JOURNAL OF BIOMECHANICS

ELSEVIER

Journal of Biomechanics 31 (1998) 677–683

The effect of intersegmental knee moments on patellofemoral contact mechanics in cycling

Jeffrey C. Wolchok, M.L. Hull*, Stephen M. Howell

Department of Mechanical Engineering, University of California, Davis, CA 95616, U.S.A.

Received in final form 30 December 1997

Abstract

The aim of this study was to evaluate the effect of bicycle pedal design on the mechanics of the patellofemoral joint. Previous research determined that for certain riders the non-driving varus and internal knee moments could be reduced by switching from fixed to free floating pedals (Ruby and Hull, 1993). It was postulated that the presence of varus and internal knee moments during fixed pedal cycling may adversely affect patellofemoral joint contact mechanics which could lead to the development of anterior knee pain. To investigate the effect of pedal design the hypothesis that varus and internal intersegmental knee moments significantly increase patellofemoral contact pressure, contact area and contact force was tested. To test this hypothesis cycling loads were simulated *in vitro* using a six-degree-of-freedom load application system (LAS). Using the LAS, varus moments ranging from 0–20 Nm and internal knee moments ranging from 0–10 Nm were applied simultaneously with quadriceps force at knee flexion angles of 60 and 90 degrees. Patellofemoral contact patterns were measured using pressure sensitive film. An applied 10 Nm internal moment significantly increased both contact area by 16% and contact force by 22% at 90° of flexion. The application of a 20 Nm varus moment modestly yet significantly increased contact area by 6% and contact force by 5%. When applied in combination, varus and internal knee moments increased contact area and force by as much as 29% and 28% respectively. The mean contact pressure was not significantly increased by either of the two moments. The results suggest that non-driving intersegmental knee moments subject the patellofemoral joint to loads and contact patterns which may accelerate the development of chondromalacia. © 1998 Elsevier Science Ltd. All rights reserved.

Keywords: Cycling; Patellofemoral; Quadriceps muscle; Contact pressure; Pressure sensitive film; Varus moment; Internal axial moment

1. Introduction

One of the most common injuries from cycling is anterior knee pain (Weiss, 1985; Holmes et al., 1991). Although there are several possible causes, degeneration of the patella cartilage as a result of repeated overloading is a probable cause when athletic activity is involved (Insall, 1982). When this degeneration reaches advanced stages on the underside of the patella, the condition is called chondromalacia (Insall, 1982; Mow and Soslowsky, 1991; Goodfellow et al., 1976).

In those cases where overloading of the patella occurs, the increased loads can be attributed to many factors (Powell, 1986), one of which is malalignment between the bicycle and rider mechanical system (Francis, 1986). The commonly used fixed pedal constrains the foot in five degrees of freedom, and allows only one rotational degree of freedom around the pedal spindle. If the cyclist's leg and foot do not naturally travel along the path of the pedal, then constraining the foot to this path will produce additional constraint loads at the pedal/foot interface (Ruby and Hull, 1993). Therefore, in addition to the loads required for propulsion, the constraint loads will be transferred by the joints of the lower limb (Ruby and Hull, 1993; Hull and Ruby, 1996).

Research has shown that the constraint loads can be reduced through the use of free floating pedals which allow the foot to rotate in either abduction/adduction rotation (Wheeler et al., 1995; Boyd et al., 1997) or inversion/eversion rotation (Boyd et al., 1997). As a

* Corresponding author. Tel.: 916 752 6220; fax: 916 752 4158; e-mail: mlhull@ucdavis.edu

0021-9290/98/$19.00 © 1998 Elsevier Science Ltd. All rights reserved.
PII S0021-9290(98)00075-X

consequence of the constraint load reduction, the non-driving varus and internal intersegmental moments transmitted through the knee are reduced (Ruby and Hull, 1993; Boyd et al., 1997).

