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Multi-joint Compensatory Effects of Unilateral Total Knee Arthroplasty during High-Demand Tasks

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Abstract

Patients with total knee arthroplasty (TKA) demonstrate quadriceps weakness and functional limitations one year after surgery during daily tasks such as walking and stair climbing. Most biomechanical analyses of patients after TKA focus on quadriceps function and rarely investigate other lower-extremity muscles or high-demand ambulatory activities of daily living. The purpose of this investigation was to quantify lower-extremity muscle forces in patients with unilateral TKA during high-demand tasks of pivoting and descending stairs. Five patients with unilateral TKA and five age and sex-matched controls performed three bilateral high-demand tasks: 1) step down from an 8-inch platform, 2) inside pivot: 90° direction change toward planted limb, and 3) outside pivot: 90° direction change away from planted limb. Subject-specific musculoskeletal simulations were created in OpenSim to determine joint angles, moments, and lower-extremity muscle forces. The results indicate that patients with TKA adopt compensatory strategies at both the hip and knee. Patients with TKA demonstrated increased hip external rotation, decreased knee flexion, decreased quadriceps force, and decreased hip abductor force in all three tasks. These strategies are likely a result of quadriceps avoidance, which may stem from instability after TKA or a habitual strategy developed during the late stages of osteoarthritis.

Keywords

musculoskeletal modeling; TKR; knee osteoarthritis; rehabilitation

INTRODUCTION

Over 700,000 total knee arthroplasty (TKA) surgeries are performed annually in the United States (Wier et al. 2009) and this number is expected to exceed over 3.5 million by the year 2030 (Kurtz et al. 2007). TKA is most frequently performed to relieve pain and improve overall function in patients suffering from end-stage knee osteoarthritis (OA) (Mizner et al. 2005; Walsh et al. 1998). However, many patients continue to demonstrate impairments and

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CONFLICT OF INTEREST

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functional limitations one year after TKA surgery during daily tasks such as walking, turning, and stair climbing (Dawson et al. 1998; Farquhar et al. 2008; Konig et al. 2000; Walsh et al. 1998). Specifically, patients with TKA demonstrate lower functional scores in timed clinical tests (e.g. timed up and go test, 6-minute walk test, stair-climbing test) compared to healthy controls, and with tests requiring high-demand tasks such as stairs and pivot turns showing the greatest limitation (Bade et al. 2010; Yoshida et al. 2008). Patients after TKA also exhibit movement asymmetries between limbs during gait, most notably reduced knee flexion angles and moments, and reduced quadriceps strength (Farquhar et al. 2008; McClelland et al. 2007; Milner, 2008). Furthermore, there is evidence that asymmetric hip extension and adduction moments exist before and after TKA between the surgical and nonsurgical limbs (Chang et al., 2005; Farquhar et al., 2008; Mündermann et al. 2005). Whether these asymmetries are residual compensatory strategies to avoid pain from OA, or developed after TKA, they can contribute to increased loading on the nonsurgical limb, long-term functional disability, and prolonged muscle weakness (Christiansen & Stevens-Lapsley, 2010; Shakoor et al. 2002).

Research examining lower extremity muscle weakness and functional disability after TKA has largely focused on the quadriceps. It is well documented that after TKA, patients demonstrate quadriceps weakness (Bade et al. 2010; Mizner & Snyder-Mackler, 2005; Stevens-Lapsley et al. 2010) and often adopt a “quadriceps avoidance” strategy by reducing the knee extensor moment on the surgical limb (McClelland et al. 2007; Milner, 2009). Apparent muscle weakness may extend proximal and distal to the knee as well. Previous investigations have demonstrated reduced hip abductor (gluteus medius) force following unilateral TKA and an association between low hip abductor strength and decreased functional performance scores (Alnahdi et al. 2014; Piva et al. 2011).

As with most studies of muscle strength, these studies centered on common clinical tasks of functional performance (e.g. gait speed, figure-of-eight walk test, stair-climb test, five-chair rise test, timed up and go test, strength), in which the primary outcome is the completion time or symmetry across limbs (Benedetti et al. 2003; Beyaert et al. 2008; Byrne et al. 2002; Farquhar et al. 2008; Glaister et al. 2007; Mizner & Snyder-Mackler, 2005; Walsh et al. 1998; Yoshida et al. 2013). Although these investigations provide valuable information regarding the functional deficits following TKA during controlled tasks, patients with TKA continue to self-report limitations on the surgical limb during high-demand tasks (Dawson et al. 1998; Konig et al., 2000), and the mechanisms underlying such tasks remain largely unknown. Previous research has shown that turns during walking account for 8–50% of steps taken during daily activities (Glaister et al. 2007) and are directly linked to injurious falls (Cumming & Klineberg, 1994). Because pivoting while walking is inherent in daily activity for avoiding obstacles and turning corners, understanding the underlying movement patterns during pivoting will provide insight into modifiable targets during movement retraining in rehabilitation.

The objective of this investigation was to quantify lower extremity biomechanics (kinematics, joint moments, and muscle forces) from the hip (e.g. gluteus medius) to the ankle (e.g. gastrocnemius) in patients one year after unilateral TKA during high-demand tasks (pivoting and descending stairs). We hypothesized that patients after TKA would demonstrate altered kinematics, joint moments, and lower muscle forces at the ankle, knee, and hip of the lower-extremity compared to control subjects.

METHODS

Participants

Five patients (2F, 3M, age: 70 ± 9.1 years; height: 1.7 ± 0.1 m; mass: 79.3 ± 16.6 kg) one year after unilateral TKA surgery and five healthy age- and sex-matched controls (2F, 3M, age: 69.0 ± 7.6 years; height: 1.7 ± 0.1 m; mass: 77.2 ± 15.4 kg) were enrolled in this study. Each participant provided a written, informed consent in accordance with the *blinded for peer review* prior to the start of the experimental session. Each participant visited the laboratory for one data collection in which lower extremity kinematics and kinetics were collected during high-demand tasks.

