A BIOMECHANICAL ANALYSIS OF TOTAL AND UNICOMPARTMENTAL KNEE ARTHROPLASY PATIENTS DURING STAIR NEGOTIATION COMPARED TO HEALTHY CONTROLS

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DEDICATION

To my mama…

My sweet MC, although you won't be with us in person to watch me walk across that stage when I get this diploma, I know you are with me in spirit every step of the way, as you are every day. There isn't a day that goes by that I don't miss you or long to hear your voice. I am so grateful and cherish so many memories together, especially of your recent visits to us here in Hawaii. We know one of the reasons God called us here, is because He knew how much joy it would bring to the last couple years of your life.

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ABSTRACT

Three-dimensional biomechanical gait analysis is an assessment tool that provides insight into patient functioning following a total knee arthroplasty (TKA) or a unicompartmental knee arthroplasty (UKA). Knee flexion moment is a biomechanical variable that provides insight into an individuals' willingness to load the knee joint. One challenge of the data collection process with these patients suffering from osteoarthritis is function, especially pre-operatively, is limited due to pain and fatigue which can restrict the researcher's ability to capture the required information. Additionally, how do both operative groups recover in terms of stair negotiation ability? Stairs are known to be a more challenging task that occurs with aging, and is even more challenging in osteoarthritis suffers. The degree to which an individual is able to perform the stair negotiation task in the absence of pathology remains in question. Results of this dissertation provide recommendations of the biomechanical data collection process. In patients that present with lower extremity joint pain and/or fatigue, identifying the force plate during the data collection process has limited clinical outcomes on biomechanical variables and will limit the number of redundant trials. Through using stair negotiation as an assessment tool, short-term (three months following surgery) functional ability favors those patients undergoing UKA. These UKA patients have knee flexion moments that are more similar to healthy controls. Furthermore, functional deficits in knee flexion moment remain in TKA patients out to one-year post-operatively when compared to healthy age-matched controls. Results of this dissertation also suggests that the long-term difficulty of the stair task in TKA patients is more related to the osteoarthritis pathology than the aging process as evident by the ability of all of the healthy controls participants to negotiate the stairs with ease.

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CHAPTER 1:

THE EFFECT OF TARGETING THE FORCE PLATE ON WALKING AND RUNNING BIOMECHANICAL VARIABLES

Abstract

Context: During biomechanical gait analysis research studies, participants are often instructed unware of the force plate. Biomechanists do not visually identify the force plate to combat targeting which may have a direct effect on biomechanical variables of interest. **Objective:** To evaluated the effect of visually targeting the force plate, without specifically altering step or stride length on walking and running gait kinetics and kinematics in a healthy population. **Design:** Experimental. **Setting:** Biomechanics laboratory **Participants:** Twenty-one young community dwelling adults (males=12), 20-39 years old (mean years ± SD: 24 years ± 4.23) volunteered for this research study. **Intervention:** Three dimensional gait kinematics (240 Hz) and kinetics (960 Hz) were collected on participants as they performed walking at self-selected velocity and running at a velocity of 1 meter/second. They performed both of these activities twice, once while unaware of the force plate and again after the force plate was identified to them. **Main Outcome Measures:** The effect of targeting on spatiotemporal parameters and lower extremity kinematics and kinetics. **Results:** During the target walking condition, participants walked at a faster velocity (p=0.018) and produced significantly higher vertical ground reaction force (vGRF)(p=0.033) and anterior/posterior ground reaction forces (A/P GRF) (p=0.021). Additionally, hip angle (p=0.003) and knee adduction moment (KAM) (p=0.018) were significantly greater. During running, A/P GRF was significantly lower (p=0.008) and vGRF was nearly significantly lower (p=0.059) in the targeting condition. Ankle angle (p=0.054) and knee flexion angle at initial contact (p=0.052) were greater at a level approaching significance in the running targeting condition. **Conclusions:** Identification of the force plate did result in some changes in the gait variables examined. However, differences did not create clinical significant differences in the gait variables examined. In walking, observed differences between conditions were small and may be attributed to differences in walking velocity. Running velocity was controlled for in the current study which resulted in decreased gait variability between conditions. Identification of the force plate may be beneficial in populations where fatigue or pathology limit the ability perform multiple trials without significantly changing gait characteristics.

Introduction

Participants in gait research studies are often given instructions to "walk with your head and eyes forward at a comfortable pace" and to "walk as naturally as possible"[1]. There is usually no mention of the force plate to participants to avoid targeting. Targeting the force plate can be defined as identifying the outline of the force plate to the participant, prior to collecting data, so that they can visually guide their steps to deliberately contact within the force plate boundary, potentially altering their natural gait in attempt to accomplish this task[2,3]. To combat targeting, biomechanists do not visually identify, instruct, or draw the participants' attention to the force plate embedded in the laboratory floor. Further, trials in which a participant is observed to alter their stride in order to target the force plate are commonly excluded.

The effect of targeting in walking and running gait parameters has been previously researched with limited consensus[1-4]. Targeting the force plate has been reported to have no significant impact on spatiotemporal parameters during walking[1,4]. However, in running, significant differences in spatiotemporal parameters were found between the short and long strides when compared to the normal stride lengths, demonstrating that subjects made adjustments to their stride to strike the force plate. Significant differences in foot, shank and thigh angle were found at initial contact in these three conditions though no differences were found in the angles of the foot, shank and thigh during the propulsion phase[3]. No differences have been reported between normal and targeting conditions for ground reaction force (GRF) vectors and the timings of the forces in walking or running gait studies[1-4].

The use of appropriate targeting instruction may decrease the number of trials necessary during the data collection period preventing fatigue to the participants who consistently miss the force plate during typically conducted gait trials that prohibit targeting. However, it is not clear from previous research if the use of appropriate targeting strategies during gait trials may be employed without affecting the validity of the results. Identification of such strategies would serve to make the collection of gait data more efficient. Therefore, the purpose of this study is to identify the effect of visually targeting the force plate, without specifically altering step or stride length on walking and running gait kinetics and kinematics in a healthy population.

Methodology

Participants

Twenty-one young adults (males=12), 20-39 years old (mean years ± SD: 24 years ± 4.23), volunteered for the one-time data collection which was conducted to assess the effects of targeting on biomechanical walking and running GRF kinetics and kinematics. Body mass of the participants was 71.58 kilograms ± 16.90, height was 1.702 ± 0.09 meters and body mass index was 24.4 ± 4.04. Inclusion criteria for all participants included: 1) between 20 and 40 years of age, 2) no history of lower extremity injury or surgery within the past six months, 3) no medical history of neurological disorders 4) able to run continuously for 10 minutes and 5) no expectation of pregnancy. Prior to the study, all participants signed informed consent approved by the Institution's Committee on Human Studies. Due to the nature of this study, participants were not informed prior to data collection that the purpose of this study is to examine force plate targeting strategies without biasing the results. Therefore, the informed consent form did not include identification of the comparison between non-targeting and targeting conditions that will take place.