Though it has been demonstrated that pedal design can affect the intersegmental knee moments, it is not known how these moments affect patellofemoral joint mechanics. Changes in the alignment of the quadriceps and patellar tendon (Q-angle), which both varus and internal moments can produce, will alter patellofemoral contact pressure, area, and force patterns (Fithian et al., 1994; Huberti and Hayes, 1984; Yang and Hefzy, 1994). Therefore, the presence of these moments during fixed pedal cycling may adversely affect patellofemoral contact mechanics which might explain the anterior knee pain common in cyclists. Thus, the hypothesis that varus and internal knee moments typically developed using fixed pedals significantly increase mean patellofemoral contact pressure, contact area, and contact force was tested.

2. Materials and methods

2.1. Experiments

Five left legs and three right legs were obtained from eight human donors. The donors consisted of 4 men and 4 women with an average subject age of 62 years and a range of 45 years to 84 years. Specimens were procured within 72 h postmortem and were sealed in a plastic bag and stored at $-15°C$ until testing. Only specimens with no articular degeneration and no prior soft tissue damage were selected.

A computer controlled, load application system (LAS) was used to control the varus and internal moments, the quadriceps force, and the knee flexion angle (Bach and Hull, 1995). The system allowed unconstrained movement of the joint in five degrees of freedom, and adjustability in flexion angle over the full physiological range. Pneumatic rotary actuators (Turn-Act, Louisville, KY) were used to apply the varus and internal moments. A pneumatic linear actuator (Mead, Chicago, IL) simulated the effect of the tension produced by the quadriceps muscle group.

Because the patellofemoral contact pressure is sensitive to the direction of the quadriceps force (Huberti and Hayes, 1988), the actuator was used in conjunction with a custom quadriceps fixture (Fig. 1). Utilizing a method that was employed in numerous other studies (Huberti and Hayes, 1984, 1988; Lewallen et al., 1990; Marder et al., 1993), a physiological quadriceps angle (Q-angle) was reproduced by aligning the pull of the quadriceps muscle with a rod extending from the femoral medullary canal. To transfer the load from the cylinder to the muscle tendon complex at least 11 cm of quadriceps tendon were sewn to a nylon tube strap using two #2

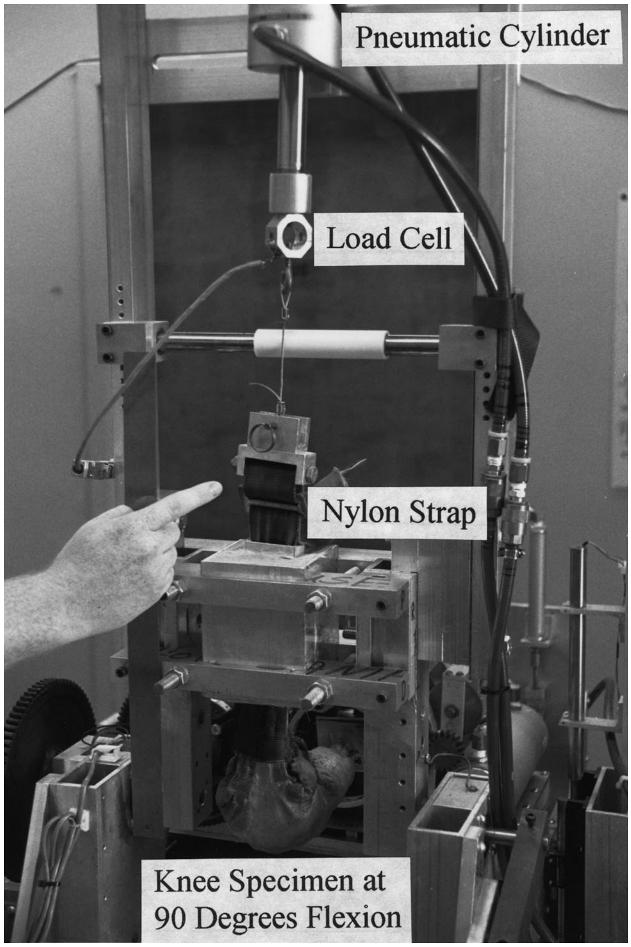

Fig. 1. Quadriceps fixture illustrating the attachment of the quadriceps tendon with the pneumatic cylinder as well as the alignment of the quadriceps force. The intramedullary rod is hidden behind the nylon muscle strap.

polyester, braided sutures (Ti-cron, Davis and Geck, Danbury, CT). This construct could reliably transmit forces up to 1600 N.

