UKA closely preserves natural knee kinematics in vitro.

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1 **Abstract:**

3 **Purpose:**

4 It is assumed that unicondylar knee arthroplasty (UKA) features kinematics close to the
5 natural knee. Clinical studies have also shown functional benefits for UKA. There is to
6 date only little biomechanical data to support or explain these findings. The purpose of
7 this study was to investigate whether UKA is able to preserve natural knee kinematics.

8 **Methods:**

9 Six fresh frozen full leg cadaver specimens were prepared to be mounted in a kinematic
10 rig with six degrees of freedom for the knee joint. Three motion patterns were applied
11 before and after medial UKA: passive flexion-extension, open chain extension, and
12 squatting. During the loaded motions, quadriceps and hamstrings muscle forces were
13 applied. Infrared cameras continuously recorded the trajectories of marker frames rigidly
14 attached to femur, tibia and patella. Prior computer tomography allowed identification of
15 coordinate frames of the bones and calculations of anatomical rotations and translations.

16 **Results:**

17 Native kinematics were reproduced after UKA in all the specimens. In the unloaded knee
18 and during open chain extension, femoral rollback patterns after UKA were very close to
19 those in the native knee. During squatting, the medial femoral condyle after UKA tended
20 to be more posteriorly and superiorly with flexion and there was less tibial internal
21 rotation. The tibia was found to be in more valgus after UKA during all motion patterns.

22 **Conclusion:**

23 As ligaments, lateral compartment and patellofemoral anatomy are preserved with UKA
24 the unloaded knee closely resembles native kinematics. The slight kinematic changes that
25 were found under load are probably due to loss of the conforming medial meniscus and to
26 the mismatch in geometry and stiffness introduced by UKA. These patterns resemble
27 those found in knees with significant loss of function of the medial meniscus.

29 **Keywords:** Knee Kinematics, Biomechanics, Unicondylar Knee Arthroplasty, Rollback,
30 UKA

Introduction:

Kinematics is the description of the motion of objects (such as a femur or a tibia) without considering the underlying driving forces behind it. Relative motion between two objects may include translations, rotations or combinations of both.

The knee joint may easily be mistaken for an articulating hinge. A milestone for the understanding of normal knee kinematics in relation to joint geometry was contributed by the group of Pinskerova and co-workers [11,14]. The medial femoral condyle sitting on a concave tibial plateau was described not to move anteroposteriorly with flexion. The lateral femoral condyle tended to roll backwards with flexion on a relatively flat tibia. This combination equates to internal rotation of the tibia around a medial pivot with flexion [11,14]. These findings were confirmed by in vivo three-dimensional weight bearing kinematics using fluoroscopy and computed tomography [16].

The last years have deepened the knowledge of kinematics before and after implantation of total knee arthroplasty (TKA). Although TKA is usually considered an extremely successful operation with excellent long-term survival of implants, there is agreement that native knee kinematics cannot easily be restored [19,27,28].

It is assumed that unicondylar knee arthroplasty (UKA) features kinematics closer to the natural knee than TKA. Functional benefits for UKA over TKA have been described in clinical studies [17,22]. However, there are only few biomechanical studies to support the findings and promising results from clinical series [10].

Suggs et al used a robotic testing system to report on the important role of the ACL for success of UKA [24]. The important role of the ACL was highlighted by another study focusing on ACL reconstruction with UKA using a surgical navigation system with manual stability testing over the knee [7]. Another paper reported on a comparison between passive knee kinematics and stability before and after UKA and TKA, also using a navigation system [5]. They showed that UKA increased varus-valgus stability in extension and restored internal-external rotation of the knee during passive motion. Two studies dealt with in vivo kinematics of UKA in small patient groups [1,3]. Some of the quoted studies have described that there is significant interindividual variation in

1 kinematics following both UKA [1,3] and TKA [28], which might be partially due to the
2 individual anatomy as well as to surgical and implant related factors.

3 Nevertheless, an efficient in vitro testing model is highly desirable to study the influence
4 of different implant types, sizes and position within the same knee as this comparative
5 data cannot easily be achieved via in vivo methods such as those mentioned above. The
6 results from fluoroscopic analyses, particularly single plane, of UKA are limited in
7 accuracy and may not be able to show subtle differences [25] that may exist between
8 UKA and native knee kinematics.

