

Tibiofemoral Joint Contact Force in Deep Knee Flexion and Its Consideration in Knee Osteoarthritis and Joint Replacement

Takeo Nagura,¹ Hideo Matsumoto,¹ Yoshimori Kiriya,¹
Ajit Chaudhari,² and Thomas P. Andriacchi²

¹Keio University and ²Stanford University

The aim of the study was to estimate the tibiofemoral joint force in deep flexion to consider how the mechanical load affects the knee. We hypothesize that the joint force should not become sufficiently large to damage the joint under normal contact area, but should become deleterious to the joint under the limited contact area. Sixteen healthy knees were analyzed using a motion capture system, a force plate, a surface electromyography, and a knee model, and then tibiofemoral joint contact forces were calculated. Also, a contact stress simulation using the contact areas from the literature was performed. The peak joint contact forces ($M \pm SD$) were 4566 ± 1932 N at 140 degrees in rising from full squat and 4479 ± 1478 N at 90 degrees in rising from kneeling. Under normal contact area, the tibiofemoral contact stresses in deep flexion were less than 5 MPa and did not exceed the stress to damage the cartilage. The contact stress simulation suggests that knee prosthesis having the contact area smaller than 200 mm² may be problematic since the contact stress in deep flexion would become larger than 21 MPa, and it would lead damage or wear of the polyethylene.

Key Words: knee joint loading, motion analysis, computer model

The tibiofemoral joint experiences significant mechanical loads in daily activities. Since direct measurement of the knee joint force is technically and ethically difficult, most of the in vivo studies have used a combination of motion analysis and modeling to calculate the joint force. Estimated tibiofemoral joint contact force ranged from 2–4 times body weights (1500–2500 N) in walking (Morrison, 1970; Taylor et al., 2004) to 5–7 times body weights (3500–6000 N) in rehabilitation exercises such as squat (Dahlkvist et al., 1982; Ellis et al., 1979; Escamilla et al., 1998; Lutz et al., 1993; Toutoungi et al., 2000; Wilk et al., 1996). However, these studies did not analyze knee motion larger than 140 degrees. Current studies revealed that the tibiofemoral joint decreases its contact area in deep flexion. At 133 degrees of flexion, the contact between the femur and tibia places at the posterior edge on the tibial plateau and the lateral femoral condyle may be regarded as being posteriorly subluxed at full flexion (Nakagawa et al., 2000; Li et al., 2004). Thus, the medial compartment, especially the medial meniscus, carries most of the mechanical load in full flexion motion. This characteristic may explain why degenerative tear is most frequently seen in the posterior portion of the medial meniscus, while there is a lack in joint loading data during such knee flexion. In addition, deep knee bending is considered as one of the risk factors to cause tibiofemoral osteoarthritis. The recent epidemiological studies of large populations in Europe, the United States, and China have shown the strong relationship

Nagura, Matsumoto, and Kiriya are with the Department of Orthopedic Surgery, Keio University, Shinjyuku, Tokyo, Japan; Chaudhari is now with the Department of Orthopaedics, Ohio State University, Columbus, OH 43210; and Andriacchi are with the Department of Mechanical Engineering, Stanford University, Stanford, CA 94305-4038.

between tibiofemoral osteoarthritis and deep knee bending (Coggon et al., 2000; Felson et al., 1991; Zhang et al., 2004). However, it is not clear how the joint loads in deep flexion play a role in leading to the degenerative change. A recent *in vitro* study (Thambyah et al., 2005) suggested that the joint contact stress in simulated deep flexion may reach the damage limits of cartilage; however, loading conditions were derived from different studies and knee flexion was limited to 120 degrees. Thus, there is a need to analyze the tibiofemoral joint loading in the motions that include full range of flexion.

Mechanical load at the knee in high flexion becomes an important issue in current total knee arthroplasty (TKA). Many “high-flexion” designs are commercially available to satisfy the patients’ expectations (Li et al., 2004; Weiss et al., 2002). However, elevated contact stress on the polyethylene leads to destructive wear process. Many studies have shown that the contact stress on the implant exceeded the yield point of the polyethylene inlay under the physiological loads (Chapman-Sheath et al., 2003; Harris et al., 1999; Kuster et al., 1997; Stukenborg-Colsman et al., 2002; Nakayama et al., 2005). In particular, the contact stress in 90 degrees or more may be considerable for the implant. One reason is thought to be limited contact areas that result in higher contact stress in that flexion with most of the current prosthesis designs (Chapman-Sheath et al., 2003; Stukenborg-Colsman et al., 2002; Nakayama et al., 2005). Therefore, evaluation of the tibiofemoral joint force in deep flexion is necessary to simulate the contact stress and to provide rationale about how much contact area is required in the prosthesis to sustain the mechanical loads.

The aim of this study was to evaluate the tibiofemoral joint forces on healthy knees in deep flexion up to 155 degrees. To consider the mechanical effect in the joint, a contact stress simulation was also performed. We hypothesize that the joint force should not become large enough to damage the joint under normal contact area, but should become deleterious to the joint under the limited contact area.

