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Patellofemoral and tibiofemoral forces in cyclists and triathletes: effects of saddle height

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Abstract

High compressive load applied to the patellofemoral joint at great knee flexion angle (e.g. >60º of flexion), as usually observed in cycling may provide a link between body position on the bicycle and knee joint forces. The aim of the present study was to compare knee joint forces in different saddle heights. Right pedal force and lower limb kinematics were collected for 24 competitive cyclists and triathletes at self-selected preferred, high (-10º of knee flexion angle from preferred), low (+10º of knee flexion angle from preferred) and optimal (25º of knee flexion angle) saddle heights in submaximal pedalling cadence and workload trials (3.45 ± 0.6 W/kg, 90 ± 2 rpm and 163 ± 33 J). Patellofemoral compressive and tibiofemoral anterior-posterior and compressive force were computed by inverse dynamics and compared for different saddle heights via effect sizes. Patellofemoral compressive force (5-13%) and tibiofemoral compressive force (1-7%) were not substantially affected by changes in saddle height. Tibiofemoral anterior shear force decreased at low saddle heights (4-6% of the preferred height) compared to optimal (35%) and high saddle heights (53%). Greater knee flexion angles were observed for lower saddle heights (8-34%). Knee flexion angle was significantly affected by changes in saddle height, which may indicate that using joint kinematics to assess saddle height effects may be useful to anticipate overload in knee joint.

Keywords: knee forces, bicycle, knee angle, inverse dynamics, pedal forces, lower limb kinematics

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Introduction

Knee pain is one of the most reported reasons for elite cyclists to reduce training and performance (Clarsen et al. 2010). The most likely explanation for that is the compressive force been repetitively applied at the patellofemoral cartilage (Callaghan 2005). Furthermore, the application of physiological load on the patellofemoral cartilage after the start of progressive degeneration can compromise the integrity of the cartilage matrix (Cohen et al. 2001). These reasons raise a concern that high compressive load applied to the patellofemoral joint at great knee flexion angle (e.g. >60º of flexion), as usually observed in cycling (Ericson and Nisell 1987), may provide a link between body position on the bicycle and knee joint forces.

The saddle height has been shown to affect the kinematics of the knees, with lower saddle height leading to greater flexion angle (Nordeen-Snyder 1977; Tamborindeguy and Bini 2011). Thus, using a lower saddle height may result in greater compressive patellofemoral force (*see* Figure 1) due to greater force required by the quadriceps muscle group (Ericson and Nisell 1987). Also, an increase in tibiofemoral anterior-shear and compressive components would be expected because of the greater quadriceps force using a lower saddle height, which would overload the anterior-posterior cruciate ligaments and the menisci, respectively. However, conflicting effects of saddle height have been shown on tibiofemoral forces (Ericson and Nisell 1986; McCoy and Gregor 1989; Tamborindeguy and Bini 2011) and patellofemoral compressive force (Ericson and Nisell 1987; Tamborindeguy and Bini 2011).

Previous studies were limited to small sample sizes (up to 10 participants) of non-cyclists. To date, no study has been found in which a reasonable sample size of athletes with competitive experience in cycling are assessed. Also, all previous researchers have determined the saddle height based on lower limb length methods (e.g. trochanteric leg length), which has been shown to result in difference in knee joint kinematics (Peveler et al. 2005). Therefore, to assess saddle height effects on knee forces depends on the most similar possible knee kinematics across the subjects, otherwise between-subjects variability may compromise final conclusion. This is also likely to explain the conflicting findings of previous studies. There is also concern that only large changes in saddle height (e.g. ~4% of the reference saddle height) (Gonzalez and Hull 1989) would affect lower limb forces and moments, which is unrealistic to usual changes performed in bicycle configuration assessment.

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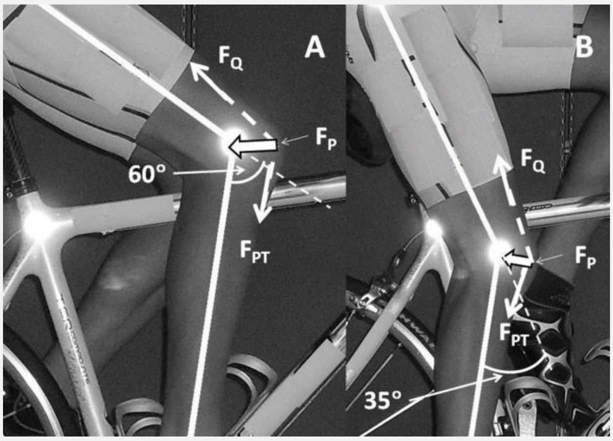

Figure 1. Schematic illustration of 25º of change in the knee flexion angle from 60º (A) to 35º (B) that should theoretically decrease patellofemoral compressive force (FP). Arrows indicate quadriceps muscle force (FQ) and patellar tendon force (FPT).

