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Are instrumented knee forces representative of a larger population of cruciate-retaining total knee replacements?

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Abstract

Background—It is not known if the loads and motions reported for instrumented knees are generalizable to a larger population of total knee replacement (TKR) patients. The purpose of this study is to: 1) report axial implant force data for chair and stair activities for a population of cruciate-retaining TKR patients and 2) compare the population forces to those measured with instrumented TKRs.

Methods—Twenty-three subjects with a cruciate-retaining TKR underwent motion analysis during stair ascending, stair descending, chair sitting, and chair rising activities after informed consent in this IRB approved study. Axial TKR forces were calculated using a previously validated computational model. Differences between the mean and variability of population instrumented TKR peak forces and force impulses were tested using t-tests and Levene's test.

Results—Peak axial forces were 3.06, 2.74, 2.65, and 2.60 kN for stair ascent, stair descent, chair rising, and chair sitting, respectively. Force impulses were 123.3, 123.4, 153.5, and 154.0 kN*% activity cycle for stair ascent, stair descent, chair sitting, and chair rising, respectively. Population TKR and instrumented TKR peak forces were different for stair ascent ($p=0.03$) and stair descent ($p=0.03$) in the second half of the activity cycles. The variability of the peak forces and impulses were not different ($p=0.106$ to $p=0.99$)

Conclusion—The forces and motions presented in this study represent cruciate-retaining TKR patients and could be used for displacement-driven knee wear testing. The forces are similar to those in the literature from instrumented prostheses of an ultra-congruent cruciate-sacrificing TKR.

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Keywords

total knee replacement; axial force; musculoskeletal model; stair and chair activities

Introduction

Total knee replacement (TKR) is a widely successful procedure for the treatment of end-stage osteoarthritis. Rates of implantation are increasing dramatically [1–4], and to such an extent that even small rates of failure can become large numbers of revision surgeries [5]. Thus, it is important to continually improve the longevity of TKRs.

An important tool for preclinical testing of TKR design enhancements is total knee wear simulation. Wear simulators require knowledge of the loads and motions experienced by the knee during daily living. Recently, with the availability of data from instrumented total knees, there has been a lot of interest surrounding knee simulator standards and their physiological relevance. The International Organization for Standardization (ISO) released an update to the displacement standard for knee wear testing [6]. In addition, the American Society for Testing and Materials (ASTM) initiated a workgroup to evaluate how multiple activities should be incorporated into a new wear testing standard for TKRs [7] and published a guide of TKR loading profiles under different activities for force control knee wear testing [8]. Recommendations for standardized testing waveforms for multiple activities were recently made from instrumented knee loads and motions measured during level walking, stair ascending and descending, standing up and sitting down, jogging on a treadmill, knee bend, and one-legged stance [9]. The data has been used in wear simulations [10–12]. Incorporating multiple activities into knee wear testing is essential, as a study of TKR patient activity levels and wear has shown that nearly 30% of wear may be attributed to chair and stair activities [13].

Motion waveforms for primary walking kinematics (knee flexion angle) are available from two groups, Bergmann et al. and D’Lima et al [14,15]. Knee load waveforms recommended for knee wear testing for walking, stair, and chair activities have been reported as the average of eight patients by one group [9]. The instrumented implant used by Bergmann et al. is an ultra-congruent cruciate-sacrificing design [14]. It is not known if the loads and motions reported for instrumented knees are generalizable to a larger population of TKR patients, or what effect using inputs that are not population specific for testing of a particular TKR design will have on wear testing.

The purpose of this study is to: 1) to report axial implant force data for chair and stair activities for a population of TKR patients implanted with a modern cruciate-retaining implant design and 2) compare the population forces to patient-specific forces measured with instrumented total knee replacements. This study will help determine if force waveform recommendations are representative of a larger population of TKR patients providing insight into considerations for wear simulation, numerical modeling, and implant design.