Motion Analysis

Sixty-four reflective markers were placed on each participant. Markers placed on distal and proximal bony landmarks were used to define each segment. Eight infrared cameras (Vicon, Centennial, CO) were used to record the spatial positions of the markers during the tasks. Ground reaction forces were collected from two force platforms (2000 Hz sampling frequency) embedded in the laboratory floor (Bertec, Columbus, OH).

Each participant performed a series of three bilateral high-demand tasks: 1) step down from an 8-inch platform (Figure 1a), 2) inside pivot: 90° change in direction toward the planted limb (Figure 1b), and 3) outside pivot: 90° change in direction away from the planted limb (Figure 1c). Participants performed each task three times. Trials were averaged across the three trials for group comparisons. No instructions were provided to participants regarding the speed at which to complete each task. It has been previously reported that slower walking speed is associated with lower hip and knee joint moments (Kirtley & Whittle, 1985; Mündermann et al. 2004), and likely the same effect is true for high-demand tasks. However, because this subset of patients with TKA was highly active, we assumed that the speeds throughout each task would be similar to that of age- and sex-matched controls. To test this limitation, we compared trial time across groups with two-tailed independent t-tests and found no significant differences (Step Down: $P = 0.82$, Inside Pivot: $P = 0.34$, Outside Pivot: $P = 0.43$). Marker and force platform data were filtered using a 4th order lowpass Butterworth filter (6 Hz and 20 Hz cutoff frequency, respectively).

Musculoskeletal Modeling

Kinematics, inverse dynamics, and muscle forces were calculated using subject-specific musculoskeletal models (29 degrees of freedom, 100 muscle tendon actuators) in OpenSim (Arnold et al. 2010) that were scaled according to each subject's anthropometric measurements. In detail, the musculoskeletal model was scaled using ratios of the distance between markers placed at bony landmarks on each subject, and virtual markers placed at corresponding locations on the model. The pelvis was scaled using relative distances between markers placed on the left and right anterior superior iliac spine (ASIS) and left and right posterior superior iliac spine (PSIS). The thighs were scaled using relative distances between markers placed on the ASIS, greater trochanters, and the medial and lateral femoral epicondyles. The shanks were scaled using relative distances between markers placed on the medial and lateral femoral epicondyles and the medial and lateral malleoli. The feet were scaled using relative distances between markers placed on the medial and lateral malleoli, heel, proximal and distal first metatarsal, and fifth metatarsal. In addition to scaling of the segment geometry, the knee within this model was modified to include three rotational and two translational degrees of freedom. Specifically, the Walker knee (Walker et al. 1988) within the model was modified to include the varus/valgus and internal/external rotations measured from the participants during each of the activities. The splines that specify varus/valgus and internal/external rotation within the Walker knee were redefined for each subject and activity using each subject's kinematic data. Anterior/posterior and superior/inferior translations were determined as a function of flexion angle using Walker's polynomial equations. The geometry of the quadriceps muscles (rectus femoris, vastus lateralis, vastus intermedius, vastus medialis) of the OpenSim model was refined to insert on the tibial tubercle with via points placed on the superior and inferior poles of the patella (Demers et al. 2014). This enabled resultant quadriceps forces to be properly included in the calculation of tibiofemoral contact forces in OpenSim. Distal via point locations of the quadriceps were adjusted to assure that the moment arm of these muscles matched measurements of patellar tendon moment arm (Krevolin et al. 2004). Finally, the mass of each segment was adjusted to match the measured mass of the subject. Each segment was scaled using the same proportion as they were in the generic model, as previously described (Arnold et al. 2010). Once scaled, the inertial tensors of each segment were adjusted accordingly to coincide with the scaled masses. For patients with TKA, maximum isometric strength of the muscles in each subject-specific model were scaled to match experimentally collected subject-specific maximum isometric joint torques in knee flexion and extension. Experimental isometric torques were produced without twitch or burst superimposition because maximum muscle force production is mainly linked to muscle volume at greater than one year post TKA surpassing the effect early postsurgical activation deficits (Meier et al. 2009).

Lower-extremity kinematics (hip, knee, and ankle) during the three high-demand tasks were calculated by minimizing the sum of weighted squared error between experimental marker positions and corresponding theoretical markers in the model. A success criterion of the kinematics was specified by a marker error of less than 2 cm with an RMS error of less than 1 cm (OpenSim User Guide). Joint moments were calculated during the three high-demand tasks with an inverse dynamic solution (Winter, 2009). For this investigation, a flexion torque is defined as that which is required to flex the joint; likewise, for extension, add/abduction, int/external rotation, and dorsi/plantarflexion moments.

Static optimization was used to estimate the muscle forces required to generate the kinetics measured during the three high-demand tasks (Anderson & Pandy, 2001). First, a residual reduction algorithm was performed to improve dynamic consistency between measured external forces and the forces produced by the model by making minor perturbations to the torso center of mass and model kinematics (Delp et al. 2007b). Second, the muscle forces required to reproduce the joint moments were estimated by solving a static optimization problem at each time frame that minimized the square of muscle activation. The success criterion of the residual reduction algorithm and static optimization was specified by a peak residual force applied to the model lower than 20 N (OpenSim User Guide). For use in validation of the muscle force predictions, joint loads at the knee were calculated in OpenSim from the muscle forces and inverse dynamics.

Model Validation

Quantitative validation of predicted muscle forces was achieved through three methods: 1) comparison of predicted muscle moments to moments calculated using inverse dynamics, 2) comparison of estimated muscle activation to previously published recordings of muscle electromyography (EMG), and 3) comparison of predicted knee joint contact loads to previously published *in vivo* knee contact loads. Muscle forces and joint moments were calculated for both groups during one gait cycle at 1.0 m/s (healthy controls: left heel strike to left heel strike; TKA cohort: involved heel strike to involved heel strike).

