Isolated patellofemoral arthroplasty reproduces natural patellofemoral joint kinematics when the patella is resurfaced

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Abstract

Purpose

The objectives of this in vitro project were to compare the dynamic three-dimensional patellofemoral kinematics, contact forces, contact areas and contact pressures of a contemporary patellofemoral prosthetic implant with those of the native knee and to measure the influence of patellar resurfacing and patellar thickness. The hypothesis was that these designs are capable to reproduce the natural kinematics but result in higher contact pressures.

Methods

Six fresh-frozen specimens were tested on a custom-made mechanical knee rig before and after prosthetic trochlear resurfacing, without and with patellar resurfacing in three different patellar thicknesses. Full three-dimensional kinematics were analysed during three different motor tasks, using infrared motion capture cameras and retroflective markers. Patellar contact characteristics were registered using a pressure measuring device.

Results

The patellofemoral kinematic behaviour of the patellofemoral arthroplasty was similar to that of the normal knee when the patella was resurfaced, showing only significant ($p < 0.0001$) changes in patellar flexion. Without patellar resurfacing, significant more patellar flexion, lateral tilt and lateral rotation was noticed. Compared to the normal knee, contact pressures were significantly elevated after isolated trochlear resurfacing. However, the values were more than doubled after patellar resurfacing. Changes in patellar thickness only influenced the antero-posterior patellar position. There was no other influence on the kinematics, and only a limited influence on the contact pressures in the low flexion angles.

Conclusion

The investigated design reproduced the nor- mal patellofemoral kinematics acceptable well when the patella was resurfaced. From a kinematic point of view, patellar resurfacing may be advisable. However, the substantially elevated patellar contact pressures remain a point of concern in the decision whether or not to resurface the patella. This study therefore not only adds a new point in the discussion whether or not to resurface the patella, but also supports the claimed advantage that a patellofemoral arthroplasty is capable to reproduce the natural knee kinematics.

Keywords Patellofemoral arthroplasty, Kinematics, Contact pressure, Patellar resurfacing

Introduction

Although patellofemoral arthroplasty (PFA) has been regularly used for more than three decades, its place in the treatment of isolated patellofemoral arthritis remains at present still controversial [24, 48, 49]. This is due to the inconsistent clinical results published in literature, mainly related to patellar tracking and catching problems in the first generation of 'inlay' patellofemoral implants [1, 2, 4, 9, 11–13, 18–21, 26, 27, 29–32, 36, 37, 41, 42]. The second generation of 'onlay' patellofemoral prostheses, based on the anterior and trochlear femoral cuts of the total knee arthroplasty (TKA), was expected to reduce the incidence of patellar maltracking and instability problems [30]. Furthermore, introduction of better instrumentation made it easier to optimize the rotation of the trochlear component. This should result in a better reproduction of the natural patellofemoral kinematics and as such in bet- ter functional outcomes and survival rates. The published medium term follow-up results are at least promising [16, 20, 28, 39]. The progression of tibiofemoral osteoarthritis, reported with ranges from 0 to 22 %, is currently the most important known reason for late failure of PFA [18, 19, 38]. Nevertheless, the most frequent 'indication' for early revision remains unexplained pain [5]. Although the published high revision rate might be partially due to inadequate surgical selection and the ease of the revision procedure, it has encouraged much surgeons to stick to TKA for the treatment of isolated patellofemoral cartilage degeneration [35].

The design improvements of the second generation of patellofemoral prostheses already significantly reduced complication rates, and subgroup analysis suggested a relation between revision rate and implant design. Therefore, PFA has gained importance and may be the preferential treatment for isolated patellofemoral destruction in the middle-aged population [14]. Although there is a clear correlation between implant design and complication rate, there is still no general consensus on the ideal trochlear 'onlay' design [23, 25].

The claimed advantages of PFA, such as a less invasive procedure, less bone and tissue destruction, less blood loss, a shorter operation time and shorter rehabilitation, are without any doubt legitimated [49]. However, there is a lack of studies supporting the mentioned potential advantage of 'more normal' knee kinematics. Most of the published studies concerning PFA are pro-or retrospective clinical studies on outcomes and complications. There are only a few biomechanical studies investigating the patellofemoral kinematics of patellofemoral replacements (PFR), mainly in the sagittal plane [17, 34].

The aim of this project was therefore to measure and compare the full 3D patellofemoral kinematics, patellar contact force, contact area and contact pressure before and after placing an isolated patellofemoral prosthesis with a contemporary, 'modern', design, and to study the influence of patellar resurfacing and patellar thickness. The null hypothesis was that PFA reproduces the natural patellofemoral kinematics acceptable well, but induces significant changes in contact mechanism.

Materials and methods

Six unmatched lower leg specimens from Caucasian subjects (one female, five male, median age 80.5 years, range 78–91 years) were disarticulated at the hip and frozen at -20 °C. The tested specimens had no signs of previous surgery, traumatic bone or ligament lesions. None of the specimens had major arthritic damage at the level of the knee, nor abnormal anatomical features or malalignment, which might be responsible for abnormal patellofemoral tracking. Three frames, with on each frame four reflective marker spheres (NDI, Waterloo, Ontario, Canada), were rigidly fixed to the frozen specimens, one at the level of the femoral diaphysis, one in the proximal tibia at a minimum of 10 cm distal to the joint and one onto the anterior aspect of the patella. A volumetric CT scan was performed (Siemens Somatom Definition Flash, Siemens AG, Erlangen, Germany). The images were obtained at 120 kV and 200 mAs + Care Dose, with a pitch of 0.8 mm per revolution, 1 mm slice thickness and slice increment, 1.0 s rotation time and B70–B30 reconstruction kernel.

