Patellofemoral compressive force and stress during the forward and side lunges with and without a stride

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Abstract

Background. Although weight bearing lunge exercises are frequently employed during patellofemoral rehabilitation, patellofemoral compressive force and stress are currently unknown for these exercises.

Methods. Eighteen subjects used their 12 repetition maximum weight while performing forward and side lunges with and without a stride. EMG, force platform, and kinematic variables were input into a biomechanical model, and patellofemoral compressive force and stress were calculated as a function of knee angle.

Findings. Patellofemoral force and stress progressively increased as knee flexion increased and progressively decreased as knee flexion decreased. Patellofemoral force and stress were greater in the side lunge compared to the forward lunge between 80° and 90° knee angles, and greater with a stride compared to without a stride between 10° and 50° knee angles. There were no significant interactions between lunge variations and stride variations.

Interpretation. A more functional knee flexion range between 0° and 50° may be appropriate during the early phases of patellofemoral rehabilitation due to lower patellofemoral compressive force and stress during this range compared to higher knee angles between 60° and 90°. Moreover, when the goal is to minimize patellofemoral compressive force and stress, it may be prudent to employ forward and side lunges without a stride compared to with a stride, especially at lower knee angles between 0° and 50°. Understanding differences in patellofemoral compressive force and stress among lunge variations may help clinicians prescribe safer and more effective exercise interventions.

Keywords: Knee pain; Knee kinetics; Knee biomechanics; Patellofemoral rehabilitation

1. Introduction

Patellofemoral rehabilitation results in millions of dollars (USA) in medical cost each year, and employing exercise therapy during patellofemoral rehabilitation has demonstrated beneficial, positive changes in cost effectiveness, pain severity, and functional disability (van Linschoten et al., 2006). Moreover, the quality of patellofemoral rehabilitation is paramount in determining the health and well being of the patient, and a faster return to function (van Linschoten et al., 2006). Consequently, judicious thought
should be given in choosing patellofemoral rehabilitation exercises.

Patellofemoral pain syndrome (PFPS), otherwise known as anterior knee pain, is the most common cause of knee pain in the outpatient setting, and accounts for 25–30% of all knee pathologies treated (Devereaux and Lachmann, 1984; Dixit et al., 2007; Fredericson and Yoon, 2006). PFPS primarily affects younger (typically the 18–40 age range) active individuals, both athletes and non-athletes (Dixit et al., 2007; Fredericson and Yoon, 2006; LaBella, 2004) and both males and females (Dehaven and Lintner, 1986), although older individuals can also be affected. Because the etiology of PFPS is poorly understood and multi-faceted, it remains one of the most difficult clinical challenges in rehabilitative medicine (Wilk et al., 1998). Although patellofemoral rehabilitation after injury can be a long and arduous process, the use of appropriate exercises can improve this process by decreasing rehabilitation time and improving function (Boling et al., 2006; Heintjes et al., 2003; Natri et al., 1998; Witvrouw et al., 2004; Witvrouw et al., 2000).

Weight bearing exercises are frequently employed during patellofemoral rehabilitation, and are specific to many functional activities such as walking, running, and jumping (Boling et al., 2006; Heintjes et al., 2003; Natri et al., 1998; Witvrouw et al., 2004; Witvrouw et al., 2000). The use of weight bearing exercises have been shown to be effective, both in short and long term outcomes, in decreasing anterior knee pain and enhancing functional performance (Boling et al., 2006; Heintjes et al., 2003; Natri et al., 1998; Witvrouw et al., 2004; Witvrouw et al., 2000). Clinicians use these types of exercises to minimize anterior knee pain and muscle loss, strengthen hip and thigh musculature, enhance balance and stability, and minimize the risk of future injuries and associated costs of health care (van Linschoten et al., 2006). Understanding how patellofemoral compressive force and stress (force per unit patella contact area) vary among weight bearing exercises allows physical therapists, physicians, and trainers to prescribe safer and more effective knee rehabilitation treatment to patients recovering from knee injury or surgery. For example, if an exercise generates greater patellofemoral compressive force and stress between 60° and 90° knee flexion compared to 0–30° knee flexion, the 0–30° knee flexion range can be initiated at an earlier stage in patellofemoral rehabilitation, while the 60–90° knee flexion range can be initiated at a later stage in the rehabilitation process. There may also be differences in patellofemoral compressive force and stress between different exercises over a specific knee flexion range.

