of evaluating the biomechanics of CG translation of the partial squat, we calculated the associated inertial forces imposed on the knee.

Methods and Materials

Subjects

Ten volunteer subjects (5 men and 5 women) were recruited from a large Midwest university (mean height 168.1 ± 23.0 cm, mean weight 72.9 ± 12.4 kg, age range 22–35 years). Because we found no studies that had previously reported on the safety of the lunge for subjects with balance deficits, we chose to conduct our initial investigation with healthy individuals inexperienced in performing the forward lunge. Subjects reported no previous history of knee injury, and no pathology was found with Lachman, posterior drawer, and varus–valgus examination. All testing was completed by an experienced physical therapist by standard methods. An informed-consent form was signed by all subjects before testing.
**Data Collection**

Kinematic data were collected using 2 Panasonic S-VHS recorders (model AG-455P; Panasonic Corp, Secaucus, NJ) equipped with 12:1 variable-speed-control power-zoom lenses and digital autofocus (focal length 5.6 to 67 mm) at a video speed of 60 Hz. The data-collection setting was arranged so that each camera was at a 30° angle relative to a perpendicular line extending from the activity plane, providing an effective camera angle of 60°. Retroreflective markers 0.64 cm in diameter were fixed to palpable body landmarks to estimate the rotational centers of the ankle, knee, hip, and shoulder. The resulting kinematic data were stored and analyzed using an Ariel Performance Analysis System (APAS; Ariel Dynamics Inc, Trabuco Canyon, CA). Ground-reaction forces were collected using a Bertec force platform (Bertec Corp, Columbus, OH). Anthropometric data (eg, segmental weights and moments of inertia) were estimated from tabular data from Winter. The CG was calculated by the APAS system as the weighted sum of the segmental masses.

Subjects completed 3 trials of a forward right-leg-led lunge with their arms either crossed across the chest or held horizontally in front of them (3 of each style). All trials were randomly assigned. Subjects began the lunge from a fundamental standing position with the feet approximately shoulder width apart. The step distance was prerehearsed, equal to 75% of the malleolus-to-trochanter length for each subject. The forward (anterior) foot placement was on the force platform, with the weight shifted to the anterior foot. The knee of the anterior leg was flexed to approximately 30°, consistent with early postrehabilitation exercise recommendations. We did not attempt to include the contributions of the cocontracting quadriceps and hamstrings muscles, because it has been demonstrated via direct measurement using a surgically implanted force transducer on the anteromedial bundle of the ACL that the full forward lunge does not produce greater loads on the ACL than a step-up, a step-down, a 1-legged sit-to-stand, or a traditional 2-legged squat.

To determine the effect of lunge style on the anteroposterior translation of the CG, the horizontal distance between the stationary foot (located over the lateral malleolus) and the horizontal projection of the center of gravity (CGD) was divided by the subject’s standing height (see Figure 1). Thus, a body position producing a greater anterior translation of the CG would produce a greater ratio of CG shift (CGD) to height.

A 3-segment anatomical model was used to estimate net reaction forces of the knee. A free-body diagram of this model (see Figure 2) was developed after the methods of Winter. Resultant forces were interpreted relative to the proximal tibia, similar to the method used by Heijne et al (see Figure 3). In this model, a net positive shear force would be anteriorly directed, and a net positive compression force superiorly directed, relative to the proximal tibia. A trigonometric resolution of the horizontal ($R_x$) and vertical ($R_y$) net reaction forces at the knee gives the shear components

\[
F_{S-X} = \cos \theta \times R_x \\
F_{S-Y} = \sin \theta \times R_y
\]
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and compression components

\[ F_{C-X} = \sin \theta \times R_x \]

\[ F_{C-Y} = \cos \theta \times R_y \]

where \( \theta \) is the angle of inclination relative to the compression axis. Noting the anatomical direction of each component, the total inertial shear force can be calculated by

\[ F_{S-T} = (\cos \theta \times R_x) - (\sin \theta \times R_y) \]

and the total inertial compression force by

\[ F_{C-T} = (\sin \theta \times R_x) + (\cos \theta \times R_y) \]

**Data Analysis**

Kinematic (positional) data were smoothed using a double-pass Butterworth recursive filter. A cutoff frequency of 6 Hz was determined using the point of linearization of the third derivative of the raw data.\(^{12,13}\) An inverse dynamic model\(^{14}\) was

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**Figure 1** — The horizontal distance from the anterior point of the weight-bearing foot to the vertical projection of the body’s center of gravity (CGD) of an individual performing a partial forward lunge with arms extended. The ratio of CGD to standing height was used to compare the anterior translation of the center of gravity of 2 styles of forward lunge.
used to compute net reaction inertial forces at the ankle and knee from smoothed kinematic and kinetic data. Compressive and shear components of the net reaction forces of the knee were computed as described in the preceding section.

