Journal of Orthopaedic Research 20 (2002) 881–886

Journal of Orthopaedic Research

www.elsevier.com/locate/orthres

Mechanical loads at the knee joint during deep flexion

Takeo Nagura, Chris O. Dyrby, Eugene J. Alexander, Thomas P. Andriacchi *

Division of Biomechanical Engineering, Department of Mechanical Engineering, Stanford University, Durand Building, 227 Stanford, CA 94305-3030, USA

Abstract

There is a lack of fundamental information on the knee biomechanics in deep flexion beyond 90°. In this study, mechanical loads during activities requiring deep flexion were quantified on normal knees from 19 subjects, and compared with those in walking and stair climbing. The deep flexion activities generate larger net quadriceps moments (6.9–13.5% body weight into height) and net posterior forces (58.3–67.8% body weight) than routine ambulatory activities. Moreover, the peak net moments and the net posterior forces were generated between 90° and 150° of flexion.

The large moments and forces will result in high stress at high angles of flexion. These loads can influence pathological changes to the joint and are important considerations for reconstructive procedures of the knee. The posterior cruciate ligament should have a substantial role during deep flexion, since there was a large posterior load that must be sustained at the knee. The mechanics of the knee in deep flexion are likely a factor causing problems of posterior instability in current total knee arthroplasty. Thus, it is important to consider the magnitude of the loads at the knee in the treatment of patients that commonly perform deep flexion during activities of daily living.

Keywords: Knee biomechanics; Deep flexion; Gait; Total knee arthroplasty

Introduction

Knee function in deep flexion is an important consideration for activity of daily living. However, there are only a limited number of studies reporting biomechanics of deep knee flexion beyond 90°. Hefzy et al. [8], used biplanar radiographs to evaluate knee kinematics in deep flexion during Moslem praying. They showed that the motion of the femur did not reveal femoral rollback on the tibia during the motion beyond 135° of flexion. Another kinematic study, by Dyrby et al. [7], showed the relationship between deep knee flexion and internal/external rotation during deep squatting. However, no information about dynamic loads on the knee was available in these studies. The only study that revealed the dynamic knee stress during deep knee flexion was done by Dahlkvist et al. [5]. They calculated the joint and muscle forces from the data collected from six subjects performing squatting and rising from a deep squat using a force plate, a cine film system and EMG. The estimated tibio-femoral joint forces were between 4.7 and 5.6 times body weight in vertical direction and 2.9–3.5 times body weight in horizontal direction. These forces are much larger than the forces during normal walking [18,19,22,23], though direct comparisons of these studies are hampered by the limited number of subjects in Dahlkvist's study and the difference in data acquisition techniques.

Thus, the lack of fundamental information is apparent in the consideration of dynamic knee mechanics during the activities requiring deep flexion such as squatting or kneeling. Squatting and kneeling are common activities among many populations. However, there are concerns among certain workers that these activities increase the risk of knee disorders, including arthritis and meniscal injuries [11,28]. Furthermore, dynamic loading in deep flexion is another important consideration in the evaluation and design of total knee anthroplasty (TKA). As surgical technique and prosthesis design have developed, the range of motion (ROM) after TKA has improved enough to permit patients more than 100° of flexion [10,13,30], sometimes enough to perform squatting or kneeling [12,21,25]. Although the clinical results of these patients were successful in the short term, there are concerns regarding possible mechanical failure with these cases in the long-term follow up [12,25]. Moreover, there is also a concern about instability in deep flexion with both types of

* Corresponding author. Tel.: +1-650-723-8024; fax: +1-650-725-1587.

E-mail address: tandriac@stanford.edu (T.P. Andriacchi).

0736-0266/02/$ - see front matter
PII: S0736-0266(01)00178-4

prostheses, posterior cruciate ligament (PCL) substituting and PCL retaining prostheses [6,17,20,26,29]. In addition, the function of the PCL in stabilizing the knee during deep flexion is not well understood [3,9,15].

