The effects of conventional and oval chainrings on patellofemoral loading during road cycling: an exploration using musculoskeletal simulation.

Jonathan Sinclair[1], Philip Stainton[1], & Benjamin Sant[1]

1. Centre for Applied Sport & Exercise Sciences, Faculty of Health & Wellbeing, University of Central Lancashire, Lancashire, UK.

Contact Details:

Dr. Jonathan Sinclair

Centre for Applied Sport & Exercise Sciences

Faculty of Health & Wellbeing

Preston

Lancashire

PR1 2HE

e-mail: jksinclair@uclan.ac.uk

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Abstract

PURPOSE: The aim of the current investigation was to utilize a musculoskeletal simulation approach to resolve muscle forces during the pedal cycle, in order to specifically examine the effects of chainring geometry on patellofemoral loading during cycling.

METHODS: Fifteen healthy male recreational cyclists rode a stationary cycle ergometer at a fixed cadence of 70 RPM in two chainring conditions (round and oval). Patellofemoral loading was explored using a musculoskeletal simulation and mathematical modelling approach. Differences between chainring conditions across the entire pedal cycle were examined using 1-dimensional statistical parametric mapping and patellofemoral force experienced per 20 km was explored using a paired samples t-test.

RESULTS: No significant (P>0.05) differences in patellofemoral force or stress were found throughout the pedal cycle between chainring conditions. It was also shown that no significant (P>0.05) differences in patellofemoral force per 20 km joint were evident (round 38576.40 N/kg·s & oval = 35637.00 N/kg·s).

CONCLUSIONS: The current analysis found no effects of chainring geometry, on the forces experienced by the patellofemoral joint during the pedal cycle.

Introduction

During linear road cycling using traditional circular chainrings, the application of tangential force is lowest when the crank in the vertically aligned, either at 0 or 180° of the pedal cycle, and maximal when the crank is horizontally aligned (1). The points during the pedal cycle where tangential force is lowest are typically referred to as upper and lower dead points (2). In an attempt to improve road cycling performance and maximize the application of effective force during the pedal cycle, oval chainrings were introduced, whereby the axes of the chainring are not perpendicular (3). This shape means that the moment arm of the force being applied to the chain is reduced at the dead points of the pedal cycle but increased when the crank is horizontally aligned (3). This optimizes the period of the pedal cycle in which tangential force is produced, and correspondingly reduces the time spent in the upper and lower dead points (4).

Quantitative analyses investing performance parameters with oval chainrings have shown inconsistent findings. Hintzy et al., (4) showed that peak power output was significantly higher when using a non-circular chainrings during short duration maximal spring cycling. Hintzy & Horvais, (5) similarly found that higher maximal aerobic power was attained when using a non-circular chainring during maximal incremental tests. Horvais et al., (6) examined mechanical and physiological parameters during 8 minute submaximal and 8 s maximal tests. During the submaximal test the oval chainring produced lower crank torques at 0° and 180° and greater torques at 90° of the pedal cycle. During the sprint test, the biceps femoris exhibited a longer burst of activation in the oval chainring condition. Conversely, Cordova et al., (2) showed that there were no significant differences in physiological responses during an incremental test until exhaustion. Similarly, Peiffer & Abbiss, (7) found that there were no differences in physiological and performance parameters between oval and round chainrings during a 10 km cycling time trial. Finally, Dagnese et al., (8) similarly showed that there were no significant differences in lower extremity muscle activation magnitude between oval and round chainrings.

Further to this, Bisi et al., (9) showed that oval chainrings altered lower extremity joint kinetics, with reductions of 6% in the knee joint moment, which they identified may have implications for chronic injury prevention at this joint. Importantly the knee joint is the musculoskeletal structure most susceptible to chronic pathology in cyclists (10). Specifically, patellofemoral pain is the most frequently experienced condition, affecting 36% of all regular cyclists' and accounting for more than 57% of all time-loss injuries (11). Despite the incidence of patellofemoral pain in cyclists it has received a paucity of attention in scientific literature in relation to other athletic disciplines. Therefore, further exploration of this condition is clearly warranted in cycling specific analyses.

Patellofemoral pain is initiated by activities that place frequent and excessive mechanical loads at the joint (12, 13). Therefore, quantification of patellofemoral loading is important in cycling specific activities as we seek to understand more about this condition and the potential mechanisms that may be important to prevent the high incidence of patellofemoral pain. Although validated mathematical models of the patellofemoral joint are available in biomechanical literature (14, 15), they typically require inverse joint dynamics to resolve muscle kinetics as input parameters into the musculoskeletal algorithm. Whilst this is suitable for movements which involve full foot contact with a force platform, this is not available for cycling specific analyses, which may help to explain the lack of scientific attention concerning to patellofemoral pain in road cycling.