As in other studies, (Huberti and Hayes, 1984, 1988; Lewallen et al., 1990; Marder et al., 1993), low-range pressure sensitive film (Fuji Photo Co., Ltd., Japan) was used to measure patellofemoral contact area and mean pressure. To both protect the film from synovial fluids and reduce the effect of any humidity variations, the film was sealed between two layers of polyethelene (Huberti and Hayes, 1984; Marder et al., 1993). Because the measurement of patellofemoral contact pressure is altered by lateral and medial incision of the capsule (Huberti and Hayes, 1988), the pressure sensitive film was inserted through an opening in the superior portion of the capsule just posterior to the quadriceps tendon.

Following alignment in the LAS using the functional axes approach (Berns et al., 1990; Bach and Hull, 1995) and preconditioning to the maximum values of the applied loads, each specimen was subjected to a series of nine load conditions which were all combinations of 0, 5, 10 Nm internal and 0, 10, 20 Nm varus moments. The moments incorporated the range of values measured in a clinical study between top dead center and a 90° crank angle during 250 W fixed pedal cycling (Ruby and Hull, 1993). All nine load conditions were randomly applied with the knee in 60° and 90° of knee flexion. To gain consistency in the pressure measurements, the varus and internal moments were ramped up to their peak values over a 20 s period and then held there for 15 s (Rudert et al., 1988). For each combination of flexion angle and each load condition, two trials were conducted to provide pressure imprints on two separate films.

Due to limitations in the load capacity of the quadriceps connection and the working range of the pressure sensitive film (2.5–7 MPa), one quarter of the average maximum isometric quadriceps moment measured clinically was applied (Haffajee and Moritz, 1972; Lindahl et al., 1969). The values were 35.3 Nm and 26.3 Nm at 60° and 90°, respectively. This resulted in an average quadriceps load of 1360 ± 160 N (SD) at 60° and an average quadriceps load of 1260 ± 120 N at 90°.

2.2. Data analysis

Once contact patterns were collected and converted to grey-scale values using a flatbed scanner (Hewlett Packard Scanjet IIcx, San Jose, CA), imaging software (Image 1.5, NIH, Bethesda, MD) was used to determine the contact area and mean pressure. Prior to analysis the contact pattern was filtered and thresholded to eliminate the fainter background stains. Once thresholded, the software ignored gray scale values less than 2.5 MPa (the film threshold) when calculating area and mean grey-scale values. The thresholding values were determined from the 2.5 MPa calibration stains. The pixels whose grey-scale values were above the thresholding value were counted to determine the contact area.

To determine mean contact pressure, in-house calibration curves were developed using a calibration procedure similar to that used in other studies (Huberti and Hayes, 1984, 1988; Lewallen et al., 1990). Eight calibration films were made, beginning at 2.5 MPa and extending up to 7.5 MPa. The patellofemoral contact force was calculated as the product of contact area and mean contact pressure. To reduce the variability inherent to the measurements, the dependent variable values (i.e. contact area, mean pressure, and contact force) obtained from the two films exposed at each combination of flexion angle and load condition were averaged.

The effects of loading on patellofemoral contact pressure, area, and force were determined using a three factor repeated measures analysis of variance. The main factors were as follows:

subject $n = 8$,
varus moment (Nm) $j = 3$ (0, 10, 20)
internal moment (Nm) $k = 3$ (0, 5, 10)

To eliminate subject-to-subject variability, only within subject effects were investigated. The statistical model included all of the main factor effects as well as the interaction effect of combined internal-varus loading. In addition to the repeated measures analysis, post hoc contrast decompositions were run to determine which levels of internal and varus loading produced significant increases in the contact pressure, area, and force. Separate analyses were run for 60° and 90° of knee flexion. All statistical analyses was conducted using Statistical Application Software (SAS, Cary, NC). The mean and standard errors were used to describe the findings.

