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Original Research

Femoral maltorsion influences both patellofemoral and tibiofemoral contact pressures. A biomechanical evaluation

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A R T I C L E I N F O

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Femoral maltorsion
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Biomechanics
Patellofemoral pressures
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Malalignment
Anteversion

A B S T R A C T

Introduction/objectives: To investigate the effect of femoral maltorsion on both the patellofemoral and the tibiofemoral contact pressures.
Methods: Experimental biomechanical study on 10 embalmed human cadaveric knees (mean age 40.2 years ± 9.5). Krackow sutures were placed in the vastus medialis, vastus lateralis, rectus femoris, vastus intermedius, and hamstrings. A custom-made test apparatus capable of independently loading the quadriceps and hamstring muscle groups and providing ground reaction forces during a simulated squat maneuver and under direct axial compression loading was used. A ground reaction force of 1000 N was applied to the distal tibia and scaled loads of 218N and 80N were applied to the quadriceps and hamstrings, respectively, at 0°, 30°, 60°, and 90° of flexion. Pressure measurements were performed by intra-articularly placed sensors (77.3 × 77.3 mm, SFM6000CXR2 Sensor) and using Snowforce 3 interpretation software. After testing of the native knee, supracondylar femoral osteotomy was performed. Pressure measurements were again made in each knee compartment, at each of the degrees of rotation evaluated.
Results: Medial aspect of the patella showed an increase of contact pressure with external femoral rotation from 10° to 30° compared with 0°. The strongest effect was measured at 30° of knee flexion ($p = 0.005$) with 30° of external rotation ($p = 0.004$) with a value of 2.140 ± 0.1832 Mpa. With internal femoral rotation there is an increment of contact pressure in the lateral aspect of the patella, with the strongest effect at 30° of flexion ($p = 0.0059$) with 30° of internal rotation ($p = 0.0002$) with a value of 1.352 ± 0.08166 Mpa. The medial tibiofemoral contact pressure showed an increment from 10° to 30° of external rotation compared with the native state. The highest pressure was shown at 90° of knee flexion ($p = 0.0006$) and 30° of external rotation ($p = 0.004$) with a value of 1.636 ± 0.01878 Mpa. The lateral tibiofemoral contact pressure increased compared with the control group more with internal than with external rotation. The highest pressure was shown at 90° of flexion ($p < 0.0001$) and 30° of internal rotation ($p < 0.0001$) with a value of 1.432 ± 0.004051 Mpa.
Conclusions: Femoral malrotation influences patellofemoral and tibiofemoral contact pressure. Femoral external rotation may result in worse knee biomechanics than internal rotation.
Level of evidence: III.

Abbreviations: FM, Femoral anteversion; KOA, Knee osteoarthritis; VMO, Vastus medialis obliquus; VL, Vastus lateralis; RF, Rectus femoris; RFVI, Vastus intermedius; PVC, Polyvinyl chloride; LPF, Lateral patellofemoral; MPF, Medial patellofemoral; LTF, Lateral tibiofemoral; MTF, Medial tibiofemoral; EFR, External femoral rotation; IFR, Internal femoral rotation; MPFL, Medial patellofemoral ligament.

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What are the new findings?

- Transverse plane alignment influences both patellofemoral and tibiofemoral contact pressure.
- Femoral external rotation may result in worse knee biomechanics than internal rotation.
- Femoral maltorsion might be a risk factor for the development of knee osteoarthritis.
- Derotational osteotomy might prevent the development of early knee osteoarthritis by reducing intra-articular pressure forces.

INTRODUCTION

Femoral maltorsion (FM) is a recognized although overlooked cause of patellofemoral pain and instability as well as hip disorders such as femoroacetabular impingement in the young population [1–5]. FM can have not only an important role in the genesis of pain and instability but also in disability, pathological gait, and in the development of knee osteoarthritis (KOA), which is one of the leading causes of disability in adults [6–10].

Although KOA is predominantly the result of an unfavorable biomechanical environment at the joint, FM remains an under-recognized factor in patellofemoral cartilage degeneration in young patients [8, 10]. A deformity in the coronal plane will cause excessive stress on the knee joint that will cause damage to the articular cartilage and ultimately KOA [9,11,12]. However, the connection between femoral FM and KOA is less studied [3,6,7,13,14].

The purpose of this study was to investigate the effect of FM on the patellofemoral and the tibiofemoral contact pressures in a biomechanical model using human cadaveric knees and a custom-made test apparatus capable of applying axial compression forces to simulate weight-bearing. We hypothesized that there would be a correlation between FM and increased intra-articular knee pressures in tibiofemoral and patellofemoral compartments.