Material and methods

Twenty-three subjects (10M/13F) with a NexGen cruciate-retaining total knee system (Zimmer Inc., Warsaw, IN) were tested after informed consent in this IRB approved study. The average age was 61.7 ± 7.0 years old, BMI was 30.9 ± 7.7 , and measurements were taken at least 12 months post-op (41.8 ± 29.7). All measurements reported as mean $\pm$ standard deviation (SD). All participants were able to function independently without assistive devices and were active in the home and/or workplace.

Motion testing took place in the Rush Motion Analysis Laboratory. A four-camera optoelectronic system (Qualysis, Gothenburg, Sweden) tracked marker motion. Ground reaction forces were recorded using an embedded force plate (Bertec, Columbus, OH). Knee flexion-extension angles, anterior-posterior femoral translation, and internal-external femoral rotation angles were measured using the point cluster technique and marker set [16,17]. Inverse dynamics were used to calculate the external moments and intersegmental forces at the knee from the kinematics and ground reaction forces [18].

During motion testing, subjects performed three trials each of stair ascent, stair descent, chair sitting, and chair rising activities. For stair ascent and descent, a three-step staircase unit (8.5 in step height) was positioned with the middle step on a force plate. Subjects took two normal walking steps to approach the staircase before ascending and were instructed to continue walking away from the staircase after descending. Kinematics were recorded for all steps; external moments were calculated from steps one to two for stair ascent and from steps two to one for stair descent. Subjects did not use a handrail for either stair activity. For chair rising and sitting, an armless chair with adjustable seat height was strategically placed at the perimeter of the force plate so that only the tested foot was detected. The seat height was adjusted so that the patient's knee was flexed to 90 degrees when in the seated position.

For stair ascent and stair descent activities, the activity cycle started when force was first registered on the force plate. The end of the activity cycle was defined at the point of maximum knee extension after the patient had placed their foot on the subsequent step, indicating the end of swing phase. For chair rising and sitting, the patient first lifted their foot off the force plate and then placed it back down before starting the activity. This allowed for the start of the activity cycle to be defined when force was first registered on the force plate. The end of the chair rising and sitting activity was defined when the knee reached maximum extension and flexion, respectively.

To calculate axial forces during the activities, we used a previously developed numerical model [19] which calculates a solution space of three-dimensional contact forces for both medial and lateral compartments of the tibial plateau. Quasi-static equilibrium is solved for by balancing internal and external forces and moments at 100 instances of an activity cycle. Internal moments and forces are transferred across the knee joint via contact forces, muscle forces, and soft tissue forces, while external forces and moments are transferred to the body via ground reaction forces. Inputs to the model include subject kinematics and kinetics, maximum physiological lower limb muscle forces from a musculoskeletal model [20,21], and a contact path between the tibia and femur. The relationship between medial and lateral

knee forces was assumed to vary with the external knee adduction moment [22,23]. The initial relationship between the medial and lateral knee forces and external knee adduction moment was assumed to be the same as what we have previously reported for level walking [22]. If forces were not obtained by the knee model, the relationship was iteratively varied until force solutions were obtained. Knee kinematics were used to calculate the tibiofemoral contact path with previously developed software [24]. The contribution of passive structures to moment equilibrium was included for the frontal and transverse planes if the sum of the muscle moments was not great enough to balance the external moments. The parametric model calculates a range of contact force solutions resulting from variation of relative muscle activation levels. The model has been extensively validated against instrumented total knee axial forces in a blinded and unblinded fashion [22,25].

Axial force profiles for each activity will be reported as the mean of all solutions at each of 100 instances of the activity cycle, averaged across all available trials for each activity. Results for the entire population of TKR patients will be reported as the mean $\pm$ 1 SD of all subject means. Peak forces (two for each half of the activity cycle for stair activities and one for chair activities) and force impulse will be calculated for each activity. Force impulse is a single value for each activity equal to the area under the loading curve during an activity cycle. To compare our results to instrumented TKRs, individual subject curves were downloaded from the knee joint test loads found online at www.orthoload.com [9,14]. All available trials were used (8 for stair activities, 7 for chair activities). Impulse and peak forces between the model and instrumented TKR will be compared using independent samples t-tests. Variability of the impulse and peak forces will be compared using Levene's test for Equality of Variances.

