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The Impact of Hip Implant Alignment on Muscle and Joint Loading during Dynamic Activities

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Abstract

Background—Component alignment is an important consideration in total hip arthroplasty. The impact of changes in alignment on muscle forces and joint contact forces during dynamic tasks are not well understood, and have the potential to influence surgical decision making. The objectives of this study were to assess the impact of femoral head/stem and cup component placement on hip muscle and joint contact forces during tasks of daily living and to identify which alignment parameters have the greatest impact on joint loading.

Methods—Using a series of strength-calibrated, subject-specific musculoskeletal models of patients performing gait, sit-to-stand and step down tasks, component alignments were perturbed and joint contact and muscle forces evaluated.

Findings—Based on the range of alignments reported clinically, variation in head/stem anteversion-retroversion had the largest impact of any degree of freedom throughout all three tasks; average contact forces 413.5 (319.1) N during gait, 262.7 (256.4) N during sit to stand, and 572.7 (228.1) N during the step down task. The sensitivity of contact force to anteversion-retroversion of the head/stem was 31.5 N/deg for gait, which was similar in magnitude to anterior-posterior position of the cup (34.6 N/m for gait). Additionally, superior-inferior cup alignment resulted in 16.4 (4.9)° of variation in the direction of the hip joint contact force across the three tasks, with the most inferior cup placements moving the force vector towards the cup equator at the point of peak joint contact force.

Interpretation—A quantitative understanding of the impact and potential tradeoffs when altering component alignment is valuable in supporting surgical decision making.

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Keywords

hip arthroplasty; joint mechanics; muscle force; implant alignment; musculoskeletal modeling

1. INTRODUCTION

In total hip arthroplasty (THA), alignment of the femoral components and acetabular cup influences the mechanics of the joint, including the functional range of motion of hip articulation and the joint positions in which impingement can and cannot occur (Patel et al., 2010). Further, component alignment has been associated with poor clinical outcomes such as impingement (Renkawitz et al., 2012), dislocation (Higa et al., 2011), increased liner wear and fracture, osteolysis (Kennedy et al., 1998), edge loading (Kwon et al., 2012) and increased metal ion in the blood (Harris, 2012). Alignment includes implantation parameters of the three dimensional position of the stem and cup relative to the femur and pelvis, respectively, as well as orientation in version and abduction (Widmer and Zurfluh, 2004). With surgical aims of improving joint mobility and restoring the ability to safely perform activities of daily living (Vissers et al., 2011), the mechanics of the joint are a critical consideration, with many studies assessing the effects of component alignment on range of motion and the likelihood of impingement (Petrella et al., 2009). Component alignment also directly affects hip joint loading during typical activities of daily living, with changes in femoral anteversion capable of resulting in 30% increases in hip joint contact force (Heller et al., 2001).

Native hip anatomy following THA is typically not fully restored compared to the contralateral hip (Tsai et al., 2014). A wide range of variation has been reported in the placement of the femoral component in particular. Wines and McNicol (Wines and McNicol, 2006) used CT measurements to show femoral anteversion averaged 16.8 (11.1)° with a range from -15.0° retroversion to 45.0° anteversion. Higher amounts of femoral version following THA have been associated with pain and decreased quality of life (Liebs et al., 2014). In addition, vertical elevation of the femoral component, which results in leg length changes, has been shown to vary between -2.5 to 12.6 mm relative to the contralateral side (Tsai et al., 2014). Placement of the acetabular component has shown smaller amounts of variability in comparison to the femoral component, but cup placement has a direct influence on the location of the center of rotation of the hip and can also result in leg length changes. Cup position variability has been shown to be similar in each degree of freedom and can vary by up to approximately 10 mm (Tsai et al., 2014). Recently, computer-aided surgery systems that improve the placement of the cup relative to native hip geometry have become available for use during THA (Renkawitz et al., 2009). However, these systems have not been widely adopted because they can add considerably to both cost and time of the surgery and, currently, there is no assistive surgical technology for the placement of the stem.