The purpose of this study was to analyze the biomechanics of deep knee flexion beyond 90°, by quantifying the dynamic loads on normal knees from 19 subjects during activities requiring deep flexion. The kinematics and joint kinetics during the activities were directly compared with those during level walking and stair climbing, so that the characteristic biomechanical features during highly flexed activities can be identified.

Material and methods

The mechanics of the knee were studied in two different groups of subjects. The subjects in the deep flexion group consisted of nine women and ten men having a mean age of 29 years (range 21–37), a mean height of 1.70 m (range 1.54–2.04), and a mean weight of 637 N (range 401–1133). The subjects in the ambulation group consisted of ten women and ten men having a mean age of 26 years (range 21–40), a mean height of 1.70 m (range 1.61–1.87), and a mean weight of 681 N (range 511–889). None of the subjects had a history of knee pain or injury.

After Institution Review Board approval and informed consent, the subjects were instrumented with six retro-reflective markers (CFTC, Chicago, IL, USA): one each at the most superolateral aspect of the iliac wing, most lateral aspect of the greater trochanter, most lateral aspect of the joint line of the knee, lateral malleolus, lateral calcaneus, and lateral head of the fifth metatarsal. The limb position was obtained from the markers using an opto-electronic system (Pro-Reflex mcu240, Qualysis, Savedalen, Sweden), and the ground reaction force was obtained with a force plate (Type 4060H, Bertec, Colombus, OH, USA). The data were collected at a frequency of 120 Hz.

The subjects in the deep flexion group performed all the tasks related to standing from the kneeling positions or returning to the kneeling positions from erect posture (Fig. 1). Four motions were analyzed using two different kneeling positions defined by each double leg or single leg motion: (a) double leg rise—start from a deep kneeling position (Fig. 1A), then stand up using both legs (Fig. 1B); (b) double leg descend—descend to a deep kneeling position from an erect posture, using a reverse motion of double leg rise; (c) single leg rise—start from a kneeling position (Fig. 1C), then step with the markered leg on the ground, then stand up using one leg (Fig. 1D); and (d) single leg descend—descend to a kneeling position from an erect posture, using a reverse motion of single leg rise. Only the markered leg was placed on the force plate during each motion and the knees were placed outside of the plate. Data were not acquired during the time when both knees were on the ground. The subjects attempted to put all of their weight on the markered leg during the single leg motions. Two trials out of three trials (the first trial was used to familiarize the subject) were analyzed for both legs in each deep flexion activity. The subjects in the ambulation group performed level walking on a 10 m walkway, and stair climbing by stepping onto 25.5 cm platforms. Two single platforms were prepared for the stair-climbing test, and the first step onto the platform was recorded. Two trials for each leg were analyzed for this group.

The kinematics and kinetics of the knee were obtained using an inverse dynamics approach. In this technique, the net forces and net moments were calculated from measurements of limb segment size, body mass and ground reaction forces. The methods for the calculation have been previously described [1,2]. The net knee force represents net inter-segmental force acting on the knee, and the net knee moment represents net inter-segmental moment about the knee, which is resisted by the contraction of flexor/extensor muscles. The forces and moments were normalized to percent body weight (%BW) and percent body weight times height (%BW × Ht) respectively. A local anatomical coordinate system on the tibia was used to represent the force directions. The long axis of the tibia was defined as superior–inferior axis.