Lunges are common weight bearing exercises used by athletes and other individuals with healthy knees to train the hip and thigh musculature. Therapists and trainers also use lunges and similar weight bearing exercises during patellofemoral rehabilitation for PFPS patients to allow them to recover faster and return to function earlier (Boling et al., 2006; Heintjes et al., 2003; Natri et al., 1998; Witvrouw et al., 2004; Witvrouw et al., 2000). Lunges can be done with varying techniques, such as forward or side lunges, as well as lunging with a stride (striding forward or sideways with one leg and returning back to upright position with feet together), or lunging without a stride (keeping both feet stationary throughout the forward or side lunge movements). However, patellofemoral compressive force and stress generated while performing forward and side lunges with and without a stride is currently unknown.

Patellofemoral compressive force, which can elevate subchondral bone stress (Besier et al., 2005), is a plausible cause of pain in individuals with PFPS. Patellofemoral stress can result in cartilage degeneration and a decrease in the ability of the cartilage to distribute patellofemoral compressive force (Besier et al., 2005). Therefore, it is important for clinicians to understanding the magnitudes of patellofemoral compressive force and stress generated during different weight bearing rehabilitation exercises. The purpose of this study was to compare patellofemoral compressive force and stress between forward and side lunges with and without a stride. It was hypothesized that patellofemoral compressive force and stress would increase as knee flexion increased, would be greater with a stride compared to without a stride, and would be greater in the forward lunge compared to the side lunge.

2. Methods

2.1. Subjects

Eighteen healthy individuals (9 males and 9 females) without a history of patellofemoral pathology participated with an average (SD) age, mass, and height of 29(7) y, 77(9) kg, and 177(6) cm, respectively, for males, and 25(2) y, 60(4) kg and 164(6) cm, respectively, for females. In addition, all subjects were required to be able to perform all exercises pain free and with proper form and technique for 12 consecutive repetitions using their 12 repetition maximum (12 RM) weight.

To control the electromyographic (EMG) signal quality, the current study was limited to males and females that had average or below average body fat, which was assessed by Baseline skinfold calipers (Model 68900, Country Technology, Inc., Gays Mill, WI, USA) and appropriate regression equations and body fat standards set by the American College of Sports Medicine. Average (SD) body fat was 12(4)% for males and 18(1)% for females. All subjects provided written informed consent in accordance with the Institutional Review Board at California State University, Sacramento, USA, which approved the research conducted and informed consent form.

2.2. Exercise descriptions

2.2.1. Forward lunge

Each subject performed the forward lunge (Fig. 1a) both with and without stride. The starting and ending positions
for the forward lunge with a stride were the same, which involved standing upright with both feet together. From the starting position for the forward lunge with a stride, the subject held a dumbbell weight in each hand and lunged forward with the right leg toward a force platform at ground level. The dumbbell weight used during the forward lunge with stride was normalized by each subject’s 12 RM weight, and this same weight was used during the forward lunge without stride. The mean (SD) mass of both dumbbells during the forward lunge was 49(11) kg for males and 32(8) kg for females. At right foot contact the right knee slowly flexed until maximum right knee flexion of 90–100° was obtained as the left knee made contact with the ground. From this ending position the subject immediately pushed backward off the force platform and returned to the starting position. A metronome was used to ensure that the knee flexed and extended at approximately 45°/s. Each subject was instructed to use as long a stride length...
as was comfortable. A tester ensured that the stride was long enough so that at the lowest position of the lunge the stride knee was maintained over the stride foot without translating forward beyond the toes. The average (SD) stride length (measured from left toe to right heel) during the forward lunge of 89(4) cm for males and 79(6) cm for females. During the forward lunge, a longer stride length is commonly preferred by individuals compared to a shorter stride length to ensure that the stride knee does not progress beyond the toes at the lowest position of the forward lunge (Fig. 1a). The forward lunge without stride was performed the same as the forward lunge with stride with the exception that both feet remained stationary throughout each repetition during the forward lunge without stride. That is, from the lowest position of the forward lunge shown in Fig. 1a, the subject simply fully extended both knees, and then flexed both knees back to return to the lowest position of the forward lunge shown in Fig. 1a.