Procedure

All biomechanical analyses were conducted the University Gait Laboratory. Walking and running gait were collected using 29 retroreflective markers placed on the thorax, pelvis and lower extremities and four marker arrays on the thigh and shank segments. Kinematic data were collected with a Vicon motion capture system and Vicon Nexus software (Vicon, Inc., Centennial, CO) at 240 Hz and time synchronized with kinetic data collected at 960 Hz collected from on force plates (Advanced Mechanical Technology Incorporated, Boston, MA) embedded flush with the floor. All kinetic data were smoothed using a Butterworth filter with a 10 Hz cut-off and processed using Visual 3D (C-Motion, Inc., Germantown, MD).

Once the markers were applied, participants traversed a four-meter data collection field under multiple conditions in non-standardized running shoes. The first condition, non-targeting, took place with the participant unaware of the force plate. Under this condition, they were instructed to walk at a "comfortable speed" and to walk with their "head up and eyes looking at the 'X'" located on the laboratory wall 15.5 meters away. Walking velocity was recorded using infrared timers (Speed Trap II, Brower Timing Systems, Draper, UT, USA). For the purpose of this study, only trials in which the dominant foot successfully lands within force plate boundary, without a visible change in gait in attempt to target the force plate will be included. Trials were repeated until three successful walking trials had been collected. After the successful walking trials, normal non-targeting running trials were then collected as participants ran at 4 meter/second ± 10% through the data collection field. They again were asked to run with their "head up and their eyes looking ahead at the 'X' on the wall." Successful running trials followed the same protocol for walking trials. Three successful running trials in which the participants' entire dominant foot contacts the force plate were collected.

The data collection protocol for the targeting condition replicated the first condition for both walking and running trials except that the force plate was identified to participants. They were instructed to "walk or run as normal as possible and to be sure that your entire dominant foot lands within the force plate boundary without changing your stride." Two trained gait biomechanics researchers observed all gait trials to assess stride alteration, as indicated by the participants shuffling their feet, lunging or breaking their stride in any other way to target the force plate by the participants. If the two biomechanics researchers agreed that a gait alteration occurred during the trial, the trial will be repeated. Those trials in which the examiners visually observed stride alterations, the participants were asked if they felt that they made contact with the force plate "in-stride." If they responded "yes" to that question, the trial was counted and used for the purpose of data analysis. If the answer was "no" it was not contacted "in-stride" the trial was discarded and repeated. Participants completed the minimum number of trials necessary to obtain three acceptable trials for both walking and running conditions. No other verbal cues were given to ensure consistency between all participants. Kinematic data were collected at 240 Hz and time synchronized with kinetic data collected at 960 Hz[5]. All kinematic and kinetic data were smoothed using a Butterworth filter with a 10 Hz cut-off and ground reaction force was filtered using a 50 Hz cut-off frequency[5]. External joint moments were calculated using inverse dynamics based on marker trajectories and kinetic data which was also filtered using a 10 Hz cut-off frequency.

Statistical Analysis

Multiple, Student t-tests were used to assess changes in biomechanical parameters between the two conditions, non-targeting and targeting, for walking and running. Alpha levels were set at $p<0.05$ for all analyses. All statistical analyses were conducting using SPSS version 23.0 (IMB, Armonk, NY, USA). All moments reported are external.

Results

Descriptive statistics for walking and running gait variables are presented in Table 1 and Table 2, respectively. During walking targeting trials, targeting of the force plate that resulted in visible stride alterations was identified by one of the trained biomechanists, in 21 of the 72 (29%) of the total walking trials. Similarly, during running targeting trials, 22 of the 72 (31%) total trials were identified as containing visible stride alterations. These trials were included in the data analysis because the participant felt that they hit the force plate "in-stride" without alteration. Additionally, the total number of trials need to complete three successful trials were statistically lower during targeting for both walking ($p \leq 0.01$) and running ($p \leq 0.01$) conditions.

During walking targeting condition, participants walked at a faster velocity ($p \leq 0.05$) and produced significantly higher vertical ground reaction force (vGRF)($p \leq 0.05$) and anterior/posterior ground reaction forces (A/P GRF) ($p \leq 0.05$). Additionally, hip angle ($p=0.003$) and knee adduction moment (KAM) ($p \leq 0.05$) were significantly greater in walking targeting conditions. During running, A/P GRF was significantly lower ($p \leq 0.01$) and vGRF was nearly significantly lower ($p \leq 0.05$) in the targeting condition. Ankle angle ($p \leq 0.05$) and knee flexion angle at initial contact ($p \leq 0.05$) were greater at a level approaching significance in the running targeting condition.

Table 1.1 Descriptive Statistics for Walking and Target Walking Variables

	Walk			Target Walk			
	Mean	±	SD	Mean	±	SD	P-value
Spatiotemporal parameters							
Loading Rate (seconds)	0.15	±	0.02	0.15	±	0.03	0.097
Stance (seconds)	0.66	±	0.03	0.65	±	0.05	0.200
Walking Velocity (m/s)	1.31	±	0.11	1.40	±	0.15*	**0.018**
Stride Width (m)	0.11	±	0.01	0.11	±	0.02	0.649
Stride Length (m)	2.25	±	1.40	2.27	±	1.47	0.809
Number of Trials	5.57	±	1.47	4.48	±	1.13**	**0.002**
Kinematic and Kinetics							
Hip Angle (degrees)	36.92	±	6.40	38.31	±	6.51**	**0.003**
Knee Angle (degrees)	1.71	±	4.99	2.31	±	5.23	0.143
Peak Knee Flexion Moment (Nm/kg)	0.74	±	0.18	0.77	±	0.25*	**0.037**
Knee Adduction Moment (Nm/Kg)	0.45	±	0.13	0.50	±	0.12*	**0.018**
Ankle Angle (degrees)	1.70	±	4.99	2.25	±	5.27	0.196
A/P GRF (N/kg)	1.92	±	0.35	2.13	±	0.14*	**0.021**
Vertical GRF (N/kg)	10.99	±	0.53	11.27	±	0.58*	**0.033**

SD = standard deviation; m/s = meters per second; m = meters;

Nm/kg = newton meters per kilogram; A/P Anterior/Posterior, N/kg = newtons per kilogram

* = significant difference between conditions ($p \leq 0.05$).

** = significant difference between conditions ($p \leq 0.01$).

Table 1.2 Descriptive Statistics for Running and Target Running Variables

	Run			Target Run			
	Mean	±	SD	Mean	±	SD	P-value
Spatiotemporal parameters							
Loading Rate (seconds)	0.16	±	0.01	0.10	±	0.01	0.256
Stance (seconds)	0.22	±	0.02	0.23	±	0.02	0.200
Run Velocity (m/s)	3.70	±	0.48	3.59	±	0.58	0.505
Stride Width (m)	0.28	±	0.04	0.19	±	0.05	0.303
Stride Length (m)	3.88	±	0.43	4.08	±	0.42	0.411
Number of Trials	9.80	±	3.06	4.90	±	1.41**	**0.000**
Kinematic and Kinetics							
Hip Angle (degrees)	50.36	±	6.50	51.32	±	6.76	0.229
Knee Angle (degrees)	2.36	±	14.46	5.79	±	10.06	0.052
Peak Knee Flexion Moment (Nm/kg)	2.74	±	0.70	2.87	±	0.47	0.209
Knee Adduction Moment (Nm/Kg)	0.89	±	0.31	0.86	±	0.32	0.559
Ankle Angle (degrees)	2.40	±	14.41	5.79	±	10.06	0.054
A/P GRF (N/kg)	3.53	±	0.52	3.14	±	0.63**	**0.008**
Vertical GRF (N/kg)	25.21	±	2.19	24.78	±	2.14	0.059

SD = standard deviation; m/s = meters per second; m = meters;

Nm/kg = newton meters per kilogram; A/P Anterior/Posterior, N/kg = newtons per kilogram

* = significant difference between conditions ($p \leq 0.05$).

