Forces and Moments on the Knee During Kneeling and Squatting

Jonisha P. Pollard, William L. Porter, and Mark S. Redfern

Euler angle decomposition and inverse dynamics were used to determine the knee angles and net forces and moments applied to the tibia during kneeling and squatting with and without kneepads for 10 subjects in four postures: squatting (Squat), kneeling on the right knee (One Knee), bilateral kneeling near full flexion (Near Full) and bilateral kneeling near 90° flexion (Near 90). Kneepads affected the knee flexion (p = .002), medial forces (p = .035), and internal rotation moments (p = .006). Squat created loading conditions that had higher varus (p < .001) and resultant moments (p = .027) than kneeling. One Knee resulted in the highest force magnitudes and net moments (p < .001) of the kneeling postures. Thigh-calf and heel-gluteus contact forces decreased the flexion moment on average by 48% during Squat and Near Full.

Keywords: restricted workspace, high flexion, thigh-calf contact, heel-gluteus contact

Low-seam coal miners are required to work in low seam heights (1.1 m [42 inches] or less), which restrict their postures, forcing them to kneel, crawl, and squat to perform work. Low-seam coalmines report 10 times as many lost-time injuries related to these postures as high-seam mines, with 75% of these injuries being to the knee (Peters et al., 2001). Knee injuries are also the most prevalent, accounting for 17% of total injury costs to mining companies—more than any other body part, including the lower back (Gallagher et al., 2009). Many low-seam coal miners wear kneepads to protect their knees from the uneven mine floor, which may contain sharp, jagged rocks. Although kneepads may protect the knees from cuts and scrapes, their effect on the internal knee structures is unknown. Castagno (2004) found no indications that kneepads affect long-term gait patterns. However, their effect on knee biomechanics is not known, and it is likely that kneepads affect kneeling posture and thereby knee loading.

Other occupations, such as carpet and floor layers, also require prolonged kneeling and report increased knee injuries (Tanaka et al., 1982). Thun et al. (1987) concluded that "workers who kneel to perform their jobs inflict chronic trauma to their knee joints." Carpet layers are at an increased risk for a variety of knee disorders, including cartilage damage due to frequent and prolonged kneeling and squatting in addition to the use of a knee kicker to stretch carpet (NIOSH, 1990; Habes et al., 1994; Thun et al., 1987; Rytter et al., 2008; Holmstrom et al., 1995). The risk for development of knee osteoarthritis is also increased in occupations requiring kneeling, squatting, climbing, and heavy physical workloads (Manninen et al., 2002; Schneider, 2001).

Previous studies have shown increased knee forces when kneeling in high flexion and squatting. Due to variability in computation methods, knee forces reported in literature exhibit wide variability and have been reported to be as high as 7 times body weight in deep flexion (Nagura et al., 2002; Dahlkvist et al., 1982; Perry et al., 1975; Escamilla et al., 1998; Komistek et al., 2005). However, many of these studies have not examined full-range flexion, and all have neglected the contact between the thigh and shank (thigh-calf contact) that occurs in high flexion.

Thigh-calf contact may have a significant effect on knee forces and moments. Caruntu et al. (2003) modeled the behavior of the knee in a deep squat (165° flexion) to determine the effect of thigh-calf contact. The authors found that neglecting thigh-calf contact resulted in a 700 N overestimation in quadriceps force and a 50% underestimation in medial collateral ligament forces. However, the authors never reported the magnitude or location of the thigh-calf contact force used in their model. Zelle et al. (2007) measured thigh-calf contact pressure distributions using a pressure mapping system. Subjects were instructed to perform two activities. In the first activity, subjects descended from erect standing to a squatting position. In the second, subjects descended from erect kneeling to full flexion with their ankles dorsal flexed.

Jonisha P. Pollard *(Corresponding Author)* and William L. Porter are with the National Institute for Occupational Safety and Health, Office of Mine Safety and Health Research, Pittsburgh, PA. Mark S. Redfern is with the Department of Bioengineering, University of Pittsburgh, Pittsburgh, PA.

Mean knee angles for the final squatting and kneeling postures were 151.8° and 156.4° respectively. Results showed exponentially increasing contact pressures with increasing knee flexion and maximal forces at terminal flexion. Resultant thigh-calf contact forces were shown to be >30% body weight and located within 17 cm of the epicondylar axis with contact occurring more proximal to the tibia when squatting compared with kneeling. In a later investigation, Zelle et al. (2009) suggested that thigh-calf contact considerably reduced the joint forces during high-flexion squatting. Results showed compressive force to decrease from 4.89 to 2.90 times body weight when accounting for thigh-calf contact. In addition, authors found maximum forces to occur at the initiation angle of thigh-calf contact instead of at maximal flexion. This implies that knee forces may be greater in lower flexion kneeling postures than in full-range kneeling and squatting due to thigh-calf contact.