The influence of component position on joint loading can be assessed non-invasively with the use of the musculoskeletal modeling software platforms, such as OpenSim (Delp et al., 2007) or Anybody (AnyBody Technology, Aalborg, Denmark). Musculoskeletal modeling is used to calculate joint kinematics and moments, as well as intersegmental joint loads and

muscle forces. Musculoskeletal simulation offers valuable data to clinicians and researchers assessing pathological conditions and understanding human movement. Simulation of human movement has significantly impacted approaches to clinical treatment of osteoarthritis (Fregly et al., 2007) and total joint replacement (Gaffney et al., 2015; Navacchia et al., 2016), as well as basic science related to the understanding of movement progression and control during dynamic tasks (Anderson et al., 2004; Neptune et al., 2009; Zajac et al., 2002). There have been a number of impactful innovations in simulation methods from sophisticated subject-specific models with highly accurate anatomic detail (Arnold et al., 2010), to creation of efficient forward dynamics simulations using computed muscle control (Thelen and Anderson, 2006) that make it possible to address variation in component positioning across multiple patients.

Musculoskeletal modeling has been previously used to assess how variation in the location of the hip joint center can result in changes to the lines of action and moment generating capabilities of the hip muscles (Delp and Maloney, 1993). However, the extent of the impact of positioning THA components on hip joint loading during daily activities is not known. Quantifying relationships between component alignment, muscle forces and joint loading can support surgical practice in assessing the tradeoffs that exist in joint mechanics and loading when altering component placement for considerations of bone quality or fixation. Accordingly, the objectives of this study were to assess the impact of femoral head/stem and cup component alignment on hip joint contact forces and muscle forces during tasks of daily living and to identify which alignment parameters have the greatest impact on hip joint loading.

2. METHODS

2.1 Patient Data Collection

A cohort of five patients who had undergone THA to treat end stage osteoarthritis (2 M: 60.5 (9.2) yrs, 94.5 (9.9) kg; 3 F: 61.3 (9.1) yrs, 72.2 (8.4) kg) performed through a posterolateral approach, were selected from a larger prospective study that included 26 patients. Patients were eligible if they were between the ages of 45 and 80 years, had no history of uncontrolled hypertension or diabetes, body mass index $<40 \text{ kg/m}^2$, no additional orthopaedic pathology, or neurologic disorders that impaired daily function. Each patient provided written, informed consent and participated in a laboratory testing session that was approved by the Colorado Multiple Institutional Review Board. The laboratory testing session was performed a minimum of 10 weeks post-operatively and averaged 11.7 ± 1.4 weeks for these five patients.

Patients were fitted with 32 reflective markers used to define anatomical landmarks for 3D motion capture. Following a standing static trial, patients were instructed to perform three activities of daily living. Activities consisted of gait at a self-selected pace, a sit-to-stand task in which patients stood from a chair 43 cm in height and achieved a fully upright posture, and a step down task from a height of 20 cm. Each task was performed onto a Bertec (Columbus, OH, USA) force platform embedded in the floor with their surgical limb while force data was collected at 2000 Hz and an 8 camera Vicon motion capture system (Centennial, CO, USA) collected at 100 Hz (Judd et al., 2016).

Isometric strength of the hip flexors, extensors, and abductors, as well as the knee flexors and extensors, was assessed using an electromechanical dynamometer (HUMAC NORM, CSMI Solutions, Stoughton, MA, USA) connected to a Biopac Data Acquisition System (Biodex Medical Systems, Inc., Shirley, NY, USA) running AcqKnowledge software (Biodex Medical Systems, Inc., Shirley, NY, USA). Strength was measured in the affected limb. For hip flexor and extensor strength assessment, participants were positioned supine with the hip flexed to 40°. Hip abductor strength was measured while participants were positioned side-lying with 0° of hip flexion/extension and 0° of hip abduction/adduction. Knee extensor and flexor strength was measured in a seated position with a shoulder harness and waist strap for stabilization; patients were placed in 85° of hip flexion and 60° of knee flexion for the measurement (Judd et al., 2016). Patients underwent 3 trials of maximal effort contractions and the highest of the three was used in further analysis.

2.2 Musculoskeletal modeling

Patient-specific lower extremity muscle strength calibration was performed using a musculoskeletal model that included detailed hip musculature (Shelburne et al., 2010) to be used in simulations of each task of daily living. Muscles and wrapping were added to a generic musculoskeletal model with 10 rigid bodies, 23 degrees of freedom, and 92 actuators (Arnold et al., 2000; Arnold and Delp, 2005; Delp et al., 2007, 1990). Analysis focused on muscles surrounding the hip that included: gluteus medius, gluteus maximus, gluteus minimus, rectus femoris, semimembranosus, semitendinosus, and tensor fasciae latae. The dimensions of each segment in the model were scaled so that the distances between the virtual markers on the model matched the distances between the experimental markers. The dimensions of the body segments, mass properties (mass and inertia tensor) of the segments, and the elements attached to the body segments, such as muscle actuators and wrapping objects were all scaled. In addition, for each patient-specific model, moment arms and maximum isometric torques were calculated for flexion/extension, internal/external rotation, and adduction/abduction of the hip. Calibration of muscle maximum isometric parameters was performed by increasing or decreasing the maximum force for each muscle to minimize differences between model-predicted and measured preoperative maximum isometric joint torques in hip flexion, extension, and abduction, as well as knee flexion and extension. Muscles in each group were all scaled by the same factor to maintain the strength ratios between muscles of the same group.