Results

Of the 23 subjects, data collection of stair activities was successful for 22 subjects for stair ascent, and all 23 subjects for stair descent. Data was successfully collected for 21 subjects for chair activities. Three subjects were excluded because their BMI was greater than 40. Higher BMIs result in higher measurement error associated with skin markers; a cutoff BMI of 40 was chosen as it is the definition of class 3 obesity [26]. The knee model was not able to calculate force curves for any of the activities for one individual. For another, the model was not able to calculate forces during chair sitting. In all, knee joint axial forces were obtained for 18 subjects for stair ascent, 19 subjects for stair descent, 17 subjects for chair rising, and 16 subjects for chair sitting.

Knee flexion angles, femoral internal rotation angles, and femoral anterior-posterior displacement during all activities for all processed trials can be seen in Figure 1. Flexion, adduction, and internal rotation external moments for the same are depicted in Figure 2. Moments were normalized to percent body weight times height (%BW*height). Activity forces were reported in kilonewtons (kN) for comparison to the OrthoLoad data recommended for knee simulator wear testing. Force curves were time-normalized, therefore force impulse was reported as a factor of percent activity cycle (kN*% activity cycle).

Stair ascending and descending had a characteristic waveform with two local maxima, one in the first half of the activity cycle and one in the second half of the activity cycle (Figure 3A–B). Stair ascent peak forces were 3.06 ± 0.67 kN in the first half and 1.97 ± 0.62 kN in the second half of the activity cycle (Table 1). Stair descent peak forces were 2.74 ± 0.70 kN in the first half and 2.31 ± 0.64 kN in the second half of the activity cycle. Force impulses over the entire activity were 123.3 ± 24.2 kN*% activity cycle and 123.4 ± 33.8 kN*% activity cycle for stair ascent and descent respectively (Table 2).

The initial medial-lateral force distribution ratio (originally tuned for level walking) did not result in solutions for chair activities and was interactively changed to obtain more solutions. The final medial-lateral weight distribution ratio was a constant 50%—meaning an equal percentage of load passed medially and laterally through the knee joint. For chair rising and sitting, the peak forces were 2.65 ± 0.84 kN and 2.60 ± 0.77 kN (Figure 3C–D and Table 1) and the force impulses were 153.5 ± 41.2 kN*% activity cycle and 154.0 ± 36.0 kN*% activity cycle respectively (Table 2).

Mean values of peak forces and force impulses were comparable between study population TKR and instrumented (OrthoLoad) data (Tables 1 and 2). The only significant differences between the study population TKR peak forces and impulses and the instrumented TKR measured peak forces and impulses were for the second peak stair ascent force ($p=0.03$) and the second peak stair descent force ($p=0.03$). There were no differences in the variability of the peak forces and impulses between the study population TKR and instrumented TKR measurements as measured by Levene's test ($p=0.106$ to $p=0.99$).

Discussion

The first purpose of this study was to report axial implant forces for chair and stair activities for a population of TKR patients implanted with a modern cruciate-retaining implant design. Axial forces reached an average of 3.1 kN and 2.7 kN during stair ascent and descent, and an average of 2.7 kN and 2.6 kN for chair rising and sitting, respectively. Normalized to body weight, the average peak forces are 3.3 ± 0.5 BW, 3.0 ± 0.4 BW, 2.9 ± 0.8 BW, 2.9 ± 0.6 BW during stair ascent, stair descent, chair sitting, and chair rising, respectively.