Given that the static optimization tool computes muscle forces to match generalized forces computed from an inverse dynamic solution (while minimizing the objective function), a comparison between joint moments estimated by the sum of muscle forces crossed with their moment arms and joint moments calculated during inverse dynamics was performed. The average RMS errors for hip flexion, hip adduction, and knee flexion was 4.47 ± 1.57 N/m, 6.95 ± 3.24 N/m, 2.20 ± 1.57 N/m, respectively, for healthy controls and was 7.43 ± 2.30 N/m, 15.85 ± 6.70 N/m, 2.94 ± 0.84 N/m, respectively, for the TKA cohort (Figure 2). Low average RMS error, as well as similar peak magnitudes and temporal characteristics of waveforms of joint moments indicate overall accuracy of estimated muscle forces.

Comparison to previously published experimentally collected muscle activation data is a common validation method within the field of musculoskeletal modeling (Thelen & Anderson, 2006; Hicks et al. 2014). Estimated muscle activations were compared to muscle EMG in healthy adults walking at 1.0 m/s, as reported by Hof et al. (2005). Timing and overall waveform characteristics throughout the gait cycle compare closely with experimental EMG results (Figure 3). Qualitative assessment of estimated activations follow the documented progression of muscle activation during walking (Anderson & Pandy, 2003; Liu et al. 2008), beginning with hamstrings activation at heel-strike, quadriceps activation during weight acceptance, and finally activation of the plantar flexor muscles through toe-off.

Given that muscle forces are the main contributor to joint load during activities of daily living (Shelburne et al. 2006; Kim et al. 2009; Winby et al. 2009), comparison of predicted joint load to in vivo measurements made with instrumented implants was used as validation of muscle force predictions (Lu et al. 1997; Lin et al. 2010; Nikooyan et al. 2010). The estimated peak contact force at the knee in the superior-inferior direction for our TKA cohort (2.08 ± 0.30 (xBW)) was similar to recordings from an instrumented knee implant (2.43 ± 0.23 (xBW)) (Kutzner et al. 2010). In addition, temporal patterns of peak knee joint contact force throughout the gait cycle correspond well between predicted and measured forces (Figure 4), providing additional validation of the musculoskeletal model.

Analysis

Biomechanical analyses were isolated during the stance period during both pivot tasks (heel strike to toe off) and during the landing period during the step down (leading-limb foot strike to trailing-limb foot strike). Maximum and minimum kinematics were compared across groups using a two-tailed independent *t*-test. Lower-extremity muscles for this analysis were chosen by their primary functional role at each degree of freedom of the lower extremity (Table 1). To ensure that body mass did not enforce a false significance on dependent variables (occurs when normalized data does not cross through the origin in a linear regression (Curran-Everett, 2013)), peak joint moments were compared across groups using an analysis of covariance (ANCOVA) with mass as the covariate. Level of significance was set at $\alpha = 0.05$ for all inferential tests. Because the inferential analysis included a group-to-group comparison in each dependent variable for every task, a correction for inflated Type I error was not required. In addition, Cohen's d effect size was calculated (small effect: $d \leq 0.2$, medium effect: $0.2 < d < 0.8$, large effect: $d \geq 0.8$) (Portney & Watkins, 2000). Statistical analysis was performed using SPSS 22.0 (SPSS, Inc.).

RESULTS

During all tasks, patients with TKA demonstrated increased hip external rotation, decreased knee flexion, decreased quadriceps force, and decreased hip abductor force. The results section (text, tables, and figures) is organized to highlight significant differences found in the kinematics, inverse dynamics, and muscle forces in all three tasks.

Step Down

In the TKA group, the peak knee flexion angle was 11.2° smaller ($P = 0.01$, $d = 2.4$), the peak knee extensor moment was 0.38 N*m/kg smaller ($P = 0.01$, $d = 2.6$), the peak gluteus maximums force was 0.11 N/BW smaller ($P = 0.04$, $d = 1.1$), the peak iliacus force was 0.41 N/BW larger ($P < 0.01$, $d = 2.2$), the peak psoas force was 0.41 N/BW larger ($P = 0.01$, $d = 1.9$), and the peak vastus lateralis force was 0.68 N/BW smaller ($P = 0.02$, $d = 2.1$) in comparison to healthy controls (Table 2, Figure 5).

Inside Pivot

In the TKA group, the peak hip external rotation angle was 14.4° larger ($P = 0.02$, $d = 1.9$), the peak knee varus angle was 12.6° smaller ($P < 0.01$, $d = 5.0$), the peak knee valgus angle was 4.4° larger ($P = 0.01$, $d = 2.5$), the peak hip external moment was 0.29 N*m/kg larger ($P = 0.03$, $d = 1.8$), the peak knee extensor moment was 0.38 N*m/kg smaller ($P = 0.01$, $d = 2.6$), peak knee internal moment was 0.20 N*m/kg larger ($P = 0.04$, $d = 1.5$), the peak biceps femoris longhead force was 0.21 N/BW larger ($P = 0.02$, $d = 2.2$), the peak medial gastrocnemius force was 0.65 N/BW larger ($P = 0.05$, $d = 1.8$), the peak gluteus minimus force was 0.12 N/BW larger ($P = 0.04$, $d = 1.5$), the peak iliacus force was 0.27 N/BW larger ($P = 0.02$, $d = 2.0$), the peak psoas force was 0.29 N/BW larger ($P = 0.04$, $d = 1.6$) in patients with TKA in comparison to healthy controls (Table 3, Figure 6).

Outside Pivot

In the TKA group, peak knee extension angle was 8.3° smaller ($P = 0.04$, $d = 1.5$), the peak knee varus and valgus angles were 7.7° and 6.9° smaller ($P = 0.01$, $d = 2,4$; $P < 0.01$, $d = 3.7$, respectively), the peak hip extension moment was 0.30 N*m/kg larger ($P = 0.04$, $d = 1.7$), the peak hip internal rotation moment was 0.14 N*m/kg larger ($P = 0.01$, $d = 2.1$), the peak knee external rotation moment was 0.17 N*m/kg larger ($P = 0.03$, $d = 1.2$), and the peak knee ankle dorsiflexion moment was 0.47 N*m/kg smaller ($P = 0.03$, $d = 1.5$), the peak iliacus force was 0.80 N/BW larger ($P < 0.01$, $d = 4.0$), and the peak psoas force was 0.96 N/BW ($P < 0.01$, $d = 3.9$) in patients with TKA in comparison to healthy controls (Table 4, Figure 7).