Methods

Sixteen healthy subjects (seven women and nine men) with no history of knee pain or injury were tested at the gait laboratory. All subjects read and signed a consent form, which was approved by

Institution Review Board at Stanford University. The subjects had a mean age of 32 years (range 27–42), a mean height of 1.70 m (range 1.52–1.80), and a mean weight of 597 N (range 430–855). An opto-electronic motion capture system (Qualysis, Svedalén, Sweden) with a multicomponent force plate (Bertec, Columbus, OH) and a four-channel surface EMG (Synergy Lab, Sensory Motor Performance Program, Chicago, IL) were used to capture synchronized motion, force, and EMG data at 120 Hz respectively. To reduce the test burden on a subject, only the left leg was tested. The markers were placed at superolateral aspect of the iliac wing, lateral aspect of the greater trochanter, lateral joint line of the knee, lateral malleolus, lateral calcaneus, and lateral head of the fifth metatarsal. The three-dimensional joint kinematics, net (external) knee moments, and net (external) knee forces were calculated using a six-marker link model (Nagura et al., 2002). In the model, a local anatomical coordinate system on the tibia was used to represent the force directions. The long axis of the tibia was defined as the superior-inferior axis. The link model included the assumption that no axial rotation occurred about the long axis of each segment. The activities of four limb muscles (rectus femoris, vastus medialis, hamstrings, and medial head of gastrocnemius) on the left limb were recorded with two electrodes (ConMed, Utica, NY) placed over each muscle belly with the centers 2 cm apart. The activity of the hamstrings was recorded at the mid point between lateral and medial hamstrings. The signal level was normalized to the signal during the maximum voluntary contraction (MVC), which was performed as indicated in the text (Lacote et al., 1987). The average signals of the rectus femoris and vastus medialis were used to indicate the quadriceps activity.

The subjects performed the trials of 10 m of level walking, stair climbing onto the two platforms 25.5 cm in height each, rising up from a kneeling position with one leg (kneeling), and rising up from a full squatting position with both legs (full squat). In the kneeling activity, the flexion angle was approximately 90 degrees during a kneeling position with both knees on the floor. Then, the subjects lifted the left leg from the ground, and stood up using that leg. In the full squat activity, the subjects started in a full squatting position with both knees on the floor, and then stood up with both legs. The maximum knee flexion was approximately

100 degrees during the kneeling, and 155 degrees during the full squat. After several practices, one trial was recorded in each activity. Stance phase of each activity was selected for analysis.

A statically determinant knee model was used to compute tibiofemoral joint compressive/shear forces, muscle forces, and the ligament forces around the knee. This 2-D planar knee model was based on anatomical measurements of eight cadaver knees, and included three muscle groups (the quadriceps, hamstrings and gastrocnemius), the lateral and medial collateral ligaments, and the cruciate ligaments (Schipplein et al., 1991). Lines of action for the three muscle groups were determined from 13 muscles (4 muscles in the quadriceps, 6 muscles in the hamstrings, and 3 muscles in the gastrocnemius) in the lower limb. Inputs to the model were

knee flexion angle, net knee moments, net knee forces, and ratio of knee extensor/flexor activity level (Figure 1). The net knee moments and forces were determined by the motion analysis data. The net knee moments consisted of net flexion-extension moment and net abduction-adduction moment. The net knee forces consisted of axial and anterior-posterior loads. The model allowed the tibial-femoral contact point to change with knee flexion. The forces due to the three muscle groups resisted the net flexion-extension moment. Normalized EMG signals of the quadriceps, hamstring, and gastrocnemius were adopted to determine the ratio of knee extensor/knee flexor activity level. The root mean squares of the EMG signals from the three muscles were averaged over 0.5 second, and activity levels of the hamstrings and gastrocnemius relative to the quadriceps were