The purpose of this study was to quantify the net forces and moments applied to the tibia and to determine the knee angles associated with kneeling and squatting. In addition, the impact of different kneeling and squatting postures and kneepad use on the tibial forces and moments were examined. Thigh-calf and heel-gluteus contact forces were also measured and incorporated into the biomechanical model. This information may be useful in developing guidelines for mining and other occupations requiring frequent and prolonged kneeling.

Methods

Subjects

Ten subjects (7 male, 3 female) recruited from the National Institute for Occupational Safety and Health (NIOSH) in Pittsburgh, PA, participated in this study. The average ± *SD* of age, weight, and height were 34 ± 17 years, 683 ± 98 N, and 169 ± 8 cm, respectively. Subjects were healthy, with no history of knee surgery. Before participating, each subject read and signed an informed consent form approved by the NIOSH Human Subjects Review Board.

Experimental Design

A split-plot design was used to evaluate the effect of posture and kneepad on the angles, forces, and moments at the right knee joint. Subjects performed four postures: 1—Squatting (Squat), 2—bilateral kneeling near 90° flexion (Near 90), 3—bilateral kneeling near full flexion (Near Full), and 4—kneeling on the right knee (One Knee) during three kneepad conditions: (1) no kneepad, (2) articulated, and (3) nonarticulated (Figure 1). Testing was blocked by kneepad condition, which was randomized, and, within these blocks, the testing order of postures was also randomized. One trial per kneepad condition was collected for each posture.

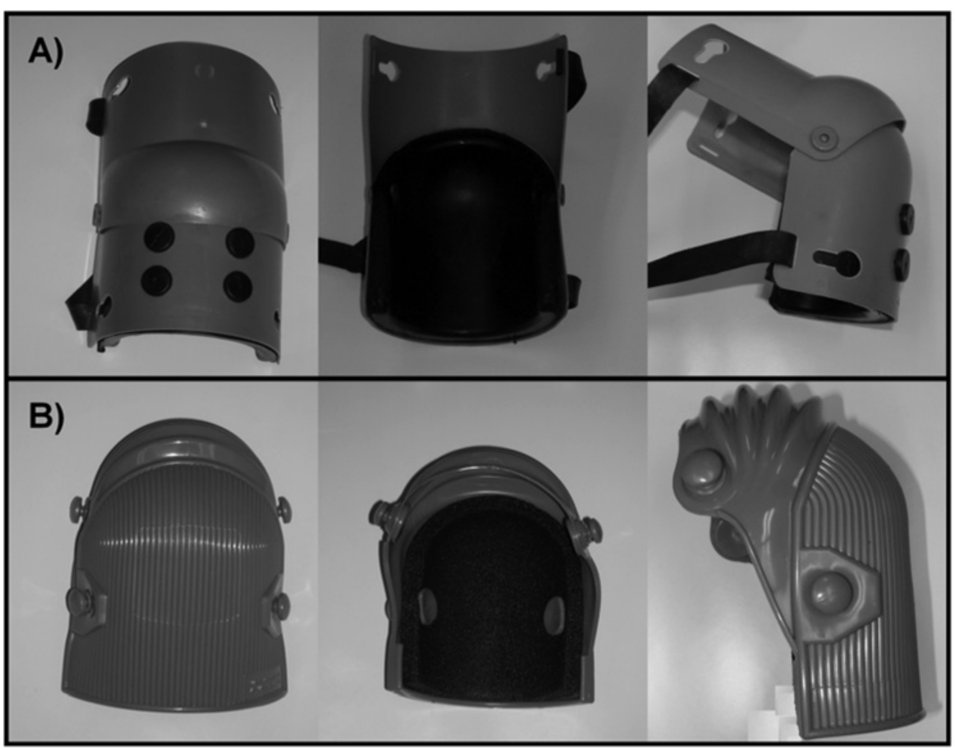

Figure 1 — Kneepads used in study. Outer shell, inner padding, and side views of articulated kneepad (A) and nonarticulated kneepad (B).

Preliminary Measurements

Preliminary measurements were taken before subject testing, including subject height, weight, and thigh-calf and heel-gluteus contact forces. Pilot testing showed standard repeatability error within 5% body weight for repeated measures of contact forces with and without kneepads. Therefore, contact force measurements were suitable for collection before the start of testing. Pressure profiles of the forces applied normal to the tibia and heel were obtained using the ClinSeat pressure sensing system (Tekscan, South Boston, MA). The sensor pad was placed on the subject's lower leg (from popliteal cavity to heel), and the subject was instructed to squat and kneel near full flexion. The distance from the top of the uppermost sensing unit (placed in the popliteal) to the lateral epicondyle of the femur was then measured and recorded. Pressure data were then collected for 5 s at 4 Hz. Using the supplied software (Advanced ClinSeat, Tekscan, South Boston, MA), the thigh-calf and heel-gluteus contact areas were selected and the average total force and center of pressure (COP) for a 3 s period were determined. The moment arms (distances from knee joint center to COPs along the long axis of tibia) of these contact forces were calculated by adding the distance from the COP to the uppermost sensing unit to the distance from the uppermost sensing unit to the lateral epicondyle of the femur. Thus, these contact forces only contributed to anterior forces and sagittal moments.