In summary, saddle height effects on patellofemoral and tibiofemoral forces in cyclists with competitive experience in cycling are not clear to date. Therefore, the aim of the present study was to compare knee joint forces of cyclists with competitive experience in cycling and triathlon riding in different saddle heights. It was hypothesised that saddle height would have a large influence on knee flexion angle but not on knee forces. The reason for this is that when saddle height is varied, knee forces may be balanced by changes in the hip and ankle kinetics, without a specific effect at a single joint. In other words, even with large changes in joint kinematics, either muscle capacity to generate power at the hip, knee and ankle joints may not be substantially affected or individual changes in muscle capacity to generate power (e.g. lower knee joint extensors power) may be balanced by hip and/or ankle joint muscles when saddle height is changed (e.g. increase hip and/or ankle joint power).

Materials and methods

Participants

Twenty four cyclists with competitive experience in cycling and triathlon without reports of knee pain or injury volunteered for the study. This sample size was accepted based on expected mean and within subjects variability reported for saddle height effects on pedal forces and joint kinematics (Diefenthaeler et al. 2006), calculated using Eng (2003) equations and bigger than previous studies with similar design (Diefenthaeler et al. 2006; Ericson and Nisell 1986, 1987). The characteristics of the 24 cyclists were (average ± SD) 39 ± 11 years of age, 75 ± 15 kg of body mass, 177 ± 8 cm of height, and 7 ± 3 weekly hours of training. Prior to the study, the participants were informed about possible risks and signed a consent form approved by the ethics committee of human research where the study was conducted.

Data collection

On the evaluation session, the athletes were interviewed to assess their history of training and injuries related to bicycle riding. After the interview, anthropometrics (height and body mass) were measured according to the International Society for Advancement of Kineanthropometry protocols (Marfell-Jones et al. 2006). Cyclists' bicycle saddle height and horizontal position were measured to set up the stationary cycle ergometer (Velotron, Racemate, Inc) at the self-selected "preferred height" configuration. Knee joint flexion angle was measured using a goniometer while cyclists held the pedal crank at the 6 o'clock position. Firstly, the lateral femoral condyle was located and defined as the knee joint centre of rotation. A midpoint between the anterior superior iliac spine and the greater trochanter was determined to define the hip joint centre (Neptune and Hull 1995). The lateral maleolus was also marked to define the ankle joint centre. Both shafts of the goniometer were aligned to the femur (following an imaginary line connecting the knee and hip joint centres) and to the tibia (following an imaginary line connecting the knee and the ankle joint centres). The resulting angle was defined as the knee flexion angle. Saddle height was recorded when the saddle was changed from the preferred position to high (-10º of knee flexion with respect to the knee flexion angle measured at the preferred height), low (+10º of knee flexion with respect to the knee flexion angle measured at the preferred height), and to the theoretical optimal (25º of knee flexion). This last saddle height was chosen because it was advocated to reduce the risk of knee injuries (Holmes et al. 1994).

Cyclists then performed 10 minutes of warm-up cycling at 90 ± 2 rpm of pedalling cadence and workload set at 150 W on the stationary Velotron cycle ergometer using their preferred saddle height and horizontal position. Workload was then increased to match 3.4 ± 0.6 W.kg-1 (257 ± 52 W) and pedalling cadence was visually controlled at 90 ± 2 rpm for two minutes (163 ± 33 J). Data were recorded during the first 20 s of the second minute for each saddle height trial. One minute of static rest was enforced between trials with different saddle heights when changes in saddle height were conducted. The order of trials using the high, low and optimal saddle heights were randomly defined for each cyclist.

Force applied on the right pedal and right lower limb kinematics were recorded for the last 20 seconds during all aforementioned conditions. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of the cyclists at the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, anterior and posterior pedal stick. One marker was attached to the sacrum to measure the position of the cyclists when they were evaluated at the different saddle heights. Two markers were taped to the bicycle frame and used as reference for image calibration. A 2D pedal dynamometer (Candotti et al. 2007) and one high speed camera positioned perpendicular to the motion plane (AVT PIKE F-032, Allied Vision Technologies GmbH, Germany) were synchronized by an external trigger. Kinematics were recorded at 60 frames per second using AVT ActiveCam viewer software (Allied Vision Technologies GmbH, Germany) and force data were recorded at 600 Hz per channel employing a 16-bits analogical to digital converter (PCI-MIO-16XE-50, National Instruments, USA) using a custom Matlab (Mathworks Inc, MA) data acquisition script.