Peak values for the resulting inertial compressive and shear forces were calculated in Newtons (N) and normalized to body weight for ease of interpretation. Repeated-measures (trial), dependent-sample $t$ tests of mean differences were used to determine statistically significant differences between peak net inertial compressive and shear forces at the proximal tibia for the 2 styles of partial forward lunge. An a priori level of significance of .05 was used for all statistical testing.

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**Figure 2** — A sagittal-plane free-body diagram of the lower extremities used to calculate net reaction forces at the ankle and knee.
Result

CG Translation

The arms-held-horizontally-in-front partial-lunge position produced a statistically greater \( (P < .001) \) anterior shift \( (0.3307 \pm 0.0236) \) in the subjects’ CG-to-standing-height ratio than the arms-at-side style \( (0.3126 \pm 0.0151; \text{see Table 1}) \).

Peak Inertial Shear Forces

Peak net inertial compressive and shear forces are interpreted using the coordinate convention illustrated in Figure 2 (positive shear is anterior, positive compression is superior, relative to the proximal tibia). All 60 (10 subjects \( \times 2 \) styles \( \times 3 \) trials) exercise trials resulted in a net posteriorly directed inertial shear force relative to

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Table 1  Ratio of Distance From Stationary Foot to Horizontal Projection of the Center of Gravity (CGD) to Standing Height for Subjects Performing 2 Styles of a Partial Forward Lunge

<table>
<thead>
<tr>
<th>Style of lunge</th>
<th>CGD (cm):Standing Height (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arms across chest</td>
<td>0.313</td>
</tr>
<tr>
<td>Arms horizontal</td>
<td>0.331</td>
</tr>
</tbody>
</table>

There were statistically significantly different \( (P < .001) \) anterior-shift: standing-height ratios for the 2 styles.
the proximal tibia (see Table 2). Maximal (peak) inertial shear force occurred at an average knee angle of 36.7° of flexion.

No significant difference was found between the peak inertial shear forces produced by the 2 styles of partial forward lunge ($P = .856$; see Table 3). The peak inertial shear force averaged $-536 \pm 113$ N (mean $\pm$ SD) and $-538 \pm 127$ N for the arms-crossed-in-front-of-body and arms-held-horizontally-in-front partial-lunge styles, respectively.

**Peak Inertial Compressive Forces**

No significant difference was found between the peak inertial compressive forces produced by the 2 styles of partial forward lunge ($P = .931$; see Table 4). The net inertial compressive forces at the same knee angle (point of peak inertial shear) averaged $427 \pm 184$ N and $426 \pm 187$ N for the same 2 lunge styles. All peak inertial-force values (shear and compressive) were recorded during the downward eccentric phase of the forward lunge.

To provide ease of clinical interpretation of the force values, the calculated forces (N) were also presented as a percentage of body weight (% BW; see Table 2). Expressed as % BW, the peak inertial shear forces averaged $-0.75 \pm 0.13$ for the arms-across-chest position and $-0.73 \pm 0.12$ for the arms-horizontal position. The net inertial compressive forces averaged $0.60 \pm 0.25$ and $0.58 \pm 0.22$ for the same positions, respectively.

**Table 2** Net Maximal Inertial Shear Forces and the Corresponding Net Inertial Compressive Forces on the Knee Produced at the Same Knee Angle (36.7°) Averaged Across 3 Trials of a Forward Lunge

<table>
<thead>
<tr>
<th>Style of lunge</th>
<th>Peak Inertial Net Shear Force (N)</th>
<th>% Body Weight</th>
<th>Inertial Net Compressive Force (N)</th>
<th>% Body Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean  SD</td>
<td>Mean  SD</td>
<td>Mean  SD</td>
<td>Mean  SD</td>
</tr>
<tr>
<td>Arms across chest</td>
<td>$-536^a$ 113</td>
<td>$-75$ 0.13</td>
<td>$427^b$ 184</td>
<td>0.60 0.25</td>
</tr>
<tr>
<td>Arms horizontal</td>
<td>$-538$ 127</td>
<td>$-73$ 0.12</td>
<td>$425$ 187</td>
<td>0.58 0.22</td>
</tr>
</tbody>
</table>

*a Note that a negative inertial shear force would be posteriorly directed relative to the proximal tibia.