3. Results

The contact area was significantly increased in the presence of varus and internal moments at both 60° (Table 1) and 90° (Table 2) ($p < 0.05$). Regardless of flexion angle, internal moments of 5 Nm and 10 Nm resulted in contact area increases of 6% and 16%, respectively (Fig. 2). Contrasts indicated that the contact area developed by a 10 Nm internal moment was significantly larger than that for a 5 Nm internal moment at both flexion angles ($p < 0.02$) (Table 3). The area increases produced by varus moments were more modest. At 90°, 10 Nm and 20 Nm varus moments produced contact area increases of 2% and 6%, respectively, while at 60° the 10 Nm and 20 Nm varus moments each produced

Table 1
Results of the ANOVA analyses at 60 degrees of flexion. Averages and corresponding standard errors are given for each load condition. *P*-values are listed for the varus and internal main factors as well as the varus/internal interaction effect

Dependent variable	Varus moment (Nm)	Internal moment (Nm)			*p*-value
		0	5	10	0.0002
Area (cm^2)	0	2.49(0.13)	2.65(0.11)	2.88(0.12)	
	10	2.60(0.11)	2.72(0.11)	2.91(0.12)	
	20	2.60(0.09)	2.67(0.10)	2.85(0.13)	
p-value	0.003				0.28
		0	5	10	0.001
Force (N)	0	1080(80)	1180(70)	1320(90)	
	10	1160(70)	1200(80)	1300(80)	
	20	1130(70)	1170(70)	1290(90)	
p-value	0.06				0.07
		0	5	10	0.06
Pressure (MPa)	0	4.32(0.16)	4.45(0.07)	4.55(0.20)	
	10	4.42(0.17)	4.39(0.16)	4.45(0.16)	
	20	4.33(0.15)	4.38(0.15)	4.48(0.18)	
p-value	0.12				0.02

4% increases in contact area. The differences in contact area between 10 Nm varus and 20 Nm varus were significant at 90° ($p < 0.02$) but were not at 60° ($p > 0.02$) (Table 3).

Substantial area increases were developed under combined loading. A combined load of 20 Nm varus/10 Nm internal produced 14% and 29% increases in contact area over the unloaded condition at 60° and 90°, respectively. When compared to the 20 Nm varus/5 Nm internal combined load the increase in area was 5% at 60° and 12% at 90°. Because the interaction between varus and internal moments was not significant ($p > 0.05$), these increases in area are attributed to the additive effect of the constituent moments.

As a result of their direct relationship, the behavior of the contact force was similar to contact area. Internal moments significantly increased the patellofemoral contact force at both flexion angles ($p < 0.05$) and varus moments were significant at 90° ($p < 0.05$) (Tables 1 and 2). Internal moments had a greater effect on the contact force than varus moments did. At 60° the 5 Nm and 10 Nm internal moments produced 8% and 19% increases in contact force respectively, while at 90° 9% and 22% increases were measured. The contrasts revealed that differences in the contact force between 5 Nm and 10 Nm internal were significant at both 60° and 90° ($p < 0.02$) (Table 3).

The varus-internal interaction effect on contact force was not significant at either flexion angle ($p > 0.05$), so that like contact area, combined loading did produce

Table 2
Results of the ANOVA analyses at 90 degrees of flexion. Averages and corresponding standard errors are given for each load condition. *P*-values are listed for the varus and internal main factors as well as the varus/internal interaction effect

Dependent variable	Varus moment (Nm)	Internal moment (Nm)			*p*-value
		0	5	10	0.004
Area (cm^2)	0	2.40(0.11)	2.56(0.14)	2.80(0.17)	
	10	2.44(0.11)	2.67(0.16)	2.82(0.17)	
	20	2.55(0.09)	2.81(0.19)	3.00(0.18)	
p-value	0.006				0.90
		0	5	10	0.005
Force (N)	0	1030(70)	1110(80)	1220(80)	
	10	1060(80)	1160(80)	1250(100)	
	20	1080(70)	1220(100)	1310(90)	
p-value	0.02				0.90
		0	5	10	0.33
Pressure (MPa)	0	4.29(0.23)	4.34(0.22)	4.38(0.23)	
	10	4.32(0.22)	4.36(0.22)	4.44(0.24)	
	20	4.24(0.21)	4.33(0.21)	4.40(0.25)	
p-value	0.18				0.80

