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Quadriceps force during knee extension in different replacement scenarios with a modular partial prosthesis

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ABSTRACT

Background: Previous biomechanical studies have shown that bi-cruciate retaining knee replacement does not significantly alter normal knee kinematics, however, there are no data on the influence of a combined medial and patellofemoral bi-compartimental arthroplasty. The purpose of this in vitro study was to evaluate the effect of different replacement scenarios with a modular partial knee replacement system on the amount of quadriceps force required to extend the knee during an isokinetic extension cycle.

Methods: Ten human knee specimens were tested in a kinematic knee simulator under (1) physiologic condition and after subsequent implantation of (2) a medial unicondylar and (3) a trochlear replacement. An isokinetic extension cycle of the knee with a constant extension moment of 31 Nm was simulated. The resulting quadriceps extension force was measured from 120° to full knee extension.

Findings: The quadriceps force curve described a typically sinusoidal characteristic before and after each replacement scenario. The isolated medial replacement resulted in a slightly, but significantly higher maximum quadriceps force (1510 N vs. 1585 N, $P = 0.006$) as well as the subsequent trochlear replacement showed an additional increase (1801 N, $P = 0.008$). However, for both replacements no significant difference to the untreated condition could be detected in mid-flexion (10–50°).

Interpretation: When considering a bi-compartimental replacement an increase of required maximum quadriceps force needed to extend the knee has to keep in mind. However, the close to physiological movement in mid-flexion suggests that patients with a bi-crutiate retaining arthroplasty might have an advantage in knee stability compared to total knee arthroplasty.

1. Introduction

When considering knee replacement surgery, orthopaedic surgeons have the option to undertake a total knee arthroplasty (TKA) or a partial knee replacement. As reported earlier, even patients with excellent results after TKA have an altered walking pattern with less flexion, a shorter swing phase, and weaker extension strength in the operated knee (Andriacchi, 1993; Li et al., 2013).

Besides the prosthetic design, especially the loss of proprioreception and alternations in lever arms and extension moments are responsible for said abnormal muscle function after TKA (Li et al., 2013; Ostermeier and Stukenborg-Colsman, 2011). In this context the anterior cruciate ligament (ACL) is of special interest. When the ACL is sacrificed, the lever arm of the extensor mechanism is reduced due to a paradoxical anterior movement of the femur relative to the tibia during flexion, which results in higher quadriceps muscle forces required to extend the knee (Dennis et al., 1998; Heyse et al., 2010a, 2010b; Lewandowski et al., 1997; Ostermeier and Stukenborg-Colsman, 2011; Ostermeier et al., 2004).

After a period of diminishing popularity, the interest in bi-cruciate retaining partial knee replacement has increased in recent years (Zanasi, 2011). In certain patients, the medial and patellofemoral (PF) compartments are affected by osteoarthritis, but the ACL and lateral compartment remain healthy. These patients are candidates for a medial plus PF bi-compartmental knee replacement to prevent or postpone the TKA and to preserve the normal ligament structures and proprioception of the knee (Heyse et al., 2010a, 2010b; Rolston et al., 2007). The clinical experiences with new knee designs indicate a high level of function and knee kinematics in patients retaining essential features of the normal knee motion (Leffler et al., 2012; Thienpont and Price 2013).

Previous biomechanical studies have shown that bi-cruciate retaining unicondylar knee replacement does not alter normal knee kinematics during simulated stair climbing in a cadaver model (Patil et al., 2005). No significant differences in tibial axial rotation, femoral rollback or quadriceps tension were noted between the physiological and unicompartmental conditions.

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However there are no biomechanical data on the influence of a combined medial and PF bi-compartimental arthroplasty and its effect on the required quadriceps extension force.

Thus, the purpose of this in vitro study was to evaluate the effect of the different components in a modular partial knee replacement system on the amount of quadriceps force required to extend the knee during an isokinetic extension cycle.