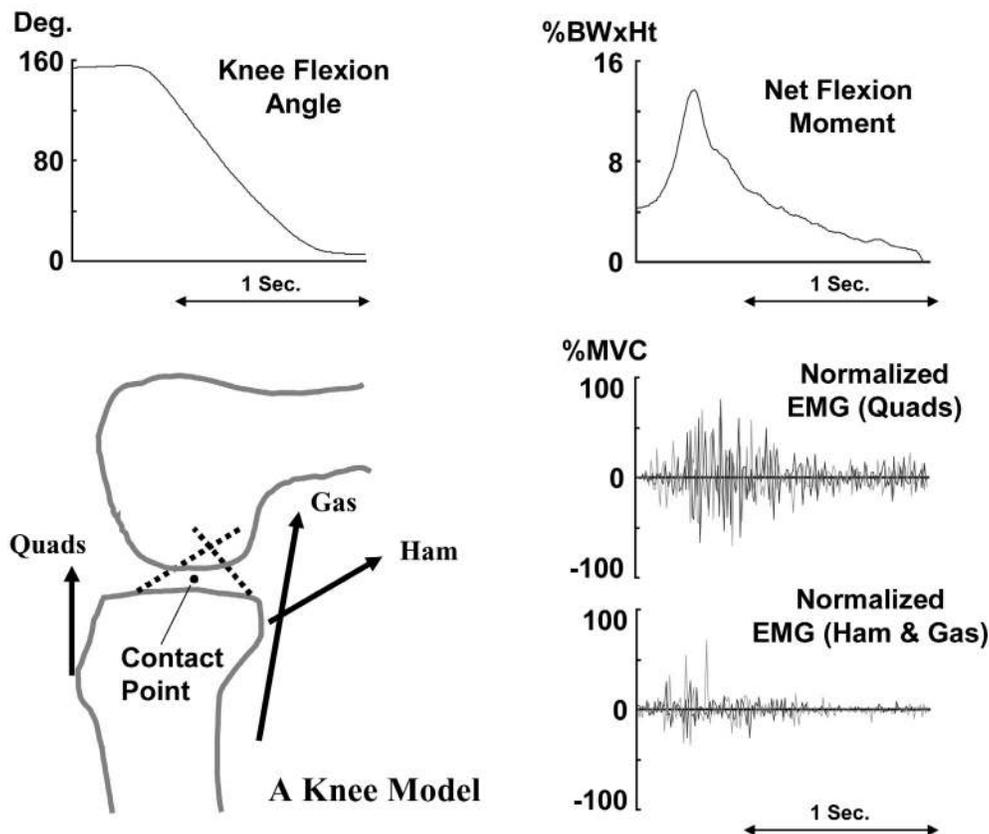


Figure 1 — A model included the three muscle groups and the ligaments at the knee. The changes in the tibiofemoral contact point with knee flexion angle, and resulting changes in moment arms and force directions of the muscles were modeled. Net extension-flexion moment and normalized EMG signals were used to determine the forces in the three muscle groups. The graphs indicate motion analysis data and EMG signals during rising from full squat in a subject. EMG signals from the quadriceps (Quads; vastus medialis = black line, rectus femoralis = gray line), and those from the hamstrings (Ham; gray line) and the gastrocnemius (Gas; black line) are indicated. *Abbreviations:* Deg. = degrees, Sec. = second, %BWxHt = % body weight times height, %MVC = % maximum voluntary contraction.

determined as the levels of co-contraction. Total tibiofemoral joint forces and the forces on the ligaments were calculated to maintain force equilibrium with the net knee forces and the muscle forces in sagittal and coronal planes. In each plane, the long axis of the tibia was defined as the superior-inferior axis. The joint forces in sagittal plane were shown as the two components of the forces: axial forces acting along the long axis of the tibia (compressive force) and the forces acting perpendicular to the axial force (shear force). Posterior shear force indicates the force tends to push the tibia to posterior direction with respect to the femur.

To see the differences in the forces, the data were evaluated and compared at every 5 degrees of flexion as the knee moved from maximum flexion to maximum extension. Peak values of knee forces were also compared among four different activities. An ANOVA with a single factor for two groups was performed each time, to test the statistical difference between any two of the activities, and p value less than 0.05 was considered as significant difference.

Results

The tibiofemoral joint contact forces became greater in higher flexion angles during the deep flexion activities (Figures 2 and 3). The peak contact forces were 4566 ± 1932 N at 140 degrees in full squat and 4479 ± 1478 N at 90 degrees in kneeling. The joint contact forces that occurred in either deep flexion activity were greater than the maximum value of the forces during walking or stair climbing.

The peaks of the joint and muscle forces occurred between 10 and 20 degrees during walking, 30 and 50 degrees during stair climbing. On the other hand, the peaks of the forces (except for the anterior shear force) occurred between 80 to 90 degrees during kneeling and 140 and 150 degrees during full squat (Table 1). The peak joint compressive force, posterior shear force, and the quadriceps force during kneeling were larger than those during walking, and those during full squat were larger than those during walking and stair climbing. There was no difference in the peak anterior shear force in each activity.