Procedure

To ensure kneeling, work area height was restricted to 1.2 m (48 inches). Four postures were selected based upon those observed to be commonly adopted in low-seam mining: Squat, Near 90, Near Full, and One Knee (Figure 2). Subjects performed each posture for three kneepad conditions: no kneepads, articulated kneepads, and nonarticulated kneepads. The kneepads tested were those most frequently purchased from mining distributors in 2007. The articulated kneepads consisted of a hard outer shell with firm inner padding. The nonarticulated kneepads consisted of a soft outer rubber shell and soft inner foam padding.

A motion capture system (Eagle; Motion Analysis Corporation, Santa Rosa, CA) was used to determine the location of the body segments as the subject assumed the four postures at a sampling rate of 60 Hz. With the use of kneepads, it was not possible to place markers on the medial and lateral epicondyles of the knee. Instead, the locations of these markers were determined from a calibration trial. Two marker sets were used: an anatomical marker set and a measured marker set (Figure 3), which were modified versions of the Cleveland Clinic marker set

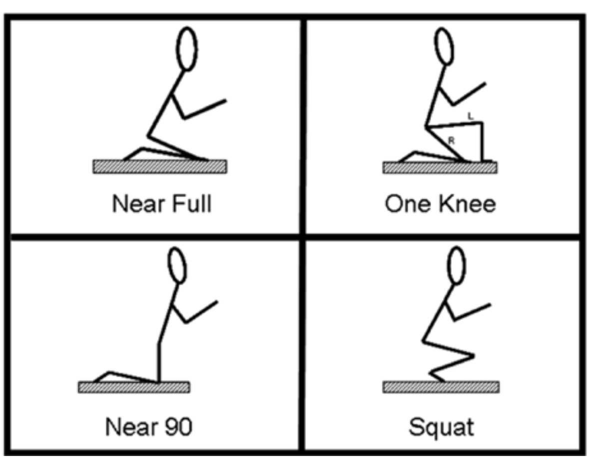

Figure 2 — Schematic showing each posture tested during the experiment. All postures were simulated in a 1.2 m (48 inch) working height.

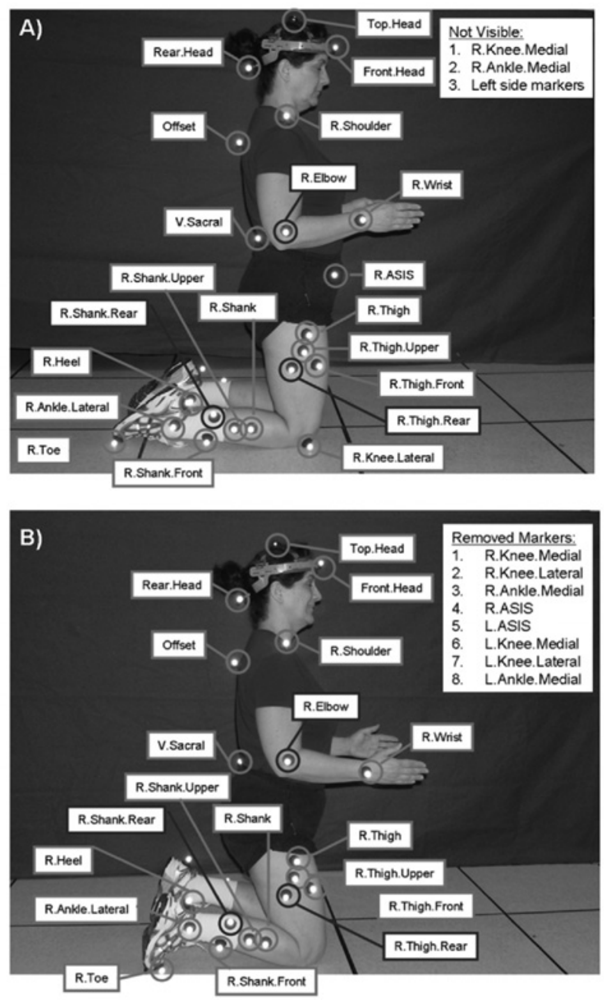

Figure 3 — Anatomical marker set (A) and measured marker set (B) in which eight markers were removed (R.Knee.Medial, R.Knee.Lateral, R.Ankle.Medial, R.ASIS (anterior superior iliac spine), L.ASIS, L.Knee.Medial, L.Knee.Lateral, and L.Ankle.Medial) (R = Right, L = Left).