METHODS

Study design and sample characteristics

An experimental biomechanical study was carried out on 10 non-paired, nonarthritic embalmed (Carbowax technique), human cadaveric knees from two anatomy departments of our institution, each one previously amputated 25 cm from the joint line proximally and distally. The mean age of the specimens was 40.2 years ± 9.5. The knees belonged to 8 male and 2 female cadavers (6 knees were right knees and 4 left knees).

Ethical considerations

The current study was approved by the Research Ethics Committee and the Research Committee of the School of Medicine of the University Hospital of the Universidad Autonoma de Nuevo Leon, with registration number: FI23-00001. All procedures performed in this study were done so in accordance with the Declaration of Helsinki and its later amendments or comparable ethical standards [15].

A board-certified knee surgeon, with experience in the field of orthopedic sports medicine (RMA), performed all preparation and surgical steps.

Prior to their inclusion in the study, the skin and subcutaneous cellular tissue were removed from each specimen, up to the exposure of the articular capsule. Particular care was taken to preserve all capsular, muscle groups, tendons, ligament structures as well as the extensor apparatus and the insertion of the hamstrings, with the intention of trying to mimic, as far as possible, the native biomechanics of the knee.

Knees were excluded if they presented by macroscopic inspection visible skin scars and/or by direct digital palpation through a suprapatellar arthrotomy data of osteoarthrosis, signs of trauma, fractures or previous surgeries, as well as obvious malalignment, and/or maltracking. X-rays were taken in anteroposterior and mediolateral projection to exclude knees with orthopedic implants, severe osteoarthritis, decreased bone density, and/or bone abnormalities or data of previous surgeries or fractures. All specimens underwent a computed tomography scan to establish patellar height, calculate Tibial Tuberosity to Trochlear Groove (TT-TG) distance, look for signs of severe osteoarthritis, or previous fractures.

Specimen preparation

Krackow sutures were placed in the vastus medialis obliquus (VMO), vastus lateralis (VL), rectus femoris (RF), and vastus intermedius (RFVI) muscle-tendon units to facilitate 3-vector quadriceps dynamic muscle loading during mechanical testing. Additionally, another vector was placed at the distal insertion of the hamstrings for independent vector loading. High-strength No. 2 sutures (Orthocore, DepuySynthes) were used, and these muscle groups were pretensioned at 20 lb (9.07 kg) for 20 min.

The femur and tibia medullary canal were tapped and the proximal end of the femur and the distal ends of the tibia and fibula were potted in a cylindrical mold using fiberglass resin (Resin pp250, Monterrey, Mexico) and polyvinyl chloride piping (10 cm length and 10 cm diameter) to obtain a smooth and stable surface that would allow the rigid support of the knees in the biomechanical device

Biomechanical setup

A custom-made test apparatus capable of independently loading the quadriceps and hamstring muscle groups and providing ground reaction forces during a simulated squat maneuver and under direct axial compression loading was used (Fig. 1). One 4.5 mm cortical screw was placed through the polyvinyl chloride tubing, fiberglass resin and shaft of each long bone to provide further stability to withstand loads during testing procedures.

A ground reaction force of 1000 N was applied to the distal tibia to simulate the pressure exerted by an average-weight adult male in a squat maneuver [16] and scaled loads of 218 N and 80 N were applied to the quadriceps and hamstrings to simulate weight-bearing resisted knee extension and flexion, respectively [17], at 0°, 30°, 60°, and 90° of flexion. The quadriceps components were pulled in physiological directions relative to the axis of the femur: 20° lateral for VL, 5° anterior for RF + VI, and 35° medial for VM. The loads of the quadriceps tendon were applied independently using a tension generated by a stepper motor redirected over a pulley system. The forces were split up between the structures as follows: 76 N for VL, 87 N for RFVI, and 55 N for VMO [18].

The knee and sensors were preconditioned with 5 cycles of loaded flexion–extension prior to testing.