Each specimen was thawed during 36 h before the start of the experiments. The three-dimensional motion capture system, composed of six infrared cameras (Vicon Motion Systems™, LA, California), was calibrated and positioned in such way that the optical reference markers were visible at all time to register the kinematic data. The specimen was prepared and embedded in metal containers with PMMA, taking care of the physiologic alignment of femur and tibia, as described in previous publications [45, 47]. The quadriceps tendon was prepared and rigidly fixed in a clamp at a distance of 6 cm above the proximal patellar pole. Lateral and medial hamstrings were isolated and sutured. In a next step, the knee joint was opened by using a classic medial arthrotomy. A K-scan 4000/9000 psi sensor (Tekscan™, South Boston, Massachusetts) was calibrated according to the instructions of the manufacturer. Previous studies revealed the accuracy of this pressure measurement device to be within 10 % for average contact pressure and area, comparable to the values for pressure-sensitive film.

However, it is thinner and allows for dynamic measurements [15, 50]. The sensor was applicated to cover the entire patellar articular surface and fixed with multiple, small, individual sutures in a way not to interfere with the articular interface (Fig. 1).

The prepared natural knee specimen was then mounted on a customized dynamic knee rig, simulating normal knee motions and loads and leaving six degrees of freedom. Each specimen performed a passive motion, an unresisted open chain motion, a resisted open chain motion and a squat movement, with loading of quadriceps and 50 N loading of medial as well as lateral hamstrings. Passive motion was induced manually with three cycles from extension to maximal flexion, with the femoral container mounted on the rig. During the open chain movement, the leg was extended at a fixed speed from about 110° of flexion to about -20° of extension by pulling on the quadriceps tendon with variable load while the ankle was hanging free. Near full extension, the quadriceps load reached values between 60 and 100 N, depending on the weight of the lower leg. The resisted open chain was performed in the same way, with a 3 kg weight fixed to the tibial container at the distal end. Close to full extension, quadriceps load reached values around 300 N. For the squat, the tibial container was also mounted in the rig. The hip height was programmed as a function of time, thus controlling flexion of the knee between 20° and 120° of flexion, while applying a variable quadriceps force to induce a vertical constant ankle load of 130 N (Fig. 2). During this closed chain knee motion, the quadriceps load was highest at 110° of flexion with values around 1,400 N. The six infrared cameras dynamically registered the motion of the marker frames on femur, tibia and patella as a function of time. Meanwhile, calibrated load cells recorded quadriceps and ankle loads and moments, and patellofemoral contact area and pressure.

In a following step, a trochlear resurfacing was performed, using the trochlear component of the Journey PFJ™ (Smith–Nephew™, Memphis, TN, USA), without resurfacing the patella. The rotational alignment and varus– valgus positioning of the implant were determined, based on the surgical instrumentation sets provided by the company, combining the tibial and femoral referencing technique. The goal was to mimic the standard in vivo operative technique. The author has a large clinical experience with this type of prosthesis, avoiding a learning curve. In a first step, the rotation and valgus–varus position of the trochlear component were determined by using the tibiofemoral alignment guide. This guide was positioned referencing a line perpendicular to the longitudinal mechanical tibial axis. The position of the femoral (anterior) cutting guide was then checked with visual reference to the surgical epicondylar and

AP femoral axes. Subsequently, the cutting block was fixed, and the anterior cut was made. The TekscanTM remained in place, and the joint was again carefully closed. All measurements of kinematics and contact mechanism were repeated again. In the next steps, patellar resurfacing was performed, using a biconvex button centred on the ridge, starting with a 3 mm under resection (overstuffing), followed by a reconstruction of the natural patellar thickness and by a 3 mm over resection (thinning). After each step, the Tekscan™ was positioned in the same place, the joint was closed and the knee was tested while performing the four different motor tasks.

A second post-test CT was then performed to confirm the unchanged positions of the reference frames and to con- trol the rotational position of the trochlear component.

Data processing

To analyse kinematics, the pre-test CT data were loaded and analysed using Mimics 11.02 and its MedCAD module (Materialise, Haasrode, Belgium). Surface reconstructions of femur, tibia and patella were created, and the relevant bony landmarks were identified. Based on these landmarks, relevant axes and planes on tibia, femur and patella were determined, as described in previous publications and the work of Belvedere [7, 45–47]. These coordinate systems were then used to convert the marker trajectories, as measured with the camera system, in an anatomically meaningful description of patellofemoral kinematics. Accuracy and precision of the motion analysis system, used for the kinematic recordings of the markers, were on the order of 0.2 mm. The six degree-of-freedom patellofemoral joint motion was described as motion in terms of rotation about and translation along a combination of femoral and patellar axes, according to the articular convention proposed by Bull et al. [8]. The kinematic results were presented as a function of tibiofemoral flexion angle, every 5°. Total contact pressures and areas were measured and statistically analysed every 0.4° of flexion, but presented as a function of the knee flexion angle, every 4°.