9
10 Therefore, the aim of this project is to determine how knee kinematics before and after
11 medial UKA compare using a validated kinematic rig for cadaver testing under different
12 conditions [27]. A first focus was put on the rollback pattern of the femoral condyles. A
13 second focus was put on the relative orientation of femur and tibia in terms of rotation
14 and valgus angle.

15 It was hypothesized that there would be no major differences in rollback patterns between
16 native knee and after UKA during passive motion. In the loaded knee, it was also
17 hypothesized that there could be some more anteroposterior translation between tibia and
18 femur in the medial compartment and, as a consequence, less tibial internal rotation
19 during squatting, due to a rather flat design of the polyethylene inlay of a UKA in
20 comparison with the stabilizing and centering function of an intact meniscus. Finally, it
21 was hypothesized that all kinematics changes due to UKA would be consistent among
22 specimens.

Material and Methods:

Three left and three right fresh frozen human legs (mean age 74.2 ± 15.2 years, 5 male, 1 female) were disarticulated at the level of the hip. All had functional ligaments and showed no major deformities of the knee.

The experiments were performed according to a previously reported and validated methodology. More detailed information on the underlying science can be retrieved from this publication. For ease of understanding, the main features of the methodology are described hereafter. It was shown that our technique is sufficiently accurate and precise to detect differences in translations (and lengths) and rotations of less than 2 mm and 2 degrees respectively [27].

Three frames with four spherical infrared reflective markers each were rigidly attached to the tibia, the femur, and the patella. Volumetric computed tomography scans on a sixty four-row helical multidetector computed tomography (MDCT) scanner (General Electric Lightspeed VCT; General Electric, Milwaukee, WI, USA) with a slice thickness of 1.25 mm were made of the frozen specimens. For surface reconstruction and identification of osseous landmark the Mimics 13.1 software and its Med-CAD Module interface (Materialise, Haasrode, Leuven, Belgium) were used.

At the day of testing, the thawed knees were transected 32 cm proximal and 28 cm distal to the knee joint. Skin and subcutaneous tissue were removed preserving the articular capsule, ligaments, and tendons. Femur and tibia were embedded in holding containers with polymethylmethacrylate, properly aligned in frontal and sagittal planes. The medial and lateral hamstrings tendons were prepared for attachment to constant load springs (50 N each). The quadriceps tendon was prepared to be clamped to a motor. Springs and motor were fixed to the knee rig in such a way to replicate the moment arms of the muscles. The knees were mounted in a kinematic rig serving as a dynamic knee simulator system (Figure 1).

The system allows for recording motions and forces in a knee joint during squatting and open chain motions. The virtual hip joint can move up and down, thus flexing and extending the knee joint. A virtual ankle joint has all rotational degrees of freedom and can slide medio-laterally. The flexion angle of the knee is the only degree of freedom directly controlled by moving the hip vertically with constant speed. All other degrees of

1 freedom of the knee are left free and allow motion according to individual anatomic
2 features or different testing conditions such as the presence of implants.

3 The specimens were subjected to three motion patterns: passive flexion-extension cycles,
4 an open chain extension with 3 kg of load hung to the distal tibia, and a squat between
5 30° and 120° of flexion with a constant vertical ankle force of 136 N as an equivalent to
6 the ground reaction force (Figure 1).

7 Six MX40+ infrared cameras continuously recorded the trajectories of the markers
8 attached to the tibia, femur, and patella during all activities. The measured marker
9 trajectories were processed in Nexus (Vicon Motion Systems, Oxford, UK) using a
10 specifically designed data processing pipeline to reconstruct and label the trajectories and
11 fill any gaps in the data. Trajectories were filtered using a Woltring filtering routine and
12 the data were exported as C3D files for further processing in Matlab (R2010b, The
13 MathWorks, Natick, MA, USA).

14 Based on the CT, 3D models of tibia and femur were made and bony landmarks
15 identified to determine coordinate frames for the bones. The marker trajectories were
16 transformed to anatomical meaningful rotations and translations as proposed by Grood
17 and Suntay [9]. Tibial axial rotation, tibia position in the coronal plane (varus-valgus),
18 and translations of the femoral condyle centers in anteroposterior (AP) and mediolateral
19 (ML) direction, were obtained as a function of flexion angle (as in Figures 3 – 6).

20 After testing the native knee, a medial UKA was performed and the same tests were
21 redone. Each specimen received an Accuris UKA system (Smith & Nephew, Memphis,
22 TN, USA). The knees were operated using a minimal-invasive medial parapatellar
23 approach using a modular metal-backed tibia with a conventional polyethylene (PE) inlay
24 and an Oxinium femoral component. Bone cuts were made aiming for a perfectly
25 balanced joint with a 9 mm polyethylene (PE) inlay.