Biomechanical testing

Each knee underwent two trials for each of the following conditions: Native knee; 10° internal derotational osteotomy; 20° internal

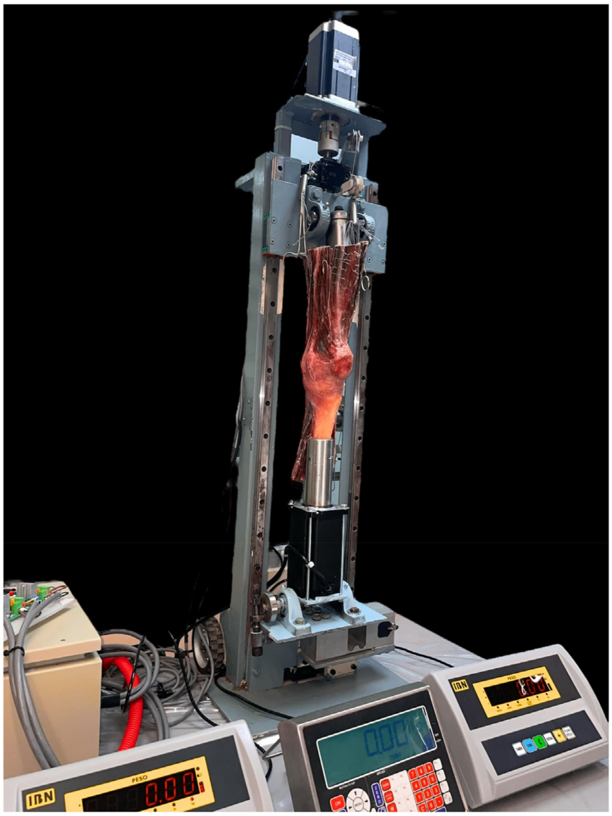

Fig. 1. Photograph of the setup of the human cadaveric knee assembled in the biomechanical device used in the present study.

derotational osteotomy; 30° internal derotational osteotomy; 10° external derotational osteotomy; 20° external derotational osteotomy; 30° external derotational osteotomy.

The tibiofemoral and patellofemoral joints were divided into 2 compartments each, thus obtaining the following: lateral patellofemoral, medial patellofemoral, lateral tibiofemoral, and medial tibiofemoral.

The control group was considered as the native knee with application of axial load and simulated muscle loads without any degree of rotation.

Pressure measurements in each compartment were performed by intra-articularly placed sensors (77.3 × 77.3 mm SFM6000CXR2 Sensor, Kitronyx, Inc®, Seoul, Korea) with the Baikal II controller (Kitronyx Inc®, Seoul, Korea) and using Snowforce 3 interpretation software (Kitronyx Inc®, Seoul, Korea). Holding sutures were attached to the medial and lateral end of the sensors to avoid movement during the test.

It was confirmed by inspection and palpation that sensors would cover the entire patellofemoral joint (Fig. 2), in the case of the femorotibial joint, each compartment was evaluated separately (Fig. 3).

After testing of the first state (native knee), osteotomy techniques were performed. A longitudinal incision was made over the iliotibial band, the vastus lateralis was separated by blunt dissection from the fascia lata, and then elevated anteriorly for the placement of the plate on the lateral side. The septum inter-muscular is detached from the femur at the level of the osteotomy with a curved raspatory. A direct medial approach was performed at the level of the osteotomy site for placement of the second plate on the medial side.

The site of supracondylar femoral osteotomy was defined under fluoroscopic guide, looking for a cut perpendicular to the anatomical axis of the femoral diaphysis. Subsequently, two threaded K-wires were inserted completely parallel to each other, one placed proximal and the other distal to the osteotomy site. Osteotomy was performed using an oscillating saw of 0.6 mm diameter. The objective was to create static rotational deformities of the femur.