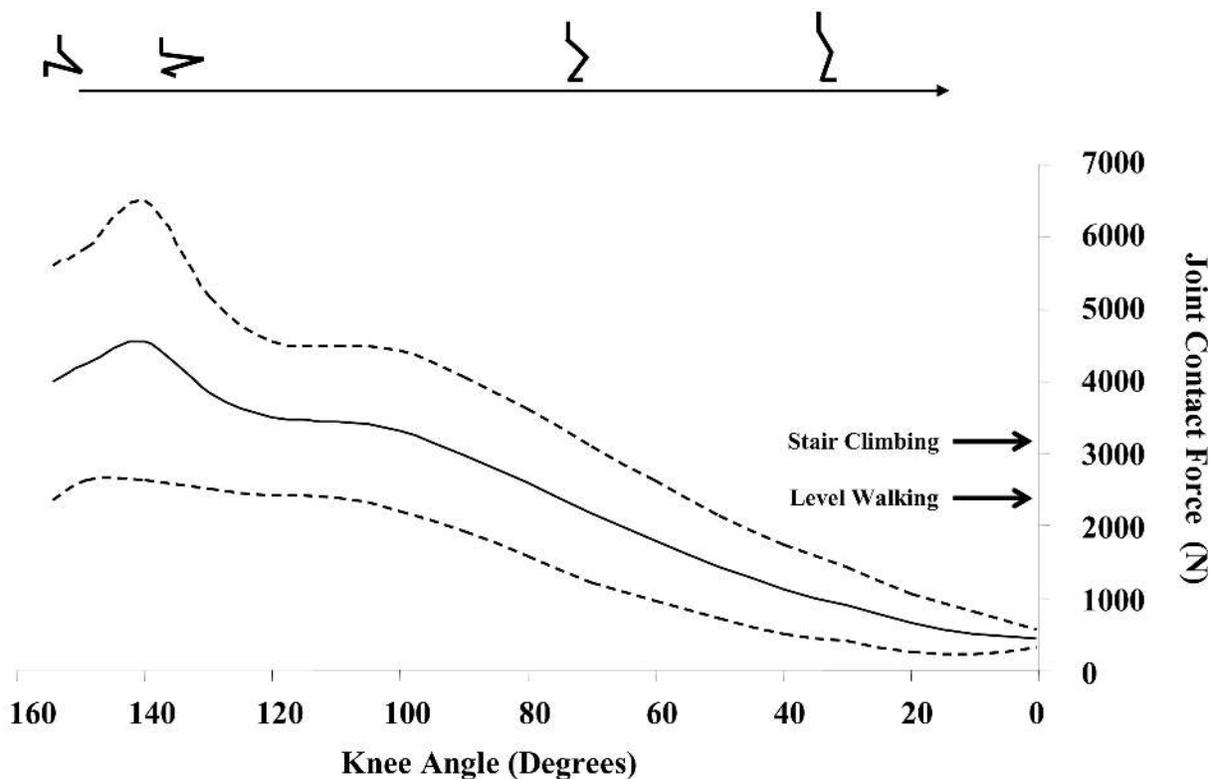


Figure 2 — Joint contact force at the tibiofemoral joint during rising from full squat. A solid line indicates an average over 16 subjects and the dashed lines indicate ± 1 SD. Arrows indicate mean peak joint contact forces during stair climbing and level walking. Stick figures indicate the sagittal image of the limb during the motion.