2. Methods

The experimental set-up and test cycle was used in this study were the same as previously reported by Ostermeier et al. It is described to simulate an isokinetic extension cycle of the knee, which allows an approximation of loadings close to the magnitude of the physiological forces and moments about the knee. In this study, ten fresh frozen adult knee specimens of nearly the same size (mean age = 75 (64–85) years, 5 male and 5 female) fulfilling the criteria of an intact periarticular musculature and an intact capsule and ligament apparatus were used. Furthermore, no significant osteoarthritis was present. The specimens were prepared according to standard protocols with transection of the femur and tibia 25 cm proximally and distally to the knee joint line. The skin and subcutaneous tissue were removed preserving the muscles, articular capsule, ligaments and tendons. The specimens were mounted into our in-vitro knee simulator in which the said isokinetic flexion–extension movements were simulated (Fig. 1). The specimens were positioned with the femur fixed horizontally and the patella facing downwards. The femoral and tibial bone stumps were fixed with bone cement in metal sleeves to reproduce the same positioning on every test cycle. The tibia was attached to the simulator at mid-length by means of a linear rotational bearing. This allowed axial sliding and turning as well as rotation transverse to the axis of the tibia. The bearing itself was attached to a swing arm that allowed motion in the varus/valgus plane. The characteristics of this arrangement were described to give complete freedom of motion of the joint, with the exception of flexion–extension, which is determined by the position of the swing-arm. The swing arm was equipped with a strain-gauge-based load-measuring device that allowed continuous monitoring of a torsional moment applied to the tibia. Tibial movement was provoked by the coordinated activation of 3 hydraulic cylinders, which were attached to the specimens' tendons by special clamps. By this, quadriceps muscle force and a co-contraction of the hamstring muscles were simulated as well as an external flexion moment was applied. One complete test cycle simulated an isokinetic extension cycle from 120° knee flexion to full extension. The quadriceps cylinder thereby applied sufficient force to the quadriceps tendon in a closed-loop control cycle to generate a constant knee extension moment of 31 Nm. The hamstring cylinder simulated co-contraction of the hamstring muscles with a constant co-contractive flexion force of 100 N. Initially, the swing arm was activated to bring the specimen into a position of 120° of flexion. In our standard protocol, the quadriceps cylinder was then activated in feedback control to provide a constant net joint extension moment by applying the constant extension moment at the swing arm. The joint moment was measured by the load cell in the swing arm, allowing continuous control of quadriceps force throughout the complete motion to maintain the nominal extension moment of 31 Nm, as reported earlier. This constant extension moment was resisted by a constant swing arm flexion moment, which was generated by a third hydraulic cylinder, creating an isokinetic extension movement. Measurement of the resulting quadriceps was performed using a load cell (Hottinger Baldwin Messtechnik GmbH, Darmstadt, Germany, accuracy 0.1 N) attached between the tendon clamp and the quadriceps cylinder. The data was collected at a frequency of 10 Hz. The degree of knee flexion was recorded using a custom-made voltage goniometer attached to the tibial swing arm at a frequency of 10 Hz and with an accuracy of 0.05°. All test cycles were run at room temperature.

The quadriceps forces of all specimens were at first measured in the native physiological joint. Subsequently, a medial unicompartmental knee replacement (Sigma® PFC High Performance Partial Knee; DePuy Orthopaedics, Kirkel, Germany) was implanted by the same surgical team without bone cement according to the manufacturer's guidelines. The prosthesis system offers a fixed polyethylene inlay. The tibia base-plate was implanted according to the anatomical joint line and posterior tibial slope. The femoral component position was adjusted to the tibial component in flexion and extension. The knee capsule and soft tissues were readapted and the specimen was remounted in the simulator. The test cycle was repeated as for the physiological knee.

That followed an additional implantation of the trochlear component (Sigma® PFC High Performance Partial Knee Trochlear Component; DePuy Orthopaedics, Kirkel, Germany) in every specimen. This again was performed by the same surgical team according to the manufactures' protocol without bone cement and without an additional patella resurfacing. The patella was freed from osteophytes and readjusted carefully to the trochlear groove. After implantation and closure of the capsule, the specimen was remounted in the simulator and the test cycles were repeated identically. The mean quadriceps forces of all test cycles were calculated.