2.2.2. Side lunge
Each subject performed the side lunge (Fig. 1b) both with and without a stride. The starting and ending positions for the side lunge with a stride were both the same, which involved standing upright with both feet together. From the starting position for the side lunge with a stride, the subject held a single dumbbell weight down between the legs and lunged sideways with the left knee remaining fully extended and the right leg moving toward a force platform at ground level. The dumbbell weight used during the side lunge with stride was normalized by each subject’s 12 RM weight, and this same weight was used during the side lunge without stride. On the average (SD), the dumbbell mass used during the side lunge was 34(9) kg for males and 20(5) kg for females. At right foot contact the right foot was turned out approximately 30–45° relative to the left foot (subject’s preference) and the right knee flexed slowly at approximately 45°/s until the right knee flexed approximately 90–100° (Fig. 1b) as the left knee remained fully extended. From this ending position the subject then pushed backward off the force platform and returned to the starting position. A metronome was used to ensure that the knee flexed and extended at approximately 45°/s. Each subject was instructed to use as long a stride length as was comfortable. A tester ensured that the stride was long enough so that at the lowest position of the lunge the stride knee was maintained over the stride foot without translating forward beyond the toes. The average (SD) stride length (measured from inside of left heel to inside of right heel) during the side lunge was 94(5) cm for males and 83(3) cm for females. Using a longer stride length compared to a shorter stride length during the side lunge is commonly preferred by individuals as it allows the left knee to remain straight and the right knee to flex approximately 90–100° and remain over the foot. The side lunge without stride was performed the same as the side lunge with stride with the exception that both feet remained stationary throughout each repetition during the side lunge without stride. That is, from the lowest position of the side lunge shown in Fig. 1b, the subject simply fully extended the right knee (left knee was already extended), and then returned back to the lowest position of the side lunge by flexing the right knee.

2.3. Data collection
Each subject came in for a pre-test one week prior to the testing session. At that time the experimental protocol was reviewed and the subject was given the opportunity to ask questions. In addition, each subject’s 12 RM was determined for the forward and side lunges by utilizing the most weight they could lift for 12 consecutive repetitions. Moreover, each subject’s stride length (as previously defined) was measured for the forward and side lunges. Blue Sensor (Ambu Inc., Linthicum, MD, USA) disposable surface electrodes (type M-00-S) were used to collect EMG data. These oval shaped electrodes (22 mm wide and 30 mm long) were placed in a bipolar electrode configuration along the longitudinal axis of muscle, with a center-to-center distance of approximately 3 cm between electrodes. Prior to positioning the electrodes over each muscle, the skin was prepared by shaving, abrading, and cleaning with isopropyl alcohol wipes to reduce skin impedance. As previously described, (Basmajian and Blumenstein, 1980) electrode pairs were then placed on the subject’s right side for the following muscles: (1) rectus femoris; (2) vastus lateralis; (3) vastus medialis; (4) medial hamstrings (semitendinosus and semitendinosus); (5) lateral hamstrings (biceps femoris); and (6) gastrocnemius.

EMG data were collected at 960 Hz using a Noraxon Myosystem EMG unit (Noraxon USA, Inc., Scottsdale, AZ, USA). The amplifier bandwidth frequency was 10–500 Hz, the input impedance of the amplifier was 20,000 kΩ, and the common-mode rejection ratio was 130 Db. EMG signals were processed through an analog to digital (A/D) converter by a 16 bit A/D board.

Spheres (3.8 cm in diameter) covered with 3 M™ reflective tape were attached to adhesives and positioned over the following bony landmarks: medial and lateral malleoli of the right foot; upper edges of the medial and lateral tibial plateaus of the right knee; posterolateral greater trochanters of the left and right femurs; lateral acromion of the right shoulder; and third metatarsal head of the right foot.

A six camera Peak Performance motion analysis system (Vicon-Peak Performance Technologies, Inc., Englewood, CO, USA) was used to collect 60 Hz video data. Each subject performed each lunge variation with their right foot on an AMTI force platform (Model OR6-6-2000, Advanced Mechanical Technologies, Inc.) and their left foot on the ground (Fig. 1a and b), with force platform data collected at 960 Hz. Once the electrodes were positioned, the subject warmed up and practiced the exercises as needed, and data collection commenced. Video, EMG, and force platform data were electronically synchronized and collected as each subject performed in a randomized manner one set of three
continuous repetitions (trials) during the forward lunge with stride, forward lunge without stride, side lunge with stride, and side lunge without stride.