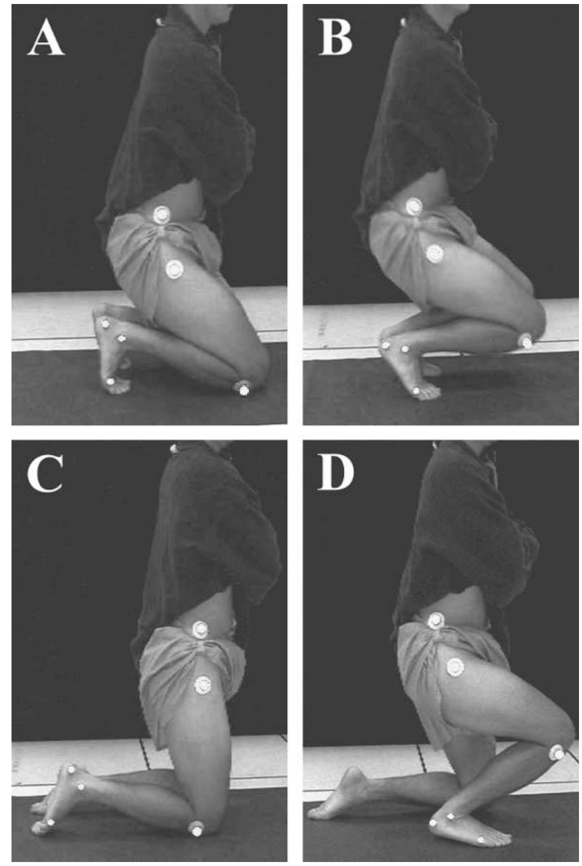

Fig. 1. The deep flexion activities: (A) deep kneeling position in double leg motions, (B) rising/descending phase in double leg motions, (C) kneeling position in single leg motions, and (D) rising/descending phase in single leg motions.

One trial was randomly chosen from each subject to calculate the statistics in each activity. To see the differences in the kinetics at different angles of flexion, the net forces and moments during the double/single leg rise (descend) were evaluated at every 5° of flexion, as the knee moved from maximum flexion (extension) to maximum extension (flexion). Peak values of knee parameters were also compared among six different activities. An analysis of variance (ANOVA) with a single factor for two groups was performed each time, to test the statistical difference between any two of the activities. Levels of significance are indicated in the text.

Results

During each deep flexion activity, the net flexion moments (sustained by net quadriceps moments) were dominant throughout the motions, while the net anterior–posterior forces were directed posteriorly. During double leg descend and double leg rise, the net moments and the net posterior forces increased along with the increase in flexion angle, and reached a peak before/after the maximum knee flexion (Fig. 2A and B). The net inferior forces were minimum at or near the maximum flexion. In the double leg motions, the contact between thigh and calf, which might affect the force and moment calculations, occurred when the knees were flexed beyond 140°. During single leg descend and single leg rise, the net moments and net posterior forces reached a peak between 70° and 100° of flexion (Fig. 3A and B). The net inferior forces were largest when the knee was in extension, while the forces were smallest when the knee was around 120° of flexion.

The double leg rise/descend produced a larger moment than the single leg rise/descend (Table 1). In particular, the net moment during double leg rise was nearly two times larger than that during single leg rise ($p < 0.001$). In contrast, net inferior forces during double leg rise/descend were about 1/2 of those during single leg rise/descend ($p < 0.001$).

There were significant differences in the peak values of sagittal knee kinematics and kinetics between the deep flexion activities and the ambulatory activities (Table 1). The magnitudes of ROM, the net flexion moment, and the net posterior force in the deep flexion group were significantly larger than those in walking or stair climbing ($p < 0.001$), except for the moments in single leg motions that were not significantly different from those in stair climbing ($p > 0.05$). All the magnitudes of net inferior force in the deep flexion group were significantly smaller than those in walking or stair climbing ($p < 0.001$). In addition, the average flexion angles at the maximum net flexion moments during the deep flexion activities were significantly larger than those in walking or stair climbing ($p <$ 0.001, Fig. 4).

Net adduction moments in double leg motions (1.2%BW × Ht in double leg rise and 0.9%BW × Ht in double leg descend) were smaller than those in walking (3.6%BW × Ht, $p < 0.001$) or stair climbing (2.7%BW × Ht, $p < 0.001$). The net adduction moments in single leg motions (2.4%BW × Ht in single leg rise and descend) were smaller than those in walking ($p < 0.005$). All the magnitudes of net rotational moments in the deep flexion group were less than 1.0%BW × Ht and were not significantly different from walking or stair climbing ($p > 0.05$). Net lateral forces on the knee during double leg motions (5.8%BW in double leg rise, 5.1%BW in double leg descend) were smaller than those in walking (14.4%BW, $p < 0.001$) or stair climbing (12.6%BW, $p <$ 0.001), while those during single leg motions (12.4%BW in single leg rise and 12.1%BW in single leg descend) were not significantly different from those in walking or stair climbing ($p > 0.05$).