*b Note that a positive inertial compressive force would be superiorly directed relative to the proximal tibia.

**Table 3** Dependent-Sample, Repeated-Measures $t$ Test of Differences Between 2 Styles of a Partial Forward Lunge on Average Peak Net Inertial Shear Force (N) Across 3 Trials

<table>
<thead>
<tr>
<th>Style of lunge</th>
<th>N</th>
<th>Mean</th>
<th>$SE_M$</th>
<th>df</th>
<th>t</th>
<th>$P$ (2-tailed)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arms across chest</td>
<td>30</td>
<td>−535.8$^a$</td>
<td>20.6</td>
<td>29</td>
<td>−.183</td>
<td>.856</td>
</tr>
<tr>
<td>Arms horizontal</td>
<td>30</td>
<td>−538.2</td>
<td>23.1</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

*a Note that a negative inertial shear force would be posteriorly directed relative to the proximal tibia.
The results of this experiment revealed that performing a partial lunge with the arms extended forward horizontally has the desired effect of greater anterior translation of the CG relative to one done with the arms held across the chest. This position, which enables the individual to provide additional balance support using the extended arms, was also found to be clinically safe relative to forces imposed on the knee, with the inertial shear forces posteriorly directed (–538 ± 127). Shifting the CG anteriorly, however, by raising the arms horizontally in front of the body did not produce significant changes in the net inertial shear force of the knee relative to the more conventional arms-across-chest style (–536 ± 113). The primary clinical significance of these findings is the confirmation of the efficacy of the partial lunge in producing additional balance demands. This is because of the anterior translation of the CG versus the more often-used squat. Note that the subjects in this study did not counter the beneficial effects of anterior translation (CG) by adopting strategies that negated the desired balance demands. Thus, the statistical difference between the 2 styles of partial lunge is not as clinically relevant as the subjects’ proper use of these exercises.

All net inertial shear forces calculated in this study were found to be posteriorly directed (non-ACL-loading) relative to the proximal tibia. Stuart et al analyzed the muscle activity and intersegmental forces about the tibiofemoral joint during 2 different squatting exercises and the lunge using a model that included electromyography (EMG), video, and force-plate measurements. Using an inverse dynamic model, they reported results similar to ours (see Table 5 for a comparison of study findings), finding net posterior shear forces for 3 closed-kinetic-chain (CKC) exercises (tibia relative to femur) throughout both the flexion and the extension phases.

The net inertial compressive forces for the 2 styles of partial forward lunge were also found to be nonstatistically significantly different (427 ± 184 v 425 ± 187). This is important because it has been suggested that additional compressive loads on the knee might attenuate ACL strain through increases in muscle activity. Although we did not measure muscle activity (action potential via EMG) during the forward lunge, increased muscle contraction would be the most likely candidate for balancing the increased extension moment at the knee created by the anterior shift in the CG. Clearly, our subjects did not use inertial changes (eg, speed of movement) to counter this anterior translation in the CG, suggesting an increase in hamstring activity.

### Table 4

<table>
<thead>
<tr>
<th>Style of lunge</th>
<th>N</th>
<th>Mean</th>
<th>SEM</th>
<th>df</th>
<th>t</th>
<th>P (2-tailed)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arms across chest</td>
<td>30</td>
<td>426.6</td>
<td>33.5</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Arms horizontal</td>
<td>30</td>
<td>425.8</td>
<td>34.2</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>df</th>
<th>t</th>
<th>P (2-tailed)</th>
</tr>
</thead>
<tbody>
<tr>
<td>29</td>
<td>.087</td>
<td>.931</td>
</tr>
</tbody>
</table>
Table 5  Comparison of Net Reaction Forces at the Knee for Studies That Quantified Knee Forces During a Squat or Lunge Activity