This study has been approved by the ethical committee of the University of Leuven (ID-number NH019-2010-04-02).

Statistical analysis

Linear mixed models were used to analyse the kinematic differences between the measurements. Random effects accounted for correlation between repeated observations within the same specimen. The models included a random intercept for specimen (general or measurement specific) and a random slope for flexion angle. The evolution of motion over an increasing flexion angle was modelled with linear-, quadratic- or splines-based models. The model with the best fit (lowest Akaike information criterion) was selected and used for inference. In a first step, the interaction between kinematic measurement and flexion angle was tested using a likelihood ratio test. In case of significant interaction, pairwise differences between measurements were analysed at different flexion angles. In case of non-significant interaction, a likelihood ratio test was performed for a main effect. In case of an overall effect, all pairwise differences between settings were further tested. The evolution of the pressure or area over increasing flexion angles was modelled through restricted cubic splines using five knots. Interactions between the setting and flexion angle were considered, allowing to model different evolutions for the measurements. Bonferroni step-down correction for multiple testing was performed. Five percentage of significance level was assumed.

An additional statistical reliability analysis was carried out on the native patellofemoral kinematic measurements of the first six specimens. This relative measure for reliability yields values between 0 and 1 with values close to 1 indicating highly reliable/repeatable measurements.

All analyses were performed using the SAS package, version 9.2 of the SAS System for Windows.

Results

Patellofemoral kinematics of the Journey PFA (Fig. 3).

During *squat motions*, significant differences in patellar flexion ($p < 0.0001$), patellar rotation ($p < 0.05$) and patellar tilt ($p < 0.001$) were noticed after isolated trochlear resurfacing: more patellar flexion, more lateral rotation in the mid-flexion range and more lateral tilt in the low- and midflexion range after isolated trochlear resurfacing. Table 1 gives the mean differences in patellar flexion (Table 1a), rotation (Table 1b) and tilt (Table 1c) at different tibiofemoral flexion angles. After additional patellar resurfacing, no significant differences in patellar rotational motions were measured compared to the native knee, except more patellar flexion in the low flexion range (<40°) and more patellar extension in the high flexion range (>80°) in the knees with a PFA. Overall, there were no significant differences in patellar translations.

During *open chain motion*, the same trends were noticed, but only the difference in lateral tilt between the natural knee and the knee after trochlear resurfacing was statistically significant

($p < 0.0001$). Lateral patellar tilt was increased in the resurfaced knee (Table 2). There were no significant differences in patellar rotations and translations once the patella was resurfaced. During *passive motion*, not any significant difference in patellar motion was measured between the native and the resurfaced knee.
Both 2 mm overstuffing and 2 mm thinning of the patellofemoral compartment did not induce changes in the patellofemoral kinematic patterns, with exception of the expected changes in antero-posterior patellar position.

Patellar contact force, contact area and contact pressure after PFA.
When performing a *squat motion*, the average patellar contact pressures were statistically significant ($p < 0.0001$) and increased over almost the whole flexion range when a trochlear resurfacing was performed, both without and with patellar resurfacing (Fig. 4). However, the differences reached clinically meaningful values after additional patellar resurfacing (Table 3). The average contact area was significantly ($p < 0.0001$) reduced after patellar resurfacing beyond 35° of knee flexion, whereas no significant differences were noticed when an isolated trochlear resurfacing was performed, with exception of a reduction in contact area ($p < 0.05$) in the flexion range between 50° and 60°. The contact force was not significantly different.
During *open chain motions*, findings for patellar pressure were the same as during squatting. However, patellar contact area, as well as patellar contact force, were, both without and with patellar resurfacing, significantly ($p < 0.0001$) reduced.
Overstuffing the anterior compartment, by a 2 mm under resection of patellar bone, created higher contact pressures, but differences were only significant in the low flexion range (<40°) during squatting ($p < 0.001$) and between 50° and 70° of knee flexion during resisted open chain motion ($p < 0.05$).

Rotational position of the trochlear component.
The rotational alignment of the trochlear component was measured on surface reconstructions based on the post- op CT scan. Determining the plane of the anterior cut appeared to be very difficult due to bone loss after removing the component and scattering from the marker frame. The measured rotational positions have therefore to be interpreted with care. With reference to the condylar centre line, the femoral component showed a mean internal rotation of 0.2° ±

4.5°. With reference to the anatomical transepicondylar line and to the posterior condylar line, the component showed a mean internal rotation of 5.3° ± 3.6° and 0.2° ± 4.7°, respectively.

Discussion

The present study supported the assumption that the investigated anatomical patellofemoral implant design reproduced the natural patellofemoral kinematics, provided the patella is also resurfaced. Only changes in patellar flexion were found, with significant less patellar flexion beyond 80° of tibiofemoral flexion in knees with a PFA. Without patellar resurfacing, significant differences in patellar flexion, rotation and tilt were measured, compared to the normal knee. However, as previously published for TKA, patellar resurfacing resulted in a significant increase in average patellar pressure and a decrease in contact area [22]. These changes were not noticed when the patella was not resurfaced. As such, our hypothesis can be accepted.