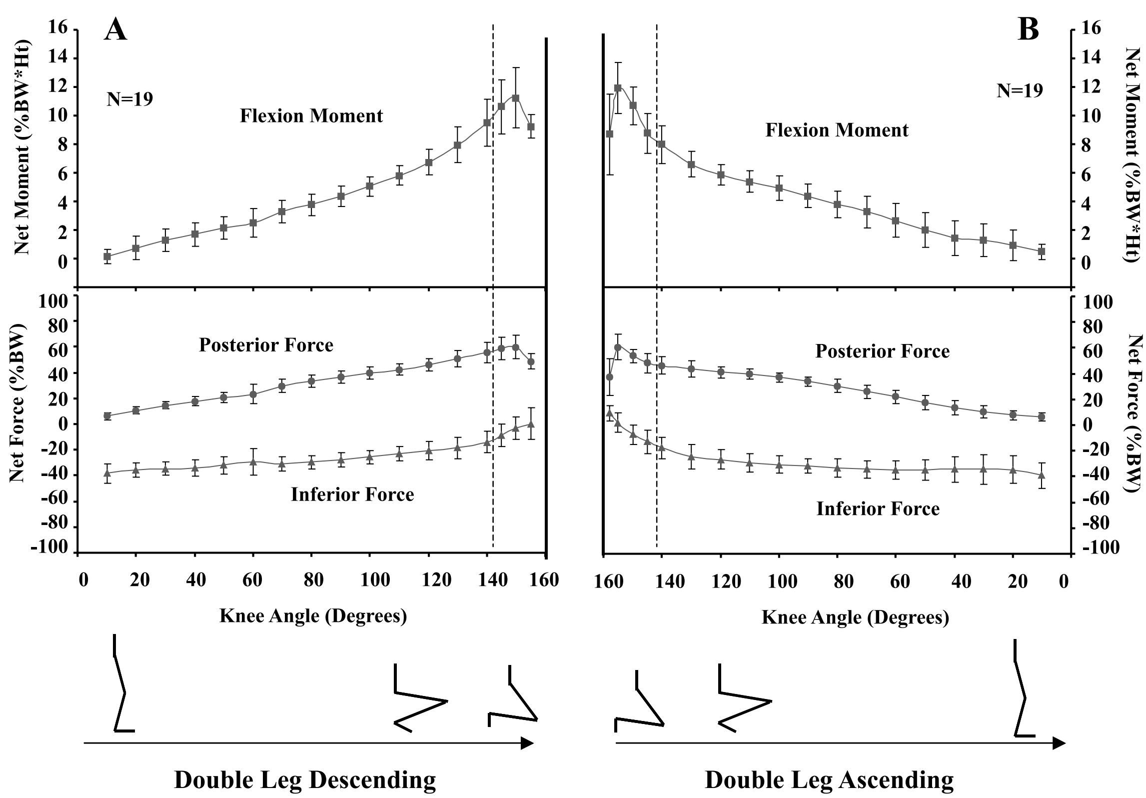

Fig. 2. Mean and standard deviation of net flexion moment (■), net posterior force (●), and net inferior force (▲) at each flexion angle during double leg descend (A) and during double leg rise (B). Stick figures indicate sagittal image of the limb during each motion. A dotted line indicates the point where the contact between thigh and calf should begin (end) during double leg descend (rise). %BW = % body weight, %BW × Ht = percent body weight times height.