** = significant difference between conditions ($p \leq 0.01$).

Discussion

Despite some significant differences, the most important result of this study was that there were limited clinical differences in walking and running gait characteristics between targeting and non-targeting conditions. Additionally, significantly fewer trials were required in both targeting conditions. These results, consistent with previously reported findings[4], indicate that identifying the force plate during biomechanical assessment of certain populations suffering from severe pathology or that fatigue easily, may be appropriate based on the limited effect of targeting on gait variables.

Although each participant was instructed to not purposely alter stride during the targeting condition in order to contact the identified force plate, purposeful stride alteration was observed in 29% of walking trials and 31% of running trials during the targeting condition. Visual inspection of the data determined no presumed differences between those with in-stride trials and trials identified as stride alterations during the targeting condition. These findings are similar to those reported by Wearing et al.[4] and Verniba et al.[1], who reported no significant differences between non-targeting and targeting walking trials. Based on this evidence, trials with visible stride alterations during the targeting conditions were included in the statistical analysis in this study.

Significant differences in biomechanical variables between conditions at the hip and knee during walking were limited in the current study. These differences consisted of two degrees of increased hip flexion and an increase of .05 Nm/kg in KAM in the targeting condition. Previous research has identified increases in both of these variables with increased walking velocity[6,7]. Participants in the current study walked at a faster velocity, perhaps due to more familiarity with the data collection procedure, in the targeting condition compared to non-targeting trials. Therefore, the differences observed in walking trials in the current study may be attributed to increase in walking velocity and not due to targeting. These differences were not found between conditions during running trials, possibly due to running velocity being controlled at 4 m/sec.

Similarly, during walking trials, A/P GRF and vGRF were significantly higher during targeting condition though there were no differences in spatiotemporal parameters. These results differ from previous research showing no differences in GRF magnitude when walking velocity was controlled[2,4]. However, walking velocity in the current study was significantly greater in the targeting condition, which has previously been associated with increases in GRF[8]. In running trials, only A/P GRF was affected, but was significantly lower during targeting. Although not significantly different between conditions, stride length in the targeting condition was slightly decreased during non-targeting which may account for the observed difference in A/P GRF.

Significantly fewer trials were performed in both targeting conditions which is advantageous when conducting biomechanical studies using participants suffering from pathology, pain, fatigue or recovering from surgery. For instance, in patients suffering from knee osteoarthritis (OA), the gait variable KAM is a variable of interest, with higher values of KAM representing increased joint loading and progression of knee OA[9]. Mobility in these patients is often limited due to pain and inflammation which affects the knee joint. The ability to decrease the number of redundant trials needed to evaluate KAM would be beneficial in this population during gait analysis.

Conclusion

Identification of the force plate did result in some targeting by participants however, it did not create clinically significant differences in the gait variables examined. Observed differences in walking between conditions were small and may be attributed to differences in walking velocity. Running velocity was controlled for in the current study which resulted in decreased gait variability between conditions. The targeting condition limited the number of redundant trials in both walking and running. Therefore, identifying the force plate may be beneficial in populations where fatigue or pathology limit the ability perform multiple trials without affecting gait characteristics.

CHAPTER 2:

A SHORT TERM POST-OPERATIVE BIOMECHANICAL ANALYSIS DURING STAIR NEGIATION IN PATIENTS UNDERGOING TOTAL OR UNICOMPARTMENTAL KNEE ARTHROPLASTY COMPARED TO HEATLHY CONTROLS

Abstract

Context: Total knee arthroplasty (TKA) is the surgical procedure performed in patients with osteoarthrosis (OA) present throughout the entire knee joint. When the OA is contained to the medial compartment, a unicompartmental knee arthroplasty (UKA) is performed. During walking gait, post-operative functioning favors the UKA over TKA. However, questions remain regarding the function of arthroplasty patients during an increasingly demanding tasks like stair negotiation. Therefore, the purpose of this study is to compare sagittal plane knee biomechanics during stair negotiation between TKA and UKA patients compared to healthy controls.
Objective: To compare the stair negotiation in TKA and UKA patients to age-matched controls. **Design:** Longitudinal gait analysis. **Setting:** Biomechanics laboratory. **Patients:** Fourteen TKA patients, nine MR (seven males,11 knees) implant and five SR implant (four males, eight knees) were compared to 30 controls (15 males, 15 knees). **Intervention:** Patients randomly received either SR (GetAroundKnee™, Stryker Orthopedics, Mahwah, NJ) or MR (Balanced Knee® System, Ortho Development Corporation, Draper, UT) implants. All UKA patients received the Oxford Unicompartmental Implant Oxford® Partial Knee Implant (Zimmer Biomet Orthopedics, Warsaw, IN). All arthroplasty patients underwent 3D motion gait analysis during a three-step staircase pre-TKA and post-TKA at six-weeks and three-months. Control data was collected at a one-time data collection. Multiple multivariate general linear model tests were used to compare variables of interest, with an alpha level at $p<0.05$, to determine differences between arthroplasty groups and controls at each time period. **Main Outcome Measures:** Knee flexion angle (PKFA), knee flexion moment (PKFM), vertical ground reaction force, trunk forward flexion, trunk side bending, strength and time measurements. **Results:** Pre-operative deficits were present in both TKA and UKA groups during stair ascent and stair descent ($p<0.05$). Six-weeks following surgery both PKFM was significantly decreased ($p<0.01$) during stair negotiation in both TKA and UKA groups. During stair ascent three-months after surgery, the TKA group had statistically ($p<0.01$) decreased Max vGRF and PKFM as well as statistically decreased trunk forward flexion and time to complete task. Compared to the UKA group which demonstrated ($p<0.01$) a decreased PKFM and increased time to complete stair ascent. During stair descent significant ($p<0.01$) deficits remained in the TKA group compared to controls. Whereas the UKA group had no significant differences in any biomechanical variable compared

to controls **Conclusion:** Results of this study indicate that short-term post-operative stair negotiation function favored the UKA patients as they were similar to healthy controls as soon as three-months following knee replacement surgery. Total knee arthroplasty patients displayed more compensatory motions including decreased knee extensor strength, decreased PKFM as well as an increased in both trunk compensatory motions and increased time to negotiate stairs suggesting that important deficits remain during stair negotiation tasks in TKA patients three-months post-operatively. Regaining knee extensor strength is recommended as the focus of rehabilitation programs in post-operative TKA patients. Improvements in post-operative function in UKA patients may be attributed to the minimally invasive procedure which decreases recovery time[10-12] and the surgical process which retains cruciate ligaments[13] which may contribute to an increase in function during highly demanding tasks. Unicompartmental knee arthroplasty may be used in some cases as an alternative to TKA and favorable functional short-term post-operative outcomes may be expected.