DISCUSSION

The purpose of this investigation was to determine the multi-joint effects on lower extremity biomechanics in patients with unilateral TKA during three high-demand tasks. Compared to healthy controls, patients with TKA demonstrated altered kinematics and joint moments, as well as lower muscle forces, at the knee and the hip. During all three tasks, patients adopted compensatory movement strategies that reduced loading on the surgical limb by externally rotating the hip and reducing knee flexion angles, which resulted in lower gluteus medius and quadriceps forces. These findings indicate the long-term TKA deficits in knee and hip function, particularly in the gluteus medius muscle, which is an important muscle for stabilization during high-demand tasks.

Patients with TKA demonstrated a knee “stiffening strategy” (reduced knee flexion excursion, reduced knee varus angles, reduced knee extension moment, reduced quadriceps force) in comparison to controls during all three tasks. This strategy is likely an effect of quadriceps avoidance, in which patients with TKA reduce the demand on the knee extensors (Blaha, 2004; Davidson et al. 2013; Mizner et al. 2005; Slemenda et al. 1997; Stevens-Lapsley et al. 2010; Walsh et al. 1998), and may stem from instability after TKA, muscle weakness following surgery, or a habitual strategy developed during the late stages of OA. However, because preoperative movement strategies of this population are unknown, it cannot be determined from our data if these differences are a result of the TKA or habits developed during the end-stage of OA. Our results indicate that to compensate for reduced knee extensor demand, patients with TKA increased hip extensor demand in order to successfully ambulate, which is consistent with previous findings (McGibbon & Krebs, 2002).

Patients with TKA demonstrated differences in muscle forces surrounding the hip during all three tasks in comparison to healthy controls. These findings are consistent with previous investigations that demonstrated reduced hip abduction moments and hip abductor in patients with TKA (Alnahdi et al. 2014; Chang et al. 2005; Mündermann et al. 2005). Patients abducted and externally rotated the hip on the surgical limb at heel strike, which resulted in reduced recruitment potential to the posterior and lateral fibers of the gluteus medius. Previous studies have shown that an abducted and externally rotated hip reduces activation of the gluteus medius (Nyland et al. 2004). In response to lower gluteus medius force and quadriceps avoidance, patients in the current study had higher force in hip extensor muscles, which is consistent with previous findings (Lewis et al. 2007). An increase in hip extensor muscle force may be a compensatory strategy adopted at the hip to successfully complete the task.

During the step down, patients with TKA demonstrated increased hip extension and external rotation, as well as decreased knee flexion and valgus angles. These kinematics corresponded to reduced joint moments at the hip and knee throughout the landing period. Consequently, peak quadriceps force (vastus lateralis, vastus medialis, vastus intermedius) and gluteus medius force were smaller and peak biceps femoris long head and iliacus force were larger in patients with TKA in comparison to healthy controls. These results may indicate that patients with TKA adopt a strategy, largely at the hip, which may be used to avoid the use of the quadriceps and reduce load on the surgical knee.

In contrast to the step down, patients with TKA adopted an anticipatory strategy at the beginning of the inside pivot by using greater hip abduction and external rotation. This may be a strategy to reduce rotation on the surgical knee during the direction change. In addition, patients with TKA demonstrated smaller knee flexion angle excursion, smaller knee valgus angles, smaller knee extension moments, and smaller hip flexion and rotational moments throughout the pivot. These differences resulted in reduced peak gluteus maximus, gluteus medius, and quadriceps forces and may indicate that patients with TKA adopt a strategy to increase the role of the hip extensors to reduce the overall demand on the knee extensor muscles, which is consistent with previous findings (Farquhar et al. 2008).

In contrast to the previous two tasks, patients with TKA demonstrated a compensatory strategy at the knee, in addition to the hip, by adopting smaller knee flexion angles and larger peak hip external rotation throughout the direction change of the outside pivot. Reduced hip abduction, knee extension, and plantarflexor moments coupled with increased hip flexion and internal rotation moment late in the pivot correspond to our observation that patients with TKA “fell out” rather than “pushed out” of the pivot on the surgical limb. Therefore, patients with TKA increased force in the hip flexors (psoas, tensor fasciae latae, rectus femoris) to successfully complete the outside pivot.

The compensatory movement strategies adopted by patients with TKA create asymmetric kinematic and joint loading conditions, which may have long-term adverse healthy and function effects. Increased loading of the contralateral joints may increase the risk of future development of contralateral knee OA (Shakoor et al. 2002). Therefore, movement retraining following TKA should emphasize symmetric movement patterns by emphasizing increased quadriceps use on the surgical limb and limit hip compensation.

Several limitations to this investigation should be considered. First, this investigation had a small sample size that consisted of patients who self-reported as active and satisfied with their physical function following TKA. The multi-joint effects of TKA during high-demand tasks may be more pronounced in lower functioning individuals. Second, the musculoskeletal model was validated during gait because EMG and joint loading experimental data is widely described and the task is repeatable. However, validation during walking does not necessarily assure the model is validated during the high-demand activities. Third, using reflective markers to track pelvis motion and calculate hip kinematics is problematic due to the difficulty of accurate marker placement and tracking on palpable bony landmarks. However, using reflective skin-based markers to calculate hip kinematics during dynamic tasks remains the current standard of practice (Kadaba et al. 1990; Anderson & Pandy, 2001; Mündermann et al. 2005; Arnold et al. 2006; Farquhar et al. 2008; Lewis & Garibay, 2014). Fourth, because pre-operative knowledge of movement patterns of this population is unknown, we cannot separate the effects of habitual movement patterns developed during OA or those resulting from TKA. However, with an eye toward improved rehabilitation, it is most relevant for us to identify outstanding deficits in patients with TKA after surgery, regardless of patterns established pre-operatively. Fifth, the effects of uncertainty in our input parameters (e.g. marker position, marker motion artifact, variability in body segment parameters, variability in muscle parameters) were not taken into account in our musculoskeletal simulations. A previous investigation demonstrated that the effect of movement artifact had the greatest overall impact on results computed within a musculoskeletal modeling workflow (Myers et al. 2014). Finally, the effect of knee replacement component alignment on the results is unknown. It has previously been reported that component alignment can affect functional outcome following TKA (Fehring 2000; Stulberg et al. 2002).