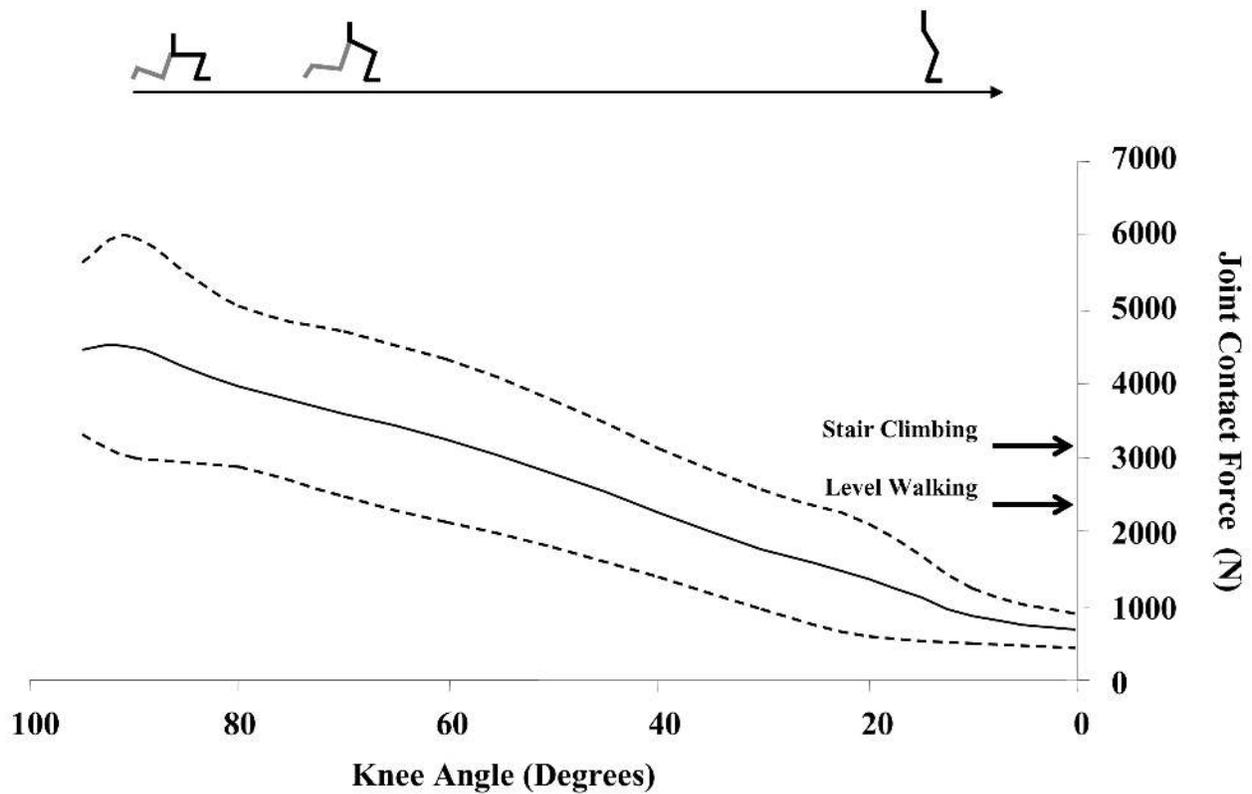


Figure 3 — Joint contact force at the tibiofemoral joint during rising from kneeling. A solid line indicates an average over 16 subjects and the dashed lines indicate ± 1 SD. Arrows indicate mean peak joint contact forces during stair climbing and level walking. Stick figures indicate the sagittal image of the limb during the motion.

Table 1 Peak Tibiofemoral Joint Forces, Quadriceps Force and Knee Angle During Each Activity (Mean, SD)

Activity	Compressive Force	Anterior shear force	Posterior shear force	Quadriceps force
Walking				
×BW	4.0 (1.2)	0.6 (0.2)	0.3 (0.1)	2.5 (0.6)
N	2355 (659)	355 (90)	188 (66)	1450 (287)
Knee angle	at 17.5°	at 16.8°	at 13.4°	at 17.5°
Stair climbing				
×BW	5.3 (1.7)	0.2 (0.2)	0.6 (0.3)	3.9 (1.2)
N	3096 (999)	146 (125)	346 (165)	2280 (758)
Knee angle	at 50.8°	at 50.8°	at 33.4°	at 50.8°
Kneeling				
×BW	6.3 (1.3)*	0.2 (0.2)	1.7 (0.5)* **	4.5 (1.0)*
N	3709 (977)*	111 (116)	995 (312)* **	2658 (777)*
Knee angle	at 83.5°	at 40.9°	at 88.0°	at 83.5°
Full squat				
×BW	7.3 (1.9)* **	0.1 (0.1)	4.9 (1.6)* **	4.5 (1.1)*
N	4470 (1825)* **	66 (52)	3005 (1339)* **	2768 (1085)*
Knee angle	at 146.3°	at 10.9°	at 143.7°	at 146.3°

Note. ×BW = times body weight.

*Statistically different from walking. **Statistically different from stair climbing.

To evaluate the tibiofemoral contact stress that possibly occurs during deep flexion activities, a contact stress estimation has been performed in three different conditions with typical contact areas, simulating average contact stress on a natural knee (1150 mm²) (Fukubayashi & Kurosawa, 1980), a knee without the menisci (520 mm²) (Fukubayashi & Kurosawa, 1980), and a total knee prosthesis (200 mm²) (Chapman-Sheath et al., 2003; Nakayama et al., 2005) (Figure 4). The contact stress analysis revealed that the average stress during deep flexion in a natural knee or a knee without menisci were less than 10 MPa, whereas the average stress in a knee with a 200-mm² contact area became larger than 21 MPa during both kneeling and full squat.

Discussion

In this study, we used motion capture techniques and a statically determinant model that have been used in the previous published works (Nagura et al., 2002; Schipplein et al., 1991). The estimated joint forces during walking show good agreement with the in situ forces measured by prostheses implanted in the patients (Lu et al., 1998; Taylor et al., 2004), and support the validity of our model. The previous in vivo studies reported a combination of high compressive force and posterior shear force at the tibiofemoral joint during various deep flexions (Table 2). Most authors stated that very small anterior shear forces occurred on the knee during deep flexion. Our results also agreed with their results. Although there were up to 36% differences in the calculated forces between the present study and

the studies by Dahlkvist and colleagues (1982) and Wilke and colleagues (1996), the knee flexion range in the activities (up to 155 degrees versus 140 and 100 degrees respectively) and subjects' body size (597 N versus 732 and 912 N respectively) varied in the studies, and those variations should explain the differences in the calculated force.

The contact stress on the joint is determined by the mechanical load and the area of the contact. The reported average contact area of a natural knee joint ranges from 105 to 2013 mm² (Fukubayashi & Kurosawa, 1980; Kettelkamp & Jacobs, 1972; Maquet et al., 1975; Thambyah et al., 2005). It is consistent with the fact that the knee joint has smaller contact area in flexion than in extension. Maquet and coworkers (1975) evaluated the tibiofemoral contact area between 0 to 90 degrees and the area was decreased to 57% at 90 degrees compared to 0 degrees. Thambyah and colleagues (2005) reported that the contact area in deep flexed position was 58% of the maximum contact area measured in the position simulating toe-off of the gait. Removal of the menisci significantly reduced the contact area to 40–50% (Fukubayashi & Kurosawa, 1980; Kettelkamp & Jacobs, 1972). Based on a simulation, the contact stresses in a natural knee or a knee without menisci were less than a stress level (15–20 MPa) that damages cartilage at the joint (Clements et al., 2001) (Figure 4). The results support our hypotheses and do not agree with the results by Thambyah and coworkers (2005), who reported the deleterious stress (>20 MPa) to damage the cartilage in deep flexion. Further study with more detailed contact analysis is required to discuss the