However, advances in musculoskeletal modelling have led to the development of bespoke software which allows skeletal muscle force distributions to be simulated during movement using motion capture based data (16). To date, such approaches have not yet been utilized in cycling specific analyses. The aim of the current investigation was therefore to utilize a musculoskeletal simulation approach to resolve muscle forces during cycling to examine the effects of chainring geometry on patellofemoral loading during the pedal cycle. A study of this nature may provide important clinical information regarding the effects of different chainring technology on the susceptibility of road cyclists to patellofemoral pain.

Materials & methods

Participants

Fifteen male recreational cyclists, who habitually utilized round chainrings for their training volunteered to take part in this study. Cyclists were required to have at least 2 years of road cycling experience and be free from musculoskeletal pathology at the time of data collection. The mean characteristics of the participants were; age 28.11 ± 5.11 years, height 1.80 ± 0.10 m and body mass 75.10 ± 8.22 kg. The procedure utilized for this investigation was approved by the University of Central Lancashire, Science, Technology, Engineering and Mathematics, ethical committee (Ref: 511) and all participants provided written informed consent

Procedure

Participants rode a stationary cycle ergometer SRM 'Indoor Trainer' (SRM, Schoberer, Germany) for 10 minutes at a fixed cadence of 70 RPM using a 52x15 gear ratio. To ensure that the current investigation examined only the effects of the different chainrings, the set-up parameters were constructed in accordance with previous recommendations (17), and standardized between the two conditions. Cycling shoes (Northwave Sonic 2 Plus Road Shoes, Northwave, Italy), pedals (Look Keo Classic 2, Look, Cedex, France) and cleats (Look Keo Grip, 4.5˚ float, Look, Cedex, France) were consistent across all trials, and adjusted so that the 1st metatarsal head was positioned superior to the pedal spindle (18). The participants were provided with continuous visual feedback regarding their cadence, which was visible via the SRM head unit (Powercontrol V, SRM, Schoberer, Germany).

The participants rode in two conditions one with a traditional round chainring (SRM power, SRM, Schoberer, Germany) and one using an oval shaped chainring (Osymetric, standard, USA), with a crank length of 172.5mm. To prevent any order effects in the experimental data, the order in which participants rode in each chainring condition was counterbalanced and a standardized rest period of 10 minutes was allowed between trials. The ergometer setup was organized based on each participant own preference and maintained between the two chainring conditions.

Kinematic information from the lower extremity joints was obtained using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) using a capture frequency of 250 Hz. To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet retroreflective markers were placed at the C7, T12 and xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. In addition to these the foot segments were tracked via the calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked using the PSIS and ASIS markers and the thorax segment was tracked using the T12, C7 and xiphoid markers. Static calibration trials were obtained with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters/markers. A static trial was conducted with the participant in the anatomical position in order for the anatomical positions to be referenced in relation to the tracking markers, following which those not required for dynamic data were removed.

Processing

Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). Marker data were smoothed using a cut-off frequency 12 Hz using a low-pass Butterworth 4th order zero-lag filter; this was established using residual analysis similar to Sinclair et al., (19).

Data from five pedal cycles in each chainring condition were exported from Visual 3D into OpenSim 3.3 software (Simtk.org). The five extracted pedal cycles were obtained during minutes 4-6 of the experimental protocol, and the pedal cycle itself was delineated in accordance with Sinclair et al., (19). A validated musculoskeletal model (gait2392) with 8 segments, 19 degrees of freedom and 92 musculotendon actuators (Delp et al., 2007) was used to resolve muscle kinetics during the pedal cycle. The model was scaled for each participant using the anthropometrics and segment inertial properties generated from the static trial to account for the dimensions of each athlete. We firstly performed a residual reduction algorithm (RRA) within OpenSim, this utilizes the inverse kinematics that were exported from Visual 3D. The RRA calculates the joint torques required to re-create the dynamic motion. The RRA calculations produced route mean squared errors <2°, which correspond with the recommendations for good quality data. Following the RRA, the computed muscle control (CMC) procedure was then employed to estimate a set of muscle force patterns allowing the model to replicate the required kinematics (20). The CMC procedure works by estimating the required muscle forces to produce the net joint torques.