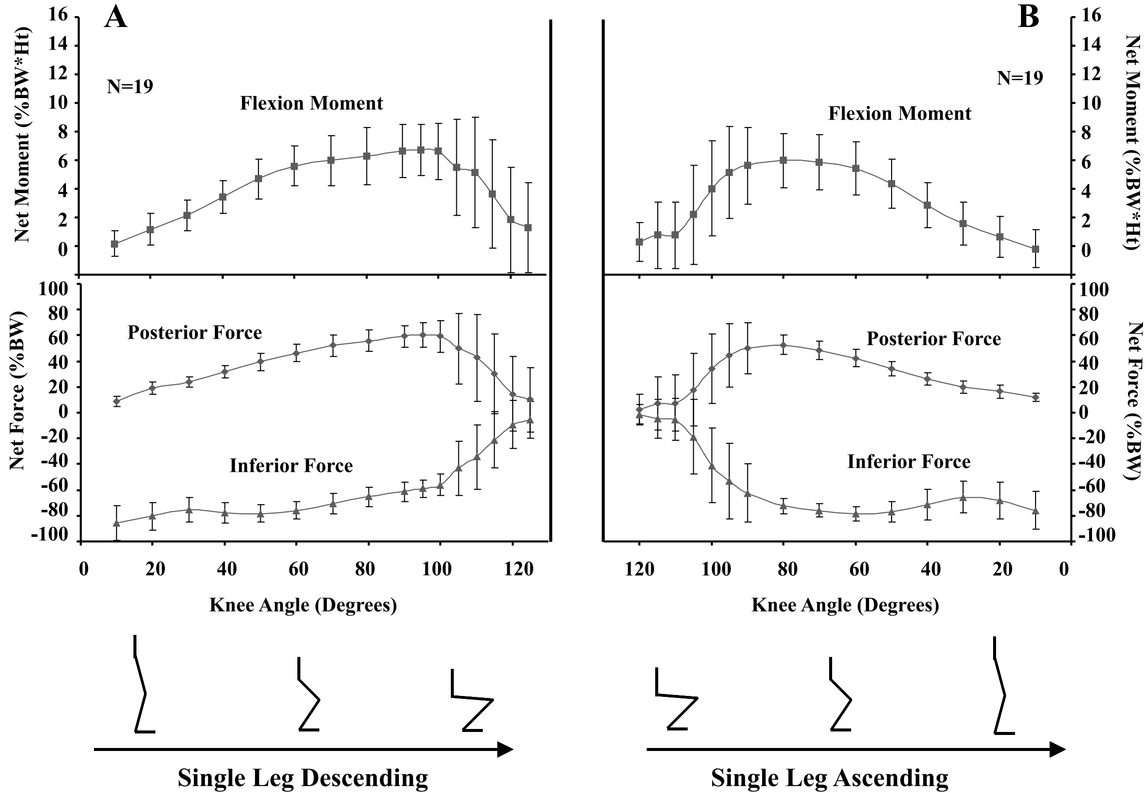

Fig. 3. Mean and standard deviation of net flexion moment (■), net posterior force (●), and net inferior force (▲) at each flexion angle during single leg descend (A) and during single leg rise (B). Stick figures indicate sagittal image of the limb during each motion. %BW = % body weight, %BW × Ht = percent body weight times height.

Table 1
Knee kinematics and kinetics in sagittal plane (mean, SD)

Activity	ROM (degrees)	Net flexion moment (%BW × Ht)	Net posterior force (%BW)	Net inferior force (%BW)
Double leg rise	150.4 (7.2)	13.5 (2.2)	62.8 (9.9)	50.7 (9.7)
Double leg descend	148.3 (6.7)	11.4 (2.1)	58.4 (9.1)	47.3 (7.5)
Single leg rise	129.3 (6.8)	6.9 (2.1)	58.3 (9.4)	93.6 (5.0)
Single leg descend	122.1 (6.2)	8.0 (1.7)	67.8 (10.1)	90.4 (7.8)
Walking	65.4 (5.7)	4.0 (1.8)	38.8 (6.0)	112.3 (8.3)
Stair climbing	86.5 (5.6)	7.5 (1.4)	47.1 (8.0)	102.9 (5.9)

ROM = range of motion, %BW × Ht = percent body weight times height, %BW = percent body weight.