Overstuffing the patellofemoral joint by increasing the patellar thickness with 2 mm had no other influence on the patellofemoral kinematics than the expected change in AP position. The average patellar pressures were, however, increased in a limited flexion range. It is not clear to which extent the elevated pressures may explain the clinical observation of a greater incidence of anterior knee pain when the anterior compartment has been build up [2]. However, a large retrospective clinical study of Pierson et al. [40] of 1100 TKAs did not endorse this vision. Others could not find a relation between an increase in antero-posterior dimension of the anterior compartment after PFA and range of motion or clinical outcome [33]. There is, however, a lack of clinical studies focusing on that subject.

This study was performed using a specific patellofemoral implant (Journey PFJ™, Smith–Nephew, Memphis, TN, USA), which can be considered as an 'anterior cut prosthesis' implant design. Results can therefore not be generalized, as there is a wide variation in geometric design criteria between the different prostheses, available on the market. Design features, such as orientation, depth of the trochlear groove and geometry of the patellar buttons, can have a major influence on patellofemoral kinematics and contact mechanism [10]. The tested device completely replaces the anterior joint compartment of the knee, similar to a total knee prosthesis (TKP). The trochlear design has the same characteristics as the trochlear configuration of the Genesis 2™ TKP (Smith–Nephew™, Memphis, TN, USA). In contrast to the Avon™ patellofemoral prosthesis from Stryker™, which is the most clinically investigated patellofemoral design from the second generation, the currently used design has an asymmetric trochlear groove, which should make it more anatomical [2, 23, 38, 39]. The trochlear groove is lateralized and deepened, and the relative broad lateral flange has a proximal extension on to the anterior femur. This is responsible for a fast trochlear engagement of the patella in the early flexion range [3, 30]. The available instrumentation technique provides a better control on the femoral rotational position. So far, there is only one study, recently published, that reports the clinical and radiological results after PFA with this design at short-term (2 years) follow-up [6]. The authors noticed an overall significant improvement of all scores, with however a greater and more continuous benefit for the group of patients that needed a combined surgical intervention for instability. This was in contrast to previous publications, reporting a greater success rate in patients with patellofemoral destruction secondary to trochlear dysplasia, or reporting no influence of the initial pathology on the outcome [4, 44].

To our knowledge, the literature on patellofemoral kinematics of knees with a PFA is limited to two studies, measuring the sagittal plane kinematics of, respectively, the Avon™ (Stryker™) and the FPV (Wright Medical Technology ™) patellofemoral prosthesis [17, 34]. The first mentioned study [17] examined the patellar tendon angle of 12 patients during different activities, using fluoroscopic technology [17]. Measurements were done during active knee extension (open chain motion), active knee flexion (squatting) and a step-up exercise in a sequential static way at 10° intervals. No significant differences were found between the kinematics of knees with a PFA and normal knees, except for a slight elevation of the PTA in knee with PFA, which was contributed to a small anterior displacement of the distal patellar bone (patellar extension). The exact reason for this finding remained unclear. Our study also showed a pattern with less patellar flexion after PFA (with patellar resurfacing), compared to the normal knee, but only in the deep flexion range, whereas more patellar flexion was found in the early flexion range. The cross-over point was localized at about 50° of knee flexion. This finding seems logical and can probably be explained by the geometry of the biconvex button. Once the most distal part of the rounded button touches the anterior prosthetic trochlear flange at engagement of the trochlea, it might indeed induce more patellar flexion than when the differently shaped distal surface of the natural patella touches the same point. But once the knee moves to deeper flexion, the patellar contact point on the button moves more proximal and consequently ‘rides’ up the rounded button, which inevitably drives the patellar bone to a somewhat more extended position compared to the natural knee. However, as the patellar button reaches the end of the trochlear component at a deeper flexion angle, its distal tip will tilt posteriorly once it leaves the groove, consequently driving the patellar bone again to a more flexed position. This is the cantilever effect, described by Monk et al. [34]. This phenomenon compensates partially for the trend towards more extension caused by the biconvex button as it makes contact with the trochlea. We indeed measured a maximum average difference of 5.7° ± 2.4° at 100° of flexion. However, in deeper flexion the average difference decreased, reaching 3.0° ± 1.9° at 120° of flexion.

The study on kinematics of the FPV prosthetic device had a comparable fluoroscopic-based methodology, using the PTA and patellar flexion angle in relation to the knee flexion angle [34]. The knee was investigated during step-up and lunge motions. In contrast to the previous study using the Avon™ prosthesis, the PTA was lower in the knees with a PFA at high flexion angles for the lunge exercise. For the step-up exercise, there was only a significant difference in PTA at 50° of flexion. Those findings were contributed to a deeper patella position in the trochlear groove, either by a deeper trochlea or by a thinning of the patella. These changes in the sagittal plane were not supported by the current study, although the cantilever theory of Monk can partially explain our findings. It should be noted that the use of the PTA as measurement tool for the relative motion of the femur in the antero-posterior plane is controversial [43]. A second point of criticism on both studies is that conclusions are made based on comparison with a group of non-paired healthy knees of individuals with a much lower mean age than the patients of the PFA group. Considering the high inter-subject variability in patellofemoral kinematic patterns and certainly in absolute values, one should be careful with interpreting the results. An in vitro study certainly cannot perfectly reproduce daily live motions, but on the other hand, it allows to compare two or more situations within the same specimen.