26 For each motion test, patterns and ranges of motion of internal tibial rotation, varus angle
27 as well as medial translation was compared before and after UKA. The translations of the
28 femoral condyle centers, projected on the tibial plateau and scaled to the maximum AP
29 width of the native tibial plateau, were also compared before and after UKA. After
30 alignment based on the specimen's tibial anatomical axes, these projected translations
31 were then overlaid on the axial view reconstructed from the tibial plateau of a typical

specimen, with and without UKA. To allow direct comparison, all motion patterns were resampled at intervals of one degree of flexion and within a common range of knee flexion for all knee specimens.

Statistical analysis

Average motion patterns and standard deviations were calculated for each test. A paired student's t-test was used to look for statistical significant differences at each flexion-extension angle. Finally, to analyze and interpret the kinematic consistency between specimens with and without UKA, a similar approach was applied as described earlier [18] using within- and between-group coefficients of multiple correlation (CMC) [15]. This is a statistical measure to quantify the similarity of time-normalized kinematic waveforms. Its dimensionless aspect allows comparison of consistency across the different conditions. In case of poor kinematic consistency of two different test conditions for a given parameter result in between-group CMC having lower values than the corresponding within-group CMC [8,18,23]. The closer the CMC is to zero, the less differences there are between two curves.

Results:

The overall, qualitative impression of knee kinematics (tibial top views Figure 2) shows physiologic femoral rollback patterns of the native knee joint with passive flexion and squatting with substantial interindividual differences in rollback patterns (Figure 2).

These were largely reproduced after UKA.

This first impression was confirmed by a more quantitative analysis of translations (antero-posterior and superior-inferior) and rotations (internal-external and varus-valgus).

The AP translation of the lateral femoral condyle centre (FLCC) was virtually equal before and after UKA as demonstrated by low $CMC_{within}-CMC_{between}$, especially for passive and squat motion (Table 1). On the medial side, the AP translation of the femoral condyle centre (FMCC) after UKA matched the native knee very closely during passive and open chain motion (Figure 3, Table 1). During squatting, the FMCC was positioned consistently more posteriorly with UKA for all specimens and over the whole flexion cycle. $CMC_{within}-CMC_{between}$ clearly shows this effect for the whole curve, while statistical testing on separate flexion angles did not show significance (Figure 3). The open chain testing showed increased stability of the lateral compartment compared to the medial compartment (Figure 3). This is also indicated by the higher CMC_{within} values for FLCC as compared to FMCC, especially for the AP motion (Table 1). Even this somewhat atypical behavior was well reproduced after UKA. The medial femoral rollback ranged from 2.4 mm (during squatting) up to 9.6 mm (during passive motion) and the lateral from 2.4 mm (in open chain) up to 24 mm (during passive motion).

The first column in Table 2 documents the anteroposterior range of motion of the FMCC presented as a fraction of tibial AP width (average and standard deviation for the six specimens) in the three motion cycles. The data shows that the medial compartment becomes increasingly more stable in AP direction under load. This is true for both the native joint and after UKA. However the difference between AP excursion of FMCC before and after UKA is not significant for any of the motion cycles, AP excursion of FMCC systematically demonstrates the largest $CMC_{within}-CMC_{between}$ (Table 1). During squatting FMCC moves anteriorly rather than posteriorly on average as demonstrated by the largest $CMC_{within}-CMC_{between}$ (Table 1).

Figure 4 shows that the FMCC is positioned consistently higher (approximately 3 mm on average) after UKA for all three motion cycles. The difference is statistically significant between 30° and 94° of flexion during passive motion, between 23° and 74° of flexion (with some gaps) during open chain extension and between 34° and 84° of flexion during squat. At 0° flexion and around 90° flexion, height of the FMCC was close to its position in the native knee, showing that the knees were well balanced. On the lateral side, the height of the FLCC before and after UKA remains virtually equal for all motion cycles over the entire flexion range as demonstrated by systematically very low and stable values for $CMC_{within} - CMC_{between}$ over all three motion cycles (Table 1). Differences between native knee and after UKA were most significant in the coronal plane, with knee joints being in less varus following surgery (Figure 6 and Table 1). No significant differences were found before and after UKA in terms of mediolateral translation of the condyle centres (data not shown).