Baseline simulations for each patient performing the three activities were constructed using the patient-specific scaled model and corresponding measured kinematics and ground reaction forces to predict hip joint contact forces (JCFs) and muscle forces using static optimization, in which the sum of muscle activation squared was minimized (Anderson and Pandy, 2001). Baseline refers to the prescribed neutral hip implant alignment that was created by exactly replicating the joint center of rotation in the patient-specific scaled model with the cup placed in 15° of anteversion and 40° of inclination and the stem in 10° of anteversion. Results from the baseline simulations were compared to data collected from patients implanted with telemetric hip implants performing the same three tasks (Bergmann et al., 2010).

Two types of evaluation were performed to assess the impact of changes in *femoral head/stem alignment* and *cup alignment* from the neutral alignment on the JCFs and muscle forces. Input ranges for each component placement degree of freedom were based on the 3 SD range in component placement relative to native geometry reported by Tsai et al. (levels reported in Fig. 1)(Tsai et al., 2014). Femoral alignment simulations considered differences in the position and orientation of the head/stem relative to the femur. In these analyses, the location of the cup center was maintained, while the femur position and associated muscle attachments were either translated (1 mm increments) or rotated (1° increments). Cup alignment simulations considered differences in cup position, but did not vary orientation. The cup center was translated in 1 mm increments along each degree of freedom. In these analyses, the thigh (including the femoral head/stem), shank and foot segment positions and muscle attachments were translated relative to the pelvis to the same extent as the cup.

2.3 Data Analysis

For each patient, JCFs and individual muscles forces were predicted through the task cycle with perturbations to each alignment variable. To identify the contributions of specific alignment degrees of freedom to muscle and joint loading, a range was defined which represented the maximum to minimum force, corresponding to the limits of the perturbation, averaged at 100 points through the task cycle. The mean and standard deviation of the range for JCF and muscle force were averaged for the 5 patients and reported for perturbations in each alignment variable. Muscle forces were evaluated both individually and as an overall average; the latter was used when considering relative contributions of alignment perturbations. For all three tasks, only the portion of the movement cycle when the foot of the surgical limb was in contact with the ground was considered as this represented higher loading than when the foot is not in contact with the ground and facilitated comparisons between activities. For the sit-to-stand task, the movement cycle began at the point of zero chair contact. The sensitivity of hip joint contact force and muscle force was calculated using the change in force per change in alignment for each degree of freedom. The impact of positioning was also assessed on the direction of the hip joint contact force by calculating the average range of the direction of the vector in three planes. For visualization purposes, the vector was transformed to the center of a cup and overlaid on CT data for a representative patient with THA. The direction of the vector relative to the cup edge can provide insight into the potential for edge loading or dislocation.

3. RESULTS

After calibration to the isometric strength data, patient-specific models with corresponding applied kinematics established the baseline JCF and muscle forces for each patient. In general, the patient JCF profiles showed similar trends (Fig. 2), although inter-subject variability results in differences in the force magnitude. For reference, the data for the five patients in the current study were comparable to the cohort of patients implanted with telemetric implants (Bergmann et al., 2010) for each of the three tasks with good agreement (Fig. 2).

Ranges of JCFs and muscle forces resulting from perturbations in the various head/stem and cup degrees of freedom provided a relative ranking of contribution (Fig. 3) and identified the most important variables for further investigation. Variation in head/stem anteversion-retroversion had the largest impact of any degree of freedom throughout all three tasks; average JCF ranges were 413.5 (319.1) N during gait, 262.7 (256.4) N during sit to stand, 572.7 (228.1) N during step down (Fig. 3). Head/stem anteversion-retroversion also had the greatest influence on average muscle force (across all muscles) with the largest ranges occurring during step down; average muscle force ranges were 137.5 (43.1) N for anteversion compared to 70.4 (14.1) N for anterior-posterior, 48.6 (10.9) N for superior-inferior, 66.7 (9.2) N for medial-lateral and 69.5 (22.6) N for abduction-adduction. Perturbations increasing anteversion resulted in the highest muscle forces, which were greatest in the gluteus medius, as a prime mover muscle during step down. However, it should be noted that changes in individual muscle forces were sensitive to the alignment degree of freedom that affected its moment arm.