To our knowledge, this is the first study to quantify multi-joint lower-extremity biomechanics in patients with TKA during high-demand tasks. All previous research involving patients with TKA has investigated gait (sagittal plane motion) or used timed-clinical tests, either of which fail to fully capture the range of task patients with TKA are required to overcome to restore full function. Our results indicate that patients with TKA adopt movement strategies at the hip, knee, and ankle that are likely a result from quadriceps avoidance, which may stem from instability after TKA, weakness of the quadriceps muscles, or a habitual strategy developed during the late stages of osteoarthritis. Furthermore, these movement strategies have long-term effects on the lower-extremity muscles beyond the quadriceps. Therefore, these patients may benefit from a rehabilitation strategy that considers the multi-joint effects of TKA and aims to restore normal movement patterns of the entire limb, which may be accomplished by targeting muscles in addition to the quadriceps.

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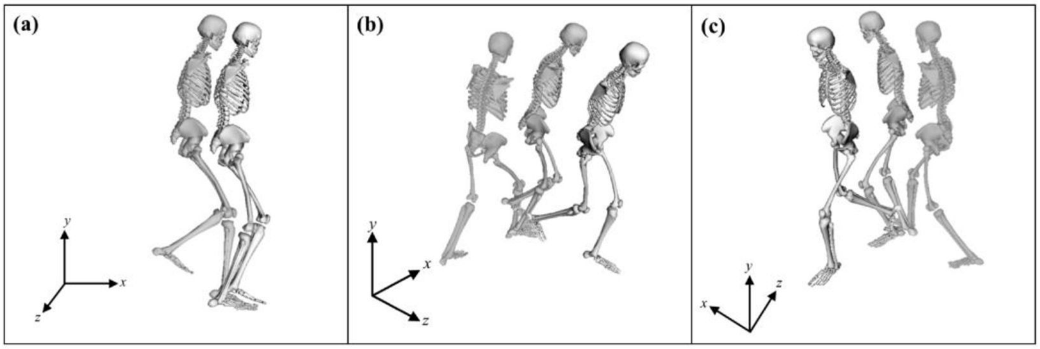

Figure 1.
Three bilateral tasks performed by each participant are (a) step down, (b) inside pivot, and (c) outside pivot.

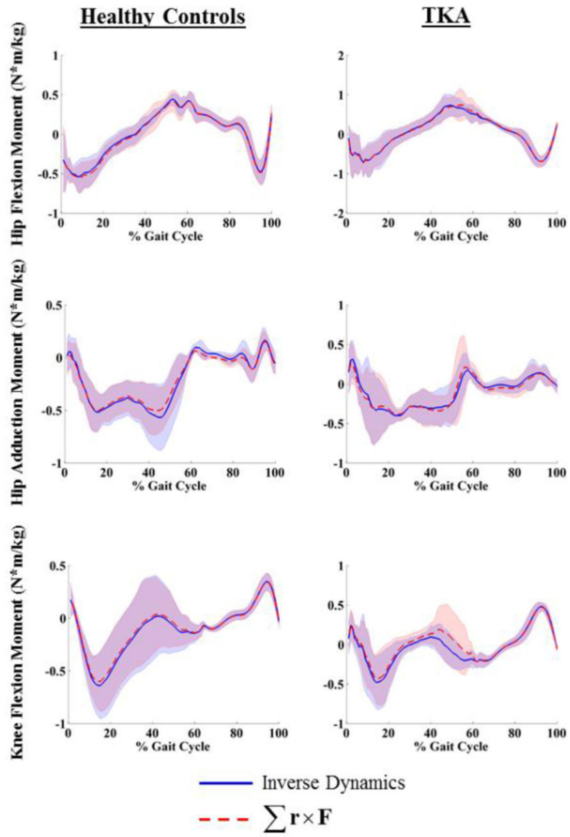

Figure 2.

Mean ± 1 SD of joint moments computed using inverse dynamics and the sum of the muscle moments (moment arm x muscle force) for healthy controls and patients with TKA during gait. Differences between the two indicate the use of joint reserve actuators, and therefore, comparison of joint moments between these two methods served as a validation of the musculoskeletal model.

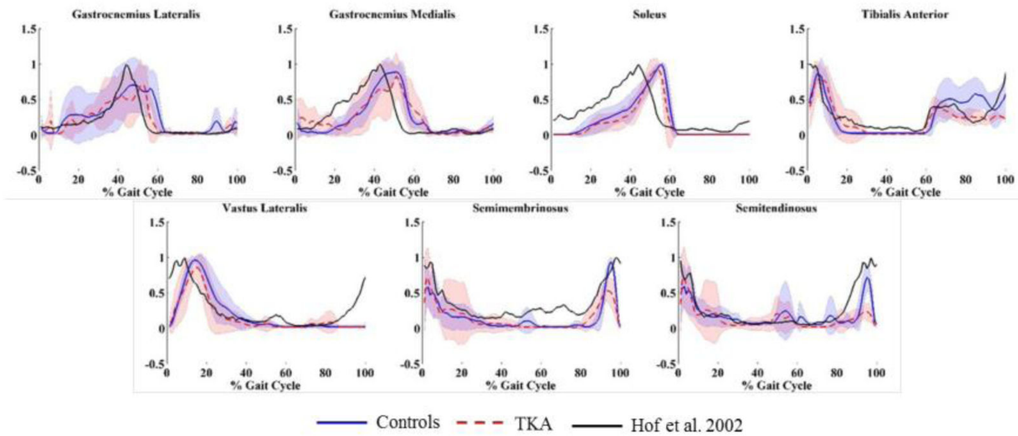

Figure 3.