(Kirtley, 2006). The anatomical marker set was composed of the measured marker set with eight additional anatomical markers. Calibration data of the combined marker sets were collected to acquire the anatomical marker locations relative to the measured markers. During actual testing, the anatomical markers were removed. The estimated locations of the anatomic landmarks were accurate to within an average of 9% for all postures.

Two force plates (Advanced Mechanical Technology, Inc., Watertown, MA) were used to capture the ground reaction forces on the right leg: one plate under the knee and one under the foot. Force plate data were collected at 1020 Hz using the software provided by the Motion Analysis Corporation (EvaRT 5.0, Motion Analysis Corporation, Santa Rosa, CA) through an analog-to-digital board (PCI-6071E, National Instruments, Austin, TX).

Before each trial, the subject was shown a diagram (similar to Figure 2) and asked to assume the posture. Minimal verbal feedback was given unless the subject was assuming a posture with knee angles grossly different from what was expected. Motion capture and force plate data were collected for 10 s. Each trial was performed once and the subject was provided seated rests between trials. Motion data were low-pass filtered using a 4th-order Butterworth to 15 Hz.

Data Analysis

Anatomical and measured coordinate systems were created for the thigh and shank from the calibration trial using the methods from Pollard (2008). The anatomical system allowed the location of the ankle joint center (AJC), knee joint center (KJC), hip joint center (HJC), and lower leg center of mass to be determined. The measured coordinate system was used to estimate the location of the anatomical markers during testing. The AJC was assumed to be midway between the medial and lateral malleoli of the tibia (measured by the medial and lateral ankle markers), the KJC was assumed to be midway between the medial and lateral epicondyles of the femur (measured by the medial and lateral knee markers) and the HJC was approximated using regression equations proposed by Bell et al. (1990).

The anatomical coordinate system of the shank was constructed such that the Z-axis was oriented in the superior direction of the right tibia (AJC to KJC vector), the Y-axis was in the anterior direction (from the cross product of the AJC to KJC vector and medial epicondyle to lateral epicondyle vector), and the X-axis was in the lateral direction (from the cross product of the Y and Z axis vectors). Similarly, the anatomical coordinate system of the thigh was constructed. The rotation matrix from the anatomical thigh to the anatomical shank coordinates was created to determine the Euler angles. Knee angles were estimated for each posture using Euler angle decomposition, with the decomposition order being first about the X-axis (included flexion angle), followed by the Y-axis (varus/valgus), and the Z-axis (internal/external rotation).

Forces and moments at the knee were calculated using ground reaction forces, lower leg weight, thigh-calf contact force ($F_{t/c}$), and the heel-gluteus contact force ($F_{h/g}$). The center of mass location and weight of the shank (W_{shank}) were determined using equations from Clauser et al. (1969) adjusted to use the KJC and AJC in this model (Hinrichs, 1990). Ground reaction forces at the foot (F_1) and knee (F_2) were also included. External force diagrams for all postures are shown in Figure 4.

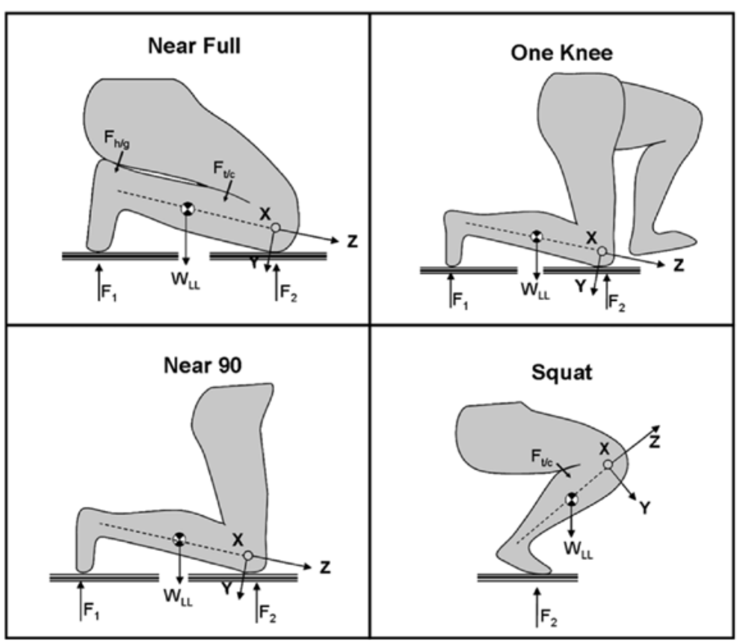

Figure 4 — External force diagrams with respect to the anatomical shank coordinate system (ASCS). The ASCS is oriented such that the X-axis is in the lateral direction, the Y-axis is in the anterior direction, and the Z-axis is in the superior direction of the right tibia. Note that squatting includes the thigh-calf contact force, as does kneeling near full flexion in addition to the heel-gluteus contact force.