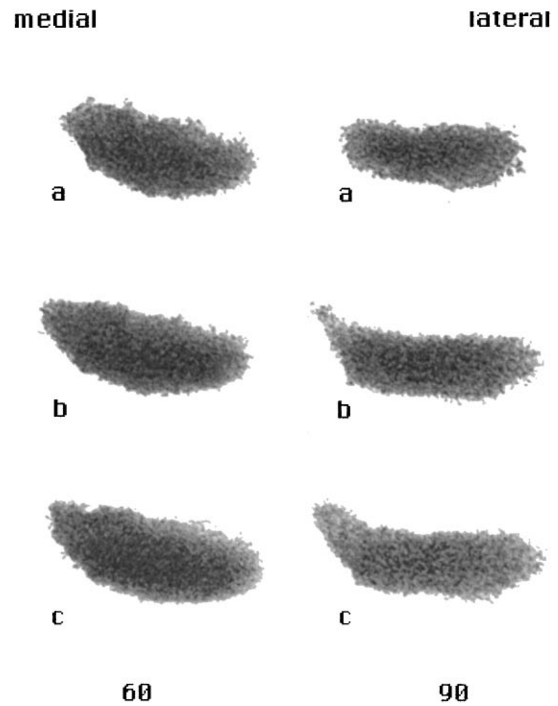

Fig. 2. Patellofemoral contact patterns illustrating the increase in contact area that developed at 60° and 90° when the internal moment was increased from (a) 0 Nm to (b) 5 Nm (c) 10 Nm.

Table 3
Results of the varus and internal load contrast decompositions. *P*-values are given for each comparison corresponding to row and column moment values. Bold and italics type correspond to contrast results obtained at 90° and 60° of knee flexion respectively. To control for experimentwise error, significance levels were scaled to 0.02 using the Bonferroni adjustment. Contrasts that yielded significant differences between levels are underlined

Dependent variable	Internal moment (Nm)	Contrast decomposition *p*-values for internal moments of: (Nm)			Varus moment (Nm)	Contrast decomposition *p*-values for varus moments of: (Nm)		
		0	5	10		0	10	20
Area	0		**0.03**	**0.006**	0		**0.42**	**0.008**
	5	0.0001		**0.005**	10	0.003		**0.007**
	10	0.0002	0.002		20	*0.11*	*0.05*	
Force	0		**0.04**	**0.007**	0		**0.25**	**0.01**
	10	0.0004		**0.004**	10	**0.04**		**0.007**
	20	0.001	0.006		20	**0.16**	**0.08**	

measurable contact force increases. At 60°, a 20 Nm varus/10 Nm internal combined load developed a 1290 N contact force while at 90° this same combined load produced a 1310 N contact force. These contact forces represented increases of 19% and 28% over the unloaded condition and 8% and 13% over the 10 Nm varus/5 Nm internal combined load at 60° and 90°, respectively.

Although contact area and force were affected by intersegmental knee moments the mean contact pressure was not. Neither varus nor internal moments significantly increased the mean patellofemoral contact pressure ($p > 0.05$). This was true at both 60° and 90° (Tables 1 and 2). The mean contact pressure never exceeded 4.55 MPa for any load condition. The varus-internal interaction was significant at 60° ($p < 0.05$), however at 90° no significant interaction effect was found ($p > 0.05$).

4. Discussion

The aim of this study was to investigate how the varus and internal knee moments developed during fixed pedal cycling affect patellofemoral joint mechanics. Because contact area, mean contact pressure, and contact force play a possible role in the development of chondromalacia, identifying knee moment levels which significantly increase any of these quantities would be important from an injury prevention standpoint. To measure these quantities and determine significant increases, a cadaver study was performed using pressure sensitive film. The use of pressure sensitive film as well as the *in vitro* aspect of the study raised several methodological issues that could affect the results of this study.