Considering the observed kinematic changes in the prosthetic knee without patellar resurfacing, one expects a redistribution of the pressure towards the lateral and distal part of the patella. On average, the increase in patellar contact pressure is limited to a maximum of 2.9 ± 0.5 MPa, which is in contrast to elevations up to 8.5 ± 2.4 MPa after placing of a biconvex button. The pressure distribution has to be further investigated and analysed. If this assumption would be confirmed in future research projects, performing a lateral retinacular release might be an option in situations where the patella is or cannot be resurfaced in order to reduce the lateral tilting and re-distribute the pressures.

In addition to the use of one specific design and the in vitro set-up, this study had some other limitations. The amount of specimens tested was limited. We only tested a biconvex patellar button. Resurfacing can also be done with an anatomical asymmetric button, which might influence the kinematic behaviour. However, from the kinematic standpoint, there is apparently no need to consider an asymmetrical button. Nevertheless, the possible impact on contact pressure and especially pressure distribution of such a button justifies further investigation.

The findings of this study have two clinical implications. First of all, the ability to reproduce the natural patellofemoral kinematics by using an isolated patellofemoral design makes this type of procedure more attractive for the treatment of isolated patellofemoral arthrosis, compared to TKA. And secondly, the better kinematic reproduction with additional patellar resurfacing adds a point in the clinical discussion whether or not to resurface the patella in knee arthroplasty.

Conclusion

From a pure biomechanical standpoint, resurfacing the patella in the PFA does not seem to be advisable, as it significantly increased the average patellar contact pressures. It is not clear whether and to which extent this increased contact pressure may influence the pressure in the underlying bone and may play a role in the onset of persisting anterior knee pain. However, from a pure kinematic point, the natural patellar tracking seems to be better reproduced with a resurfaced patella, as the absence of resurfacing induced more patellar flexion, more lateral patellar rotation and more lateral tilting. Despite the excellent patellofemoral kinematic figures of this design, longer-term clinical studies are necessary to validate a positive impact on the complication rate and survival and to prove the durability of the good short-term clinical results.

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Conflict of interest

Some of the authors have a sporadic consultancy agreement (lectures) with Smith&Nephew.

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Captions

Table 1: Mean (SD) patellar flexion (°) (a), patellar rotation (°) (b) and patellar tilting (°) (c) for squat motion in the native knee and mean differences (with SD and p values) after both isolated trochlear resurfacing (PFJ) and standard patellofemoral arthroplasty with patellar resurfacing (PFJ + P neutral), compared to the native knee, at different tibiofemoral flexion angles.

ns not significant * $p < 0.05$, ** $p < 0.001$, *** $p < 0.0001$

Table 2: Mean (SD) patellar tilting (°) for open chain motion in the native knee and mean differences (with SD and p values) after both isolated trochlear resurfacing (PFJ) and standard PFA with patellar resurfacing (PFJ + P neutral), compared to the native knee, at different tibiofemoral flexion angles.

ns not significant * $p < 0.05$, ** $p < 0.001$, *** $p < 0.0001$

Table 3: Mean (SD/p values) differences (MPa) in patellofemoral contact pressures between the native knee, the knee with trochlear implant and the knee with trochlear and patellar implant, at different tibiofemoral flexion angles from 30° to 105°.

ns not significant * $p < 0.05$, ** $p < 0.001$, *** $p < 0.0001$

Figure 1: Natural knee specimen with Tekscan covering the patella.

Figure 2: Knee specimen mounted in knee rig (with Tekscan covering the patellar surface) to perform a squat movement.

Figure 3: Patellar rotation (°) and translation (mm) plots as function of the tibiofemoral flexion angle, for the squat motion, comparing the native knee, after opening and closing the joint (native opened), the knee after trochlear resurfacing without patellar resurfacing (PFJ) and the knee after PFA with patellar resurfacing (PFJ + P neutral). The error bars depict the standard deviation.