An important focus was put on the rotation of the tibia. Internal tibial rotation showed no differences before and after UKA in the passive motion (Figure 5). “During open chain, between 64° and 77° of flexion, the native knee rotated internally when going into extension from 0° to 1.5° (SD 11°) of internal rotation while the UKA knee remained almost stable between 3.3° and 3.9° (SD 11°) of internal rotation. This difference was statistically significant ($p < 0.05$).

In squatting, there was a tendency for less internal rotation of the tibia after UKA. CMC analysis outcome showed that there was less internal rotation of the tibia following UKA.

1 **Discussion:**

2 The most important finding of this in vitro biomechanical study was that knee kinematics
3 after UKA are indeed close to those of a native knee especially under passive motion.
4 There were however some small but nevertheless significant differences, particularly in
5 terms of inferior-superior position of the medial femoral condyle and varus-valgus tilting
6 of the tibia. There was also less internal tibial rotation and a more posterior position of
7 the femoral medial condyle during squatting. The kinematic findings were highly
8 consistent over the tested specimens.

9
10 Based on the measurements made, the bony geometry of a native medial femoral condyle
11 and tibial plateau seems to be functionally restored by the implant. Due to difficulties in
12 reproducing an accurate contact point with UKA, the PE inlay of most UKA is however
13 relatively flat, differing from the more conforming wedge shape of the medial meniscus.
14 The importance of an intact medial meniscus for knee kinematics has recently been
15 described in a couple of biomechanical studies [2,20].

16 Allaire et al described kinematics in cadaver knees at four knee-flexion angles with a
17 (repaired) posterior root tear of the medial meniscus and after total medial meniscectomy.
18 In their study, root tears and especially a total meniscectomy led to less internal tibial
19 rotation with flexion and a little more varus. They also described a more posterior
20 position of the medial femoral condyle after total meniscectomy. Biomechanics were
21 described to return to normal after surgical repair of posterior root tears [2]. Most of those
22 kinematic effects in a knee with significant damage to the medial meniscus mirror the
23 findings made after UKA. Loss of the stabilizing and conforming function of the medial
24 meniscus results in less internal tibial rotation and more posterior position of the medial
25 femoral condyle with flexion especially under loaded conditions. This becomes evident
26 when high forces are transmitted over the knee joint as in a squat. Under less straining
27 activities such as passive motion of the knee, the remaining preserved structures of the
28 knee provide sufficient stability to maintain normal kinematics after UKA.

29 However, the tests also revealed some other (statistically significant) differences between
30 native and UKA kinematics which cannot be explained solely by the absence of the
31 medial meniscus and loss of conformity of the medial compartment. The femoral medial

condyle center of the knees with UKA was positioned more superiorly and, probably as a consequence of this, the tibia was more in valgus. There are two plausible reasons for this result, one geometrical the other biomechanical.

The curvature of the femoral component (R' in Figure 7a) might be slightly smaller than the curvature of the native femoral condyle (R in Figure 7a) for example. Because the UKA is balanced at 0° and 90° , the femoral component will be flush with the native articular surface at those flexion angles, but it will protrude at all intermediate flexion angles as shown in Figure 7. This effect will lead to a more superior position of FMCC, in any motion cycle whether loaded or unloaded. This is also what was found during the tests. On the other hand, this explanation requires that femoral component curvature is consistently smaller than the native condyle curvature for the six specimens, which seems improbable. One would rather expect the component curvature to be smaller in some specimens and larger in others.

Another important parameter is the mismatch in stiffness which is introduced by UKA between the native articular surface (cartilage 5 to 12 MPa) on the lateral side versus a UHMWPE insert (850 MPa) on an Oxinium femoral component (97 GPa,[12] medially. When the knee is loaded, the medial compartment deforms less after UKA than in the native joint (as illustrated in Table 1 by decreasing pre- vs. post-UKA consistency in terms of the IS motion of FMCC with increased loading, i.e. when going from passive, over open chain to squat motions). This has three consequences. First of all, in loaded conditions, after UKA, the femur will be more superior than in the native knee. Secondly, due to uneven compression on medial and lateral side, the tibia will tilt towards a more valgus position. And finally, the PCL remains more taught than in the loaded native knee and therefore the femoral medial condyle center will be held more posteriorly (Figure 7b).