When measuring contact pressures in diarthrodial joints using pressure sensitive film sandwiched between polyethylene sheets, one concern is that the transducer may be too stiff to conform to the compound curvature of the articular surfaces thus producing an artifact due to wrinkling. When wrinkled, the pressure sensitive film shows obvious signs, namely red lines and streaks indicating where the film has been wrinkled or folded. This is especially true for the sensitive low-range pressure film which was used in this study. Visual inspection of the films following insertion into the patellofemoral joint and application of quadriceps and moment loads indicated a smooth continuous region of contact with no evidence of wrinkling. Thus, the film had sufficient elasticity to conform to the compound curvatures over the relatively small area of contact developed in this application.

Because the pressure transducer could not make dynamic measurements, discrete joint angles were selected. The flexion angles of 60° and 90° were selected because they correspond to regions of the crank cycle where muscle stresses (Redfield and Hull, 1986) as well as varus and internal moments (Ruby and Hull, 1993) were near maximum, and therefore are of greatest interest from an injury standpoint. The crank angle region encompassed by these knee angles extends from 40° (90° flexion) to 90° (60° flexion) where top dead center is the crank reference position and 90° corresponds to the crank forward and horizotal.

During the crank angle region of interest noted above, six intersegmental knee loads are transmitted through the joint. These loads typically include varus, internal and extension moments and posterior, compressive, and medial forces (Ruby and Hull, 1993). Although all three of the intersegmental knee moments were applied, none of the intersegmental knee forces were applied. It is possible that the application of these forces may affect the response of the knee joint to the applied moments which in turn could affect the mechanics of the patellofemoral joint.

Of the three intersegmental forces, the compressive force had the greatest potential to affect contact patterns measured in this study, not only because of its large magnitude (150–325 N) but also because it increases tibial-femoral friction which can clearly inhibit the internal tibial rotations and possibly varus rotations. While this is true for tibial rotations that occur within the laxity region of the knee when the joint is otherwise unloaded, this study investigated joint mechanics under active quadriceps loads as high as 1600 N. Because a majority of this load is transferred through the knee as a compressive load, the effect of the additional pedal compressive load on tibial rotations would be relatively minor. This assumption was verified through preliminary testing.

Because the LAS could not simulate the hamstring and gastrocnemius muscles, the *in vitro* leg extension moment was developed using the quadriceps muscle alone. However, during cycling all three muscle groups are active during some portion of the crank cycle. EMG data (Jorge and Hull, 1986) indicates that the omission of the hamstring and gastrocnemius muscles should not limit the accuracy of the simulation at a knee flexion angle of 90° (40° crank angle) because at this stage only the quadriceps are active. However at 60° (90° crank angle) the hamstring and gastrocnemius muscles are both active (Jorge and Hull, 1986). Inclusion of the hamstring and gastrocnemius muscles would introduce additional compressive forces which would affect patellofemoral joint mechanics by inhibiting the tibial rotations to some degree.

Owing to the strength limitation of 1600 N in the attachment to the quadriceps tendon, only about 25% of the maximum extensive moment that the quadriceps muscle is capable of developing at the knee was applied. The question arises as to whether the corresponding quadriceps forces were representative of steady-state endurance cycling. The 25% of maximum extensive moments (35 Nm and 26 Nm at 90° and 60° flexion, respectively) corresponded to those computed at the knee for the subjects in Ruby and Hull (1993). Thus, the corresponding quadriceps force levels (1360 N and 1260 N at 90° and 60°, respectively) were representative of the forces actually developed during the tests which measured the changes in the intersegmental loads as affected by the fixed versus free-floating pedals.

Results from this study indicate that the application of either varus or internal moments in combination or alone can lead to increases in the patellofemoral contact force and contact area. Although cartilage injuries caused by mechanical forces are not well understood, data suggest that an increase in patellofemoral force can lead to overloading of the patellar cartilage which over time can weaken the macromolecular framework leading to cartilage fissures and eventually chondromalacia (Buckwalter, 1992). In this study contact force increases as large as 280 N were developed. Currently, it is impossible to say whether force increases of the magnitudes developed in this study are severe enough to produce injurious overloading.