Data analysis

Video files were digitized and automatic tracking of markers were conducted in DgeeMe software (Video4Coach, Denmark) for x-y coordinates over time. Kinematic data were smoothed with a digital second order zero lag band pass Butterworth filter with cut-off frequency optimized to reduce signal residual (Winter 2005). Segment kinematics of the hip, knee, and ankle joints during pedalling movement were calculated from the smoothed x-y coordinate data. Knee flexion angle was measured from segment kinematics as illustrated in Figure 1 (full extension = zero). Correction of the hip joint center based on the average coordinate between the marker on the anterior superior iliac spine and the greater trochanter was performed (Neptune and Hull 1995). The average relative horizontal position of the marker on the sacrum to the bottom dead centre of the bicycle was computed over time across ten pedal revolutions for the analysis of body position on the saddle at the four saddle heights.

Linear and angular velocities and accelerations were computed from smoothed kinematic data by a three points derivative method (Winter 2005). Pedal angle in relation to the global coordinate system was calculated to convert the forces on the pedal reference system to forces in the global reference system by means of trigonometric procedures (Marsh et al. 2000). The right lower limb was modelled as a three-segment rigid body system (thigh, shank and foot-pedal) with segment mass and centre of mass estimated according to De Leva (1996). Conventional inverse dynamics was conducted to calculate the net joint moments at the knee and ankle (Redfield and Hull 1986) using adapted scripts of van den Bogert and de Koning (1996). Patellofemoral (compressive force) was computed as described by Bressel (2001) including corrections for quadriceps-patellar tendon force ratio (Sharma et al. 2008). Tibiofemoral compressive and shear forces were computed as described by Thambyah et al. (2005). Peak patellofemoral force and tibiofemoral components (anterior shear and compressive) were calculated due to their potential relationship with load on the anterior cruciate ligament and menisci (Neptune and Kautz 2000). Knee flexion angle when the crank was at 3 o'clock and 6 o'clock crank positions were determined from segment kinematics taken from cycling motion. Knee forces and flexion angle were computed using a custom written program in Matlab (Mathworks Inc, MA) for ten consecutive crank revolutions to determine average and standard deviation for each participant.

Statistical analyses

Between cyclists means and standard deviations were reported for the tibiofemoral (anterior shear and compressive) and patellofemoral (compressive) peak force and for the knee angle when the crank was at 3 o'clock and 6 o'clock crank positions. Data normality distribution and sphericity was evaluated by the Shapiro-Wilk and Mauchly tests, respectively. Data normality was corrected for patellofemoral compressive force, tibiofemoral compressive and anterior force via logarithm transformation. Peak patellofemoral compressive and tibiofemoral (anterior and compressive) forces were normalized by the workload (in Joules) of each cyclist.

To compare the effect of the saddle height on the patellofemoral compressive force, tibiofemoral anterior shear and compressive force, knee angle at 3 o'clock and 6 o'clock crank positions of the crank, Cohen's effect sizes (ES) were computed for the analysis of magnitude of the differences, and rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen 1988).