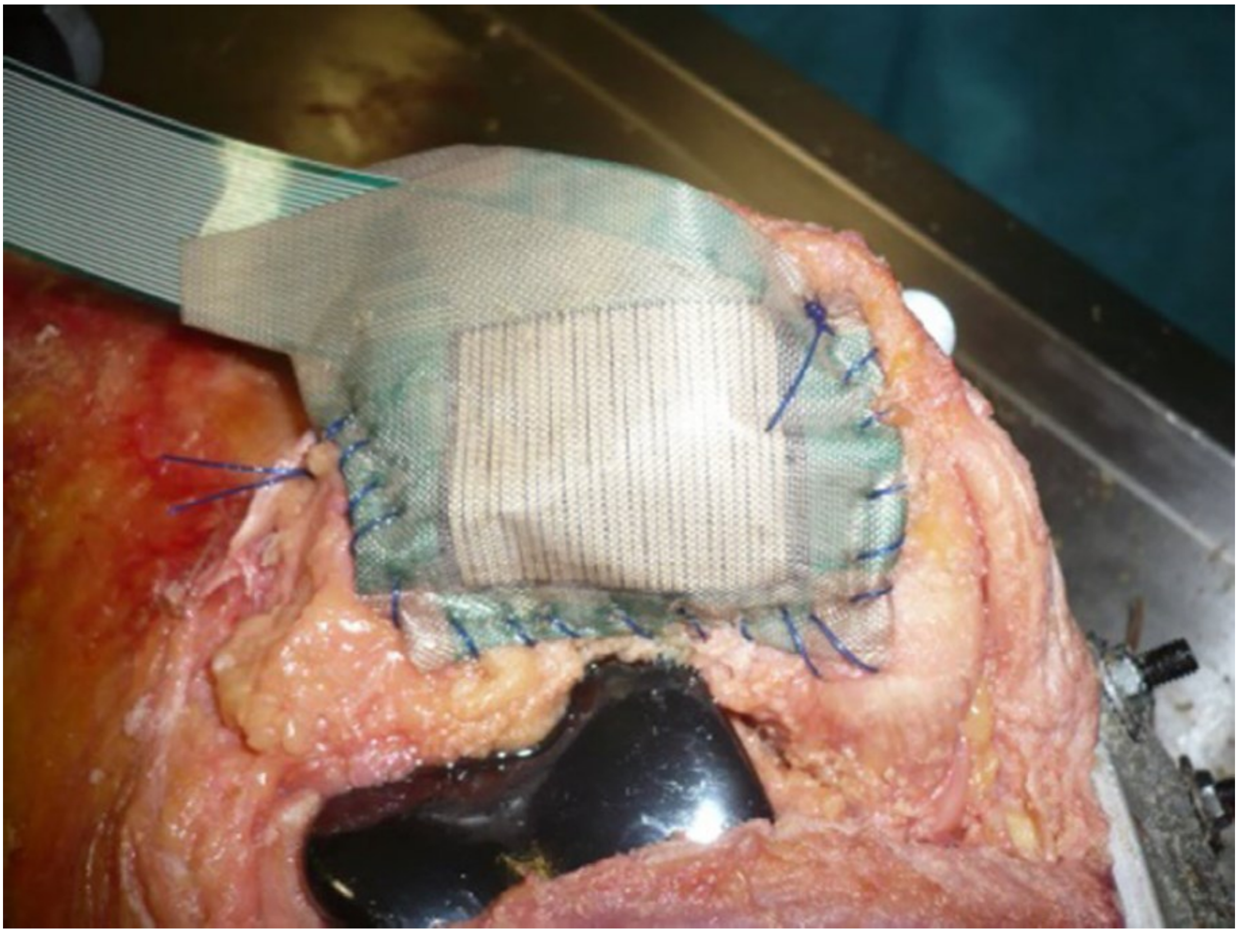
Figure 4: Patellar contact pressure (MPa) plots as function of the tibiofemoral flexion angle (°), for the native knee (native opened), the knee with a resurfaced trochlear (PFJ) and the knee

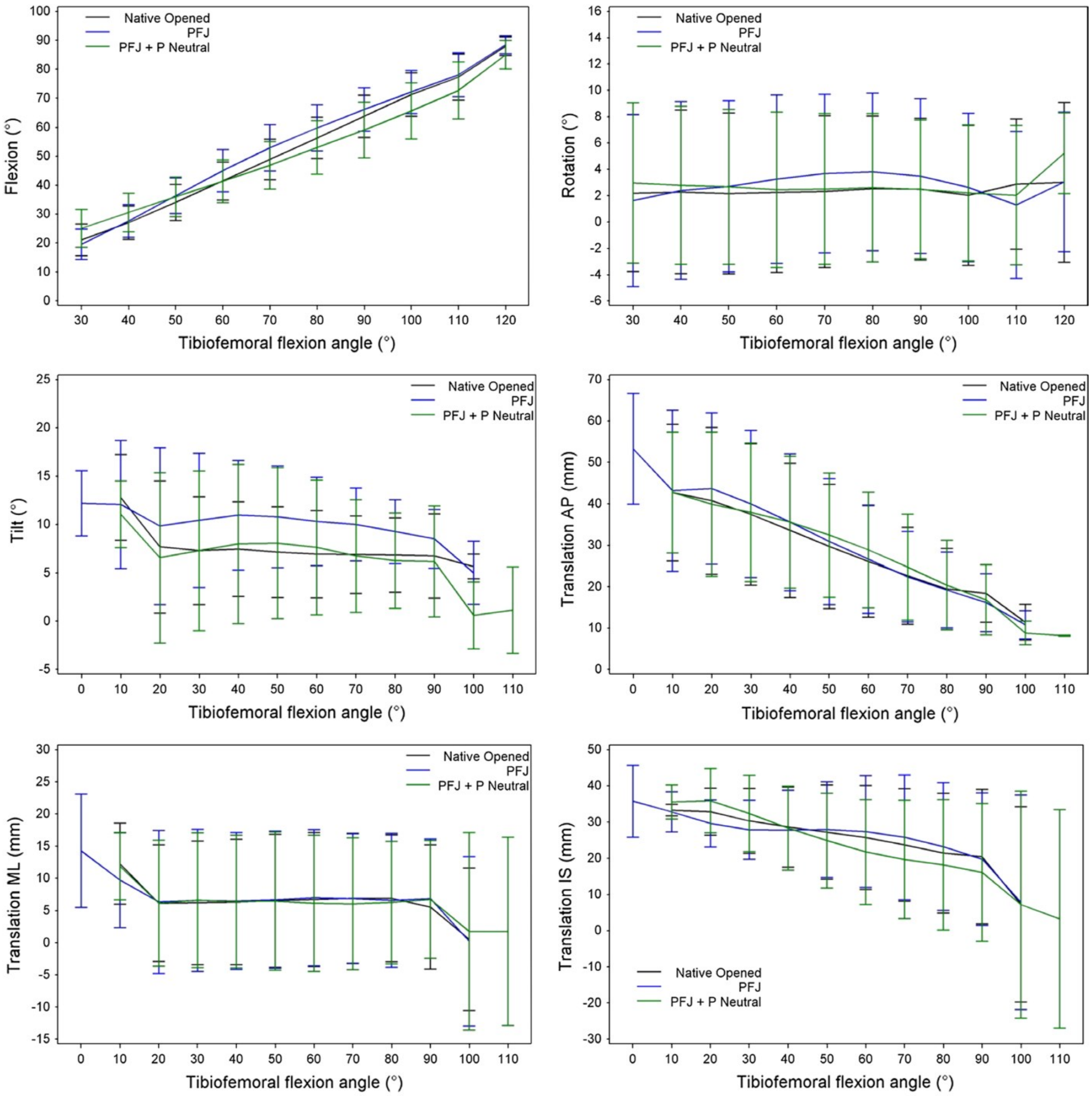
with trochlear and patellar resurfacing (PFJ + P neutral). The error bars depict the standard deviation.