Based on the output force ranges (Fig. 3), which were derived for clinical levels of variability, it appears that head/stem alignment has a greater effect on JCF and average muscle forces than cup alignment. Sensitivity, reported as changes in output with respect to 1 mm and 1 degree changes in each alignment variability (Table 1), provides an alternative, normalized comparison. The sensitivity of hip JCF to cup perturbations were 21.2 N/mm on average across all degrees of freedom, which was 1.9× greater than all other stem perturbations except for stem anteversion (31.4 N/°)(Table 1). These results also highlight the interdependencies between individual muscles and the alignment degrees of freedom, and differences in muscle recruitment with activity.

Representative data for one patient illustrated the impact of variability in stem anteversion (Fig. 4) and cup anterior/posterior position (Fig. 5) on two large hip muscles (gluteus maximus and gluteus medius) and the magnitude of the hip JCF. These alignment variables, which were the largest contributors (Fig. 3, Table 1), are of interest clinically. Perturbations in head/stem anteversion (34.2°) resulted in large differences in gluteus maximus and gluteus medius forces and corresponding JCFs (ranging from 2.1× to 4.75× body weight) in the gait and step down activities (Fig. 4). Perturbations in cup anterior-posterior position (8.1 mm) resulted ranges with less variability throughout the task; however, cup position changes at that level did influence gluteus maximus muscle force during gait, resulting in a 203.5 N range in hip JCF during the latter portion of the stance phase when the hip was in extension.

Additionally, cup alignment impacted the direction of the hip JCF vector; S/I cup position had the greatest impact across all three tasks. This was most evident in the sagittal plane (average sagittal plane angle range: Cup 11.5 (3.6)° for anterior-posterior, 16.4 (4.9)° for superior-inferior, 9.9 (3.7)° for medial-lateral; Head/stem 1.6 (2.6)° for anterior-posterior, 11.2 (4.6)° for superior-inferior, 0.6 (1.4)° for medial lateral, 14.6 (6.3)° for anteversion, and 0.9 (1.8)° for abduction-adduction) with the most inferior cup placements moving the JCF vector towards the cup equator at the point of peak JCF for patients who exhibited higher values of hip extension during gait and the step down task (Fig. 6).

4. DISCUSSION

An experimentally-based musculoskeletal modeling analysis that included perturbations in THA component alignment was performed with the goal of identifying the relative contributions of alignment parameters on joint contact and muscle forces during three dynamic activities. Quantifying changes in loading during dynamic activities is important for surgical decision making to assess placement tradeoffs in impingement, fixation/bone quality and loading. Anteversion of the head/stem was identified as having the greatest impact on the magnitude of the hip JCF. Additionally, the superior-inferior position of the cup had the greatest influence on the direction of hip JCFs even though the input ranges, derived from CT measurements of differences in component placement relative to native hip geometry, were smaller for cup than head/stem alignments. While these were the alignment parameters that had the largest impact on hip joint contact force, the input variability in each degree of freedom taken from literature resulted in a range of 224.4 N in hip JCF, on average.

Anteversion of the head/stem resulted in the largest changes in muscle forces, and should be a point of emphasis given high levels of patient variability. Heller et al., (2001) also determined that increases in femoral anteversion of greater than 20° resulted in considerable increases in hip joint loading. To avoid impingement and/or dislocation, a “target zone” of between 10°–30° is a common surgical goal but only 69% of cases result in stem placement within this zone (Reikerås and Gunderson, 2011). Alignments that avoid increases in muscle loading during daily high demand tasks may lead to earlier recovery of function by reducing muscle fatigue and the need for the development of compensatory movement patterns (Gaffney et al., 2015). For example, as changes in combined anteversion are directly associated with impingement-free range of motion, differences in orientation suggest that patients would have to internally rotate the hip and adduct to offset the differences that resulted in the highest loads simulated in this study.