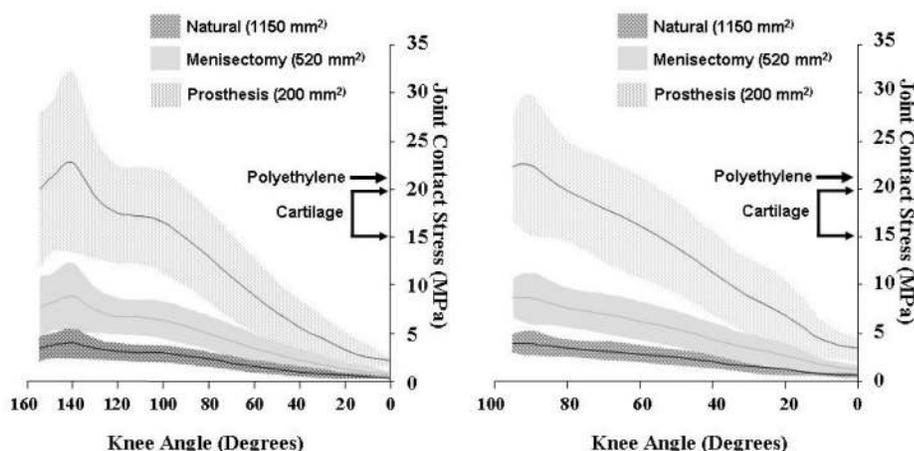


Figure 4 — A joint contact stress simulation with three contact areas under the joint contact forces during full squat (left) and kneeling (right) (average ± 1 SD, averages shown by solid lines). In the simulation, the contact areas were assumed to be constant in all range of flexion. Arrows indicate the tensile yield of polyethylene (21 MPa, Chapman-Sheath et al., 2003) and the stress levels that are known to damage the cartilage (between 15 and 20 MPa, Clements et al., 2001).

Table 2 Comparison of Cited Studies That Evaluated Knee Forces During High Flexion Activities

Activity	Knee flexion range (°)	Mean BW (N)		Mean peak tibiofemoral compressive force	Mean peak tibiofemoral anterior shear force	Mean peak tibiofemoral posterior shear force
Dahlkvist et al., 1982						
Deep squat	0–140	732 ± 79	×BW	5.5 ± 1.7		3.6 ± 0.4
			N		4018 ± 1230	
Ellis et al., 1979						
Rising from chair	0–110	582 ± ?	×BW N	5.1 ± 0.8		3.5 ± 2.6
Escamilla et al., 1998						
Squat	0–95	912 ± 145	×BW	3.4 ± 1.1	N/A	2.0 ± 1.0
			N		3134 ± 1040	
Toutoungi et al., 2000						
Squat	0–100	765 ± ?	×BW		0.1 ± 0.1	3.5 ± 1.4
			N			28 ± 36
Wilke et al., 1996						
Squat	0–100	912 ± 137	×BW	6.7 ± 1.9	N/A	2.0 ± 0.7
			N		6139 ± 1709	
Present study						
Full squat	0–155	597 ± 120	×BW	7.3 ± 1.9	0.1 ± 0.1	4.9 ± 1.6
			N		4470 ± 1825	66 ± 52

Note. ×BW = times body weight.

effect of the deep flexion loads on the joint. In particular, the distribution of the force under the loads is a key to explain the effect. At this point, there is a lack of information about the contact area near full flexion. Although the lack of information limits discussion on the relationship between osteoarthritis and deep flexion loads, one possible explanation is a stress concentration in the medial meniscus. Near maximum flexion, the contact between the femur and tibia occurs mainly at posterior edge in the medial compartment (Nakagawa et al., 2000; Li et al., 2004). The significant contact forces during full squat will result in stress concentration in the posterior horn of medial meniscus. This can lead the process to damage the meniscus and initiate the onset of the degenerative change in the tibiofemoral joint (Wilson et al., 2003; Pena et al., 2005).