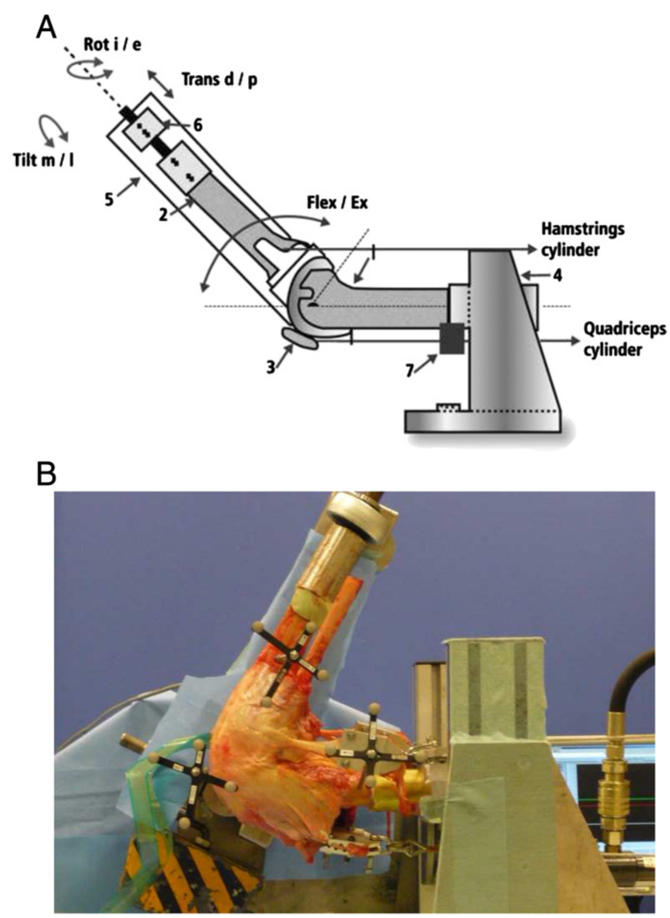

Fig. 1. A. Schematic side view of the experimental set-up; and B. Photograph of a specimen mounted in the in-vitro knee simulator. The specimen is brought from a position of 120° of flexion to full extension by applying force on the quadriceps cylinder, providing a constant joint extension moment (31 Nm) resisted by the swing arm. An additional flexion force is applied by the hamstrings cylinder. 1) femur, 2) tibia, 3) patella, 4) femur frame 5) swing arm, 6) strain gauge, and 7) load cell.

2.1. Statistics

Since no comparisons of the consecutive replacement scenarios with this modular type of prosthesis have been quantified before, no power analysis could be done. Differences in the quadriceps force between the mean values of the experimental groups were evaluated using the non-parametric paired Wilcoxon signed-rank test at a significance level of $P = 0.05$, as the forces from each group were not all normally distributed. We used SPSS for Windows (version 19) for statistical analysis.

3. Results

The typical sinusoidal quadriceps force curve reached a maximum value of 1510 N at 106° of flexion (SD = 3°, range 102–109°) in the physiological knee. Between 60° and 10° of knee flexion, a quadriceps force of less than 1000 N was required to extend the knee (Fig. 2).

The implantation of the unicompartimental medial prosthesis (UNI) resulted in a slightly, but significantly higher maximum quadriceps force of 1585 N ($P = 0.006$) at 104° (SD = 3°, 100–108°) knee flexion. Also the subsequent implantation of the trochlear component (BI) resulted in an additional increased maximum quadriceps force of 1801 N ($P = 0.008$) at 106° (SD = 6°, 94–115°) knee flexion (Table 1). Whereas not every sample showed this increase in maximum quadriceps force after implantation of an UNI (mean: 81 N, range: −22 to 237 N) this was observed in every single specimen after additional implantation of the trochlear prosthesis (mean: 217 N, range: 90 to 332 N).

The minimum quadriceps forces after implantation of the unicompartimental medial as well as after implantation of the bicompartimental medial and trochlear replacement showed no significant difference to the physiological knee (Table 1).

Reviewing the complete extension cycle we observed a significant increase in the required quadriceps force for the bi-compartimental replacement scenario except for the close to full extension movement (40°–10°). Significant differences between the physiological knee and the unicompartimental replacement only occurred in the mentioned maximum values and during mid flexion (60°) (Table 2 and Fig. 2).

4. Discussion

This study evaluates the dynamic changes in the quadriceps muscle force required to extend the knee after different replacement scenarios with a modular bi-cruciate retaining partial knee arthroplasty design. A low quadriceps extension force to extend the same extension moment was considered to be biomechanically advantageous, delivering a higher degree of efficacy of the extensor mechanism and displaying a better constraint of articulation (Ostermeier and Stukenborg-Colsman, 2011). Previous biomechanical studies have shown, that all ACL-sacrificing knee replacement systems result in altered quadriceps forces, as they do not reproduce the physiological kinematics (Ostermeier and Stukenborg-Colsman, 2011; Ostermeier et al., 2004, 2008). This effect is attributed to the changes of the physiological lever arm and the insufficient restoration of the tibiofemoral and patellofemoral trajectories. In contrast to this, Patil et al. demonstrated normal joint biomechanics after UKR implantation with regard to the tibial axial rotation or the femoral rollback (Patil et al., 2005). But in their setting the quadriceps force required to extend the knee was not determined. Also, another study by Price et al. evaluating the in vitro kinematics of UKR compared to TKA did not address this issue (Price et al., 2006).