Forces and moments applied to the tibia were calculated in three dimensions with respect to the anatomical shank coordinate system (ASCS). All resulting forces were normalized to body weight (BW) and moments were normalized to the product of body weight and height (BW*Ht) (Moisio et al., 2003).

Statistical Analyses

Statistical analyses were performed using Statistix 8.0 for Windows. A two-way (3 kneepads × 4 postures) split-plot ANOVA was performed to determine whether significant differences existed in knee angles, forces, and moments among the postures and kneepad conditions. A priori orthogonal contrasts included comparisons of the no-kneepad to kneepad-present (articulated and nonarticulated) conditions, high-flexion (Squat and Near Full) and lower-flexion (Near 90 and One Knee) postures, and Squat versus kneeling (Near 90, One Knee and Near Full). All orthogonal contrasts were tested using a t statistic. An alpha value of 0.05 was used. A priori nonorthogonal contrasts were tested to determine whether significant differences existed between the kneepad conditions using the Scheffé F test ($p < .05$).

Results

Thigh-calf and heel-gluteus contact forces and moments are shown in Table 1. All subjects had thigh-calf contact forces greater than 20%BW for Squat with a mean of 39 ± 14%BW. In the Near Full posture, seven subjects had contact between the heel and gluteal muscles, with a mean of 11 ± 6%BW. Pressure distribution profiles for Squat and Near Full are shown in Figure 5.

Significant differences were seen between postures for the included ($p < .001$) and internal rotation angles ($p < .001$) (Figure 6). When compared with the kneeling postures, Squat showed higher flexion and internal rotation ($p < .001$). A priori orthogonal contrasts showed significant differences in knee flexion between the high-flexion and lower-flexion postures ($p < .001$) and between Near 90 and One Knee ($p < .001$). Differences in internal rotation angles were seen between high-flexion and lower-flexion postures ($p < .001$) and between Squat and Near Full ($p < .001$). The no-kneepad condition showed higher knee flexion (included angles = 35 ± 24°) than the nonarticulated kneepad condition (included angles = 41 ± 26°) ($p = .033$). No significant differences were seen between the articulated (40 ± 26°) and nonarticulated kneepads.

Posture had a significant effect on tibial forces (Figure 7). When compared with the kneeling postures, Squat showed significantly higher superior and lower posterior and resultant forces ($p < .001$). Lower-flexion postures showed greater posterior, inferior, and resultant forces than the high-flexion postures ($p < .001$). Squatting showed significantly higher medial ($p = .002$), superior ($p < .001$), and resultant forces ($p < .001$) and lower posterior forces than Near Full ($p < .001$). One Knee showed significantly higher medial, posterior, inferior, and resultant force magnitudes than Near 90 ($p < .001$). The use of kneepads affected the medial forces ($p = .036$) with the no-kneepad condition (–6.1 ± 8.1%BW) having

Table 1 Thigh-calf and heel-gluteus contact forces, as a function of percent body weight (%BW), and sagittal moment, as a function of body weight times height (%BW*Ht) contributions for the Squat and Near Full postures

Subject	Squat		Near Full			
	Thigh-Calf Contact		Thigh-Calf Contact		Heel-Gluteus Contact	
	Force	Moment	Force	Moment	Force	Moment
1	45	4.9	31	2.7	13	3.4
2	24	2.8	19	1.5	8	2.0
3	20	1.8	32	3.0	3	0.8
4	62	8.6	53	6.7	No contact	
5	57	5.1	8	0.5	16	3.7
6	50	5.1	39	3.9	5	1.3
7	36	3.2	26	2.5	10	2.4
8	33	3.2	13	1.2	20	4.8
9	27	2.4	23	2.4	No contact	
10	33	2.3	37	2.6	No contact	
Mean (*SD*)	39 (14)	3.9 (2)	28 (13)	2.7 (1.7)	11 (6)	2.6 (1.4)