Introduction

Individuals with end stage osteoarthritis (OA) present throughout the entire knee joint undergo a total knee arthroplasty (TKA). When the OA is contained to the medial compartment of the joint, unicompartmental knee arthroplasty (UKA), or partial knee joint replacement is performed. Longevity of the UKA implant has been reported to be similar to TKA[14]. Post-operative function favors the UKA surgical procedure and is attributed to a minimally invasive procedure which decreases recovery time[10-12] and patients report fewer post-operative complications[15]. It is estimated that approximately 75% of the knee during UKA surgical procedures remains untouched which leads to a more natural knee motion and improvements in range of motion, walking gait and improvements in overall patient satisfaction following surgery[11,16]. In addition to the minimally invasive procedure, the cruciate ligaments remain untouched during a UKA procedure, which contain mechanoreceptors that contribute to joint proprioception, neuromuscular control and play an important role in functional stability of the knee joint[13].

Prior to surgical procedures, physical function and mobility is low in both UKA and TKA patients, in fact, no pre-operative differences have been reported[15]. However, after surgery, improvements in the ability to kneel, negotiate stairs and improvements in physical activity involvement favors the UKA procedure[17,18]. Biomechanical studies evaluating walking gait in post-UKA individuals, have reported that sagittal plane knee flexion/extension range of motion and knee flexion moments are no different when compared to healthy controls which have been attributed to the retention of the anterior cruciate ligament (ACL)[10,19,20]. Wiik et al investigated downhill walking and reported the UKA patient's knee mimicked the normal knee compared to TKA patients performing the same task[21]. However, biomechanical questions remain regarding UKA patient function with an increase in demanding physical activity like stair negotiation.

Stair negotiation was used in this biomechanical analysis to assess post-operative function in both TKA and UKA patients because it has been shown to be a strong predictor of functional decline in older adults[22] and is a more difficult task which places a high demand on knee extensor musculature[23]. Following surgery, both TKA and UKA patients experience strength deficits[24,25] which can make negotiating stair more difficult following surgery, but the extent to which this weakness affects patient function compared to healthy controls is limited. Therefore, the purpose of this study is to compared sagittal plane knee biomechanics during stair negotiation between those patients undergoing TKA or UKA compared to healthy age-matched controls.

Methodology

Participants

A randomized, longitudinal design was conducted consisting of OA patients undergoing either a TKA (n=15, 20 knees) or UKA (n=7, 9 knees) arthroplasty for the treatment of OA. A gait assessment was performed within one week prior to arthroplasty and post-arthroplasty at six-weeks and three-months. Inclusion criteria for arthroplasty patients included: 1) under 75 years of age, 2) no previous history of lower extremity fracture, osteotomy, or joint replacement, 3) undergoing an unilateral or bilateral knee joint arthroplasty for the treatment of osteoarthritis, and 4) able to walk without an aid. The same board certified orthopedic surgeon screened each patient and perform all TKA procedures. Prior to enrollment in the study, all patients and healthy controls signed informed consent forms approved by the Institution's Committee on Human Studies.

Patients were screened for inclusion within the study and underwent the first data collection prior to surgery. The TKA patients were randomly allocated to receive either a Single-radius (SR) implant (GetAroundKnee™, Stryker Orthopedics, Mahwah,NJ) or a Multi-radius (MR) implant (Balanced Knee® System, Ortho Development Corporation, Draper, UT). The Oxford Unicompartmental Implant Oxford® Partial Knee Implant (Zimmer Biomet Orthopedics, Warsaw, IN) was used for UKA patients. All implants used for this study were approved by the FDA and will be used in accordance with their FDA approval. Additionally, 30 age-matched community members were recruited by word of mouth to serve as healthy, age-matched controls. Inclusionary criteria for the healthy controls included: 1) no history of lower extremity joint surgery, 2) no history or treatment of diagnosed arthritis, 3) no diagnosed neurological or balance disorders and 4) no physical activity restrictions from their physician.

Procedures

Upon arrival to the gait laboratory anthropometric data was collected including height using a wall mounted stadiometer (Model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA) and body mass was determined using a Detecto certifer scale (Webb City Mo, USA). Shank lengths were determined from lateral joint line to the most proximal portion of the lateral malleolus and a mark was placed at 80% of shank length. These markings served as location points for the hand held dynamometer during knee extensor strength testing. This allowed for consistent placement of the handheld dynamometer, relative to each individual. The UCLA Activity Score, an ordinal survey from 1-10 to describe activity level, was used to assess self-reported overall functional and physical activity[26] and was completed by both the TKA and control participants. Higher UCLA scores indicate a higher amount of rigorous activity level, with choice #10 stating: "Regularly participates in impact sports."

All biomechanical analyses were collected using 29 retroreflective markers placed on the thorax, pelvis and lower extremities and four marker arrays on thigh and shank segments, with a Vicon motion capture system and Vicon Nexus software (Vicon, Inc., Centennial, CO). Kinematic data were collected at 240 Hz and time synchronized with kinetic data collected at 960 Hz from forces plate (Advanced Mechanical Technology Incorporated, Boston, MA) embedded flush with the floor at the bottom of the stairs and also inserted within the second step of the stairs. All kinematic data were smoothed using a low-pass Butterworth filter with a 10 Hz cut-off and ground reaction force was filtered using a 50 Hz cut-off frequency. External joint moments were calculated using inverse dynamics based on marker trajectories and kinetic data which was also filtered using a 10 Hz cut-off frequency. All data was processed using Visual 3D (C-Motion, Inc., Germantown, MD). All knee joint moments are reported as external moments. Knee and trunk flexion values are reported as a positive number. Additionally, during trunk side bending, a positive value indicates trunk motion towards the stance leg.

Stair negotiation trials followed a similar protocol performed by Vallabhajosula et al.[27]. The laboratory stairs included three steps with the following dimensions: step rise, 18 cm; step width, 46 cm; step tread, 28 cm. Each participant was barefoot and began walking at a self-selected velocity taking three steps before ascending the stairs. The second step was also instrumented with a force plate to measure the second step of the involved limb during both stair ascent and descent. Patients were instructed to walk up the stairs "as quickly and as safely as possible." Each patient was asked to take two additional steps on the stair platform to ensure a natural gait is continued through the last step and deceleration did not occur. For stair descent, patients took a step on the stair platform prior to stepping down with the involved limb. An additional three steps were taken after completion of the stair descent trials. Handrails were provided for safety but patients were instructed not to use them unless balance was compromised. If the handrails were used, the trial was discarded and not included for data analysis. Additionally, subjects who could not complete stair negotiation due to pain or unwillingness and those who did not maintain continuous movement throughout stair negotiation were included in data analysis using calculated maximal values for biomechanical variables of interest. Due to high intra-subject variability previously reported during stair climbing in the osteoarthritis population, five successful trials were averaged.

Bilateral muscle torque was then assessed using a hand held dynamometer (HHD) (Hoggan Health Industries, West Jordan, UT), performed in a gravity dependent position the muscular testing. Knee extension torque was measured while the patient was seated with the knee placed at 60° of knee flexion. The HHD was placed on the anterior shank, just proximal to the medial malleolus and secured with a strap. The patient was instructed to extend their knee, without extending their trunk. For each strength measure, the patient was asked to maximal contraction for three seconds. Following a submaximal familiarization trial, two to three maximal trials were recorded, and the peak value was used for data analysis. Patients had a 30 second rest in between trials. Pain level was also assessed after each muscular torque trial using a visual analog scale.