In addition to the overloading pathogenesis, the increase in contact area is a possible injury mechanism. Increasing the contact area places normally unloaded cartilage near the edges of the patella at risk. Around the periphery of the patella the cartilage thins to about 1 mm and therefore could be more susceptible to wear than regions along the patellar ridge where the cartilage is thicker. However, the increase in the contact area in response to the applied loads may be seen as an injury avoidance response. This response allows the contact force to be transmitted across a larger area, thereby reducing contact pressure magnitudes from what they would have been if no increase in area occurred. The long-term effects of these contact area changes are unclear.

If one accepts the idea that abnormal contact patterns and forces lead to degenerative cartilage damage (Outerbridge, 1961; Ficat and Hungerford; 1977), then the reduction in varus and internal intersegmental moments that can result from the use of free-floating pedals may be seen as beneficial from an injury avoidance standpoint. Unfortunately, not every rider experiences reductions in these intersegmental moments through the use of free-floating pedals. In fact, some riders experience increases in the moments (Boyd et al., 1997). Currently, the only way to identify riders who experience intersegmental load reductions is through the use of a six load component pedal dynamometer. Using the dynamometer it is possible not only to determine the magnitudes of the load reduction but also to determine which types of free-floating pedal configurations may help reduce the intersegmental knee loads. Because each rider experiences different reactions to the free-floating pedals (Ruby and Hull, 1993; Boyd et al., 1997), some riders may need pedals that allow either abduction/adduction rotation or inversion/eversion rotation while some riders may need both to achieve substantial intersegmental load reductions. Without this testing it is impossible to say whether a rider suffering from chondromalacia should switch from fixed to free-floating pedals.

Comparison of results with those from previous patellofemoral contact studies (Huberti and Hayes, 1984; Lewallen et al., 1990; Marder et al., 1993), was limited to the results obtained in the absence of varus and internal intersegmental loads (unloaded). The unloaded patellofemoral contact area measurements were lower than those reported by other investigators. To prevent both quadriceps attachment failure as well as saturation of the film, extension moments magnitudes were less than those used in other studies. This most likely accounts for the smaller contact areas measured here.

Mean contact pressures measured in this study for the unloaded condition fell between those reported by other investigators (Huberti and Hayes, 1984; Lewallen et al., 1990; Marder et al., 1993). Because contact pressure is most sensitive to the applied extension moment, the mean contact pressure magnitude measured in this study should be lower than those measured in other studies. However, results from prior studies indicated that the application of identical moments does not insure consistent mean contact pressures (Huberti and Hayes, 1984; Lewallen et al., 1990). Therefore other factors such as size, shape, and condition of the patella as well as image analytical techniques must be controlled if consistent contact pressures are desired.

5. Conclusions

The conclusions from this study can be summarized as follows:

(1) The application of both varus and internal knee moments significantly increased mean contact area and force.
(2) The contact area appeared more sensitive to the internal than varus moment.
(3) Neither varus nor internal moments had a significant effect on the mean contact pressure.

Acknowledgements

The authors are grateful to the Shimano Corporation and the Mercy Medical Foundation for their financial support.

References

Bach, J.M., Hull, M.L., 1995. A new load application system for *in vitro* study of ligamentous injuries to the human knee joint. Journal of Biomechanical Engineering 117, 373–382.

Berns, G.S., Hull, M.L., Patterson, H.A., 1990. Implementation of a five degree of freedom automated system to determine knee flexibility *in vitro*. Journal of Biomechanical Engineering 10, 167–176.

Boyd, T., Neptune, R.R., Hull M.L., 1997. Pedal and knee loads using a multi-degree-of-freedom pedal platform in cycling, Journal of Biomechanics, 30, 505–511.

Buckwalter, J.A., 1992. Mechanical injuries of articular cartilage. In: Finnerman, A.M., Noyes, F.R. (Eds.), Biology and Biomechanics of the Traumatized Joint: The Knee as a Model. American Academy of Orthopaedic Surgeons, Rosemont, IL, pp. 83–96.

Ficat, R.P., Hungerford D.S., 1977. Disorders of the patellofemoral joint. Williams and Wilkins, Baltimore, MD, pp. 29–32.