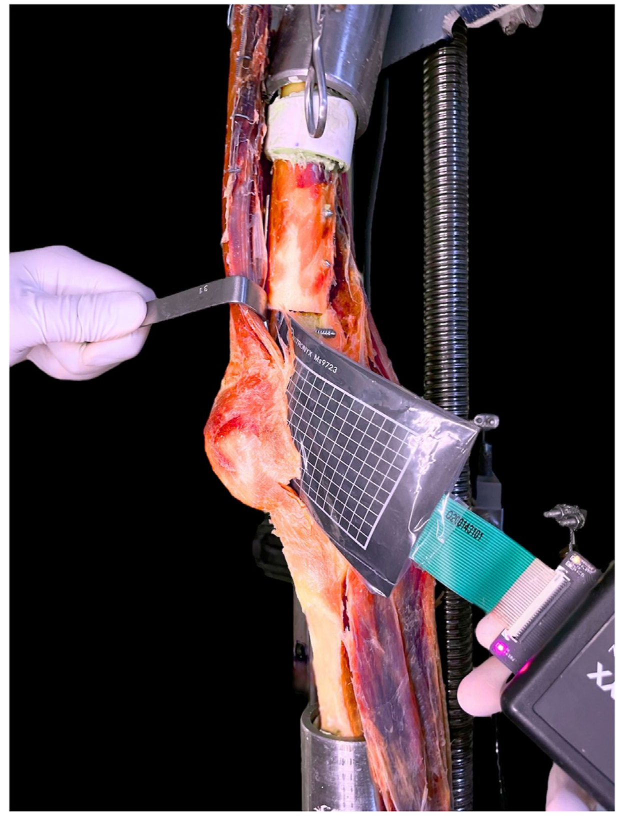

Fig. 2. Method to obtain the femoropatellar pressure.

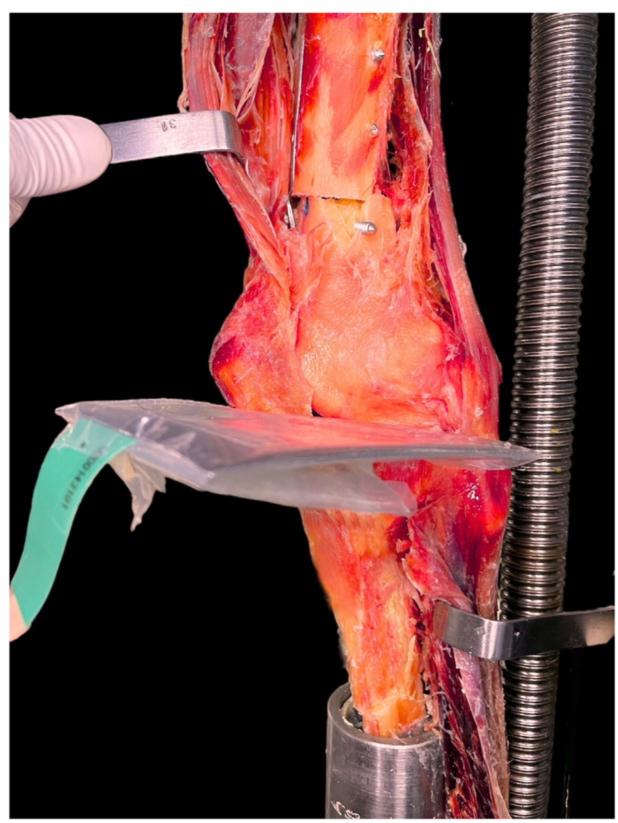

Fig. 3. Method to obtain the medial femorotibial pressure.

The distal fragment was rotated until the desired angle of rotation (10°, 20°, and 30° degrees of internal rotation of the distal fragment as well as 10°, 20°, and 30° degrees of external rotation of the distal fragment) was reached with the distal K-wire. Angulations were confirmed using a digital goniometer.

The osteotomy was fixed using a non-locking plating system, employing a lateral and medial plate to obtain absolute stability of the osteotomy site (L-buttress plate for 4.5 mm screws, Model 624, GPC Medical USA, Inc. and T-buttress plate for 4.5 mm screws, Model 645 GPC Medical USA, Inc.). Three fixation screws were placed for the distal fragment and 4 screws for the proximal fragment (Fig. 4).

Afterward, pressure measurements were again made in each knee compartment, at each of the degrees of rotation evaluated, using the same screw holes proximal to the osteotomy site.

Statistical analysis

Rigorous testing was performed on a data set with a sample size of 10 for each group. Depending on the specific nature of our data, a choice was made between repeated measures ANOVA and Friedman's tests. This choice was based on a preliminary evaluation of the distribution of the data, using the Shapiro–Wilk test to determine the most relevant statistical technique. In addition, Levene's test was applied to verify the homogeneity of variances, thus guaranteeing the validity and reliability of our statistical results. Effect sizes were estimated using partial $\eta 2$.

In those cases where the data followed a normal distribution, the results were presents as the mean ± standard deviation. On the other hand, for data that did not fit a normal distribution, the median and interquartile range were preferred.

A value of $p < 0.05$ was established as the criterion for statistically significant difference, allowing detailed comparisons between the groups studied. All this analysis was performed using GraphPad Prism version 9.

All measurements were taken twice and independently, to perform intra- and interobserver correlation tests. To assess intraobserver reliability, observers were asked to repeat measurements after the first measurement. The percentage error between the two sets of measurements was used to estimate intraobserver agreement using the intraclass coefficient correlation test. To evaluate interobserver error, the intraclass coefficient correlation test was performed to compare the two sets of measurements. For quantitative observations, intrarater and interrater reliability were calculated using the intraclass correlation coefficients (ICC) between all pairs. An ICC ≥0.8 was considered a good agreement.