Statistical Analysis

Normality was assessed using Shapiro-wilk test. To test for homogeneity of variance between standard deviations the Levene's test was used. In order to allow inclusion in statistical analyses of participants who were unable to complete stair negotiation at six-weeks following surgery, a maximal "ceiling" value was calculated for each variable of interest. Ceiling values were calculated as four standard deviations above or below the pooled TKA and UKA group mean for each variable. Each participant who was unable to complete stair negotiation was assigned the calculated ceiling value for each biomechanical variable of interest. Multiple Multivariate General Linear Model tests tested for significance. Post-hoc Tukey tests determined where significant differences existed among groups and the dependent variables. All data were analyzed using SPSS Version 22.0 with an alpha level of $p<0.05$.

Results

A total of 16 TKA (13 males, 20 knees) and seven UKA (five males, 9 knees) patients were included for data analysis and were compared to 22 (15 males, 22 knees) healthy controls (CON). Patient demographics did not different between groups and can be found in Table 2.1. Please see Appendix A for tables with all associated p-values for subsequent tables for this chapter.

Pre-operatively, compared to controls during stair ascent, the TKA group had a decreased Max vGRF (CON = 11.57 Nm/kg, TKA = 10.27 Nm/kg, $p < 0.05$). A decreased PKFM was present in both TKA (CON = 1.06 Nm, TKA = 0.60 Nm, $p<0.01$) and UKA (CON = 1.06 Nm, UKA = 0.76 Nm, $p<0.05$) groups. Additionally, there was an increased peak trunk flexion in TKA (CON = 18.51°, TKA = 27.26°, $p<0.01$) and UKA (CON = 18.51°, UKA = 25.62°, $p<0.01$), increased peak trunk side bending in TKA (CON = 2.84°, TKA = 6.18°, $p <0.01$) and UKA (CON = 2.84°, UKA = 8.02°, $p<0.01$), as well as an increased time to complete stair ascent in TKA (CON = 1.91 s, TKA = 2.96 s, $p<0.01$) and UKA (CON = 1.91 s, UKA = 3.49 s, $p<0.01$) groups. Pre-operative stair descent descriptive variables are located in Table 2.2.

During stair descent pre-operatively a decreased Max vGRF was present in both TKA (CON = 15.21 Nm/kg, TKA = 12.09 Nm/kg, $p<0.05$) and UKA (CON = 15.21 Nm/kg, UKA = 9.08 Nm/kg, $p<0.01$). A decreased in PKFM during the first 25% of stance was present in both TKA (CON = 1.32 Nm, TKA = 0.66 Nm, $p<0.01$) and UKA (CON = 1.32 Nm, UKA = 0.68 Nm, $p<0.01$) as well during the first 50% of stance in both TKA (CON = 1.39 Nm, TKA = 0.75 Nm, $p<0.01$) and UKA (CON = 1.39 Nm, UKA = 0.81, Nm, $p<0.01$) groups. The TKA group also experienced an increased peak trunk side bending (CON = 1.82°, TKA = 10.47°, $p<0.01$), time on the force plate (CON = 0.72 s, TKA = 1.59 s, $p<0.01$) and an overall increased time to complete the stair descent (CON = 1.14 s, TKA = 2.29 s, $p<0.05$) compared to the healthy controls. The UKA group had an increased peak trunk flexion (CON = 10.91°, UKA = 23.54°, $p<0.05$), increased peak trunk side bending (CON = 1.82°, UKA = 12.90°, $p<0.01$), increased time on the force plate (CON = 0.72 s, UKA = 2.22 s, $p<0.01$), time to Max vGRF (CON = 0.12 s, UKA = 0.75 s, $p<0.01$), and increased time to complete stair descent (CON = 1.14 s, UKA = 3.32 s, $p<0.01$) compared to healthy controls. When compared to the TKA group, the UKA group demonstrated an increased time to complete stair descent (TKA = 2.29 s, UKA = 3.32 s, $p<0.05$). Pre-operative stair descent descriptive statistics are located in Table 2.3.

Table 2.1 Participant Demographics

	Controls (n=22, 22 knees)			TKA (n=15, 20 knees)			TKA to CON	UKA (n=7, 9 knees)			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Age	67.3	±	4.7	65.0	±	4.8	0.333	68.0	±	3.7	0.951	0.384
Height (mm)	1.7	±	0.1	1.7	±	0.1	0.827	1.7	±	0.1	0.986	0.830
Body Mass (kg)	75.1	±	15.3	82.5	±	17.5	0.466	85.6	±	16.0	0.309	0.853

TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty,
n = number, SD = standard deviation, mm = millimeters, kg = kilograms

Table 2.2 Pre-Operative Biomechanical Variables During Stair Ascent

	Controls (n=22 knees)			TKA (n=20 knees)			TKA to CON	UKA (n=9 knees)			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (Nm/kg)	11.57	±	1.13	10.27	±	1.23*	**0.020**	10.33	±	2.61	0.108	0.993
PKFA (º)	67.10	±	5.86	69.68	±	6.95	0.404	63.91	±	6.57	0.428	0.076
PKFM (Nm/kg)	1.06	±	0.20	0.60	±	0.29**	**0.000**	0.76	±	0.52*	**0.043**	0.416
Peak Trunk Flexion (º)	18.51	±	5.81	27.27	±	5.93**	**0.000**	25.62	±	7.70*	**0.016**	0.677
Peak Trunk Side Bending (º)	2.84	±	2.27	6.18	±	3.86**	**0.005**	8.02	±	3.74**	**0.001**	0.344
Time (s)	1.91	±	0.21	2.96	±	1.30**	**0.003**	3.49	±	1.29**	**0.001**	0.378

TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty, n = number,
SD = standard deviation, Max = maximum, vGRF = vertical ground reaction force, N/kg = newtons per kilogram,
PKFA = peak knee flexion angle, ˚ = degrees, PKFM = peak knee flexion moment, Nm/kg = newton meters per kilogram,
s = seconds

* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).

Table 2.3 Pre-Operative Biomechanical Variables During Stair Descent

	Controls (n=22 knees)			TKA (n=19 knees)			TKA to CON	UKA (n=9 knees)			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	15.41	±	2.33	12.09	±	4.55*	**0.025**	9.80	±	5.63**	**0.002**	0.335
PKFA first 25% of stance (º)	27.51	±	6.75	25.18	±	10.13	0.714	20.18	±	10.34	0.120	0.714
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.66	±	0.41**	**0.000**	0.68	±	0.53**	**0.000**	0.986
PKFA first 50% of stance (º)	32.30	±	5.60	28.84	±	10.81	0.537	27.28	±	17.26	0.453	0.928
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.75	±	0.39**	**0.000**	0.81	±	0.57**	**0.001**	0.942
Peak Trunk Flexion (º)	10.91	±	6.27	19.59	±	15.66	0.080	23.54	±	16.85*	**0.040**	0.720
Peak Trunk Side Bending (º)	1.82	±	2.39	10.47	±	10.48**	**0.005**	12.90	±	12.52**	**0.005**	0.758
Time on Force Plate (s)	0.72	±	0.10	1.59	±	1.16**	**0.008**	2.22	±	1.25**	**0.000**	0.189
Time to Max vGRF (s)	0.12	±	0.04	0.36	±	0.46	0.136	0.75	±	0.68**	**0.001**	0.136
Total Time on Stairs (s)	1.14	±	0.16	2.29	±	1.60*	**0.015**	3.32	±	1.94**^	**0.000**	**0.015**
Knee Extensor Strength (lbs)	85.53	±	26.01	70.78	±	29.90	0.254	81.90	±	37.78	0.949	0.624

SD = standard deviation, TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty, n = number, Max = maximum, vGRF = vertical ground reaction force, N/kg = newtons per kilogram, PKFA = peak knee flexion angle, ˚ = degrees, PKFM = peak knee flexion moment, Nm/kg = newton meters per kilogram, s = seconds

* = significantly different than Controls ($p \leq 0.05$).