The second purpose of this study was to compare the study population TKR kinetics to instrumented TKR kinetics. The only statistically significant differences were that the peak axial forces in the second half of the activity cycle of stair ascent and descent were lower for the study population TKRs than the instrumented TKRs. Levene's test did not reveal any differences in the variability of the force waveforms between the study population TKRs and instrumented TKRs. Although there were no differences in most peak forces, force variability, and force impulses, the force waveforms appeared visually different between the two prosthesis designs for stair descent and chair sitting. For stair descent, the calculated cruciate-retaining TKR forces were lower than the instrumented knee ultra-congruent cruciate-sacrificing TKR measured forces throughout the activity cycle. Differences in implant design have previously been shown to affect stair climbing kinematics [27], which could in turn result in different forces. In addition, force differences may be due to the test protocol. The subjects in our study did not use handrails to descend stairs where the

instrumented subjects may have. Differences in the test staircase may also have affected the axial force output, as the step height in this study is about 0.5 inches taller than the step height of the staircase used in the Bergmann study. For chair sitting, the calculated cruciate-retaining TKR forces were greater than instrumented knee ultra-congruent cruciate-sacrificing TKR measured forces during the first half of the activity cycle. There were a few differences in the chair setup between our laboratory and the OrthoLoad data. Our chair testing setup used a standardized chair with a subject-specific chair height and no armrests, while the videos from OrthoLoad testing suggest that various chairs were used for testing with the instrumented TKR.

This study had several limitations. First, we could not obtain solutions for axial forces with the computational model for a few trials. The model failed to calculate forces during chair sitting for one subject and during all activities for another subject. An examination of the knee kinematics for these trials revealed large offsets for anterior-posterior displacement and internal-external rotation. This is likely an artifact due to errors associated with skin movement in marker-based gait analysis. Second, the model relies on knowledge of the ratio between medial and lateral axial forces to the total axial force passing through the knee. A prior validation study showed that the ratio between medial and lateral axial forces significantly affected force prediction results. The study also identified strategies for proper tuning of this parameter which we used in this study [28]. Although previous investigations have found that the medial-lateral force ratio is correlated with the external knee adduction moment [23,29], it is uncertain as to whether this relationship holds true for all activities and whether it may be influenced by patient characteristics such as knee alignment [30,31]. Specifically, for chair sitting and rising the medial-lateral force ratio of 50%, indicating equal force transfer between the medial and lateral sides of the tibial plateau, was necessary to obtain forces with the computational model. A previous study of instrument prostheses, although with a more constrained prosthesis design, also found the percentage of force transmitted medially through the knee to be lower for chair activities than stair and walking activities [32,31].

Recently, Rasnick et al. predicted the knee joint stair ascent forces for five patients implanted with a posterior stabilized TKR [33]. Average resultant peak contact forces were 2.8 BW and 3.9 BW for the peaks in the first and second half of the stair ascent activity cycle. The value of the first peak compares well to our study but the value of the second is considerably higher than our findings and of that found in instrumented TKRs. Additionally, the force waveform had the opposite pattern where the second peak is higher than the first peak. Our peak forces in this study also compare well to values previously reported for the eKnee instrumented TKR, a cruciate-retaining design. Peak eKnee forces have been reported as ~3.6 BW, 3.5 BW, 1.9 BW, and 2.5 BW for stair ascent, stair descent, chair sitting, and chair rising, respectively [34].

Conclusions

We present forces and motions for stair and chair activities representative of a population of TKR patients implanted with a cruciate-retaining prosthesis design which could be used for displacement-driven knee wear testing. The forces are similar to those in the literature from

instrumented prostheses of an ultra-congruent cruciate-sacrificing TKR. Although the variability in this patient cohort was not greater than that of the instrumented knee patients, solely using the recommended waveforms as force inputs may not represent wear of this implant type due to differences in knee kinematics, especially secondary kinematics (anterior-posterior translation and internal-external rotation). Although studies have investigated the difference in wear due to prosthesis design, it is unknown how using inputs for wear testing from populations of TKR patients implanted with different prostheses will affect the outcome of wear tests. However since wear tested simulator components continue to not be representative of wear measured on retrieved TKRs [35,36], work should continue towards defining accurate wear simulation inputs generalizable to multiple TKR patient cohorts who participate in a variety of daily activities.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