Mean ± 1 SD muscle activations from seven lower extremity muscles estimated from the musculoskeletal model during gait in patients with TKA (red dashed line) and healthy controls (blue solid line). Estimated activations were compared to experimentally collected EMG during gait (solid black line) previously published (Hof et al. 2002).

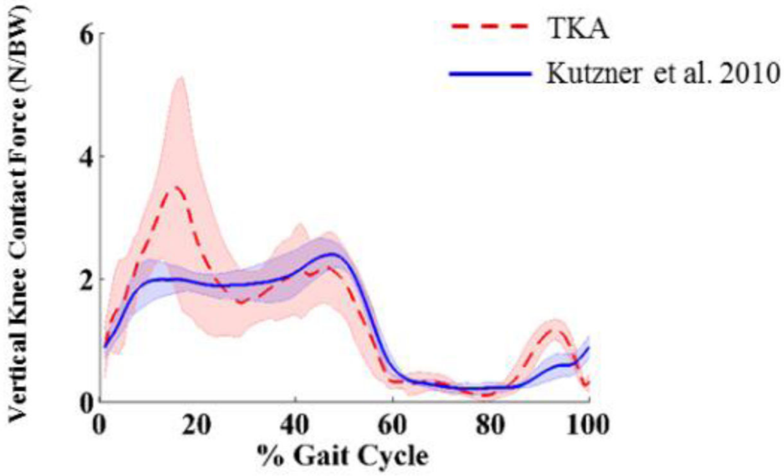

Figure 4.

Mean ± 1 SD vertical knee contact force predicted from the musculoskeletal model in patients with TKA (red dashed line) and from *in-vivo* contact forces measured from an instrumented implant (blue solid line) during gait (Kutzner et al. 2010). Comparison of peak magnitudes and temporal patterns across groups served as validation of the musculoskeletal model.

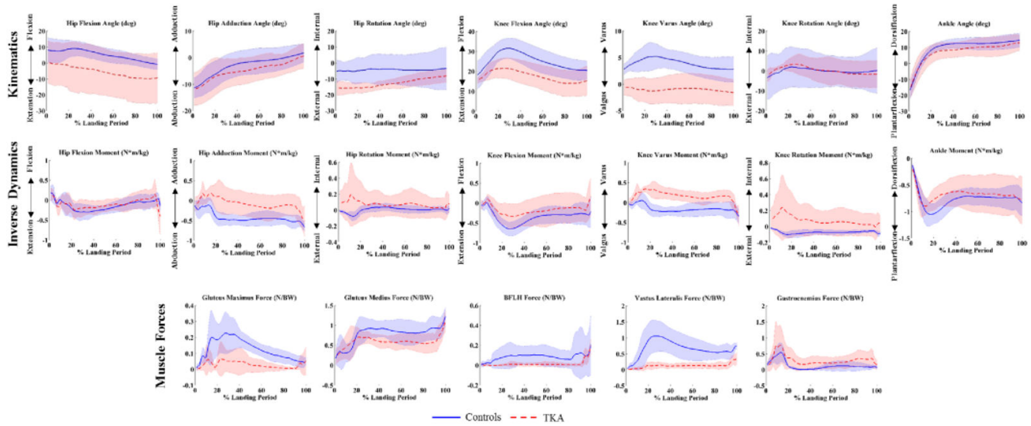

Figure 5.
Mean ± 1 SD kinematics, inverse dynamics, and estimated muscle forces during the Step Down for patients with unilateral TKA (red dashed line) and controls (solid blue line). For visualization, to standardize dependent variables across all participants, joint moments were normalized to body mass (Nm/kg) and muscle forces were normalized to bodyweight (xBW).

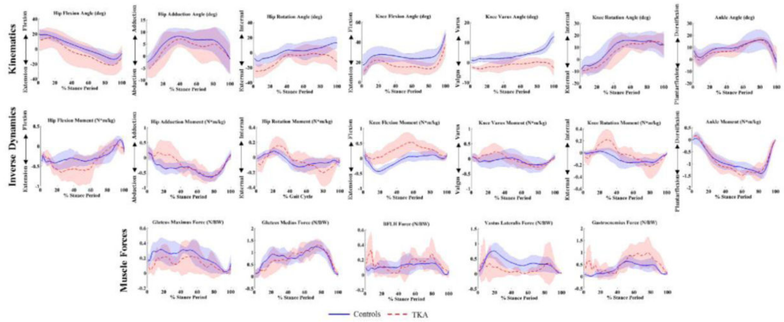

Figure 6.
Mean ± 1 SD kinematics, inverse dynamics, and estimated muscle forces during the Inside Pivot for patients with unilateral TKA (red dashed line) and controls (solid blue line). For visualization, to standardize dependent variables across all participants, joint moments were normalized to body mass (Nm/kg) and muscle forces were normalized to bodyweight (xBW).