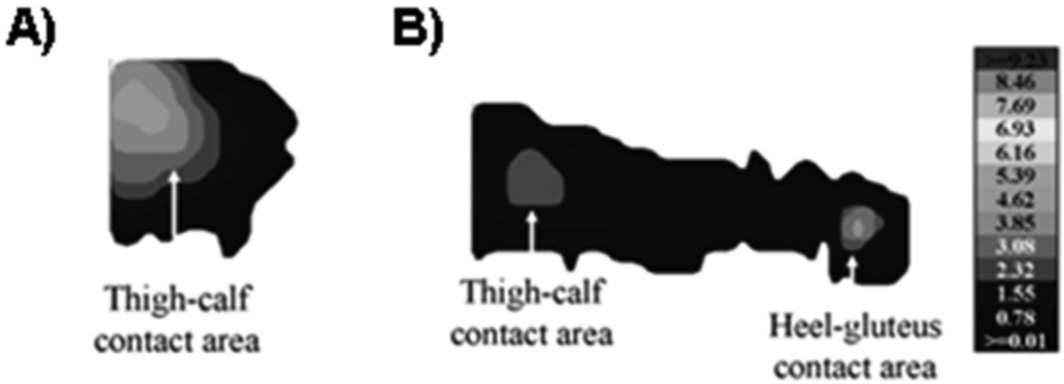

Figure 5 — Typical pressure distribution (psi) for Squat (A) and Near Full (B).

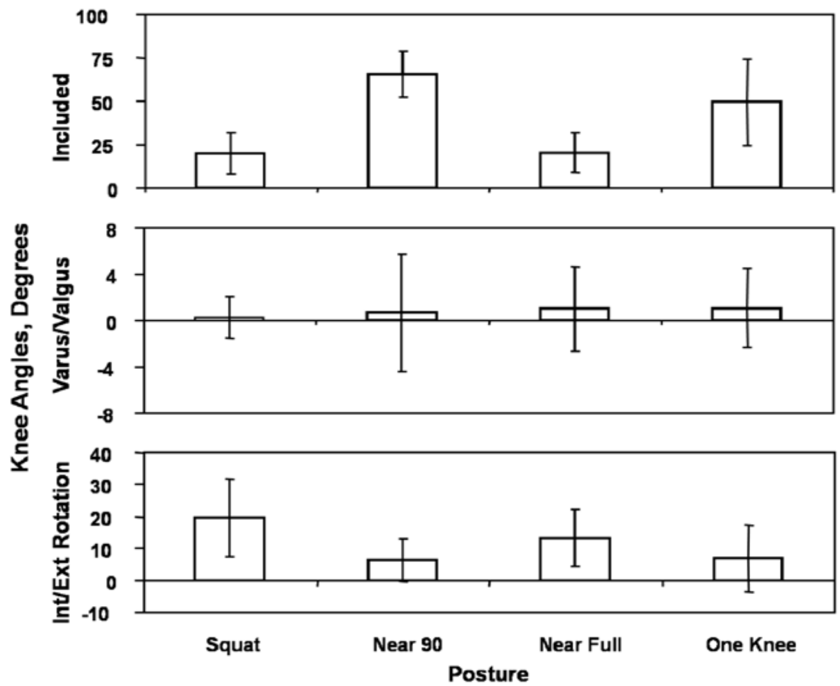

Figure 6 — Knee angles (degrees) for all postures. Varus and internal rotation angles are positive.

higher medial forces than the articulated kneepad condition (–4.3 ± 7.6%BW).

Posture also had a significant affect on tibial moments (Figure 8). Flexion moments had the greatest magnitudes for all postures. Of the postures examined, One Knee had the greatest flexion moments, Squat had the greatest varus moments and Near 90 had the greatest internal rotation moments. One Knee and Squat had the highest resultant moments with comparable magnitudes of 5 ± 4%BW*Ht and 5 ± 2%BW*Ht, respectively. When compared with the kneeling postures, Squat showed significantly lower internal rotation (p = .027) and increased varus (p < .001) and resultant moments (p = .027). High-flexion postures showed significantly higher varus (p < .001), internal rotation (p = .009), and resultant moments compared with the lower-flexion postures (p = .007). Squat showed significantly higher varus moments than Near Full (p < .001). One Knee showed significantly higher flexion (p < .001), varus (p = .012), and resultant moments and significantly lower (p < .001) internal rotation moments than Near 90 (p = .01). Kneepad use affected the internal rotation moments (p < .001), with the no-kneepad condition (0.23 ± 0.32%BW*Ht) having significantly smaller internal rotation moments than the nonarticulated (0.36 ± 0.36%BW*Ht) (p = .049) and articulated kneepad conditions (0.36 ± 0.35%BW*Ht) (p = .008).