The reported contact area of total knee prostheses is between 100 and 850 mm², and there were 30 to 50% reductions in the contact area with flexion of 110 degrees or more (Chapman-Sheath et al.,

2003; Harris et al., 1999; Stukenborg-Colsman et al., 2002; Nakayama et al., 2005). Kuster and colleagues (1997) assumed the contact stress on the prosthesis during downhill walking. They indicated that the contact stress would exceed the yield point of the polyethylene inlays (21 MPa, Chapman-Sheath et al., 2003) and recommended to have more than a 400-mm² contact area with knee prosthesis. Chapman-Sheath and coworkers (2003) reported that there was increase in the peak contact stress with flexion on mobile bearing knees under a 3600-N load. The contact stresses at 110 degrees were larger than 21 MPa in five out of nine designs. Nakayama and colleagues (2005) evaluated the contact stress at the post-cam mechanism in posterior-stabilized prostheses under a posterior force of 500 N. All knees had the peak contact stresses than 30 MPa in flexion beyond 120 degrees. Polyethylene wear can be caused by many factors and excessive contact stress produces positive results in some cases (Barbour et al., 1997). The stress threshold for poly-

ethylene damage can be variable and is not always a single value. Distribution of the pressure based on the prosthesis geometry should be evaluated to see whether the load in deep flexion is excessive. However, a small contact area in flexion will increase the contact stress and surely increases the risk for mechanical failure. The mechanical loads used in the previous contact stress analyses were less than the loads calculated in the present study. This may lead to underestimation of the contact stress on the prosthesis. In a joint with contact area less than 200 mm², the stress during both kneeling and full squat may exceed the yield point of polyethylene (Figure 4). Considering the joint contact force in high flexion, the design protocol for the knee prosthesis should include providing a larger contact area in any flexion range. On this point, we recommend a contact area of more than 200 mm² to safely sustain loads during deep flexion.

As discussed in the above sections, published contact areas were used in our contact stress simulation and the average contact stress was calculated. Using this approach, it is not possible to indicate peak contact stress or stress distribution at the joint. A 2-D knee model in this study has a generic joint anatomy and does not consider anatomical variations among the subjects. Modification of the model (i.e., a subject-specific model) or using a different modeling approach (i.e., finite element model) will be helpful to clear these limitations. Our analysis involved knees with normal function and range of motion. This fact limits the application of our results to TKA patients. Usually, patients following TKA have limitation in the extensor muscle force and range of motion. In many cases, it is not possible to perform the deep flexion described in this study. However, current improvements in surgical technique and prosthesis design have been providing better postoperative function and range of motion for patients. Many patients should be able to acquire knee function close to normal in the near future. Our data will be useful for deriving design criteria for a prosthesis that is capable for full flexion.

Acknowledgments

The authors would like to acknowledge the contributions of Chris O. Dyrby, Eng., and Eugene J. Alexander, Ph.D., for helping with the acquisition and processing of the data. The authors also appreciate Ms. Mayumi Oshidari for her help in this study. The

research materials in this study were partially supported by a grant from Zimmer.

References

- Barbour, P.S., Barton, D.C., & Fisher, J. (1997). The influence of stress conditions on the wear of UHMWPE for total joint replacements. *Journal of Materials Science. Materials in Medicine*, 8, 603-611.
- Chapman-Sheath, P.J., Bruce, W.J.M., Chung W.K., Morberg, P., Gillies, R.M., & Walsh, W.R. (2003). In vitro assessment of proximal polyethylene contact surface areas and stresses in mobile bearing knees. *Medical Engineering & Physics*, 25, 437-443.
- Clements, K.M., Bee, Z.C., Crossingham, G.V., Adams, M.A., & Sharif, M. (2001). How severe must repetitive loading be to kill chondrocytes in articular cartilage? *Osteoarthritis Cartilage*, 9, 499-507.
- Coggon, D., Croft, P., Kellingray, S., Barrett, D., McLaren, M., & Cooper, C. (2000). Occupational physical activities and osteoarthritis of the knee. *Arthritis and Rheumatism*, 43, 1443-1449.
- Dahlkvist, N.J., Mayo, P., and Seedhom, B.B. (1982). Forces during squatting and rising from a deep squat. *Engineering in Medicine*, 11, 69-76.
- Ellis, M.I., Seedhom, B.B., & Wright, V. (1979). Forces in the knee joint whilst rising from a seated position. *Engineering in Medicine*, 6, 33-40.
- Escamilla, R.F., Fleisig, G.S., Zheng, N., Barretine, S.W., & Andrews, J.R. (1998). Biomechanics of the knee during closed kinetic chain and open kinetic chain exercises. *Medicine and Science in Sports and Exercise*, 30, 556-569.
- Felson, D.T., Hannan, M.T., Naimark, A., Berkeley, J., Gordon, G., Wilson, P.W.F., & Anderson, J. (1991). Occupational physical demands, knee bending, and knee osteoarthritis: results from the Framingham Study. *Journal of Rheumatology*, 18, 1587-1592.
- Fukubayashi, T., & Kurosawa, H. (1980) The contact area and pressure distribution pattern of the knee. *Acta Orthopaedica Scandinavica*, 51, 871-879.
- Harris, M.L., Morberg, P., Bruce, W.J.M., & Walsh, W.R. (1999) An improved method for measuring tibiofemoral contact areas in total knee arthroplasty: A comparison of K-scan sensor and Fuji film. *Journal of Biomechanics*, 32, 951-958.
- Kettelkamp, D.B., & Jacobs, A.W. (1972). Tibiofemoral contact area-determination and implications. *Journal of Bone and Joint Surgery*, 54-A, 349-356.
- Kuster, M.S., Wood, G.A., Stachowiak, G.W., & Gachter, A. (1997). Joint load considerations in total knee replacement. *Journal of Bone and Joint Surgery*, 79-B, 109-113.
- Lacote, M., Chevalier, A., Miranda, A. (1987). *Clinical Evaluation of Muscle Function*. New York: JB Churchill Livingstone.
- Li, G., Most, E., Sultan, P.G., Schule, S., Zayontz, S., Park, S.E., & Rubash, H.E. (2004). Knee kinematics with a high-flexion posterior stabilized total knee prosthesis:

- An in vitro robotic experimental investigation. *Journal of Bone and Joint Surgery*, 86-A, 1721-1729.
- Lu, T.W., O'Connor, J.J., Taylor, S.J., & Walker, P.S. (1998). Validation of a lower limb model with in vivo femoral forces telemetered from two subjects. *Journal of Biomechanics*, 31, 63-69.
- Lutz, G.E., Palmitier, R.A., An, K.N., & Chao, E.Y.S. (1993). Comparison of tibiofemoral joint forces during open-kinetic-chain and closed-kinetic-chain exercises. *Journal of Bone and Joint Surgery*, 75-A, 732-739.
- Maquet, P.G., Van de Berg, A.J., & Simonet, J.C. (1975). Femorotibial weight-bearing areas. Experimental determination. *Journal of Bone and Joint Surgery*, 57-A, 766-771.
- Morrison, J.B. (1970). The mechanics of the knee joint in relating to normal walking. *Journal of Biomechanics*, 3, 51-61.
- Nakagawa, S., Kadoya, Y., Todo, S., Kobayashi, A., Sakamoto, H., Freeman, M.A.R., & Yamano, Y. (2000). Tibiofemoral movement 3: Full flexion in the living knee studied by MRI. *Journal of Bone and Joint Surgery*, 82-B, 1199-1200.
- Nakayama, K., Matsuda, S., Miura, H., Higaki, H., Otsuka, K., & Iwamoto, Y. (2005). Contact stress at the post-cam mechanism in posterior stabilized total knee arthroplasty. *Journal of Bone and Joint Surgery*, 87-B, 483-488.
- Nagura, T., Dyrby, C.O., Alexander, E.J., & Andriacchi, T.P. (2002). Mechanical loads at the knee joint during deep flexion. *Journal of Orthopaedic Research*, 20, 881-886.
- Pena, E., Calvo, B., Martinez, M.A., Palanca, D., & Doblare, M. (2005). Finite element analysis of the effect of meniscal tears and meniscectomies on human knee biomechanics. *Clinical Biomechanics*, 20, 498-507.
- Schipplein, O.D., & Andriacchi, T.P. (1991). Interaction between active and passive knee stabilizers during level walking. *Journal of Orthopaedic Research*, 9, 113-119.
- Stukenborg-Colsman, C., Ostermeier, S., Hurschler, C., & Wirth, C.J. (2002). Tibiofemoral contact stress after total knee arthroplasty: Comparison of fixed and mobile-bearing inlay designs. *Acta Orthopaedica Scandinavica*, 73, 638-646.
- Taylor, W.R., Heller, M.O., Bergmann, G., & Duda, G.N. (2004). Tibio-femoral loading during human gait and stair climbing. *Journal of Orthopaedic Research*, 22, 625-632.
- Thambyah, A., Goh, J.C.H., & De, S.D. (2005). Contact stress in the knee joint in deep flexion. *Medical Engineering & Physics*, 27, 329-335.
- Toutoungi, D.E., Lu, T.W., Leardini, A., Catani, F., & O'Connor, J.J. (2000). Cruciate ligament forces in the human knee during rehabilitation exercises. *Clinical Biomechanics*, 15, 176-187.
- Weiss, J.M., Noble, P.C., Conditt, M.A., Kohl, H.W., Roberts, S., Cook, K.F., Gordon, M.J., & Mathis, K.B. (2002). What functional activities are important to patients with knee replacements? *Clinical Orthopaedics*, 404, 172-188.
- Wilk, K.E., Escamilla, R.F., Fleisig, G.S., Barrentine, S.W., Andrews, J.R., & Boyd, M.L. (1996). A comparison of tibiofemoral joint forces and electromyographic activity during open and closed kinetic chain exercises. *American Journal of Sports Medicine*, 24, 518-527.
- Wilson, W., van Rietbergen, B., van Donkelaar, C.C., & Huikes, R. (2003). Pathways of load-induced cartilage damage causing cartilage degeneration in the knee after meniscectomy. *Journal of Biomechanics*, 36, 845-851.
- Zhang, Y., Hunter, D.J., Nevitt, M.C., Xu, L., Niu, J., Lui, L., Yu, W., Aliabadi, P., & Felson, D.T. (2004). Association of squatting with increased prevalence of radiographic tibiofemoral knee osteoarthritis: The Beijing Osteoarthritis Study. *Arthritis & Rheumatism*, 50, 1187-1192.