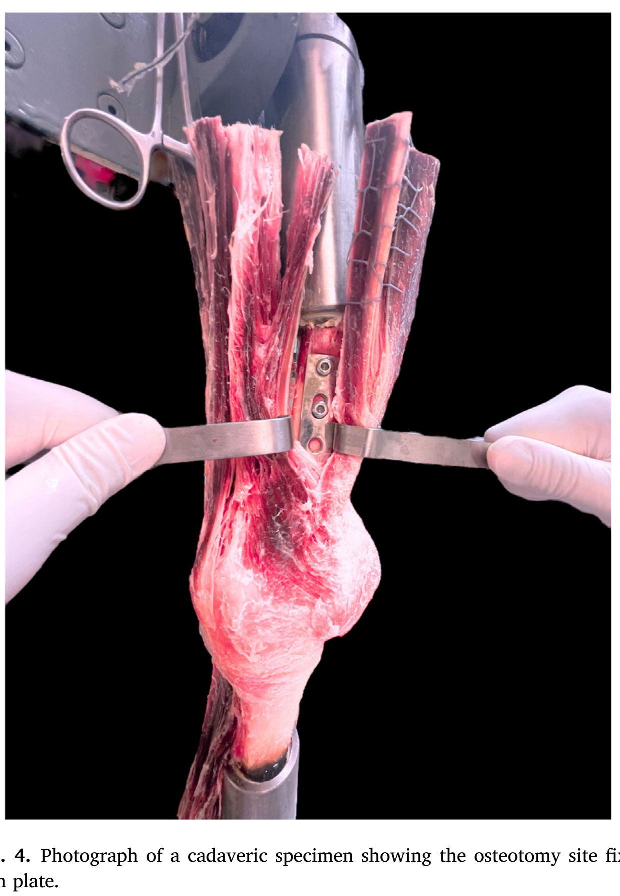

Fig. 4. Photograph of a cadaveric specimen showing the osteotomy site fixation plate.

RESULTS

Patellofemoral joint

The medial aspect of the patella showed an increase in contact pressure with progressive simulated external femoral rotation (EFR) from 10° to 30° compared with the pressure at 0°. The strongest effect was measured at 30° of knee flexion ($p = 0.005$, $\eta = 0.94$) with 30° of EFR ($p = 0.004$, $\eta = 0.94$) with a value of 2.140 ± 0.1832 Mpa and a statistically significant difference compared with the control group ($p < 0.0001$, $\eta = 0.98$), EFR of 10° ($p = 0.0028$, $\eta = 0.82$), internal femoral rotation (IFR) of 10° ($p = 0.001$, $\eta = 0.99$), IFR of 20° ($p = 0.0083$, $\eta = 0.93$), and IFR of 30° ($p = 0.015$, $\eta = 0.98$). These results show an increase in pressure on the medial facet of the patella from 20° of EFR compared with any degree of IFR (Table 1).

With IFR there is an increment of contact pressure in the lateral aspect of the patella compared with EFR with the strongest effect at 30° of knee flexion ($p = 0.0059$) with 30° of IFR ($p = 0.0002$) with a value of 1.352 ± 0.08166 Mpa and a statistically significant difference compared with the control group ($p = 0.0325$, $\eta = 0.86$) and EFR of 20° ($p = 0.0082$, $\eta = 0.90$) (Table 2).

Tibiofemoral joint

The medial tibiofemoral contact pressure showed a significant increment from 10° to 30° of EFR compared with the native state. However, medial tibiofemoral compartment contact pressure was smaller with IFR than with EFR but higher than in the control group. Tibiofemoral pressure was significantly lowest in near extension and increased gradually with knee flexion from 0 to 90° in both IFR and EFR. The highest pressure was shown at 90° of knee flexion ($p = 0.0006$, $\eta = 0.92$) and 30° of EFR ($p = 0.004$, $\eta = 0.93$) with a value of 1.636 ± 0.01878 Mpa and a statistically significant difference compared with the control group ($p < 0.0001$, $\eta = 0.99$), IFR of 20° ($p = 0.0045$, $\eta = 0.92$), and IFR of 30° ($p < 0.0001$, $\eta = 0.98$) (Table 3).