Patellofemoral loading during cycling was quantified using a model adapted from van Eijden et al., (14) in accordance with the protocol of Willson et al., (21). A key drawback of this model is that co-contraction of the knee flexor musculature is not accounted for. Taking this into account, summed hamstring and gastrocnemius forces derived from the CMC procedure were multiplied by their estimated knee joint muscle moment arms as a function of knee flexion angle (22) and then added together to determine the knee flexor torque during the pedal cycle. In addition to this the knee extensor torque was also calculated by dividing the summed quadriceps forces by this muscle groups' knee joint muscle moment arms as a function of knee flexion angle (14). The knee flexor and extensor torques were then summed and subsequently divided by the quadriceps muscle moment arm (14) to obtain quadriceps force adjusted for co-contraction of the knee flexor muscles (21). Patellofemoral force was quantified by multiplying the derived quadriceps force by a constant which was obtained by using the data of Eijden et al., (14). Finally, patellofemoral joint stress was quantified by dividing the patellofemoral force by the patellofemoral contact area. Patellofemoral contact areas were obtained by fitting a polynomial curve to the sex specific data of Besier et al., (12), who estimated patellofemoral contact areas as a function of the knee flexion angle using MRI.

Following this the patellofemoral force, muscle force and knee flexion angle data for each participant during the entire pedal cycle were extracted and time normalized to 101 data points. All joint and muscle force parameters were subsequently normalized by dividing the net values by body mass (N/kg). In addition to this, the patellofemoral force integral during the pedal cycle was obtained using a trapezoidal function. As cycling requires a uniquely recurrent movement pattern, with a significant number of pedal cycles to complete typical training/ competitive distances, the total patellofemoral force experienced per 20 km was also extracted. This was resolved firstly by quantifying the velocity of the bicycle using the gear ratio, cadence and typical wheel diameter/ tire width. Using this information (neglecting for air resistance and assuming that the velocity was uniform) the time taken to cycle 20 km could then be calculated. From this the number of pedal cycles required to complete the aforementioned distance was calculated. Finally, in accordance with Sinclair et al., (23) the patellofemoral force integral was multiplied by the number of pedal cycles necessary to cycle 20 km to extract the patellofemoral force experienced during this distance.

Analyses

Differences in patellofemoral and muscle forces across the entire pedal cycle were examined using 1-dimensional statistical parametric mapping with MATLAB 2017a (MATLAB, MathWorks, Natick, USA), in accordance with (24), using the source code available at http://www.spm1d.org/. For patellofemoral force per 20 km, descriptive statistics of means, standard deviations (SD) and 95 % confidence intervals (*95% CI*) were calculated for both chainring conditions. Differences in patellofemoral force per 20 km between chainring conditions were examined using a paired samples t-test. Effect sizes were calculated using partial eta^2 (pη^2). The alpha (α) level for statistical significance was set at the 0.05 level throughout. Discrete statistical tests were conducted using SPSS v23.0 (SPSS, USA).

Results

Table 1 and figures 1-6 present differences in muscle kinetics and patellofemoral loading as a function of the different chainring conditions.

Patellofemoral loading

No significant differences (P>0.05) in patellofemoral loading were evident across the pedal cycle as a function of the different chainring conditions (Figure 1-2). In addition, no significant (P>0.05) differences in patellofemoral force per 20 km were evident between chainring conditions (Table 1).

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@@@ FIGURE 1 NEAR HERE @@@

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Muscle kinetics

No significant differences (P>0.05) in muscle kinetics were evident across the pedal cycle as a function of the different chainring conditions (Figure 3-6).

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Discussion

The aim of the current investigation was to examine the effects of chainring geometry on patellofemoral loading throughout the pedal cycle using a statistical parametric mapping approach. To the authors knowledge this represents the first investigation to quantify the effects of different chainrings on the loads experienced by this joint throughout the pedal cycle. Given the high incidence of patellofemoral pain in road cyclists this investigation may provide important information concerning the effects of different bicycle technology regarding cyclists' susceptibility to chronic pathologies.

The key observation from the current study is that no significant differences in patellofemoral loading parameters were observed at any point during the pedal cycle as a function of the different chainring geometries examined as a part of this investigation. This opposes the proposition initiated by Bisi et al., (9) which denoted that the reduction in knee joint moment observed in the oval chainring condition may have implications for chronic injury prevention at this joint. This disagreement is likely due to the distinction between joint inverse dynamics and specific indices of joint loading; it has been shown that alterations in joint torque do not necessarily reflect changes in joint loading (25). Therefore, it can be concluded from this investigation that chainring geometry does not appear to influence patellofemoral loading during the pedal cycle.

It is proposed that this finding relates to the lack of statistical differences in muscle kinetics between the two conditions. No differences in knee flexor/ extensor muscle kinetics were observed between round and oval chainrings at any point during the pedal cycle. Importantly, Herzog et al., (26) have shown that muscles are the main determinant of joint forces. In addition, the current study showed that there were no differences in knee joint kinematics throughout the pedal cycle, between the two chainring conditions. Taking into account that patellofemoral contact area (12) and knee flexor/ extensor muscle moment arms (14, 22), were modelled as a function of the knee joint angle, provides further insight into the absence of statistical differences in patellofemoral loading between conditions.