One general limitation of this in vitro test is that only one constant moment during the whole extension cycle is simulated. This distinguishes the experimental setup from the factual varying peak extension moments during gait in vivo (Hatfield et al., 2011). We choose to apply a constant extension moment of 31 Nm, which is meant to represent a mean extension moment reached by patients over an entire isokinetic extension cycle (Ostermeier et al., 2004). The co-contraction force of the hamstrings was set to 100 Nm according to Durselen et al.'s analysis of the physiological muscle co-contractions around the knee (Durselen et al., 1995). This is to provide a sufficient additional stabilization of the knee joint. Both are simplified models, hence, the quantitative results of our study should not be translated directly to in vivo conditions. However, we consider the qualitative changes determined in this study to illustrate the mechanical effect after implantation of the various knee prosthesis systems both in vitro and in vivo.

Compared to previous measurements (Ostermeier and Stukenborg-Colsman, 2011; Ostermeier et al., 2004, 2008) with the same test set-up in our institution the determined physiological maximum quadriceps forces were similar. The displayed overall sinusoidal quadriceps force curve with the lowest values under physiological knee conditions between 60° and 10° of knee flexion showed a good correlation to previous studies, as well. However, the minimum values differed by about 20%. Equal effects were observed in several studies with the same test set up before (Ostermeier and Stukenborg-Colsman, 2011; Ostermeier

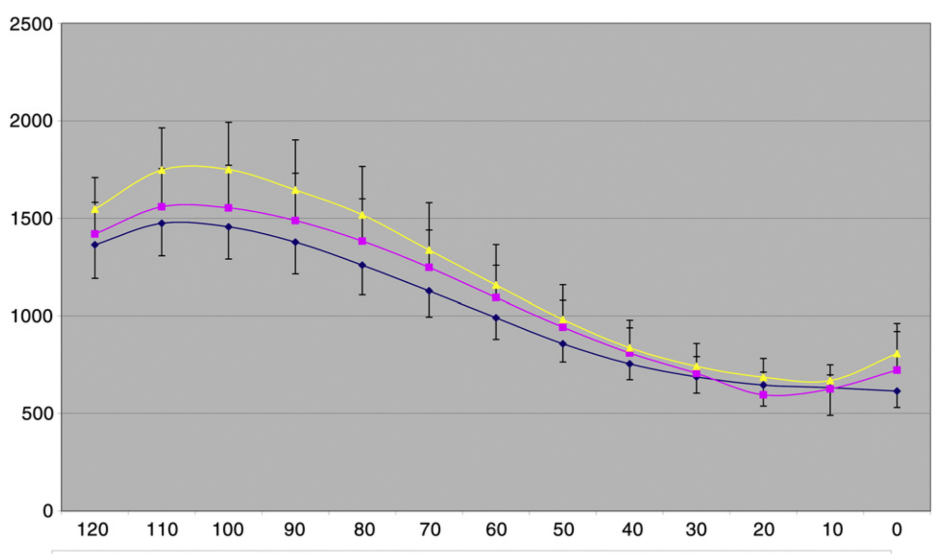

Fig. 2. Quadriceps force required to generate an extension moment of 31 Nm under (1) physiological knee conditions, (2) after implantation of a medial unicondylar prosthesis and (3) after consecutive implantation of a trochlear replacement from 120° knee flexion to full extension. Mean values, bars represent SD.

Table 1
Maximum and minimum quadriceps force to generate an extension moment of 31 Nm under (1) physiological knee conditions, (2) after implantation of a medial unicondylar prosthesis and (3) after consecutive implantation of a trochlear replacement in a modular unicompartimental knee design.

		Force* [N]	SD	*P*-value**	*P*-value***
Maximum	Physiological	1510	176		
	Medial unicompartimental replacement	1585	196	0.006	
	Medial + trochleal bi-compartimental replacement	1801	232	0.008	0.008
Minimum	Physiological	598	53		
	Medial unicompartimental replacement	605	62	0.944	
	Medial + trochleal bi-compartimental replacement	637	75	0.084	0.209

* Mean values, with standard deviation (SD).
** *P*-values for comparison to physiological native knee conditions.
*** *P*-values for comparison between the 2 types of replacement scenarios.

et al., 2004, 2008). This was explained with potential slightly different external conditions such as age or activity level of the donors, size of the specimen or even room temperature. This means, that the absolute values of the quadriceps forces cannot be directly compared to previous studies, but the effect of the prosthesis on the joint kinematics compared to the respective native condition clearly can be compared to previous data.