Discussion

The results of this study clearly demonstrate that deep flexion activities generate significant net quadriceps moments at the knee. In addition, the maximum moments occur at larger flexion angles during deep flexion activities relative to routine ambulatory activities (Fig. 4). The increase in the extensor force during deep flexion will increase the stress on the patellar tendon and joint contact forces. Thus, rehabilitation of quadriceps strength as well as procedures involving the patellar tendon or its insertion at the tibial tubercle require special consideration for a patient where routine activities of daily living include deep flexion. It is interesting to note that a patient can substantially reduce the demand on the quadriceps by using the single limb rising technique rather than the double limb rising from kneeling positions on the floor (Table 1). The differences in the quadriceps demand between single and double limb motions are believed to result from the differences in the required ROM during each motion.

Another important consideration is the increase in the net posterior force acting on the tibia during deep flexion. This component of the force increases by approximately 50% over ambulatory activities (Table 1). Particularly, the peak forces occur at approximately

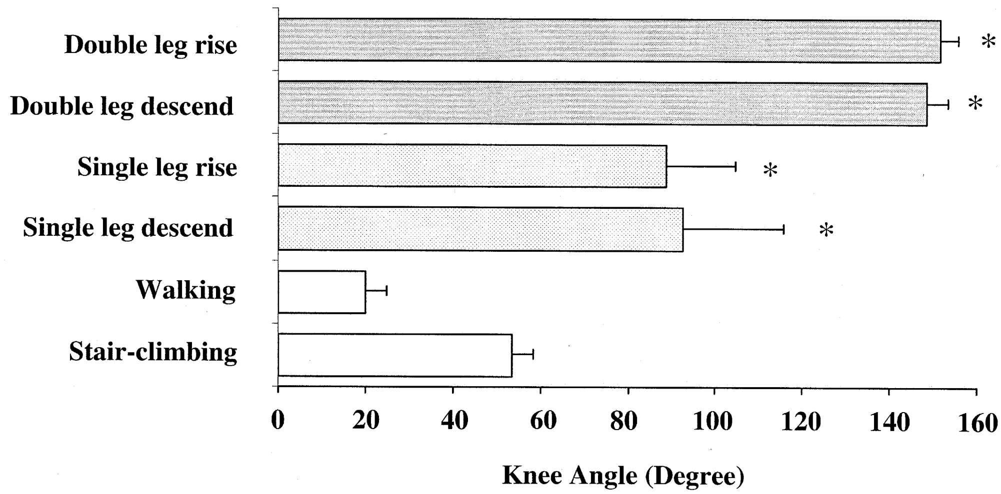

Fig. 4. The average flexion angles at the maximum net flexion moments during each activity. The asterisks (*) indicate significant differences between each deep flexion activity and walking ($p < 0.001$) or stair climbing ($p < 0.001$).

150° during double leg motions (Fig. 2). At this angle of flexion, the extensor mechanism will also apply a posteriorly directed force on the tibia since the patellar tendon is tilting posteriorly, and this would increase the total posterior force component at the knee. The combination of these forces would substantially strain any structure providing posterior stability to the knee joint. Thus, the PCL, which has been reported to contribute 94% of the posterior stability of the knee [3], should have a substantial role during rising from or descending to deep flexion.

The differences in the joint loads between deep flexion activities and ambulatory activities also provide new insight into the pathogeneses of various conditions at the knee associated with activities of deep flexion. Since the normal tibio-femoral joint decreases its contact area to 55% as the knee is flexed to 90° [14], the significant joint forces in deep flexion could influence the rate of degenerative changes to the knee over time in individuals that frequently perform activities in deep flexion [11,28]. For example, the high loads on the patello-femoral joint may explain some clinical problems that are related to deep flexion [8,16]. The large posterior forces should be an important consideration in patients with injury to the PCL [9]. "It should be noted that this study did not include stair descent. Previous studies [1] have shown that there can be larger forces sustained by the quadriceps muscles during descending relative to ascending stairs. Therefore, descending stairs should also be considered when evaluating activities that are associated with large forces generated by the quadriceps muscles."