Subsequent to completing all exercise variations, EMG data were collected during maximum voluntary isometric contractions (MVIC) to normalize the EMG data collected during each lunge variation. The MVIC for the rectus femoris, vastus lateralis, and vastus medialis were collected in a seated position at 90° knee and hip flexion with a maximum effort knee extension. The MVIC for the lateral and medial hamstrings were collected in a seated position at 90° knee and hip flexion with a maximum effort knee flexion. MVIC for the gastrocnemius was collected during a maximum effort standing one leg toe raise with the ankle positioned approximately halfway between neutral and full plantar flexion. Two 5 s MVIC trials were randomly collected for each MVIC.

2.4. Data reduction

Video images for each reflective marker were tracked and digitized in three-dimensional space with Peak Performance software, utilizing the direct linear transformation calibration method (Shapiro, 1978). Testing of the accuracy of the calibration system resulted in reflective balls that could be located in three-dimensional space with an error less than 0.7 cm. The raw position data were smoothed with a double-pass fourth order Butterworth low-pass filter with a cut-off frequency of 6 Hz (Escamilla et al., 1998). Joint angles, linear and angular velocities, and linear and angular accelerations were calculated in a two-dimensional sagittal plane of the knee utilizing appropriate kinematic equations (Escamilla et al., 1998).

Raw EMG signals were full-waved rectified and smoothed with a 10 ms moving average window (linear enveloped) throughout the knee range of motion for each repetition. These EMG data were then normalized for each muscle and expressed as a percentage of each subject’s highest corresponding MVIC trial. The MVIC trials were calculated using the highest EMG signal over a 1 s time interval throughout the 5 s MVIC. Normalized EMG data for the three repetitions (trials) were then averaged at corresponding knee angles between 0° and 90°, and were used in the biomechanical model described below.

2.5. Biomechanical model

As previously described (Escamilla et al., 1998; Zheng et al., 1998), a biomechanical model of the knee was used to continuously calculate patellofemoral forces throughout a 90° knee range of motion during the knee flexing (0–90°) and knee extending (90–0°) phases of the lunge exercises. Resultant force and torque equilibrium equations were calculated using inverse dynamics and the biomechanical knee model (Escamilla et al., 1998; Zheng et al., 1998). Moment arms of muscle forces and angles of the line of action for muscles were represented as polynomial functions of the knee flexion angle using data from Herzog and Read (1993).

Quadriceps, hamstrings, and gastrocnemius muscle forces were calculated as previously described (Escamilla et al., 1998; Zheng et al., 1998). Because the accuracy of calculating muscle forces depends on accurate calculations of a muscle’s physiological cross-sectional area (PCSA), maximum voluntary contraction force per unit PCSA, and the EMG-force relationship, resultant force and torque equilibrium equations may not be satisfied. Therefore, each muscle force $F_{m(i)}$ was modified by the following equation: $F_{m(i)} = c_i k_i k_{v(i)} A_i \sigma_{m(i)} [EMG_i/MVIC_i]$ where $A_i$ was PCSA of the $i$th muscle, $\sigma_{m(i)}$ was MVC force per unit PCSA of the $i$th muscle, $EMG_i$ and $MVIC_i$ were EMG window averages of the $i$th muscle EMG during exercise and MVIC trials, $c_i$ was a weight factor (explained below) adjusted in a computer optimization program to minimize the difference between the resultant torque from the inverse dynamics ($T_{res}$) and the resultant torque calculation from the biomechanical model ($T_{bio}$).

$$k_l = (b - (a/F_v) v)/(b + v) \quad \text{concentric}$$

$$k_e = C - (C - 1)(b + (a/F_v) v)/(b - v) \quad \text{eccentric}$$

with $F_v$ representing isometric muscle force, $v$ = velocity, $a = 0.32 F_v$, $b = 3.2 l_0/sec$, and $C = 1.8$. Muscle force from eccentric contractions was scaled up by 1.8 times the isometric muscle force $F_v$. Forces generated by the knee flexors and extensors at MVIC were assumed to be linearly proportional to their physiological cross-sectional area. Muscle force per unit physiological cross-sectional area at MVIC was 35 Ncm$^{-2}$ for the knee flexors and 40 Ncm$^{-2}$ for the quadriceps (Cholewicki et al., 1995; Narici et al., 1992; Narici et al., 1988; Wickiewicz et al., 1984).