The knee flexion angle, femoral internal rotation angle, and femoral anterior displacement for (A) stair ascent (n=18), (B) stair descent (n=19), (C) rising from a chair (n=17), and (D) sitting down on a chair (n=16). Data is reported as mean $\pm$ 1 standard deviation, time normalized to percent activity cycle. Coordinate system is as follows: negative extension positive flexion (-Ext/+Flex), negative external rotation of the femur positive internal rotation of the femur (-ER/+IR), and negative posterior displacement of the femur positive displacement of the femur (-Post/+Ant Displ).

Figure 2.

The average knee flexion, adduction, and internal rotation moments normalized to percent body weight times height (%BW*HT) and time normalized to percent activity cycle for (A) stair ascent, (B) stair descent, (C) rising from a chair, and (D) sitting down on a chair. Data is reported as mean $\pm$ 1 standard deviation. Coordinate system is as follows: negative extension positive flexion (-Ext/+Flex), negative adduction positive abduction (-Add/+Abd), and negative external rotation positive internal rotation (-ER/+IR) moments about the knee.

Figure 3. Axial loading during (A) stair ascent, (B) stair descent, (C) chair rising, and (D) chair sitting. Data is reported as mean $\pm$ 1 standard deviation time normalized to percent activity cycle. Knee model output is compared to measured forces from instrumented knee implants reported by Bergmann et al. 2014 which can be found at www.orthoload.com.

Peak forces for the study population TKRs and instrumented TKRs. Results are reported as mean $\pm$ 1 standard deviation. Independent samples t-tests were used to determine significance of differences in the means of the peak forces. Levene's inequality of variances test was used to test for differences in the variability of the peak forces.

Table 1

Activity	Study Population TKRs (kN)	Instrumented TKRs (kN)	t test p-value	Inequality of Variances f	p-value
Chair Rising	2.65 $\pm$ 0.84	2.48 $\pm$ 0.62	0.663	0.819	0.381
Chair Sitting	2.60 $\pm$ 0.77	2.44 $\pm$ 0.71	0.655	0	0.99
Stair Ascent					
<i>Peak 1</i>	3.06 $\pm$ 0.67	2.77 $\pm$ 0.57	0.358	0.032	0.861
<i>Peak 2</i>	1.97 $\pm$ 0.62	2.70 $\pm$ 0.58	0.030	0.101	0.755
Stair Descent					
<i>Peak 1</i>	2.74 $\pm$ 0.70	2.89 $\pm$ 0.62	0.626	0.638	0.438
<i>Peak 2</i>	2.31 $\pm$ 0.64	3.01 $\pm$ 0.59	0.030	0.434	0.520

Force impulse for study population TKRs and instrumented TKRs. Results are reported as mean $\pm$ 1 standard deviation. Independent samples t-tests were used to determine significance of differences in the means of force impulses. Levene's inequality of variances tests were used to test for differences in the variability of force impulses.

Table 2

Activity	Study Population TKRs (kN*% Activity Cycle)	Instrumented TKRs (kN*% Activity Cycle)	t test		Inequality of Variances	
			p-value	f	p-value	
Chair Rising	153.5 $\pm$ 41.2	137.4 $\pm$ 25.3	0.386	3.02	0.106	
Chair Sitting	154.0 $\pm$ 36.0	129.3 $\pm$ 26.3	0.137	0.382	0.545	
Stair Ascent	123.3 $\pm$ 24.2	131.8 $\pm$ 22.5	0.452	0.073	0.791	
Stair Descent	123.4 $\pm$ 33.8	146.6 $\pm$ 23.7	0.116	0.6	0.449	