The classic technique of THA (Charnley, 1970; Muller, 1970) includes medialization of the cup with a corresponding increase in femoral offset with the goal of achieving strong cup fixation and increasing abductor moment arms. Cup alignment has a direct effect on muscle moment arms and lines of action because it alters the hip joint center, with medialization targets typically in the range of 5–10mm. While cup medialization has been shown to benefit certain patients with the most appropriate anatomy for this surgical approach (McGrory et al., 1995; Terrier et al., 2014), its impact on loading during dynamics tasks had not been previously defined. While medial perturbations of the cup did not result in the greatest impact on hip joint loading, a 1 mm medial change in the position of the cup resulted in 24.1 N change in HJC (approximately 1.20% of baseline HJC) and 15.8 N on gluteus medius muscle forces (approximately 1.45% of baseline gluteus medius force) during gait. In addition, 1 mm medial changes in the position of the head/stem resulted in 10.5 N changes in HJC (approximately 0.51% of baseline HJC) and 13.1 N on gluteus medius muscle forces (approximately 1.20% of baseline gluteus medius force) during gait. Delp and Maloney (Delp and Maloney, 1993) developed a model to study how the location of the center of the hip joint affects muscle moment generation. Similar to our findings, they found that moving the hip joint center medial by 2 cm decreased abductor muscle force generation by 22%.

This is the result of increasing the abductor moment arm which required less muscle force to produce the necessary net joint moment. The alterations to muscle lengths that are caused may require significant movement adaptations. However, it has been shown that for particular patients, cup medialization can be beneficial (Terrier et al., 2014) and understanding the impact on hip joint loading can be helpful when making this surgical decision. Surgical procedures, device designs, and instrumentation that facilitate optimal cup placement can be used to limit excessive loading on the implant.

The resulting variation in muscle forces and joint contact forces were dependent on the input ranges selected for component positioning. The inputs were selected based on relevant literature and were within typical surgical practice (Tsai et al., 2014). There are multiple THA surgical approaches that have been designed in order to optimize component placement and minimize the impact of surgical variability. Weber et al., (2015) used musculoskeletal modeling to investigate how hip JCFs during gait compared between a group of THA patients who had undergone a novel computer-assisted femur first/combined anteversion approach versus patients who had undergone traditional THA. They found similar baseline hip JCFs to those of our patients throughout the gait cycle and noted that hip JCFs from the computer-assisted femur first/combined anteversion approach were closer to young healthy individuals than those from the traditional THA group. This may be an example of how surgical approach can minimize the impact of the surgical variability, with potential to result in more favorable patient outcomes. In contrast, there are cases in which a deviation from the native geometry may be warranted by the surgeon based the pathology and specific diagnosis. For example, in cases with severe bone loss or developmental dysplasia, restoring native alignment may not be possible or in the best interest of the patient and implant positions could be considered outside of the ranges that were used in this study (Dudkiewicz et al., 2002; Hansen et al., 2011) In these extreme cases, it is valuable to understand how muscle and joint loading can change per the prescribed magnitude of positional change that a surgeon may consider pre-operatively (Table 1).

There are limitations to this study that should be considered. First, the simulation of each task used baseline kinematics for each patient to calculate muscle forces and joint contact forces. Kinematics were not altered following perturbations to component alignment. While it is possible that the perturbations could result in musculoskeletal adaption, those can be complex neuromuscular responses that were outside of the scope of this study. Second, it is common to compare JCFs calculated from musculoskeletal models to those recorded from telemetric implants; however, there is not currently an in-vivo representation of the influence of moving the alignment of implant components for use in comparison to the perturbation analysis performed here. Finally, the study perturbed component alignment variables individually and thereby did not evaluate the interaction effects of combined degrees of freedom. While our approach was able to meet the goals of identifying the alignment degrees of freedom that had the greatest impact on muscle forces and joint contact forces, we recognize that interactions effects are important moving forward.

5. CONCLUSION

The identification of the important hip implant alignment variables and quantification of their impact on joint loading during high demand activities, can support surgical decision-making and instrumentation development. Considering the surgical implications of how alignment affects range of motion and fixation, an improved understanding of its impact on loading can help assess tradeoffs and lead to improved THA patient outcomes.

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Highlights

- Hip arthroplasty alignment parameters with greatest impact on loading were identified.
- Subject-specific musculoskeletal models simulating daily activities were used.
- Head/stem anteversion-retroversion had the largest impact throughout all three tasks.
- Inferior cup placements resulted in forces towards the cup equator at peak loads.
- Alignment impact supports surgical decision-making and instrumentation development.

Fig. 1. **Left/Middle)** The range of head/stem and cup alignments that were analyzed in the musculoskeletal model for each degree of freedom from Tsai et al. [2] **Right)** Patient-specific musculoskeletal model in simulation of the dynamic tasks using measured kinematics and ground reaction forces.