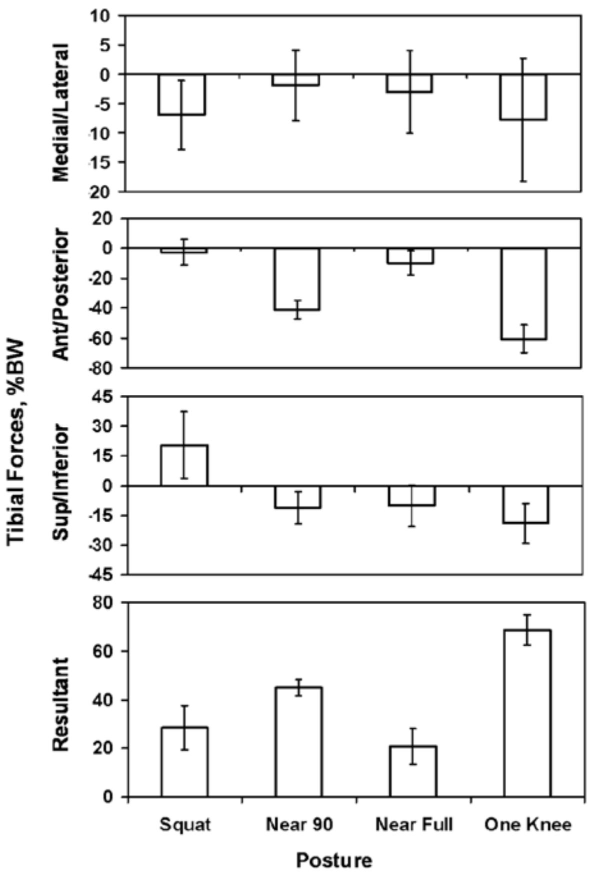

Figure 7 — Mean tibial forces normalized to body weight (BW). Lateral, superior, and anterior are positive.

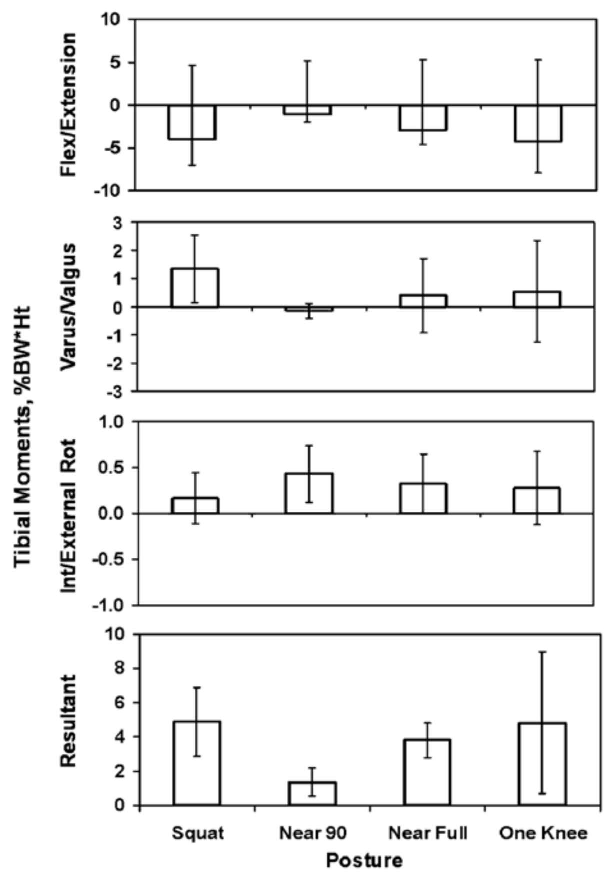

Figure 8 — Mean tibial moments normalized to body weight times height (BW*Ht). Extension, varus, and internal rotation moments are positive.

Discussion

In this study, knee angles and net forces and moments applied to the right tibia were determined while in kneeling and squatting postures with and without kneepads. Significant differences were found between squatting and kneeling postures as well as between the high-flexion (Squat and Near Full) and lower-flexion (Near 90 and One Knee) postures. The results present several important findings: (1) tibia loading when squatting is significantly different than when kneeling; (2) the highest shear loading across the knee joint (i.e., medial forces) occurred when kneeling on one knee; (3) heel-gluteus contact force significantly affects tibial forces and moments; and (4) the use of kneepads affect joint loading when kneeling.