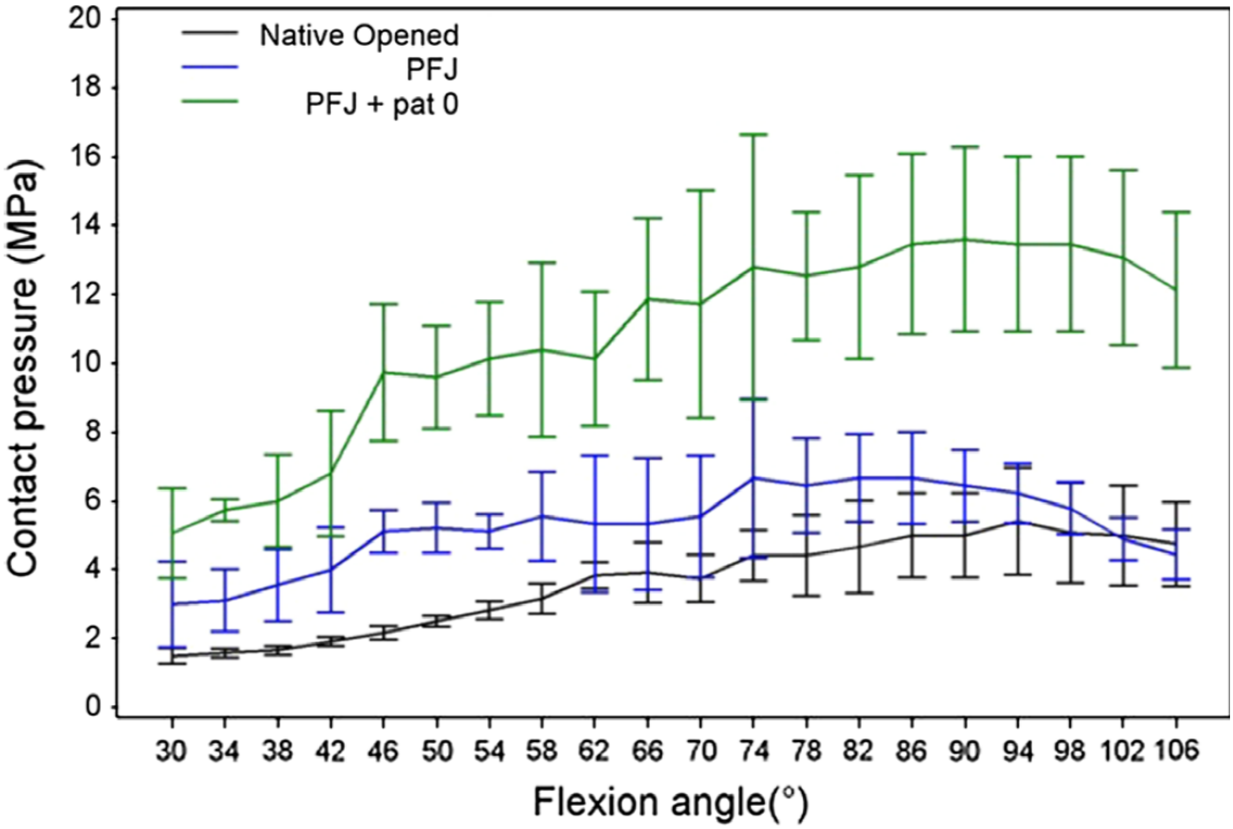
Tibiofemoral flexion	Native opened		Difference compared to native					
			PFJ			PFJ + P neutral		
	Mean	SD	Mean	SD		Mean	SD	
(a) Squat motion patellar flexion								
30	21.1	5.5	1.5	0.9	*	−4.0	1.4	***
40	27.0	5.7	−0.6	0.9	ns	−3.5	2.2	**
50	34.0	6.2	−2.3	1.0	***	−1.9	2.4	ns
60	41.4	6.5	−3.5	1.5	***	0.1	2.5	ns
70	48.8	7.0	−4.0	1.5	***	2.0	2.5	ns
80	56.3	7.2	−3.5	1.1	***	3.3	2.7	***
90	63.8	7.3	−2.3	1.5	***	4.8	2.7	***
100	71.3	7.5	−0.8	0.9	*	5.7	2.4	***
110	77.3	7.9	0.0	0.7	ns	5.6	3.0	***
120	87.9	3.2	−0.5	0.8	ns	3.0	1.9	***
(b) Rotation								
30	2.2	6.0	0.6	1.3	ns	−0.8	0.6	ns
40	2.3	6.2	−0.1	1.4	ns	−0.5	0.5	ns
50	2.2	6.1	−0.5	1.0	ns	−0.5	0.6	ns
60	2.3	6.1	−1.0	1.2	*	−0.2	1.0	ns
70	2.3	5.8	−1.4	1.0	*	−0.2	1.1	ns
80	2.5	5.6	−1.3	1.1	*	−0.1	1.3	ns
90	2.5	5.4	−1.0	1.2	*	0.0	1.6	ns
100	2.0	5.3	−0.6	1.0	ns	−0.2	2.0	ns
110	2.9	5.0	−0.1	0.9	ns	−0.9	2.6	ns
120	3.0	6.1	−0.0	1.4	ns	−2.2	3.6	*
(c) Tilt								
30	7.3	5.8	−3.2	2.7	***	−0.2	4.3	ns
40	7.5	4.7	−4.5	2.1	***	−1.3	4.7	ns
50	7.2	4.2	−5.3	1.2	***	−2.5	4.8	ns
60	7.0	4.1	−5.1	1.4	***	−2.4	3.9	ns
70	7.0	3.7	−4.4	2.0	***	−1.9	3.6	ns
80	7.0	3.2	−3.5	2.0	***	−0.6	4.0	ns
90	7.1	3.2	−1.7	1.9	*	−0.1	3.7	ns
100	7.1	3.6	−0.4	1.6	ns	0.2	3.7	ns
110	8.0	3.3	0.4	0.8	ns	−0.9	4.1	ns
120	5.4	3.6	−0.1	1.1	ns	−1.1	5.6	ns