The objective function used to determine each $i$th muscle’s coefficient $c_i$ was as follows:

$$\min f(c_i) = \sum_{i=1}^{n_i} (1 - c_i)^2 + \lambda (T_{res} - \sum_{i=1}^{n_i} T_{me})^2,$$

subject to $c_{low} \leq c_i \leq c_{high}$, where $c_{low}$ and $c_{high}$ were lower and upper limits for $c_i$, and $\lambda$ was a constant. The weight factor “$c$” was to adjust the final muscle force calculation. The bounds on “$c$” were set between 0.5 and 1.5. The torques predicted by the EMG driven model matched well (<2%) with the torques generated from the inverse dynamics. The assumptions associated with this model are (1) the torque from cruciate ligament forces were ignored (2) other forces and torques out of the sagittal plane were ignored.

Patellofemoral force was a function of patellar tendon force and quadriceps tendon force. Patellar tendon force was calculated by the quadriceps tendon force and the ratio
of the patellar tendon force and the quadriceps tendon force, as previously described (van Eijden et al., 1986; van Eijden et al., 1987). The angles between the patellar tendon, quadriceps tendon, and patellofemoral joint were expressed as functions of knee angle (van Eijden et al., 1986; van Eijden et al., 1987).

Patellofemoral stress, which was calculated every 10° between 0° and 90° knee angle, was expressed as the ratio of patellofemoral force (calculated from the biomechanical model described above) and patellar contact area (Escamilla et al., 1998; Zheng et al., 1998). Patellar contact areas were determined at 10° intervals between 0° and 90° knee angle. Contact areas from in vivo MRI data from Salsich et al. (2003) (who used both male and female subjects with healthy knees and had them perform weight bearing exercise using resistance, similar to the current study) were used at 0° (146 mm²), 20° (184 mm²), 40° (290 mm²), and 60° (347 mm²) knee angles. These four contact area values formed a near linear relationship as a function of knee angle, resulting in a line of best fit equation of $y = 3.55x + 135$ ($r = 0.98$) with $y =$ contact area and $x =$ knee angle. This line of best fit equation was used to determine contact areas at $10°$ knee angle (171 mm²), $30°$ knee angle (242 mm²), and $50°$ knee angle (313 mm²). The contact areas at 40°, 50°, and 60° knee angles were used to develop the line of best fit equation $y = 2.81x + 176$ ($r = 0.99$), which was used to determine contact areas at $70°$ knee angle (373 mm²), $80°$ knee angle (401 mm²), and $90°$ knee angle (429 mm²). Like the current study, a near linear relationship between patellar contact area and knee angles has been reported between 0° and 90° knee angles in several studies involving weight bearing exercises (Besier et al., 2005; Cohen et al., 2001; Hinterwimmer et al., 2005; Patel et al., 2003; Salsich et al., 2003).

### 2.6. Data analysis

To determine significant differences in patellofemoral forces between the two lunge variations (forward lunge and side lunge) and two stride variations (with stride and without stride), patellofemoral forces were statistically analyzed every 10° during the 0–90° knee flexing phase and the 90–0° knee extending phase using a two factor (lunge variations and stride variations) repeated measure Analysis of variance. Bonferroni t-tests were used to assess pairwise comparisons. The level of significance used was $P < 0.01$. Because each patellofemoral stress value was derived by dividing each patellofemoral force value by a constant for each lunge variation and each stride variation, patellofemoral stress values between lunge and stride variations will have the same $P$ values as the corresponding patellofemoral force values in which they were derived.

### 3. Results

Patellofemoral force and stress progressively increased as knee flexion increased and progressively decreased as knee flexion decreased (Figs. 2–7). Table 1 shows patellofemoral force values as a function of knee angle between forward and side lunges and between with a stride and without a stride. From Table 1, between 80° and 90° knee angles of the knee flexing phase and at 90° knee angle of the knee extending phase patellofemoral force and stress were greater in the side lunge compared to the forward lunge. Between 10° and 50° knee angles of the knee flexing phase and between 50° and 20° knee angles of the knee extending phase patellofemoral force and stress were significantly greater with a stride compared to without a stride. There were no significant interactions between lunge variations and stride variations.