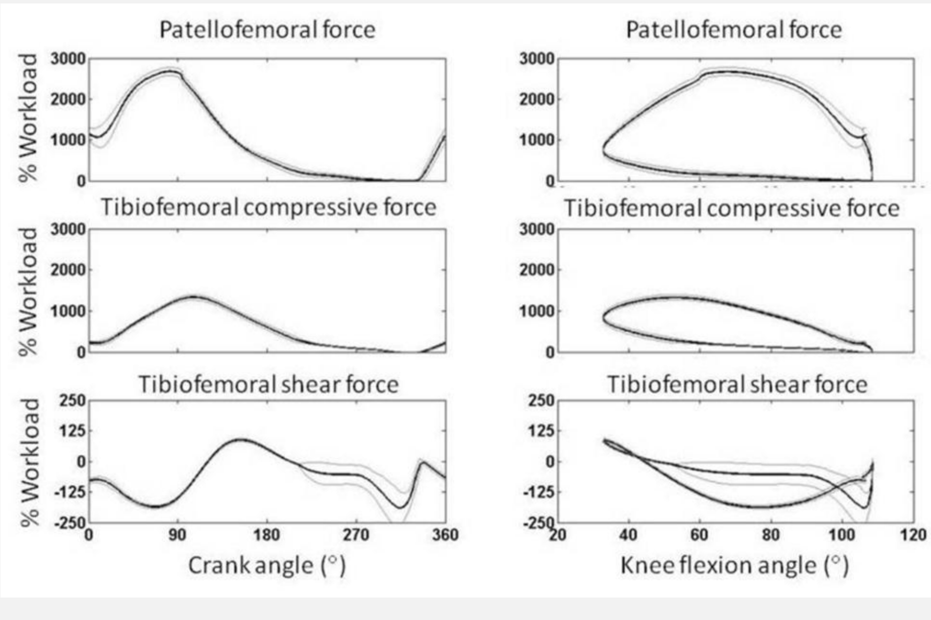

Figure 2. Ensemble data from one cyclist for patellofemoral compressive force, tibiofemoral compressive and shear components taken at the preferred saddle height trial. Force data were normalized by cyclists' workload level (151 J) and 90 rpm of pedalling cadence.

Results

Large changes in saddle height (except comparing high and optimal saddle

Table 1. Results (mean ± SD) of saddle height, patellofemoral compressive force, tibiofemoral anterior and compressive force, knee flexion angle at 3 o'clock and 6 o'clock crank positions presented for four saddle heights (preferred, high, low and optimal). Differences between saddle heights are reported as mean difference percentages along with effect size magnitudes. Percentage and magnitude of differences presented for different saddle heights.

	Optimal	High	Preferred	Low
Saddle height (cm)	89 ± 4.7 High 1%; 0.3, S *Pref 4%; 2.3, L* *Low 6%; 5.0, L*	88 ± 5.3 *Pref 3%; 2.1, L* *Low 5%; 4.8, L*	86 ± 5.0 *Low 2%; 3.8, L*	84 ± 5.2
Patellofemoral compressive force (%Workload)	1573 ± 300 High 5%; 0.3, S Pref 6%; 0.3, S Low 9%; 0.4, S	1489 ± 293 Pref 12%; 0.6, M Low 13%; 0.7, M	1673 ± 312 Low 3%; 0.1, T	1720 ± 362
Tibiofemoral anterior force (%Workload)	78 ±14 High 2%; 0.1, T *Pref 19%; 1.0, L* *Low 35%; 1.6, L*	76 ±13 Pref 17%; 0.9, M *Low 53%; 1.5, L*	63 ±17 Low 26%; 0.7, M	51 ± 21
Tibiofemoral compressive force (%Workload)	958 ± 372 High 1%; 0.1, T Pref 7%; 0.2, S Low 5%; 0.1, T	947 ± 375 Pref 5%; 0.1, T Low 3%; 0.1, T	892 ± 260 Low 2%; 0.1, T	913 ± 280
Knee flexion angle 3 o'clock crank position (°)	48 ± 5 High 3%; 0.4, S *Pref 12%; 1.5, L* *Low 18%; 1.8, L*	49 ± 5 *Pref 8%; 1.0, L* *Low 13%; 1.6, L*	53 ± 4 Low 6%; 0.7, M	57 ± 5
Knee flexion angle 6 o'clock crank position (°)	33 ± 7 High 6%; 0.3, S Pref 16%; 0.9, M *Low 34%; 1.9, L*	31 ± 6 Pref 25%; 1.4, M *Low 23%; 2.6, L*	38 ± 5 *Low 13%; 1.2, L*	44 ± 4

Abbreviations used for comparisons are preferred saddle height (Pref) and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in italics.

heights) resulted in small to moderate increases in patellofemoral force for the preferred and low saddle heights compared to the high saddle height. Large decreases in anterior tibiofemoral force were observed for the preferred saddle height compared to the optimal saddle height and for the low saddle height compared to the optimal and high saddle heights. Trivial to small changes were observed for the tibiofemoral compressive force due to changes in saddle height. Greater knee flexion angle was observed towards lower saddle heights with large differences for the 3 o'clock and 6 o'clock crank positions (*see* Table 1).

Discussion

The aim of the present study was to compare patellofemoral and tibiofemoral joint forces in athletes with competitive experience in cycling and triathlon using different saddle heights. The main findings were that patellofemoral and tibiofemoral compressive forces were not substantially affected by changes of 1-6% of saddle height. Smaller tibiofemoral anterior force was observed at the low saddle height along with greater knee flexion angles.