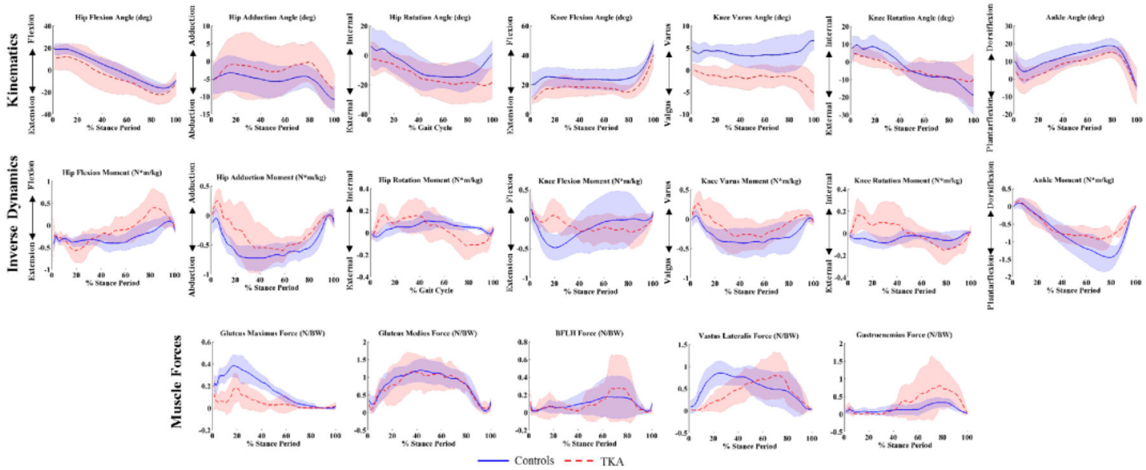

Figure 7.
Mean ± 1 SD kinematics, inverse dynamics, and estimated muscle forces during the Outside Pivot for patients with unilateral TKA (red dashed line) and controls (solid blue line). For visualization, to standardize dependent variables across all participants, joint moments were normalized to body mass (Nm/kg) and muscle forces were normalized to bodyweight (xBW).

Table 1

Lower-extremity muscles selected as dependent variables in the current analysis based upon their functional role during the three bilateral high-demand tasks.

Muscle	Functional Role
Gluteus Medius	Hip abductor, hip external rotator, hip internal rotator
Gluteus Minimus	Hip abductor, hip internal rotator
Gluteus Maximus	Hip abductor, hip extensor
Biceps Femoris Long Head	Hip extensor, Knee flexor
Iliacus	Hip flexor, hip rotator
Psoas	Hip flexor, hip external rotator
Tensor Fasciae Latae	Hip flexor, hip abductor, hip internal rotator
Rectus Femoris	Hip flexor, knee extensor
Vastus Lateralis	Knee extensor
Vastus Medialis	Knee extensor
Vastus Intermedius	Knee extensor
Gastroc Medius	Knee flexor, ankle plantarflexor

Table 2

Peak inverse kinematics, inverse dynamics, and muscle forces (mean±SD) during the Step Down in patients with unilateral TKA and controls.

	Inverse Kinematics													
	Hip						*Knee*						*Ankle*	
	Ext	Flex	Abd	Add	Ext Rot	Int Rot	Ext	Flex	Varus	Valgus	Ext Rot	Int Rot	Dors	Plant
Control	−0.8° (3.3°)	10.4° (6.0)	−11.4° (3.8°)	1.9° (3.4)	−9.7° (6.6)	1.7° (10.6°)	16.7° (5.5°)	**32.0° (5.0)**	2.1° (1.6°)	5.8° (2.3°)	−8.1° (11.7°)	5.9° (4.0°)	14.6° (4.4°)	−16.7° (5.5°)
TKA	−10.6 (15.7)	1.2° (13.1°)	−10.8° (5.9°)	2.0° (5.9°)	−17.5° (4.4°)	−8.2° (1.4)	11.9° (4.3°)	**20.9° (4.3)**	−0.1° (2.0°)	−1.3° (2.1°)	−2.8° (2.1°)	3.9° (3.6°)	11.5° (3.9°)	−16.4° (6.5°)

	Inverse Dynamics (N*m/kg)													
	Hip						*Knee*						*Ankle*	
	Ext	Flex	Abd	Add	Ext Rot	Int Rot	Ext	Flex	Varus	Valgus	Ext Rot	Int Rot	Dors	Plant
Control	−0.41 (0.14)	0.27 (0.10)	−0.68 (0.10)	0.01 (0.10)	−0.08 (0.05)	0.05 (0.05)	**−0.67 (0.17)**	0.15 (0.16)	0.11 (0.15)	−0.39 (0.15)	−0.12 (0.02)	−0.02 (0.01)	−1.16 (0.08)	−0.12 (0.04)
TKA	−0.33 (0.15)	0.25 (0.13)	−0.60 (0.20)	0.27 (0.32)	−0.01 (0.12)	0.20 (0.26)	**−0.29 (0.13)**	0.18 (0.08)	0.36 (0.24)	−0.36 (0.23)	−0.04 (0.07)	0.22 (0.31)	−1.06 0.17	−0.15 (0.04)

	Muscle Force (N/BW)											
	GlutMax	GlutMin	GlutMed	BFLH	Iliacus	Psoas	TFL	RecFem	VasLat	VasMed	VasInt	GasMed
Control	**0.28 (0.11)**	0.30 (0.12)	1.28 (0.23)	0.28 (0.28)	**0.21 (0.07)**	**0.24 (0.08)**	0.11 (0.03)	0.40 (0.08)	**1.14 (0.44)**	0.33 (0.24)	0.19 (0.08)	0.58 (0.32)
TKA	**0.17 (0.08)**	0.38 (0.10)	1.20 (0.14)	0.20 (0.08)	**0.62 (0.26)**	**0.65 (0.30)**	0.10 (0.06)	0.28 (0.28)	**0.46 (0.07)**	0.57 (0.31)	0.38 (0.30)	0.89 (0.76)

Bold indicates significant difference across groups ($\alpha < 0.05$). Shaded indicates large effect across groups ($d > 0.8$).

Table 3

Peak inverse kinematics, inverse dynamics, and muscle forces (mean±SD) during the Inside Pivot in patients with unilateral TKA and controls.