These results are important in the analysis of reconstructive procedures for patients that commonly perform deep flexion during routine activities of daily living. A greater ROM in current TKA is of particular concern for an increased risk of mechanical failure, since the mechanics of the knee in deep flexion are likely a factor causing problems of instability in TKA. As Delp et al., have shown in their computer simulation study [6], the substantial risks for posterior knee dislocation in TKA increase at deep flexion (especially at maximum flexion). In fact, many cases of posterior knee dislocations have been reported with several types of posterior substituting knee designs [4,24,27]. In addition, Pagnano et al. [20] report flexion instability following late rupture of the PCL in PCL retaining TKA. It should be noted that most patients following TKA do not perform the activities with the extended ROM described in this study. However, as higher flexion become available in TKA, higher posterior loads are expected at the knee. The significant loads in deep flexion should be regarded in the design of TKA, as well as in the treatment of TKA patients.

Several limitations should be pointed out regarding the calculation of the kinetics during the activities. The contact of posterior thigh and calf during double leg motions shares a portion of the moment and the force. Therefore the net moment and force might be overestimated since such contact was neglected in our computational model. An impingement between thigh and calf should be expected when the knees are flexed beyond 140° (depending on the thickness of the limb of the subject), while the quadriceps are sustaining gravity during the motions except for the time when the knees are on the ground (Fig. 2A and B). The continuous action of the quadriceps against gravity should reduce the effect of an impingement. In addition, given the large differences between the deep flexion loads and the ambulatory loads, the conclusion of this study should not be altered by the effect of an impingement. Another consideration regarding the methods is the estimation of the joint center position. There will be relative tibio-femoral translation during deep knee flexion, which could alter the moment calculation at the knee. In this study, the calculations of the moment were made about a fixed point at the geometric center of the knee. A sensitivity test [2] has been performed, which includes 4 cm translations (± 2.0 cm) of the joint center in the sagittal plane. The translations made a maximum of 15% changes in net knee flexion moment during single leg motions, but less than 10% changes in other parameters. The test demonstrates that the translations will affect the parameters, however they will not affect the difference between two groups of activities, except for the difference in the moment between single leg motions and stair climbing. Again, the conclusion of the study should not be influenced by this limitation. It should also be noted that a different group of subjects was used for the ambulatory activities. This was done to limit the test burden on a single subject. While this design did reduce the statistical power of the study, the differences between the deep flexion activities and the ambulatory activities were sufficiently large to show significance in most of the parameters.

In conclusion, large net quadriceps moments and net posterior forces at the knee were seen during the deep flexion activities. The results indicate that the loads on the knee during deep flexion are important considerations both for the pathogeneses of the knee and the evaluation of reconstructive procedures for the patients, such as TKA.

Acknowledgements

The authors would like to acknowledge the contribution of Ajit Chaudhari and John MacMahon, for helping with the acquisition and processing of the data.

References

[1] Andriacchi TP, Andersson GBJ, Fermier RW, Stern D. A study of lower-limb mechanics during stair-climbing. J Bone Joint Surg [Am] 1980;62:749–57.

[2] Andriacchi TP, Galante JO, Fermier RW. The influence of total knee-replacement design on walking and stair-climbing. J Bone Joint Surg [Am] 1982;64:1328–35.

[3] Butler DL, Noyes FR, Grood ES. Ligamentous restraints to anterior–posterior drawer in the human knee. A biomechanical study. J Bone Joint Surg [Am] 1980;62:259–70.

[4] Cohen B, Constant CR. Subluxation of the posterior stabilized total knee arthroplasty. A report of two cases. J Arthroplasty 1992;7:161–3.

[5] Dahlkvist NJ, Mayo P, Seedhom BB. Forces during squatting and rising from a deep squat. Eng Med 1982;11:69–76.

[6] Delp SL, Kocmond JH, Stern SH. Tradeoffs between motion and stability in posterior substituting knee arthroplasty design. J Biomech 1995;28:1155–66.

[7] Dyrby CO, Toney MK, Andriacchi TP. Relation between knee flexion and tibial-femoral rotation during activities involving deep flexion. Gait Posture 1997;5:179–80.

[8] Hefzy MS, Kelly BP, Cooke DV. Kinematics of the knee joint in deep flexion: a radiographic assessment. Med Eng Phys 1998; 20:302–7.