** = significantly different than Controls ($p \leq 0.01$).

At six-weeks post-operatively, when the TKA group was compared to controls during stair ascent all biomechanical variables of interest were significantly different ($p<0.01$). The UKA group had an increased PKFM (CON = 1.06 Nm/kg, UKA = 0.59 Nm/kg, $p<0.01$) compared to healthy controls. When compared to the TKA group, the UKA group demonstrated an increased PKFM (TKA = 0.31 Nm/kg, UKA = 0.59 Nm, kg, $p<0.05$), a decrease in peak trunk flexion (TKA = 32.27°, UKA = 23.50°, $p<0.05$), decreased peak trunk side bending (TKA = 11.51°, UKA = 4.89°), $p<0.05$) and decreased time to descend the stairs (TKA = 3.82 s, UKA = 2.59 s, $p<0.05$). Stair ascent six-week descriptive statistics are located in Table 2.4.

During stair descent six-weeks post-operatively, the TKA group had statistically significant in every biomechanical variable of interest ($p<0.01$) compared to healthy controls. The UAK group had a decreased PKFM during the first 25% of stance (CON = 1.32 Nm/kg, UKA = 0.94 Nm/kg, $p = 0.002$), as well significantly decreased PKFM during the first 50% of stance (CON = 1.39 Nm/kg, UKA = 1.02 Nm/kg, $p = 0.003$) compared to healthy controls. The UKA group experienced an increased time on the force plate (CON = 0.72 s, UKA = 1.47 s, $p<0.05$) and increased time to complete stair descent (CON = 1.14 s, UKA = 2.03 s, $p<0.01$). Six-week descriptive statistics during stair descent are located in Table 2.5.

Three months post-operatively, compared to controls during stair ascent a decreased PKFM was statistically significant in both the TKA (CON= 1.06 Nm/kg, TKA = 0.44 Nm/kg, $p<0.01$) and UKA (CON = 1.05 Nm/kg, UKA = 0.61Nm/kg, $p<0.01$). A significantly increased time to complete stair ascent was observed in both TKA (CON = 1.91 s, TKA = 2.41 s, $p<0.01$) and UKA (CON = 1.91 s, UKA = 2.28 s, $p<0.01$) groups compared to healthy controls. Additionally, the TKA group had a significantly increased trunk flexion during stair ascent compared to controls (CON = 18.51°, TKA = 26.92°, $p<0.01$). Stair ascent six- week descriptive statistics are located in Table 2.6.

Table 2.4 Six-Week Biomechanical Variables During Stair Ascent Variables

	Controls (n=22 knees)			TKA (n=19 knees)			TKA to CON	UKA (n=9 knees)			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	11.57	±	1.13	8.73	±	2.62**	**0.000**	10.50	±	1.67	0.344	0.067
PKFA (º)	67.10	±	5.86	58.55	±	10.59**	**0.004**	63.54	±	5.67	0.501	0.280
PKFM (Nm/kg)	1.06	±	0.20	0.31	±	0.31**	**0.000**	0.59	±	0.17**^	**0.000**	**0.023**
Peak Trunk Flexion (º)	18.51	±	5.81	32.27	±	10.57**	**0.000**	23.50	±	3.87^	0.247	**0.020**
Peak Trunk Side Bending (º)	2.84	±	2.27	11.51	±	7.21**	**0.000**	4.89	±	2.29^^	0.532	**0.004**
Time (s)	1.91	±	0.21	3.82	±	1.71**	**0.000**	2.59	±	0.42^^	0.266	**0.018**

TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty, n = number, SD = standard deviation, Max = maximum, vGRF = vertical ground reaction force, Nm/kg = newton meters per kilogram, PKFA = peak knee flexion angle, ˚ = degrees, PKFM = peak knee flexion moment, Nm = newton, s = seconds

* = significantly different than Controls (p ≤ 0.05).

** = significantly different than Controls (p ≤ 0.01).

Table 2.5 Six-Week Biomechanical Variables During Stair Descent Variables

	Controls (n=22 knees)			TKA (n=19 knees)			TKA to CON	UKA (n= 9 knees)			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	15.41	±	2.33	8.28	±	6.08	**0.000**	11.18	±	7.03	0.093	0.333
PKFA first 25% of stance (º)	27.51	±	6.75	18.62	±	8.92**	**0.004**	22.18	±	8.87	0.256	0.534
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.43	±	0.37**	**0.000**	0.75	±	0.52**	**0.001**	0.098
PKFA first 50% of stance (º)	32.30	±	5.60	24.51	±	9.92**	**0.009**	27.14	±	8.45	0.243	0.696
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.55	±	0.45**	**0.000**	0.82	±	0.52**	**0.002**	0.212
Peak Trunk Flexion (º)	10.91	±	6.27	23.77	±	13.82**	**0.001**	19.16	±	11.79	0.136	0.539
Peak Trunk Side Bending (º)	1.82	±	2.39	10.65	±	7.67**	**0.000**	7.28	±	7.04*	**0.054**	0.331
Time on Force Plate (s)	0.72	±	0.10	1.86	±	0.96**	**0.000**	1.47	±	0.87*	**0.025**	0.352
Time to Max vGRF (s)	0.12	±	0.04	0.70	±	0.57**	**0.000**	0.47	±	0.53	0.100	0.373
Total Time on Stairs (s)	1.14	±	0.16	2.41	±	0.98**	**0.000**	2.03	±	0.88**	**0.008**	0.387
Knee Extensor Strength (lbs)	85.53	±	26.01	51.87	±	24.62**	**0.000**	70.88	±	26.88	0.326	0.165

TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty, n = number, SD = standard deviation, Max = maximum, vGRF= vertical Ground Reaction Force, N/kg = newtons per kilogram, PKFA = peak knee flexion angle, ˚ = degrees, PKFM = peak knee flexion moment, Nm/kg = newton meters per kilogram, s = seconds, lbs = pounds

* = significantly different than Controls ($p \leq 0.05$).