<table>
<thead>
<tr>
<th>Reference</th>
<th>N</th>
<th>Age (y)</th>
<th>BW (N)</th>
<th>Weight lifted (N)</th>
<th>Knee-flexion range (°)</th>
<th>Shear-force direction acting on tibia</th>
<th>Peak tibiofemoral shear force % (BW + load)</th>
<th>Peak tibiofemoral compressive force % (BW + load)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Escamilla et al²³</td>
<td>12</td>
<td>29 ± 6</td>
<td>912 ± 145</td>
<td>1437 ± 383</td>
<td>0–95</td>
<td>posterior</td>
<td>80 ± 37</td>
<td>133 ± 44</td>
</tr>
<tr>
<td>Escamilla et al⁹</td>
<td>12</td>
<td>30 ± 7</td>
<td>917 ± 137</td>
<td>1309 ± 363</td>
<td>0–95</td>
<td>posterior</td>
<td>99 ± 36</td>
<td>154 ± 38</td>
</tr>
<tr>
<td>Hattin et al²⁶</td>
<td>10</td>
<td>23 ± 2</td>
<td>790 ± 109</td>
<td>339 ± 64</td>
<td>0–90</td>
<td>posterior</td>
<td>67 ± 52</td>
<td>367 ± 122</td>
</tr>
<tr>
<td>Stuart et al¹</td>
<td>6</td>
<td>27 ± 5</td>
<td>798 ± 76</td>
<td>223 ± 0</td>
<td>0–90</td>
<td>posterior</td>
<td>29 ± 3</td>
<td>54 ± 5</td>
</tr>
</tbody>
</table>

Abbreviations: BW, body weight.
This study was unique in that it employed a “partial” lunge and it used an arm position designed to shift the CG anteriorly. Knee-rehabilitation protocols often use exercises such as the lunge based on the hypothesis that CKC exercises do not cause injury by straining the ACL.\textsuperscript{16–18} This is the basis for prescribing exercises such as weight shifting and balance training early after ACL injury, to help achieve full range of motion and reduce muscle atrophy. Several studies\textsuperscript{1,2,19–21} have suggested that open-kinetic-chain and more demanding CKC exercises such as lunges, stair climbing, and leg presses are normally introduced around 6 to 8 weeks after ACL injury. In addition to protecting the ACL through the compressive component of force, it has been suggested that such exercises might strain the ACL less because of a more flexed hip position.\textsuperscript{2,7} Heijne et al\textsuperscript{2} measured ACL strain directly via a surgically implanted displacement transducer aligned with the anteromedial bundle of the ACL. Subjects performed a step-up, step-down, lunge, and 1-legged sit-to-stand immediately after the surgical implantation. Their results showed that the forward lunge did not produce greater strains on the ACL than the traditional 2-legged squat.\textsuperscript{22} The anterior shift of the CG produced by the forward arm position used in this study has been previously reported to produce increases in hip flexion—enhancing safety.\textsuperscript{7} The posteriorly directed shear force, nearly identical to the more common hands-across-chest lunge style reported here, support this claim.

The primary limitation of this investigation is the lack of muscle forces in the model.\textsuperscript{23} Additional study of the biomechanics of the forward lunge would benefit from more sophisticated models, using electromyographical data to estimate muscular contributions to the joint forces and moments. Although the safety of the forward lunge has been well documented, the individual muscle contributions to the net muscle forces and moments of various styles of lunge might contribute to more effective exercise prescription. The primary movers (muscles) at the tibiofemoral joint are the quadriceps, hamstrings, and gastrocnemius. Together, these muscle groups compose approximately 98% of the total cross-sectional area of all knee musculature.\textsuperscript{24} Cocontraction of these muscles produces additional compressive and shear components of force at the knee above those of inertial forces. van Eijden et al\textsuperscript{25} reported that a maximal voluntary contraction of the quadriceps alone generates forces ranging from 2000 to 8000 N, depending on the angle of knee flexion. Most often, EMG is used to estimate muscle force, using the assumption that the magnitude of force is proportional to the strength (amplitude) of the signal.\textsuperscript{13} Escamiila\textsuperscript{23} points out that this assumption is tenuous, because EMG data and muscle force frequently are not strongly correlated, particularly in dynamic movements. Because the safety of the partial lunge studied here was not the primary dependent variable, a simple knee model without muscle-force-contribution estimations was used to compare the direction and magnitude of the imposed forces.

**Conclusion**

This investigation provided evidence that a partial-range-of-motion forward lunge can be used effectively to produce the desired anterior shift of the body’s CG to create unique balance demands for rehabilitation patients. The safety of this exercise was supported by the finding of properly directed internal shear forces on the
knee using a simple 2-dimensional kinetic model. It is suggested that further study be employed to support these findings. The clinical use of the partial lunge with varying arm positions is supported.

References