[9] Höher J, Harner CD, Vogrin TM, Baek GH, Carlin GJ, Woo SLY. In situ forces in the posterolateral structures of the knee under posterior tibial loading in the intact and posterior cruciate ligament-deficient knee. J Orthop Res 1998;16:675–81.

[10] Insall JN, Lachiewicz PF, Burstein AH. The posterior stabilized condylar prosthesis. Two to four year clinical experience. J Bone Joint Surg [Am] 1982;64:1317–23.

[11] Jensen LK, Eenberg W. Occupation as a risk factor for knee disorders. Scand J Work Environ Health 1996;22:165–75.

[12] Kim J, Moon M. Squatting following total knee arthroplasty. Clin Orthop 1995;313:177–86.

[13] Maloney WJ, Schurman DJ. The effects of implant design on range of motion after total knee arthroplasty. Clin Orthop 1992;278:147–52.

[14] Maquet PG, Van de Berg AJ, Simonet JC. Femorotibial weight-bearing areas. Experimental determination. J Bone Joint Surg [Am] 1975;57:766–71.

[15] Markolf KL, Mensch JS, Amstutz HC. Stiffness and laxity of the knee: the contributions of the supporting structures. A quantitative in vitro study. J Bone Joint Surg [Am] 1976;58:583–94.

[16] Matusda S, Ishinishi T, White SE, Whiteside LA. Patellofemoral joint after total knee arthroplasty. Effect on contact area and contact stress. J Arthroplasty 1997;12:790–7.

[17] Matsuda S, Miura H, Nagamine R, Urabe K, Matsunobu T, Iwamoto Y. Knee stability in posterior cruciate ligament retaining total knee arthroplasty. Clin Orthop 1999;366:169–73.

[18] Mikosz RP, Andriacchi TP, Andersson GBJ. Model analysis of factors influencing the prediction of muscle forces at the knee. J Orthop Res 1988;6:205–14.

[19] Morrison JB. The mechanics of the knee joint in relating to normal walking. J Biomech 1970;3:51–61.

[20] Pagnano MW, Hanssen AD, Lewallen DG, Stuart MJ. Flexion instability after primary posterior cruciate retaining total knee arthroplasty. Clin Orthop 1998;356:39–46.

[21] Schai PA, Gibbon AJ, Scott RD. Kneeling ability after total knee arthroplasty. Clin Orthop 1999;367:195–200.

[22] Schipplein OD, Andriacchi TP. Interaction between active and passive knee stabilizers during level walking. J Orthop Res 1991;9:113–9.

[23] Seireg A, Arvikar RJ. The prediction of muscular load sharing and joint forces in the lower extremities during walking. J Biomech 1975;8:89–102.

[24] Sharkey PF, Hozack WJ, Booth RE, Balderstone RA, Rothman RH. Posterior dislocation of total knee arthroplasty. Clin Orthop 1992;278:128–33.

[25] Shoji H, Yoshino S, Komagamine M. Improved range of motion with the Y/S total knee arthroplasty system. Clin Orthop 1987; 218:150–63.

[26] Stiehl BJ, Komistek RD, Cloutier JM, Dennis DA. The cruciate ligaments in total knee arthroplasty: a kinematic analysis of 2 total knee arthroplastics. J Arthroplasty 2000;15:545–50.

[27] Striplin DB, Robinson RP. Posterior dislocation of the Insall/Burstein Π posterior stabilized total knee prosthesis. Am J Knee Surg 1992;5:79–83.

[28] Thun M, Tanaka S, Simth AB, Halperin WE, Lee ST, Luggen ME, Hess EV. Morbidity from repetitive knee trauma in carpet and floor layers. Br J Ind Med 1987;44:611–20.

[29] Uvehammer J, Kärrholm J, Brandsson S. In vivo kinematics of total knee arthroplasty: concave versus posterior-stabilised tibial joint surface. J Bone Joint Surg [Br] 2000;82: 499–505.

[30] Whiteside LA. Cementless total knee replacement. Nine to 11 year-results and 10-year survivorship analysis. Clin Orthop 1994;309:185–92.