In all the kneeling postures examined, the forces acting on the tibia occurred in the medial, posterior, and inferior directions. Squatting created a different loading pattern with mean forces acting in the medial, posterior, and superior directions of the tibia. When squatting, subjects were found to exhibit significant internal tibial rotation, which may be the primary cause of the increased varus moments. These varus moments increase the loading to the medial compartment of the knee. In high flexion, tibiofemoral contact occurs at the posterior aspects of the femur and tibia with greater loading on the medial compartment (Hefzy et al., 1998; Thambyah et al., 2005). Although the articular cartilage and meniscus are designed for high compressive loads in the central portion, the effects of high loading on the posterior portions of the medial compartments remains unclear. An increased risk for medial tibiofemoral narrowing has been associated with varus alignment (Sharma et al., 2001). The increased varus moments (increasing loading on the medial compartment of the tibia), high knee flexion (decreasing tibiofemoral contact area), and high posterior and inferior forces found in this study may explain the greater effect of squatting on medial compartment osteoarthritis and the significant relationship between varus moments and osteoarthritis severity (Zhang et al., 2004; Sharma et al., 1998; Cooper et al., 1994; Anderson & Felson, 1988). These previous findings, in conjunction with the current study's results, imply that deep squatting is likely to be the most detrimental high-flexion postureOf all postures examined, One Knee had the highest resultant force magnitudes. It was noted that when kneeling on the right knee, a predominant portion of the subjects' weight was transmitted through their right knee and foot. Therefore, it is believed that in the One Knee posture, the knee making contact with the floor is likely to be primarily weight bearing and the other leg may act solely to maintain balance. In effect, high posterior forces, as much as 60%BW are transmitted to the tibia in addition to flexion moments up to 5%BW*Ht. During flexion, the weight-bearing surfaces of the knee move posterior to the tibial plateaus and decrease in area which may increase the stresses transmitted to the tibia. (Maquet et al., 1975; Thambyah et al., 2005) This may create a detrimental loading condition for the articular cartilage, which is thought to be damaged by maximum shear stress (Atkinson et al., 1998). High shear forces combined with high flexion moments are believed to make kneeling on one knee the most detrimental of the kneeling postures examined.

Previous research has shown thigh-calf contact to have a significant effect on model predictions of joint forces and moments (Caruntu et al., 2003; Zelle et al., 2009). To the authors' knowledge, Zelle et al. (2007) was the first study to quantify thigh-calf contact forces. The current study expanded on this knowledge, quantifying the additional heel-gluteus contact force found when kneeling near full flexion. Although this force had smaller magnitudes than thigh-calf contact, its effect on the flexion moment was comparable. The larger moment arm of the heel-gluteus contact force acted to create an extension moment that counteracted the flexion moment created by the ground reaction forces at the knee and foot. The results of the current study imply that heel-gluteus contact may also reduce loading of knee tissues as has been shown for thigh-calf contact (Zelle et al., 2009). Although knee kinematics were determined in three dimensions, due to the limitations of the pressure sensor, thigh-calf and heel-gluteus contact forces were obtained in 2-D. These contact forces were measured normal to the tibia and thus the medial-lateral and superior-inferior shear forces caused by tissue deformation were neglected, thereby affecting the accuracy of the estimated shear forces and internal rotation and varus moments.

High-flexion kneeling and squatting create a significant change in internal joint structure orientations, such as femoral rollback and varying segment contact forces. During increased knee flexion, the tibia internally rotates changing the orientation of the knee joint center to a position that may only be determined using imaging techniques not employed in this study. In effect, this shift in knee joint center location may have caused inaccuracies in moment calculations, as the net moment was assumed to act at a point in the center of the epicondylar axis. However, the 3-D model used in this study may minimize this error by accounting for tibial rotations.

The current study is the first to imply that kneeling with kneepads affects posture and joint loading. While the smooth, even laboratory floor did not create a kneeling environment comparable to that of a coalmine, it is believed that the general findings within this study hold for other environments. However, the increase in transverse moments associated with the hard contoured outer shell kneepad may create deleterious effects that may be reduced when wearing kneepads with a softer outer shell.

The thigh-calf contact forces measured in this study (39%BW for Squat and 28%BW for Near Full) are comparable to the values found in literature. Zelle et al. (2007) reported average thigh-calf contact forces of 34.2%BW and 30.9%BW for squatting and kneeling near full flexion, respectively. However, subjects knelt with ankles dorsally flexed, which was not used by all subjects in the current study and may explain the presence of heel-gluteus contact. Foot posture has been shown to significantly affect varus knee moments during gait (Teichtahl et al., 2006; Lynn et al., 2008), and it is possible that the differences in foot posture across subjects may have increased the internal rotation of the tibia and thereby the varus moments.

Although flexion angles may be of similar magnitude in different postures when kneeling in restricted workspaces, the biomechanical stresses on the knee differ significantly. Squatting and kneeling near full flexion have similar joint flexion requirements; however, squatting was found to have much great forces and moments than kneeling near full flexion. Thus, posture of the entire body, as well as simple knee flexion angle, are important determinants of knee loading characteristics.

Reducing the loads and moments on the knee are assumed to provide a reduction in risk of injury. The results of this study lead to three key recommendations toward reducing forces and moments on the joint during kneeing. (1) Equipment that increase thigh-calf or heel-gluteus contact forces will lead to decreased loading at the knee. (2) When kneeling in high flexion, internal tibial rotations should be minimized to reduce loading on the medial meniscus and articular cartilage, thereby reducing the risk for medial compartment osteoarthritis. This could be accomplished by limiting ankle rotation. (3) Postural rotation strategies (i.e., changing postures throughout the day) can reduce any cumulative effect on tissues. Kneeling on one knee and squatting have particularly high forces and moments, and so should not be sustained for long periods of time.

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