The results presented here provide added value compared to previous experimental work and gait analysis on UKA [6,13,21]. A cadaver study was also performed by Patil et al. They limited their kinematics analysis and interpretation mainly to the tibial horizontal plane and did not simulate hamstring forces in their setup, with a more limited amount of loading conditions and simulated motion patterns. The implants used were fixed bearing UKA in both studies with relatively flat PE inlays but from different manufacturers. The sensitivity of the setup described in their publication seemed to be relatively low as it

1 surprisingly failed to show differences in kinematics before and after ACL resection with
2 UKA [21] unlike other reports underlining the importance of the ACL for normal knee
3 kinematics with or without UKA [7,19,24,28]. The statement that native knee kinematics
4 and those found after UKA are identical [21] seems to simplify the complex conditions
5 following UKA surgery. The more differentiated approach used in our study, by
6 considering different motion patterns and loading regimens of daily living may therefore
7 more adequately reveal the subtle differences in knee kinematics between native knees
8 and UKA.

9 Suggs et al. used a robotic system to move the knee while applying constant forces on
10 quadriceps and hamstrings. They found no significant effect of UKA on AP position of
11 the medial condyle. UKA mainly led to a clearly more valgus orientation at 0° and 30° of
12 flexion, but not in deeper flexion and it also put the tibia in more external rotation over
13 most of the flexion range. Overall, these findings are similar to ours, although the details
14 of where the differences are statistically significant may differ. The authors also made the
15 link between differences in tibial axial rotation and the loss of the medial meniscus after
16 UKA, but they did not attempt to explain the differences.

17 Becker et al reported that there was no difference in stability between the native and
18 UKA knees in AP translation and rotation in all knee flexion angles independent of fixed
19 and mobile bearings in their robotic cadaver setup [4], which corresponds well with our
20 findings.

21
22 There are limitations to this study. Firstly, access to cadaver specimens is obviously
23 limited and only a few specimens were used. Secondly, testing conditions often include a
24 large number of tests without applying a statistical correction for multiple testing. It was
25 felt that this is however ethically justified. Thirdly, test results may be influenced by the
26 quality of specimen affected by donor factors such as age and underlying diseases as well
27 as timing of storage and treatment during testing and preparation. Finally, the testing
28 conditions could only include a limited range of activities and motions over the knee and
29 will never reflect the full spectrum of living function. To date, the interaction between
30 flexors and extensors over the knee is not fully understood nor described. Victor et al
31 described that loading the knee with hamstrings and quadriceps reduces rotation and

1 translation compared to the passive condition. Lateral hamstring action was described to
2 be more influential on knee kinematics than medial hamstrings action [26]. In our setup
3 we used muscle forces as described in their work as it may approximate physiologic loads
4 on these muscle groups closely. We did not, however, investigate the effect of different
5 hamstring and quadriceps load combinations. We would guess that after UKA the effect
6 of presence of absence of medial hamstring load might become more visible because of
7 the loss of AP stability in the medial compartment.

8
9 It should still be acknowledged that the kinematics after UKA is indeed very close to the
10 native knee. Even for those degrees of freedom where we found statistically significant
11 differences, those differences were small enough to probably go unnoticed by the patient
12 during his daily activities. Indeed, axial rotation discrepancies before and after UKA were
13 at most 3.5° and the difference in varus after UKA was limited to a maximal 3.1° .

1 **Conclusion:**

2 The data presented in this article explain that UKA seems to resemble kinematics of a
3 native knee closely with some features of knees with a resected medial meniscus under
4 loaded conditions.

5 It was shown that the medial femoral condyle center is consistently positioned more
6 superiorly with respect to the native knee, both in unloaded and loaded conditions.

7 Despite these findings, kinematics after UKA were overall very close to the native knee.

8

9

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2 The authors wish to thank all donors, who dedicated their mortal remains to science.
3 Research, such as in this project, to deepen the understanding about function of the
4 human body would be impossible without such act of selflessness.
5 Rest blinded for review process.

6

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Captions to Figures:

Figure 1:

Setup photographs of open chain testing (left) and a squat (right): R: Rig; F: Femur; T: Tibia; *: frames with infrared reflective markers; H: hamstrings with load springs; Q: clamped quadriceps tendon.

Figure 2:

The rollback behavior before and after UKA for the tested motions for all specimens. Rollback is displayed from the projected displacement of the femoral condyle centers, overlaid on the tibial plateau, and calculated in relation to the tibial plateau AP size of the corresponding specimen. Tibial images with and without UKA are generic depictions created from a native knee and only serve as illustrations. In these plots, the medial and lateral femoral condyles centers (dots) are linked by a line whose color reflects the corresponding flexion angle, thus the progressive color gradient from full extension (darker color) to full flexion (lighter color).