Inverse Kinematics

	Hip						Knee						Ankle	
	Ext	**Flex**	**Abd**	**Add**	**Ext Rot**	**Int Rot**	**Ext**	**Flex**	**Varus**	**Valgus**	**Ext Rot**	**Int Rot**	**Dors**	**Plant**
Control	−14.1° (7.6°)	20.3° (5.5°)	−4.2° (3.1°)	10.3° (4.3°)	**−11.9° (7.5)**	14.3° (7.8°)	16.3° (9.3°)	48.5° (4.4°)	**13.5° (2.9°)**	**0.5° (1.8°)**	−8.2° (6.5°)	19.8° (5.6°)	−2.9° (5.7°)	17.4° (7.2°)
TKA	−21.9° (13.1°)	16.2° (12.2°)	−5.8° (3.3°)	9.7° (5.2°)	**−26.3° (7.7°)**	5.5° (9.5°)	9.3° (3.8°)	38.1° (10.0°)	**−4.9° (2.5°)**	**0.89° (2.0°)**	−10.6° (2.6°)	18.3° (4.0°)	−6.4° (6.3°)	16.4° (2.1°)

Inverse Dynamics (N*m/kg)

	Hip						Knee						Ankle	
	Ext	**Flex**	**Abd**	**Add**	**Ext Rot**	**Int Rot**	**Ext**	**Flex**	**Varus**	**Valgus**	**Ext Rot**	**Int Rot**	**Dors**	**Plant**
Control	**−0.51 (0.08)**	0.20 (0.07)	−0.68 (0.13)	0.23 (0.19)	−0.17 (0.07)	0.09 (0.04)	**−0.46 (0.10)**	**0.22 (0.09)**	0.16 (0.19)	−0.28 (0.20)	−0.22 (0.06)	**0.04 (0.03)**	−1.47 (0.21)	0.15 (0.11)
TKA	**−0.80 (0.22)**	0.10 (0.09)	−0.71 (0.15)	0.33 (0.27)	−0.25 (0.15)	0.18 (0.11)	**−0.08 (0.18)**	**0.58 (0.27)**	0.33 (0.19)	−0.31 0.10	−0.27 (0.16)	**0.25 (0.18)**	−1.65 (0.28)	0.17 (0.05)

Muscle Force (N/BW)

	GlutMax	**GlutMin**	**GlutMed**	**BFLH**	**Iliacus**	**Psoas**	**TFL**	**RecFem**	**VasLat**	**VasMed**	**VasInt**	**GasMed**
Control	0.43 (0.12)	**0.27 (0.06)**	1.32 (0.28)	**0.24 (0.05)**	**0.23 (0.05)**	**0.25 (0.05)**	0.09 (0.04)	0.16 (0.08)	0.81 (0.24)	0.39 (0.22)	0.20 (0.04)	**0.67 (0.24)**
TKA	0.46 (0.07)	**0.39 (0.10)**	1.43 (0.18)	**0.45 (0.12)**	**0.50 (0.16)**	**0.54 (0.24)**	0.13 (0.03)	0.25 (0.22)	0.60 (0.66)	0.39 (0.30)	0.25 (0.21)	**1.32 (0.47)**

Bold indicates significant difference across groups ($\alpha < 0.05$). Shaded indicates large effect across groups ($d > 0.8$).

Table 4

Peak inverse kinematics, inverse dynamics, and muscle forces (mean±SD) during the Outside Pivot in patients with unilateral TKA and controls.

	Inverse Kinematics													
	Hip						*Knee*						*Ankle*	
	Ext	Flex	Abd	Add	Ext Rot	Int Rot	Ext	Flex	Varus	Valgus	Ext Rot	Int Rot	Dors	Plant
Control	−16.9° (3.7°)	19.5° (4.8°)	−11.1° (3.6°)	−2.1° (5.0°)	−17.8° (9.0°)	8.4° (11.1°)	**17.3° (7.6°)**	47.4° (3.3°)	**7.3° (2.2°)**	**1.9° (2.6°)**	−19.4° (10.7°)	10.7° (5.8°)	−4.6° (2.4°)	19.1° (4.0°)
TKA	−22.6° (9.1°)	13.1° (11.2°)	−9.2° (4.5°)	2.6° (6.0°)	−25.3° (8.2°)	2.0° (9.0°)	**9.0° (2.2°)**	40.7° (8.0°)	**0.4° (1.4°)**	**−5.8° (3.7°)**	−15.3° (9.6°)	7.6° (3.5°)	−7.9° (7.2°)	16.8° (3.6°)

	Inverse Dynamics (N*m/kg)													
	Hip						*Knee*						*Ankle*	
	Ext	Flex	Abd	Add	Ext Rot	Int Rot	Ext	Flex	Varus	Valgus	Ext Rot	Int Rot	Dors	Plant
Control	**−0.53 (0.14)**	0.12 (0.13)	−0.76 (0.21)	0.09 (0.09)	**−0.05 (0.02)**	0.15 (0.07)	−0.50 (0.22)	0.32 (0.20)	0.11 (0.05)	−0.45 (0.23)	−0.10 (0.06)	**0.03 (0.02)**	**−1.48 (0.36)**	0.15 (0.12)
TKA	**−0.83 (0.19)**	0.45 (0.41)	−0.72 (0.29)	0.27 (0.19)	**−0.19 (0.09)**	0.23 (0.13)	−0.36 (0.16)	0.38 (0.13)	0.28 (0.20)	−0.39 (0.29)	−0.22 (0.08)	**0.20 (0.15)**	**−1.01 (0.26)**	0.18 (0.08)

	Muscle Force (N/BW)											
	GlutMax	GlutMin	GlutMed	BFLH	Iliacus	Psoas	TFL	RecFem	VasLat	VasMed	VasInt	GasMed
Control	0.40 (0.09)	0.30 (0.10)	1.24 (0.33)	0.26 (0.19)	**0.20 (0.12)**	**0.21 (0.14)**	0.17 (0.07)	0.30 (0.16)	0.93 (0.27)	0.32 (0.08)	0.19 (0.05)	0.38 (0.12)
TKA	0.32 (0.07)	0.45 (0.25)	1.34 (0.47)	0.39 (0.31)	**1.00 0.26**	**1.17 (0.32)**	0.23 (0.11)	0.27 (0.15)	0.97 (0.42)	0.58 (0.34)	0.27 (0.14)	1.01 (0.76)

Bold indicates significant difference across groups ($\alpha < 0.05$). Shaded indicates large effect across groups ($d > 0.8$).