To reduce the risk of overuse knee injuries and optimize cycling efficiency, cyclists are recommended to use a saddle height that elicit 25-30º of knee flexion when the crank is at 6 o'clock position (Burke and Pruitt 2003; Holmes et al. 1994). However, there was no evidence to date that using a saddle height out of this range of knee flexion angle would result in smaller knee joint forces. The cyclists from our study presented ~38º of knee flexion when the crank was close to the 6 o'clock crank position, which would suggest a greater risk of overuse injuries on the knees using the advocated guidelines. However, changes up to 6% in saddle height were conducted without substantial effects in patellofemoral compressive force. This result is in line with previous findings that changes smaller than 3% of trochanteric height do not substantially affect patellofemoral compressive force (Tamborindeguy and Bini 2011). Decreases up to 10% in saddle height resulted in substantial larger patellofemoral compressive force (Ericson and Nisell 1987) which may suggest that changes in saddle height should be greater than 6% to substantially reduce patellofemoral compressive force.

In contrary to previously expected anterior tibiofemoral shear force increased at the high saddle height. One explanation for this result was that the anterior tibiofemoral force depends on the patellar moment arm, which is expected to increase at the high saddle height, using Herzog and Read (1993) model. Therefore, assuming no substantial changes in quadriceps muscle force due to changes in saddle height up to 6% (based on small to moderate changes in patellofemoral compressive force), the effect from quadriceps muscle force to tibiofemoral anterior force would be patellar moment arm.

Apart from that, the importance of the tibiofemoral anterior shear force has been reduced in cycling because no reports of anterior cruciate ligament were indicated in a previous review study (Dettori and Norvell 2006). Low levels of strain on the anterior cruciate ligament were found during in vivo measures

(Fleming et al. 1998). The effects of changing saddle height on tibiofemoral compressive force were not substantial in the present study. This finding are in line with any reports of menisci overuse injury (Dettori and Norvell 2006). Therefore, changes in saddle height up to 6% of the preferred saddle height are unlikely to affect the anterior cruciate ligament or the menisci.

Knee joint flexion angle has been used as a reference to configure the saddle height, with especial attention to reduce the likelihood of knee injuries (Burke and Pruitt 2003; Holmes et al. 1994). In agreement with previous findings (Nordeen-Snyder 1977; Tamborindeguy and Bini 2011), an inverse relationship between saddle height and knee flexion angle was observed in the present study. The commonly recommended assessment of knee flexion angle at the 6 o'clock crank position was complemented with the analysis of knee flexion angle at the 3 o'clock crank position. Greater force is applied to the pedal at the 3 o'clock crank position (Coyle et al. 1991) compared to the 6 o'clock crank position, therefore, knee forces are more likely to be minimized if saddle height is assessed at the 3 o'clock crank position instead of the commonly used 6 o'clock crank position. The results of the present study highlighted that joint kinematics are more sensitive to changes in saddle height than knee forces. One reason is that small changes were found in pedal forces when changing saddle height (Bini et al. 2011; Ericson and Nisell 1988), which may suggest that muscle force production may be tuned to sustain workload via changes in joint kinematics.

Different knee flexion angles were found during cycling motion compared to static measures. From video analysis, the 6 o'clock crank position resulted in greater knee flexion angle than the measure taken statically at the 6 o'clock position. This finding may be explained by the lack of angular momentum at the 6 o'clock crank position during static poses and differences in ankle angle. Therefore it is expected that guidelines for saddle height configuration may be based on joint angles taken dynamically, not statically.

The study was limited to the assessment of right sagittal plane during stationary cycling in a cycle ergometer. Therefore, out of plane movements of the lower limb are not accounted in this design, which would be expected to have minor effect on sagittal plane variables (Umberger and Martin 2001). Bilateral asymmetries in pedal force and joint kinematics may have affected the conclusions, however, competitive cyclists presented differences up to 7% between bilateral pedal forces (Bini et al. 2007). Also, modelling joint kinetics (e.g. knee forces) without information on muscle activation (e.g. muscle length and activation profile) may be less reliable then forward dynamics simulations (Neptune and Kautz 2000). However, most studies have used only the kinetic-kinematic model (Bressel 2001; Ericson and Nisell 1987; Tamborindeguy and Bini 2011), which enables the comparison of our results to others of similar design.

Practical applications

Patellofemoral and tibiofemoral compressive forces are not substantially affected by changes in saddle height within a range of 4-6% of the preferred height. Tibiofemoral anterior shear force decreases by 19-53% when saddle height is decreased, however, it may not affect injury risk in cycling. Greater knee flexion angle is observed at lower saddle heights, which may indicate that using joint kinematics to assess saddle height may be useful to anticipate overload in knee joint.

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Conflict of interest
None.

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