Fithian, D.C., Hacker, S.A., Kaufman, K.R., Daniel, D.M., 1994. Quadriceps angle revisited: the effect of tibial tubercle transfer on patellofemoral contact and tibial rotations. Transactions of the Orthopaedic Research Society, Vol. 19, p. 678.

Francis, P.R., 1986. Pathomechanics of the lower extremity in cycling. In: Burke, E.R. Newsom, M.M. (Eds.), Medical and Scientific Aspects of Cycling. Human Kinetics, Champaign, IL, pp. 3–16.

Goodfellow, J., Hungerford, D.S., Woods, C., 1976. Patello-femoral joint mechanics and pathology. 2. chondromalacia patellae. The Journal of Bone and Joint Surgery 58-B, 291–299.

Haffajee, D., Moritz, U., 1972. Isometric knee extension strength as a function of joint angle, muscle length and motor unit activity. Acta Orthopaedic Scandinavica 43, 138–147.

Holmes, J.C., Pruitt, A.L., Whalen, N.J., 1991. Cycling knee injuries. Cycling Science 3, 11–14.

Huberti, H.H., Hayes, W.C., 1984. Patellofemoral contact pressure. The Journal of Bone and Joint Surgery 66-A, 715–724.

Huberti, H.H., Hayes, W.C., 1988. Contact ppressures in chondromalacia patellae and the effects of capsular reconstructive procedures. Journal of Orthopaedic Research 6, 499–508.

Hull, M.L., Ruby, P., 1996. Preventing overuse knee injuries in cycling. In: Burke, E.R. (Ed.), Hi-Tech Cycling, Chap 11. Human Kinetics, Champaign, IL.

Insall, J., 1982. Current concepts review patellar pain. The Journal of Bone and Joint Surgery 64-A, 147–152.

Jorge, M., Hull, M.L., 1986. Analysis of EMG measurements during bicycle pedaling. Journal of Biomechanics 19, 683–694.

Lewallen, D.G., Reigger, C.L., Myers, E.R., Hayes, W.C., 1990. Effect of retinacular release and tibial tubercle elevation in patellofemoral degenerative disease. Journal of Orthopaedic Research 8, 856–862.

Lindahl, O., Movin, A., Rinqquist, I., 1969. Knee extension measurement of the isometric force in different positions of the knee-joint. Acta Orthopaedic Scandinavica 40, 79–85.

Marder, R.A., Swanson, T.V., Sharkey, N.A., Duwelius, P.J., 1993. Effects of partial patellectomy and reattachment of the patellar tendon on patellofemoral contact areas and pressures. The Journal of Bone and Joint Surgery 75-A, 35–45.

Mow, V.C., Soslowsky, L.J., 1991. Friction, lubrication, and wear in diarthrodial joints. In: Mow, V.C., Hayes, W.C. (Eds.), Basic Orthopedic Biomechanics, Chap 6. Raven Press, New York.

Outerbridge, R.E., 1961. The etiology of the chondromalacia patellae. Journal of Bone and Joint Surgery 43-B, 752–757.

Powell, B., 1986. Medical aspect of racing. In: Burke, E.R. (Ed.), Science of Cycling Human Kinetics, Champaign, IL, pp. 185–201.

Redfield, R., Hull, M.L., 1986. Prediction of pedal forces in bicycling using optimization methods. Journal of Biomechanics 19, 523–540.

Ruby, P., Hull, M.L., 1993. Response of intersegmental knee loads to foot/pedal platform degrees of freedom in cycling. Journal of Biomechanics 26, 1327–1340.

Rudert, M.G., Rehak, D., Fu, F., 1988. Loading characteristics of pressensor-A pressure sensitive film. Transaction of the Orthopaedic Research Society Vol. 13, p. 226.

Weiss, B.D., 1985. Nontraumatic injuries in amateur long distance bicyclist. The American Journal of Sports Medicine 13, 187–192.

Wheeler, J.B., Gregor, R.J., Broker, J.B., 1995. The effect of clipless float design in shoe/pedal interface kinetics and overuse knee injuries during cycling. Journal of Applied Biomechanics 11, 110–141.