The lateral tibiofemoral contact pressure increased compared with the control group more with IFR than with EFR, increasing progressively with knee flexion from 0° to 90°. The highest pressure was shown at 90° of flexion ($p < 0.0001$) and 30° of IFR ($p < 0.0001$) with a value of 1.432 ± 0.004051 Mpa and a statistically significant difference compared with the control group ($p < 0.0001$), EFR of 10° ($p < 0.0001$, $\eta = 0.98$), EFR of 20° ($p < 0.0001$, $\eta = 0.96$)), EFR of 30° ($p < 0.0001$, $\eta = 0.96$), IFR of 10° ($p < 0.0001$, $\eta = 0.94$) and IFR of 20° ($p < 0.0001$, $\eta = 0.96$) (Table 4).

The results of the intraobserver and interobserver analyses resulted in good agreement (ICC >0.8) for all sets.

DISCUSSION

The most important finding of the present study was that both patellofemoral and tibiofemoral load distribution are sensitive to FM, especially in EFR, suggesting it is more damaging to the knee joint than IFR. Given that KOA onset has been attributed to overloading of the knee induced by lower limb malalignment [6–9,15], the present study might contribute to a better understanding of KOA pathogenesis.

Coronal plane deformities have been associated with excessive knee joint stress and progression to articular cartilage damage and, ultimately,

Table 1
Intra-articular pressures of the medial patellofemoral joint (MPF) joint by degrees of rotation and flexion.

Medial patellofemoral pressure (MPa)	Flexion 0°	Flexion 30°	Flexion 60°	Flexion 90°	F	*p*
Control	0.8296 ± 0.2036	1.403 ± 0.2279	1.233 ± 0.1945	1.140 ± 0.1174	21.65	<0.0001
EFR 10°	1.258 ± 0.06754	1.730 ± 0.1471	1.359 ± 0.1693	1.226 ± 0.1066	41.39	<0.0001
EFR 20°	1.329 ± 0.07345	1.952 ± 0.1333	1.542 ± 0.1331	1.251 (0.242)	21.60	<0.0001
EFR 30°	1.437 ± 0.1055	2.140 ± 0.1832	1.713 (0.181)	1.374 ± 0.1124	18.45	0.0004
IFR 10°	1.186 ± 0.04490	1.655 ± 0.08713	1.396 ± 0.1001	1.183 ± 0.04037	94.42	<0.0001
IFR 20°	1.238 ± 0.04783	1.773 ± 0.09088	1.447 ± 0.1469	1.242 ± 0.09576	57.98	<0.0001
IFR 30°	1.335 ± 0.1297	1.878 ± 0.1230	1.587 ± 0.09653	1.295 ± 0.1107	50.25	<0.0001
F	26.82	30.76	31.02	23.57		
p	<0.0001	<0.0001	<0.0001	0.0006		

EFR = external femoral rotation; IFR = internal femoral rotation.

Table 2
Intra-articular pressures of the lateral patellofemoral joint (LPF) joint by degrees of rotation and flexion.

Lateral patellofemoral pressure (MPa)	Flexion 0°	Flexion 30°	Flexion 60°	Flexion 90°	F	*p*
Control	1.049 ± 0.08994	1.278 ± 0.08453	1.178 ± 0.07892	1.102 ± 0.07857	25.59	<0.0001
EFR 10°	1.096 ± 0.05519	1.268 ± 0.1013	1.186 ± 0.07241	1.124 ± 0.05994	18.05	<0.0001
EFR 20°	1.129 ± 0.06866	1.301 ± 0.07977	1.203 ± 0.06883	1.149 (0.121)	16.65	0.0008
EFR 30°	1.135 ± 0.05609	1.328 ± 0.07630	1.231 ± 0.06460	1.162 (0.114)	15.60	0.0014
IFR 10°	1.124 ± 0.05903	1.299 ± 0.07842	1.196 ± 0.06807	1.136 ± 0.05126	18.60	0.0003
IFR 20°	1.145 ± 0.06854	1.327 ± 0.06760	1.232 ± 0.07526	1.144 ± 0.01934	28.36	<0.0001
IFR 30°	1.158 ± 0.06544	1.352 ± 0.08166	1.235 ± 0.08159	1.153 (0.054)	19.95	0.0002
F	12.26	6.743	5.506	32.36		
p	0.0029	0.0059	0.0110	<0.0001		

EFR = external femoral rotation; IFR = internal femoral rotation.

Table 3
Intra-articular pressures of the MTF (Medial tibiofemoral joint) joint by degrees of rotation and flexion.