### 4. Discussion

Both patellofemoral compressive force and stress can result in the degeneration of patellofemoral cartilage and anterior knee pain from subchondral bone, synovial plicae, infrapatellar fat pad, retinacula, joint capsule, tendons and ligaments (Biedert and Sanchis-Alfonso, 2002). Another proposed mechanism of anterior knee pain is increased patellar cartilage stress from patellofemoral compressive force leading to subchondral bone stress (Biedert and Sanchis-Alfonso, 2002), and it has been demonstrated that the subchondral bone plate is rich in pain receptors (Wojtyjas et al., 1990). Therefore, understanding differences in patellofemoral compressive force and stress among forward and side lunge variations is helpful to the clinician when prescribing therapeutic exercises to individuals with PFPS.

Contrary to our original hypothesis, patellofemoral compressive force and stress were greater while performing the side lunge compared to the forward lunge, but only at higher knee angles between 80° and 90°. From these data it can be concluded that loading the patellofemoral joint is similar between forward and side lunges except at higher knee angles, where patellofemoral loading was greater in the side lunge.

As hypothesized, patellofemoral compressive force and stress were greater with a stride compared to without a stride, but only between 0° and 50° knee angles. Performing forward and side lunges without a stride within a smaller knee angle range (eg, 0–50°) may be easier and safer to start with earlier during patellofemoral rehabilitation when the goal to minimize patellofemoral compressive force and stress, while the performing forward and side lunges with a stride may be more appropriate later during patellofemoral rehabilitation due to greater patellofemoral compressive force and stress compared to without a stride.

Patellofemoral compressive force and stress curves were similar in shape to each other due to proportional increases in patellofemoral compressive force and patellar contact area with increased knee flexion. One exception was with knee angles between 80° and 90°, which resulted in a decrease in patellofemoral stress. This occurred because although patellar contact area increased between 80° and 90°, patellofemoral compressive force did not increase...
proportionally, but instead began to plateau. This implies that patellofemoral stress decreased during forward and side lunges as maximum knee flexion was approached. These findings are consistent with patellofemoral compressive force and stress data during the barbell squat from Escamilla et al. (1998) and Salem and Powers (2001). Escamilla et al. (1998) reported that patellofemoral compressive forces increases until 75°–80° knee flexion, and then began to level off and plateau or slightly decrease. Salem and Powers (2001) reported no significant differences in patellofemoral compressive force or stress at 75°, 100°, and 110° knee flexion. It can be concluded from these squat data that injury risk to the patellofemoral joint may not increase with knee angles between 75° and 110° due to similar magnitudes in patellofemoral stress during these knee angles, with the benefit of increased quadriceps, hamstring, and gastrocnemius activity when training at higher knee angles (75°–110°) compared to training at lower knee angles (0°–75°) (Escamilla et al., 1998).

Another consideration during patellofemoral rehabilitation is what knee flexion range of motion to employ while performing lunge exercises. Because patellofemoral compressive force and stress both increased with increased knee flexion and decreased with decreased knee flexion, a more functional knee flexion range between 0° and 50° may be appropriate during the early phases of patellofemoral rehabilitation due to lower patellofemoral compressive force and stress that is generated during this range of motion. Higher knee angles between 60° and 90° may be more appropriate later in the rehabilitation process due to higher patellofemoral compressive force and stress that are generated during this range of motion. For example, during the
knee extending phase of the forward lunge, patellofemoral compressive force ranged from 66 to 1176 N between 0° and 50° knee angle and from 1577 to 2191 N between 60° and 90° knee angle (Table 1), and patellofemoral stress ranged from 0.43 to 3.76 MPa between 0° and 50° knee angle and from approximately 4.54–5.11 MPa between 60° and 90° knee angle. This same pattern of increased patellofemoral compressive force and stress with increased knee flexion has been reported during the barbell squat and leg press (Escamilla et al., 1998; Escamilla et al., 2001; Salem and Powers, 2001; Steinkamp et al., 1993; Wallace et al., 2002). These authors reported that patellofemoral compressive force and stress progressively increased from 0° to approximately 90°, peaked near 90°, and then progressively decreasing from approximately 90°–0°. Computer optimization techniques demonstrated similar results during a simulated squat (Cohen et al., 2003).