There is a clear link between excessive patellofemoral joint kinetics and the aetiology and progression of patellofemoral pain (12, 13). The current study represents the first investigation firstly to explore patellofemoral kinetics during the pedal cycle using a mathematical model that accounts for co-contraction of the knee flexor musculature but also to quantify the loads experienced by this joint during a typical cycling training/ competitive distance. The findings show that cyclists experience considerable patellofemoral loads, indeed although the peak forces during the pedal cycle (round = 27.86 & oval = 25.92 N/kg) are lower than those during the stance phase of running which range between; 31.29 - 76.4 N/kg (27-29); the cumulative loads observed during the current study (round = 38576.40 N/kg·s & oval = 35637.00 N/kg·s) over the same linear distance are larger than those experienced during running which range between; 27774.07 - 30721.33 N/kg·s (23). This is a thought-provoking statistic which helps to contextualize the high incidence of patellofemoral pain in cyclists and highlights the lack of scientific research into the patellofemoral joint in cycling. There is currently a clear requirement for both prophylactic and treatment intervention studies in cycling which are almost entirely absent in scientific literature. This will serve to address the underlying epidemiological factors associated with patellofemoral pain in cyclists and most importantly initiate a body of clinical research concerning sustained conservative treatment modalities.

Limitations & conclusions

A limitation of the current investigation is that only healthy cyclists were examined. It is currently unknown whether cyclists with patellofemoral pain differ in their joint loading in comparison to healthy athletes, but Dieter et al., (10) demonstrated that cyclists with patellofemoral pain exhibit altered muscle activation patterns compared to healthy controls. Therefore generalizations of the current observations results to cyclists with existing patellofemoral symptoms should be made with caution. A second potential drawback is that patellofemoral loading was extracted using a mathematical modelling approach. Whist this procedure was considered an improvement over previous approaches in that co-contraction of the knee flexor musculature was accounted for; individualized muscle moment arms and patellofemoral contact areas are still not available within biomechanical literature. Finally, that the current investigation examined cyclists who do not habitually ride using oval shaped chainrings, may limit the generalizability of the results, which may have differed had the riders been more familiar with this chainring condition. Therefore, it is important for the current investigation to be repeated using cyclists who habitually utilize oval chainrings, which will allow more definitive conclusions to be drawn.

In conclusion, although the effects of altering the geometry of the chainring have been investigated previously, current knowledge regarding the effects of oval chainrings on patellofemoral loading during cycling is lacking. This study consequently adds to the current literature base in the field of biomechanics by presenting a comprehensive examination of patellofemoral loading parameters during linear cycling with both round and oval chainrings.

The findings from current work show that the no differences in patellofemoral loading were evident between the two chainring conditions. This therefore indicates that chainring geometry does not significantly influence patellofemoral loading linked to the aetiology of patellofemoral pain during cycling.

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Compliance with ethical standards

Conflict of interest

We declare that we have no conflict of interest.

Ethical approval

All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and the declaration of Helsinki.

Informed consent

All of the subjects provided written consent.

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Figure labels

Figure 1: Patellofemoral force (a.), stress (b.) and (c.) sagittal plane knee angle as a function of chainring geometry.

Figure 2: Comparison of patellofemoral force (a.), stress (b.) and (c.) sagittal plane knee angle between conditions, positive values indicate that the round chainring values exceed those in the oval condition (SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines).

Figure 3: Knee extensor kinetics (a.), rectus femoris (b.), vastus lateralis (c.) and vastus medialis (d.) vastus intermedius as a function of chainring geometry.

Figure 4: Comparison of rectus femoris (a.), vastus lateralis (b.), vastus medialis (c.) and (d.) vastus intermedius between conditions, positive values indicate that the round chainring

values exceed those in the oval condition (SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines).

Figure 5: Knee flexor kinetics (a.) semimembranosus, (b.) semitendinosus, (c.) biceps femoris short head, (d.) biceps femoris long head, (e.) lateral gastrocnemius and (f.) medial gastrocnemius as a function of chainring geometry.

Figure 6: Comparison of semimembranosus (a.), semitendinosus (b.), biceps femoris short head (c.), biceps femoris long head (d.), (e.) lateral gastrocnemius and (f.) medial gastrocnemius positive values indicate that the round chainring values exceed those in the oval condition (SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines).

Tables

Table 1: Patellofemoral force per 20 km (Mean, SD & 95% CI) as a function of chainring geometry.

	Round			**Oval**				
	Mean	SD	*95% CI*	**Mean**	SD	*95% CI*	P-value	$p\eta^2$
Patellofemoral force per 20 km (N/kg·s)	38576.40	10796.83	31716.42-45436.38	35637.00	8306.64	30359.21-40914.78	0.52	0.04