Figure 3:

Translations of the Medial femoral condyle centers (FMCC) are expressed as relative to the AP size of the tibial plateau. Values for each knee were calculated using the respective specimen's dimensions, then averaged in the following graphs. Standard deviations are shown with shaded areas of corresponding color. Regions of non-significant differences are shown with faded out colors.

Figure 4:

Average inferior-superior translation between femur and tibia as a function of flexion before and after UKA for the three motion tests. Standard deviations are shown with shaded areas of corresponding color. Regions of non-significant differences are shown with faded out colors.

Figure 5:

Average internal rotation of the tibia as a function of flexion before and after UKA for the three motion tests: Standard deviations are shown with shaded areas of corresponding color. Regions of non-significant differences are shown with faded out colors.

Figure 6:

Average varus angle between femur and tibia as a function of flexion before and after UKA for the three motion tests. Standard deviations are shown with shaded areas of corresponding color. Regions of non-significant differences are shown with faded out colors.

Figure 7:

Schematic view of the effect of geometrical (a) and stiffness (b) mismatch between the native and the replaced medial compartment on height and AP position of the FMCC.

Table 1: Overview of within-group CMC, between-group CMC values and the absolute difference between both for all analysed components of pre- and post-UKA knee kinematics.

Table 2: Average, minimum and maximum values for AP excursion of FMCC, varus tibial tilting excursion and internal tibial rotation excursion. Significant differences ($p < 0.05$) are indicated with an asterisk.

Figure 1
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Figure 2
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Figure 3
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Figure 5
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Figure 6
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Figure 7
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Potential effect of overstuffing the femoral articular surface

Anterior and inferior femoral shift as a consequence of load

Table 1:

		CMC_{within}	CMC_{between}	CMC_{within} - CMC_{between}
Passive	Varus/valgus	<i>0.711</i>	<i>0.697</i>	0.014
	Tibial rotation	<i>0.593</i>	<i>0.579</i>	0.015
	AP motion of FLCC	<i>0.409</i>	<i>0.402</i>	0.007
	IS motion of FLCC	<i>0.818</i>	<i>0.809</i>	0.009
	AP motion of FMCC	<i>0.535</i>	<i>0.497</i>	0.038
	IS motion of FMCC	<i>0.841</i>	<i>0.829</i>	0.013
Open Chain	Varus/valgus	<i>0.928</i>	<i>0.858</i>	0.070
	Tibial rotation	<i>0.903</i>	<i>0.890</i>	0.014
	AP motion of FLCC	<i>0.953</i>	<i>0.898</i>	0.055
	IS motion of FLCC	<i>0.976</i>	<i>0.971</i>	0.006
	AP motion of FMCC	<i>0.892</i>	<i>0.850</i>	0.042
	IS motion of FMCC	<i>0.963</i>	<i>0.926</i>	0.037
Squat	Varus/valgus	<i>0.887</i>	<i>0.865</i>	0.022
	Tibial rotation	<i>0.862</i>	<i>0.803</i>	0.059
	AP motion of FLCC	<i>0.818</i>	<i>0.809</i>	0.009
	IS motion of FLCC	<i>0.876</i>	<i>0.862</i>	0.014
	AP motion of FMCC	<i>0.846</i>	<i>0.758</i>	0.088
	IS motion of FMCC	<i>0.858</i>	<i>0.805</i>	0.053

Table 2:

		AP motion of FMCC	Varus tilt	Internal rotation
Passive	Native	-0.16 (-0.46 – 0.04)	3.2° (-7.3° – 8.1°)	16° (9.6° – 24.3°)
	UKA	-0.16 (-0.45 – -0.06)	3.0° (-5.1° – 12.4°)	13° (9° – 19°)
Open Chain	Native	-0.11 (-0.16 – -0.06)	0.8° (-2.4° – 4.2°)	-6.6° (-14.7° – -1.1°)*
	UKA	-0.07 (-0.18 – -0.01)	1.8° (-3.8° – 6.1°)	-1° (-12.1° – 6.2°)*
Squat	Native	-0.04 (-0.15 – 0.16)	-1.7° (-5.2° – 4.5°)	6.7° (4° – 16°)
	UKA	0.04 (-0.14 – 0.25)	1.6° (-6.8° – 8°)	7.9° (3.9° – 19.9°)