Peak patellofemoral force and stress magnitudes from the current study are greater compared to weight bearing functional exercises such as walking (Heino Brechter and Powers, 2002) and going up and down stairs (Brechter and Powers, 2002), but less than other weight bearing activities, such as the squat and leg press (Escamilla et al., 1998). Escamilla et al. (1998) reported peak patellofemoral force magnitudes between 4500 and 4700 N at 90° knee angle during the 12 RM squat and leg press using healthy subjects, resulting in patellofemoral stress magnitudes between 11 and 12 MPa. These peak force and stress magnitudes during the squat and leg press are approximately 60% greater compared to peak force and stress magnitudes in the current study, which also occurred at 90° knee angle. Peak patellofemoral force and stress in healthy subjects during fast walking reportedly are approximately 900 N and 3.13 MPa, respectively (Heino Brechter and
Powers, 2002), which are approximately 2–3 times lower than the peak force and stress magnitudes in the current study. However, peak patellofemoral force and stress magnitudes in healthy subjects going up and down stairs reportedly are approximately 2500 N and 7 MPa, respectively (Heino Brechter and Powers, 2002), which are similar to the peak force and stress magnitudes in the current study. Unlike healthy subjects, patients with patellofemoral pathologies have demonstrated smaller patellar contact areas and greater patellofemoral stress during some weight bearing functional activities (Brechter and Powers, 2002; Heino Brechter and Powers, 2002). However, for both healthy subjects and patients with patellofemoral pathologies it is currently unknown what patellofemoral force or stress magnitudes, and over what time duration, can ultimately lead to patellofemoral pathology or exacerbate an existing condition. There are many factors that may contribute to patellofemoral pathology, such as: (1) imbalance or malalignment of the extensor mechanism, which can lead to lateral patellar subluxation or tilt; (2) muscle weakness, such as weak quadriceps and hip external rotators; (3) overuse or trauma; (4) muscle tightness, such as tight quadriceps, hamstrings, or iliotibial band; (5) lower extremity malalignment, such as patella alta; genu valgum, femoral neck anteversion, excessive Q angle; and excessive rearfoot pronation. It can only be surmised that relatively large patellofemoral force and stress magnitudes over time may lead to or exacerbate patellofemoral pathology, especially in individuals that exhibit some of the above factors and are predisposed to patellofemoral problems. Clinicians can use information regarding patellofemoral force and stress magnitudes among different weight bearing exercises, technique variations, and functional activities to more

![Mean (SD) patellofemoral compressive force between forward lunge with and without a stride.](image1)

![Mean (SD) patellofemoral compressive force between side lunge with and without a stride.](image2)
Table 1

Mean (±SD) patellofemoral force (N) values between lunge variations (forward lunge versus side lunge) and stride variations (with a stride versus without a stride)

<table>
<thead>
<tr>
<th>Knee angle for knee flexing phase</th>
<th>Exercise variations</th>
<th>P-value</th>
<th>Stride variations</th>
<th>P-value</th>
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</thead>
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<tr>
<td></td>
<td>Forward lunge</td>
<td>Side lunge</td>
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<td></td>
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<tr>
<td>0°</td>
<td>69 ± 62</td>
<td>46 ± 46</td>
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<td>40°</td>
<td>628 ± 236</td>
<td>573 ± 328</td>
<td>0.390</td>
<td></td>
</tr>
<tr>
<td>50°</td>
<td>1059 ± 425</td>
<td>926 ± 408</td>
<td>0.127</td>
<td></td>
</tr>
<tr>
<td>60°</td>
<td>1524 ± 550</td>
<td>1533 ± 579</td>
<td>0.847</td>
<td></td>
</tr>
<tr>
<td>70°</td>
<td>1944 ± 672</td>
<td>2009 ± 619</td>
<td>0.445</td>
<td></td>
</tr>
<tr>
<td>80°</td>
<td>2161 ± 657</td>
<td>2493 ± 702</td>
<td>0.009*</td>
<td></td>
</tr>
<tr>
<td>90°</td>
<td>2185 ± 654</td>
<td>2668 ± 788</td>
<td>&lt;0.001*</td>
<td></td>
</tr>
</tbody>
</table>

| Knee angle for knee extending phase | P-value | | |
|-----------------------------------|---------|---|
| 0°                                |        |   |
| 10°                               |        |   |
| 20°                               |        |   |
| 30°                               |        |   |
| 40°                               |        |   |
| 50°                               |        |   |
| 60°                               |        |   |
| 70°                               |        |   |
| 80°                               |        |   |
| 90°                               |        |   |

* Significant difference (P < 0.01) between lunge variations or stride variations.

effectively make informed decisions regarding which exercise they choose to employ during patellofemoral rehabilitation.