Fig. 2.

Left: Representative baseline kinematics from one patient for each task. **Right:**

Comparisons between the magnitude of hip joint reaction force between the group of Orthoload patients (grey) and the baseline hip joint contact force for the group of five patients that participated in this study (blue). Envelope shown captures all of the subject data. 23

Fig. 3. Mean and standard deviation for the range of hip joint contact forces (top) and muscles forces (bottom) when considering variability in component position in each degree of freedom. Average muscle force represents a composite of all muscle forces. Magnitude represented the average for a task and the error bars represented variation between subjects. Head/stem alignment had a larger impact on both JCF and muscle forces than cup alignment. Anteversion-retroversion was the most significant degree of freedom.

Fig. 4.

Representative gluteus maximus and gluteus medius muscle forces and hip joint contact force for a single patient during each task when considering variability in stem anteversion, selected because it had more impact than other degrees of freedom (Fig. 3). Anteversion had the greatest impact on hip joint contact forces during gait and on gluteus medius muscle force during the step down.

Fig. 5.

Representative gluteus maximus and gluteus medius muscle forces and hip joint contact force for a single patient during each task when considering variability in anterior/posterior cup position.

Fig. 6. Direction of the hip joint contact force relative to the cup for superior/inferior cup positions at the point of peak joint contact force with femur in $\sim 15^\circ$ extension during the step down task.

Table 1

Sensitivity of the joint contact force (JCF) and individual muscle forces to changes in head/stem and cup alignment. Data reported as change in output (N) to change in input as 1 mm for translations and 1° for rotations.

Task	Head/Stem						Cup			
	AP	SI	ML	ANT	ABD		AP	SI	ML	
Gait	JCF	13.8	9.8	10.5	31.5	9.0	JCF	34.6	24.5	24.1
	GMED	2.5	5.3	13.1	8.8	10.4	GMED	2.3	2.0	15.8
	GMIN	0.8	1.3	3.6	3.6	1.8	GMIN	0.7	1.4	2.1
	GMAX	2.0	3.4	3.1	2.2	4.1	GMAX	4.4	2.5	5.1
	RECFEM	8.0	2.4	6.2	16.9	5.0	RECFEM	6.6	1.5	4.1
	SEMIMEM	0.4	0.2	0.3	1.1	0.4	SEMIMEM	2.4	0.7	0.9
	SEMITEN	0.3	0.1	0.1	0.6	0.2	SEMITEN	1.0	0.2	0.4
TFL	1.3	1.1	2.5	2.8	1.7	TFL	1.1	0.8	2.1	
Step Down	JCF	12.7	15.4	14.1	26.7	7.7	JCF	35.4	17.7	22.1
	GMED	2.6	5.2	9.9	9.6	4.4	GMED	5.0	3.6	17.9
	GMIN	0.8	1.6	3.8	3.7	1.9	GMIN	1.6	1.7	1.8
	GMAX	4.3	4.6	2.3	4.2	3.6	GMAX	9.8	4.6	5.3
	RECFEM	6.0	1.3	2.9	12.8	2.2	RECFEM	1.1	0.1	1.3
	SEMIMEM	3.0	0.9	1.4	2.5	2.4	SEMIMEM	4.4	2.8	2.5
	SEMITEN	0.7	0.4	0.5	0.5	0.7	SEMITEN	2.2	0.2	1.0
TFL	1.1	1.0	1.8	2.2	1.9	TFL	0.6	0.4	1.2	
Sit to Stand	JCF	11.8	9.5	6.7	36.0	12.0	JCF	22.8	26.4	14.6
	GMED	1.2	1.0	1.0	2.0	2.4	GMED	1.1	2.3	4.4
	GMIN	0.7	0.6	0.3	0.6	1.4	GMIN	0.6	1.8	0.7
	GMAX	6.0	6.4	4.2	7.1	7.1	GMAX	4.2	11.3	10.8
	RECFEM	3.1	2.5	2.8	6.2	8.7	RECFEM	2.7	3.2	2.9
	SEMIMEM	1.8	1.7	1.0	5.0	2.6	SEMIMEM	3.1	1.5	6.9

Task	Head/Stem						Cup			
	AP	SI	ML	ANT	ABD	SEM	TFL	AP	SI	ML
SEM	0.4	0.8	0.4	1.5	0.8	SEM	TFL	1.2	0.6	1.4
TFL	0.1	0.1	0.1	0.2	0.0	TFL	TFL	0.2	0.1	0.4