Medial femorotibial pressure (MPa)	Flexion 0°	Flexion 30°	Flexion 60°	Flexion 90°	F	*p*
Control	1.033 ± 0.01880	1.042 ± 0.01924 x	1.048 ± 0.01912	1.056 ± 0.02094	196.7	<0.0001
EFR 10°	1.346 ± 0.03246	1.356 ± 0.03468	1.363 ± 0.03529	1.373 ± 0.03640	88.04	<0.0001
EFR 20°	1.540 ± 0.03059	1.549 ± 0.02902	1.560 ± 0.03059	1.570 ± 0.03040	107.8	<0.0001
EFR 30°	1.608 ± 0.01816	1.618 ± 0.01733	1.628 ± 0.01646	1.636 ± 0.01878	148.8	<0.0001
IFR 10°	1.163 ± 0.01721	1.173 ± 0.01801	1.186 ± 0.01914	1.197 ± 0.01638	191.1	<0.0001
IFR 20°	1.087 ± 0.04478	1.117 (0.034)	1.144 ± 0.01884	1.157 ± 0.02302	24.00	<0.0001
IFR 30°	1.047 ± 0.03077	1.081 ± 0.02480	1.107 ± 0.02230	1.110 (0.031)	24.00	<0.0001
F	600.8	48.00	48.00	48.00		
p	<0.0001	<0.0001	<0.0001	<0.0001		

EFR = external femoral rotation; IFR = internal femoral rotation.

Table 4
Intra-articular pressures of the lateral tibiofemoral joint (LTF) joint by degrees of rotation and flexion.

Lateral femorotibial pressure (MPa)	Flexion 0°	Flexion 30°	Flexion 60°	Flexion 90°	F	*p*
Control	0.9789 ± 0.03813	1.003 ± 0.01754	1.013 ± 0.01492 #	1.020 ± 0.01283	9.029	0.0184
EFR 10°	1.112 ± 0.01156	1.124 ± 0.009130	1.134 ± 0.008643	1.144 ± 0.008043	88.24	<0.0001
EFR 20°	1.068 (0.038)	1.104 ± 0.006896	1.116 ± 0.006949	1.128 ± 0.008036	24.00	<0.0001
EFR 30°	0.9925 ± 0.03755	1.035 ± 0.01918	1.058 ± 0.02296	1.079 ± 0.02061	48.40	<0.0001
IFR 10°	1.112 ± 0.01544	1.124 ± 0.009881	1.135 ± 0.008709	1.147 ± 0.008242	63.94	<0.0001
IFR 20°	1.303 ± 0.01242	1.316 ± 0.01165	1.328 ± 0.01200	1.341 ± 0.01077	103.4	<0.0001
IFR 30°	1.398 ± 0.01004	1.411 ± 0.008141	1.423 ± 0.007344	1.432 ± 0.004051	165.4	<0.0001
F	46.18	1300	1231	1324		
p	<0.0001	<0.0001	<0.0001	<0.0001		

EFR = external femoral rotation; IFR = internal femoral rotation.

KOA [21]. However, the link between FM and KOA is less known [8,19,20]. Previous studies have demonstrated that isolated lateral patellofemoral KOA is strongly associated with femoral anteversion in valgus knees [19]. Goutallier et al. showed that the presence of femoral anteversion results in the long-term success of valgus tibial osteotomy [8]. They reported the loss of valgus deviation in patients with decreased femoral anteversion after tibial valgus osteotomy for medial compartment KOA and the increase of valgus deviation with increased femoral anteversion, both of which led to worse functional outcomes when compared with patients with stable coronal alignment [8]. On the other hand, other studies have found no significant correlation between femoral anteversion and cartilage degeneration of the lateral patellofemoral joint [20].

Prior studies addressed patellofemoral contact pressures in FM [21–24]. Patellofemoral cartilage stresses were more sensitive to external rotation of the femur, compared with internal rotation [21]. Our study findings are in accordance with those reported by Lee et al. [23], showing that increased femoral torsion results in higher patellofemoral contact pressure on the patella facet opposite to the torsion side. However, our contribution over previous studies is that it replicated the physiological weight-bearing condition by applying ground reaction forces (axial compression forces) in cadaveric specimens with a lower mean age of decease. The present study showed that the greatest overload of the medial aspect of the patella occurs at 30° of knee flexion when there is an EFR of 30°. With an internal rotation of 30°, the greatest overload of the lateral aspect of the patella also occurs at 30° of knee flexion. From 30° onward, patellofemoral contact pressures decreased compared with the values at 30°. This is logical because the medial patellofemoral ligament relaxes from 30° of knee flexion and therefore the patellofemoral contact pressure decreases from 30° [25]. In addition, with internal femoral rotation the lateral patellofemoral compression force increases [26]. However, with external femoral rotation the medial patellofemoral compression force increases [26]. That is to say, the greatest patellofemoral pressures occur with a knee flexion typical of daily life activities.