There are some limitations in the current study. Firstly, MRI knee kinematic data have shown during the weight bearing squat that the femur moves and rotates underneath a relatively stationary patella, and if this femoral rotation is excessive it may result in an increase in patellofemoral stress on the contralateral patellar facets (Doucette and Child, 1996; Li et al., 2004; Powers, 2003). This implies that excessive medial femoral rotation may place more stress on the lateral patellar facets, while excessive lateral femoral rotation may place more stress on the medial patellar facets. Unfortunately, collecting MRI knee kinematic data while performing the lunge is not currently possible due to technology limitations, so it is unknown how much femoral rotation occurs during the lunge, how this rotation varies among individuals (healthy and pathologic), and if femoral rotation occurs in the lunge similar to how it occurs in the squat.

Another limitation is the effect of Q-angle on patellofemoral compressive force and stress. From cadaveric data during a simulated squat it was shown that an increased Q-angle significantly caused a lateral shift and medial tilt and rotation of the patella, which may increase patellofemoral stress (Mizuno et al., 2001). It is currently not feasible to effectively measure lateral shift and medial tilt and rotation of the patellar while performing the lunge. Moreover, increased medial femoral rotation, which also increases Q-angle, is also difficult to measure accurately while performing the lunge.

There are also limitations in the biomechanical model. Firstly, muscle and patellofemoral forces were estimated from modeling techniques and not measured directly, which is currently not possible in vivo. Secondly, patellofemoral stress magnitudes were measured using patellar contact area values from MRI data from the literature and were not measured directly. However, the contact areas used from the literature were determined during loaded weight bearing exercise in healthy male and female subjects, similar to the current study. Moreover, the near linear and direct relationship between contact area and knee angle has been shown to be similar among studies (Besier et al., 2005; Cohen et al., 2001; Hinterwimmer et al., 2005; Patel et al., 2003; Salsich et al., 2003). Thirdly, there are limitations regarding the magnitude of patellofemoral contact areas (and concomitant stress magnitudes), in which the literature reports a wide array of values (Besier et al., 2005; Cohen et al., 2001; Hinterwimmer et al., 2005; Patel et al., 2003; Salsich et al., 2003). Differences in patellar contact area magnitudes and concomitant patellofemoral stress magnitudes among weight bearing studies are due to many factors, such as sex, mass, measuring techniques, and loading magnitudes. Nevertheless, although patellofemoral stress magnitudes during weight bearing exercises or functional activities in the current study are approximations only and not exact values, these varying magnitudes may be helpful to clinicians in deciding interventions to employ during patellofemoral rehabilitation. Finally, the current study was limited to healthy subjects who were able to perform the lunge in the sagittal plane of motion without transverse and frontal plane motions.
Future studies are needed during the lunge and other weight bearing exercises to investigate the effects of transverse plane rotary motions and frontal plane valgus/varus motions on patellofemoral force and stress magnitudes, which may occur with individuals with patellofemoral pathology.

5. Conclusions

Patellofemoral compressive force and stress magnitudes were greater at higher knee angles and smaller at lower knee angles, were greater during the side lunge compared to the forward lunge between 80° and 90° knee angles, and were greater with a stride compared to without a stride between 10° and 50° knee angles. A more functional knee flexion range between 0° and 50° may be appropriate during the early phases of patellofemoral rehabilitation due to lower patellofemoral compressive force and stress during this range compared to higher knee angles between 60° and 90°. Moreover, when the goal is to minimize patellofemoral compressive force and stress, it may be prudent to employ forward and side lunges without a stride compared to with a stride. Understanding differences in patellofemoral compressive force and stress among lunge variations will help clinicians prescribe safer and more effective exercise interventions.

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References