Typically, the physiological quadriceps force describes a sinusoidal curve during extension as a result of the translating tibiofemoral and patellofemoral contact points und consecutive lever arm changes (Nisell, 1985; Ostermeier et al., 2004). In doing so, the force of the quadriceps muscle is at its minimum most of the time during daily activity. Interestingly, in this clinically relevant mid flexion range of motion the determined quadriceps forces between the various replacement scenarios did not differ significantly to the physiological knee kinematics. However, the determined maximum quadriceps force occurring in deep flexion showed a significant increase to the physiological knee for all the replacement scenarios.

On the one hand this indicates that even after the singular implantation of the fixed bearing medial UNI the normal knee kinematics and lever arms could not be reproduced physiologically. However, on the other hand the bi-crutiate retaining arthroplasty showed a stable flexion/extension axis with physiological quadriceps force especially in the clinical relevant mid flexion movement.

In our previous biomechanical studies on tricompartimental knee arthroplasties we did observe a slight increase in the mentioned maximum quadriceps force after implantation of a single radius femoral design prosthesis and a significant increase after implantation of a multi radius design prosthesis (Ostermeier and Stukenborg-Colsman, 2011). Additionally, we have demonstrated an unphysiological shift of the quadriceps force maximum to lower flexion angles for various multi radius knee prosthesis, earlier (Ostermeier and Stukenborg-Colsman, 2011; Ostermeier et al., 2004, 2008). This is considered to indicate a paradoxical movement of the tibiofemoral contact point following implantation of a TKA. Moreover, we have shown a significant difference in required extension force for both, single and multi radius TKA compared to the physiological knee, especially in the mid flexion movement. This is in contrast to the bi-cruciate retaining partial knee prosthesis. In this study we found the quadriceps force maximum at the same flexion angle compared to the physiological knee and the mid flexion quadriceps forces did not differ from those of the native joint, at all. This displays a minimized paradoxical movement and an almost restoration of the physiological lever arm for the bi-cruciate retaining arthroplasy.

The additional implantation of a trochlear replacement led to a significant increase of the maximum quadriceps force in deep flexion to extension movements. This might be due to changes in the patellofemoral contact points and in the patella path of motion. But again, the mid flexion scenario showed no significant difference to the physiological knee. Even though, the difference between the replacement scenarios of the unicompartimental to the bi-compartimental one was significant. This data might indicate a probably not anatomical alignment of the patellofemoral joint with this trochlear prosthesis.

Table 2
Quadriceps force to generate an extension moment of 31 Nm (1) under physiologic knee conditions, (2) after implantation of a medial unicondylar prosthesis and (3) after consecutive implantation of a trochlear replacement in a modular unicompartimental knee design.

	Force[a] [N]	SD	QF % of phys. knee[b]	*P*-value[c]	*P*-value[d]
0° knee flexion					
Physiological	615	84			
Medial replacement	723	197	118	0.3	
Medial + trochlear replacement	808	154	131	0.01	0.1
30° knee flexion					
Physiological	688	84			
Medial replacement	707	86	102	0.6	
Medial + trochlear replacement	742	118	108	0.3	0.03
45° knee flexion					
Physiological	786	79			
Medial replacement	843	131		0.16	
Medial + trochlear replacement	873	150		0.07	0.01
60° knee flexion					
Physiological	991	111			
Medial replacement	1093	166	110	0.04	
Medial + trochlear replacement	1158	207	117	0.01	0.02
90° knee flexion					
Physiological	1377	163			
Medial replacement	1487	243	107	0.06	
Medial + trochlear replacement	1645	257	118	0.01	0.02

[a] Mean values at specific knee flexion angles, with standard deviation (SD).
[b] Percentage of forces of physiologic knee conditions.
[c] Significance compared to physiologic knee conditions.
[d] Significance compared between the types of replacement.

Transferring these in vitro findings to the in vivo situation, the bi-cruciate retaining modular partial knee arthroplasty require adequate quadriceps muscle strength as under physiological conditions and indicate a better constraint of articulation and restoration of the tibiofemoral flexion/extension kinematics compared to most TKA designs. However, especially the bi-compartimental replacement scenario has a noticeable effect on the maximum quadriceps force required to extend the knee from deep flexion. On the other hand, the close to physiological movement in mid-flexion suggest that patients with a bi-cruciate retaining arthroplasty might have an advantage in knee kinematics compared to those with total knee arthroplasty.

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