** = significantly different than Controls ($p \leq 0.01$).

Table 2.6 Three-Month Biomechanical Variables During Stair Ascent

	Controls (n=22 knees)		TKA (n=17 knees)		TKA to CON	UKA (n=9 knees)		UKA to CON	UKA to TKA
	Mean	± SD	Mean	± SD	P-value	Mean	± SD	P-value	P-value
Max vGRF (N/kg)	11.57	± 1.13	10.53	± 0.49**	**0.001**	11.03	± 0.69	0.269	0.345
PKFA (º)	67.10	± 5.86	66.36	± 4.45	0.903	67.97	± 6.21	0.915	0.750
PKFM (Nm/kg)	1.06	± 0.20	0.44	± 0.16**	**0.000**	0.61	± 0.18**	**0.000**	0.076
Peak Trunk Flexion (º)	18.51	± 5.81	26.92	± 5.43**	**0.000**	22.12	± 4.59	0.230	0.091
Peak Trunk Side Bending (º)	2.84	± 2.27	4.83	± 3.86	0.088	4.23	± 1.69	0.449	0.869
Time (s)	1.91	± 0.21	2.41	± 0.33**	**0.000**	2.28	± 0.13**	**0.001**	0.435

TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty, n = number,
SD = standard deviation, Max = maximum, vGRF = vertical ground reaction force, Nm/kg = newton meters per kilogram,
PKFA = peak knee flexion angle, ˚ = degrees, PKFM = peak knee flexion moment, Nm = newton, s = seconds
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).
TKA = total knee arthroplasty, CON = Control, UKA = unicompartmental knee arthroplasty, n = number,

During stair descent at three-months post-operatively a decreased PKFM during the first 25% of stance was observed in the TKA (CON = 1.32 Nm/kg, TKA = 0.75 Nm/kg, $p<0.01$) group compared to healthy controls. Additionally, a decreased PKFM during the first 50% of stance remained in TKA (CON = 1.39 Nm/kg, TKA = 0.85 Nm/kg, $p<0.01$) group. Furthermore, the TKA group also had a statistically increased trunk flexion (CON = 10.91°, TKA = 18.72°, $p<0.01$), trunk side bending (CON = 1.82°, TKA = 6.51°, $p<0.01$), time on force plate (CON = 0.72 s, TKA = 1.38 s, $p<0.01$), overall time to descend the stairs (CON = 1.14 s, TKA = 1.88 s, $p<0.01$) and a decreased knee extensor strength (CON = 85.53 lbs, TKA = 60.87 lbs, $p<0.05$) compared to healthy controls. There were no significant differences in biomechanical variables of interest between the UKA group and the healthy controls during stair descent. Three-month stair descent descriptive statistics can be found in Table 2.7.

When assessing self-perceived activity level using the UCLA Activity Scores pre-operatively, compared to controls, significantly decreased scores were reported in both the TKA (CON = 7.32, TKA = 4.53, $p<0.01$) and UKA (CON = 7.32, UKA = 4.71, $p<0.01$) groups. At six-weeks post-operatively UCLA scores remained statistically decreased compared to controls in both TKA (CON = 7.32, TKA = 4.14 $p<0.01$) and UKA (CON = 7.32, UKA = 5.79, $p<0.05$) groups. Finally, at three-months, scores remained statistically lower in the TKA (CON = 7.32, TKA = 5.14, $p<0.01$) group. There were no differences between the healthy controls and UKA UCLA (CON = 7.61, UKA = 7.71, $p>0.05$) scores three-months post-operatively. Participant UCLA Activity Scores are reported in Table 2.8.

Table 2.7 Three-Month Biomechanical Variables During Stair Descent

	Controls (n=22 knees)			TKA (n=17 knees)			TKA to CON	UKA (n=9 knees)			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	15.41	±	2.33	14.61	±	5.62	0.814	14.29	±	4.01	0.770	0.980
PKFA first 25% of stance (º)	27.51	±	6.75	23.01	±	7.04	0.104	27.63	±	5.82	0.998	0.225
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.75	±	0.33**	**0.000**	1.05	±	0.36	0.091	0.073
PKFA first 50% of stance (º)	32.30	±	5.60	30.09	±	8.67	0.557	32.45	±	5.80	0.998	0.684
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.85	±	0.32**	**0.000**	1.13	±	0.29	0.078	0.066
Peak Trunk Flexion (º)	10.91	±	6.27	18.72	±	7.29**	**0.001**	13.12	±	2.39	0.643	0.080
Peak Trunk Side Bending (º)	1.82	±	2.39	6.51	±	4.95**	**0.000**	3.14	±	1.84	0.612	0.057
Time on Force Plate (s)	0.72	±	0.10	1.38	±	0.85**	**0.001**	0.96	±	0.19	0.499	0.132
Time to Max vGRF (s)	0.12	±	0.04	0.28	±	0.37	0.120	0.22	±	0.22	0.583	0.819
Total Time on Stairs (s)	1.14	±	0.16	1.88	±	0.95**	**0.001**	1.48	±	0.25	0.334	0.238
Knee Extensor Strength (lbs)	85.53	±	26.01	60.87	±	20.08**	**0.006**	76.21	±	26.63	0.619	0.300

SD = standard deviation, TKA = total knee arthroplasty, CON = Control,
UKA = unicompartmental knee arthroplasty, n = number, Max = maximum,
vGRF= vertical Ground Reaction Force, N/kg = newtons per kilogram,
PKFA = peak knee flexion angle, % = percentage, ˚ = degrees,
PKFM = peak knee flexion moment, Nm/kg = newton meters per kilogram, s = seconds, lbs = pounds
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).

Table 2.8 Participant UCLA Activity Scores

	Controls			TKA			TKA to CON	UKA			UKA to CON	UKA to TKA
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Pre-Operatively	7.61	±	1.67	4.53	±	1.30**	**0.000**	4.71		2.63**	**0.001**	0.972
6-Week Post-Operatively	7.61	±	1.67	4.43	±	1.16**	**0.000**	5.79		1.47*	**0.020**	0.135
3-Months Post-Operatively	7.61	±	1.67	4.93	±	1.28**	**0.000**	7.71	±	1.25^^	0.985	**0.001**

UCLA = University of California Los Angeles, TKA = total knee arthroplasty, CON = Control,
UKA = unicompartmental knee arthroplasty,
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).
^^ = significantly different than TKA group ($p \leq 0.01$).