There are few studies that have attempted to quantify the effect of FM on tibiofemoral contact pressures [13,22,27]. Using a computational model, Papaioannou et al. [27] showed increased tibiofemoral contact pressure in the medial tibiofemoral compartment in cases of EFR. Similarly, in cadaveric studies [13,22] it has been shown that EFR increases overload of the medial tibiofemoral compartment, and IFR increases contact pressures at the lateral tibiofemoral compartment. Bretin et al. [13] have shown that FM has a significant effect on mechanical axis alignment. IFR provokes valgus deviation of the mechanical axis while EFR caused varus deviation. These findings are probably due to a compensatory mechanism for the movement of the hip in the opposite direction. Therefore, external rotation deformity and decreased FAV will be compensated by internal rotation of the limb, resulting in a deviation of the medial mechanical axis that will increase the load on the medial compartment. Internal rotation deformity and increased FAV will be compensated by external rotation with a lateral deviation of the mechanical axis and a decrease of the pressures in the medial compartment [22].

The present study showed that the medial tibiofemoral compartment contact pressure was lower with IFR than with EFR but higher than in the control group. Medial tibiofemoral compartment contact pressure increased from 10° to 30° of EFR compared with the native state. The highest overloads of the tibiofemoral joint occur at 90° of knee flexion and 30° of EFR. Therefore, the tibiofemoral joint is less damaged in activities of daily living than the patellofemoral joint. From our results we can conclude that in patients with severe FM, the patellofemoral joint is damaged more in activities of daily living than the tibiofemoral joint.

Our findings suggest that femoral internal rotation shows better knee contact pressures than internal rotation. FM might be a risk factor for the development of KOA, and therefore, derotational osteotomy might play a role in preventing early KOA by reducing intra-articular pressure forces. Moreover, when repairing or replacing damaged cartilage, FM assessment and correction may impact their outcome. If we do not correct FM, the surgery on the cartilage will fail due to overload. In these cases, cartilage repair techniques should be performed combined with derotational osteotomies to off-load the repair site supporting the healing process.

Limitations

The limitations of this study include those inherent to *in vitro* testing. But our study has certain strengths compared with other studies carried out on cadavers, such as the fact that we used knees from young adults. Furthermore, our experimental model allows us to apply axial compression forces that simulate weight-bearing. Moreover, can apply loads to the quadriceps and hamstrings.

This study does not consider the dynamic behavior occurring in patients with torsional deformities. Only aspects concerning the statics of both the patellofemoral and tibiofemoral joints were investigated. In general, patients with lower limb torsional deformities tend to present kinematic compensatory mechanisms.

A stratified analysis based on different kinematic gait patterns might help identifying specific subgroups of patients who present a higher risk of joint overloading because of both altered morphology and kinematics.

CONCLUSION

In conclusion, the biomechanical findings indicate that the transverse plane alignment influences both patellofemoral and tibiofemoral contact pressure. Moreover, it has been shown that there is more tolerance in the knee to IFR (anteversion) than to EFR (retroversion).

Author contributions

All authors contributed to the study's conception and design. Material preparation and data collection were performed by RMA, CC, CCC, REEO, and SGL. The methodology was designed by GCJ, JGL, CFOG, JCM, and VSA. The formal analysis was conducted by VSA, RMA, CC, and CCC. Preparation of the draft was done out by GCJ, JGL, CFOG, REEO, JCM, and VSA. The final write-up, review, and editing was done by VSA, RMA, CC, CCC, SGL, REEO, and GCJ. The work was supervised by JGL, CFOG, JCM, VSA, and RMA. All the authors commented on the previous versions of the manuscript and then read and approved the final manuscript.

Ethics approval

The present protocol was approved by the Institutional Review Board and Research Ethics Committee of the University Hospital of the Universidad Autonoma de Nuevo Leon (U.A.N.L.) with registration number #FI23-00001.

Article summary

Femoral maltorsion may be a risk factor for the development of knee osteoarthritis (KOA). Therefore, derotational osteotomy might delay the development of early KOA by reducing intra-articular pressure forces.

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Declarations of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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