Patellar tilt open chain			Difference wrt. native opened					
Tibiofemoral flexion	Native		PFJ			PFJ + P neutral		
	Mean	SD	Mean	SD		Mean	SD	
30	7.3	5.8	−3.2	2.7	***	−0.2	4.3	ns
40	7.5	4.7	−4.5	2.1	***	−1.3	4.7	ns
50	7.2	4.2	−5.3	1.2	***	−2.5	4.8	ns
60	7.0	4.1	−5.1	1.4	***	−2.4	3.9	ns
70	7.0	3.7	−4.4	2.0	***	−1.9	3.6	ns
80	7.0	3.2	−3.5	2.0	***	−0.6	4.0	ns
90	7.1	3.2	−1.7	1.9	*	−0.1	3.7	ns
100	7.1	3.6	−0.4	1.6	ns	0.2	3.7	ns
110	8.0	3.3	0.4	0.8	ns	−0.9	4.1	ns
120	5.4	3.6	−0.1	1.1	ns	−1.1	5.6	ns

Contact pressure (MPa)			Difference wrt. native					
Flexion_angle	Native		PFJ			PFJ + pat 0		
	Mean	SD	Mean	SD		Mean	SD	
30	1.5	0.2	−1.5	1.1	***	−3.6	1.1	***
34	1.6	0.1	−1.5	0.9	***	−4.2	0.3	***
38	1.7	0.1	−1.9	1.0	***	−4.3	1.3	***
42	1.9	0.1	−2.1	1.1	***	−4.9	1.7	***
46	2.2	0.2	−2.9	0.5	***	−7.6	1.8	***
50	2.5	0.2	−2.7	0.6	***	−7.1	1.5	***
54	2.8	0.3	−2.3	0.6	***	−7.3	1.4	***
58	3.2	0.4	−2.4	1.3	***	−7.2	2.3	***
62	3.8	0.4	−1.5	1.9	***	−6.3	1.8	***
66	3.9	0.9	−1.4	1.6	***	−8.0	1.8	***
70	3.8	0.7	−1.8	1.3	***	−8.0	2.8	***
74	4.4	0.7	−2.3	1.6	***	−8.4	3.3	***
78	4.4	1.2	−2.0	0.6	***	−8.1	1.4	***
82	4.7	1.3	−2.0	0.5	***	−8.1	2.1	***
86	5.0	1.2	−1.7	0.4	***	−8.5	2.3	***
90	5.0	1.2	−1.4	0.5	***	−8.6	2.4	***
94	5.4	1.6	−0.8	1.0	***	−8.1	2.7	***
98	5.1	1.5	−0.7	1.1	***	−8.4	3.0	***
102	5.0	1.4	0.1	1.6	ns	−8.1	2.9	***
106	4.8	1.2	0.3	1.7	*	−7.4	2.2	***




Native Opened
PFJ
PFJ + P Neutral
Flexion (°)
Rotation (°)
Tilt (°)
Translation AP (mm)
Translation ML (mm)
Translation IS (mm)
Tibiofemoral flexion angle (°)



Native Opened
PFJ
PFJ + pat 0
Contact pressure (MPa)
Flexion angle(°)
0
2
4
6
8
10
12
14
16
18
20
30 34 38 42 46 50 54 58 62 66 70 74 78 82 86 90 94 98 102 106

526