When evaluating the biomechanical variable changes during stair ascent over time in the TKA group, compared to pre-operative (PRE) values, at six-weeks (6WK), the TKA group significantly decreased; Max vGRF, (PRE = 10.42 N/kg, 6WK = 8.32 N/kg, $p<0.01$), PKFA (PRE = 71.60°, 6WK = 56.83°, $p<0.01$) and PKFM (PRE = 0.58 Nm/kg, 6WK = 0.22 Nm/kg, $p<0.01$). They also significantly increased peak trunk flexion (PRE = 27.62°, 6WK = 34.29°, $p<0.05$), peak trunk side bending (PRE = 5.85°, 6WK = 12.42°, $p<0.01$) and time to ascend the stairs (PRE = 2.87 s, 6 WK = 3.92 s, $p<0.01$). During stair decent, comparing 6WK to PRE, the TKA group significantly decreased Max vGRF (PRE = 12.57 N/kg, 6WK = 7.63 N/kg, $p<0.01$), PKFA first 25% of stance (PRE = 27.57°, 6WK = 17.30°, $p<0.01$), PKFM first 25% of stance (PRE = 0.72, 6WK = 0.38 N/kg, $p<0.01$), PKFM first 50% of stance (PRE = 0.81 N/kg, 6WK = 0.49 N/kg, $p<0.01$) and experienced a decreased knee extensor strength (PRE = 67.55 lbs, 6WK = 49.93 lbs, $p<0.01$). In addition, a significant increase in time to Max vGRF was observed (PRE = 0.23 s, 6WK = 0.27 s, $p<0.01$). When comparing 3MO to PRE stair descent values, there was a significant decrease in PKFA (PRE = 71.60°, 3MO = 66.84°, $p<0.01$) and during stair descent there was a significant decrease in time to Max vGRF (PRE = 0.30 s, 3MO = 0.21 s, $p<0.05$). Comparing the 3MO to 6WK, the TKA group demonstrated improvement in every biomechanical variable ($p<0.01$) during stair ascent. During stair descent, 3MO compared to 6WK, the TKA group experienced a significant increase in; Max vGRF (6WK= 7.63 N/kg, 3MO = 15.22 N/kg, $p<0.01$), PKFM during the first 25% of stairs (6WK = 0.37 Nm/kg, 3MO = 0.82 Nm/kg, $p<0.01$), PKFA first 50% of stance (6WK = 22.88°, 3MO = 31.60°, $p<0.01$), PKFM during the first 50% of stance (PRE = 0.86 Nm/kg, 3MO = 1.01 Nm/kg, $p<0.0$) and knee extensor strength (PRE = 67.55 lbs, 3 MO = 62.62 lbs, $p<0.01$).In addition, a decreased in peak trunk flexion (PRE = 21.30°, 3MO = 18.95°, $p<0.01$), trunk side bending (PRE = 9.45°, 3MO = 6.29°, $p<0.01$) and time to Max vGRF (6WK = 0.27 s, 3 MO = 0.13 s, $p<0.01$) was present. Stair ascent and descent biomechanical variables descriptive statistics in the TKA group over time are presented in Table 2.9.

When evaluating the biomechanical changes over time in the UKA group, during stair ascent there was a significant decrease in trunk side bending (PRE = 8.02°, 6WK = 4.89°, $p<0.01$) and total time (PRE = 3.29 s, 6WK = 2.59 s, $p<0.05$). There were no significantly different values from PRE to 6 WK ($p>0.05$) during stair descent. At 3MO compared to PRE stair ascent values, the UKA group significantly decreased peak trunk side bending (PRE =

8.02°, 3MO = 4.23°, $p<0.05$) and time to ascend stairs (PRE = 3.49 s, 3MO = 2.28 s, $p<0.05$). During stair descent comparing 3MO to PRE, the UKA group significantly increased Max vGRF (PRE = 9.80 N/kg, 3MO = 14.29, $p<0.05$), PKFA first 25% of stance (PRE = 20.18°, 3MO = 27.63°, $p<0.05$), PKFM first 25% of stance (0.68 Nm/kg, 1.05 Nm/kg, $p<0.05$) and PKFM first 50% of stance (PRE = 0.81 Nm/kg, 3MO = 1.13 Nm/kg, $p<0.05$). Additionally, the UKA group demonstrated a decreased peak trunk side bending (PRE = 12.90°, 3MO = 3.14°, $p<0.05$), time on force plate (PRE = 2.22 s, 3MO = 0.96 s, $p<0.01$), time to Max vGRF (PRE = 0.75 s, 0.22 s, $p<0.05$) and total time on stairs (PRE = 3.32 s, 3MO = 1.47 s, $p<0.01$). Comparing the 3M to the 6WK values during stair ascent, the UKA group had no significantly different biomechanical variables ($p>0.05$). During stair decent, comparing 3MO to 6WK, the UKA group significantly increased PKFM first 25% of stance (6WK = 0.75 Nm/kg, 3MO = 1.05 Nm/kg, $p<0.01$) and PKFM during the first 50% of stance (6WK = 0.82 Nm/kg, 3MO = 1.13 Nm/kg, $p<0.05$). Stair ascent and descent biomechanical variables in the UKA group during stair negotiation over time are presented in Table 2.10.

Table 2.9 TKA Biomechanical Variables Over Time During Stair Negotiation (n = 16)

Stair Ascent	Pre-Operative Me	±	SD	Six-Weeks Mean	±	SD	6W to PO P-value	Three-Months Mean	±	SD	3M to PO P-value	3M to 6W P-value
Max vGRF (N/kg)	10.42	±	0.64	8.32	±	2.64	**0.005**	10.61	±	0.45	0.140	**0.002**
PKFA (º)	71.60	±	5.07	56.83	±	10.69	**0.000**	66.84	±	4.46	**0.003**	**0.001**
PKFM (Nm/kg)	0.58	±	0.20	0.22	±	0.22	**0.000**	0.45	±	0.17	0.120	**0.004**
Peak Trunk Flexion (º)	27.62	±	4.64	34.29	±	10.32	**0.017**	26.49	±	5.56	0.261	**0.011**
Peak Trunk Side Bending (º)	5.85	±	3.09	12.42	±	5.56	**0.002**	4.62	±	3.92	0.180	**0.002**
Time (s)	2.87	±	1.07	3.92	±	1.70	**0.007**	2.42	±	0.35	0.111	**0.001**
Stair Descent												
Max vGRF (N/kg)	12.57	±	4.08	7.63	±	6.42	**0.003**	15.22	±	5.67	0.119	**0.001**
PKFA first 25% of stance (º)	27.57	±	8.69	17.30	±	9.07	**0.000**	23.95	±	6.66	0.060	**0.006**
PKFM first 25% of stance (Nm/kg)	0.72	±	0.35	0.38	±	0.37	**0.002**	0.82	±	0.28	0.191	**0.000**
PKFA first 50% of stance (º)	31.13	±	9.77	22.88	±	10.00	**0.003**	31.60	±	7.48	0.820	**0.003**
PKFM first 50% of stance (Nm/kg)	0.81	±	0.34	0.49	±	0.46	**0.004**	0.91	±	0.28	0.214	**0.002**
Peak Trunk Flexion (º)	21.30	±	11.50	26.86	±	12.34	0.057	18.95	±	7.71	0.164	**0.012**
Peak Trunk Side Bending (º)	9.45	±	9.95	11.63	±	7.80	0.347	6.29	±	5.21	0.127	**0.009**
Time on Force Plate (s)	1.54	±	1.07	1.93	±	0.96	0.179	1.38	±	0.89	0.303	0.058
Time to Max vGRF (s)	0.30	±	0.38	0.74	±	0.59	**0.008**	0.21	±	0.28	**0.053**	**0.002**
Total Time on Stairs (s)	2.21	±	1.44	2.48	±	0.98	0.418	1.87	±	0.98	0.069	**0.029**
Knee Extensor Strength (lbs)	67.55	±	29.49	49.93	±	17.10*	**0.027**	62.61	±	23.29^^	0.447	**0.010**

TKA = total knee arthroplasty, 6W = six weeks, PO = pre-operative, 3M = three months, SD = standard deviation, Max = maximum, vGRF = vertical ground reaction force, N/kg = newtons per kilogram, PKFA = peak knee flexion angle, ˚ = degrees, PKFM = peak knee flexion moment, Nm/kg = newton meters per kilogram, s = seconds, lbs = pounds

* = significantly different than Pre-Operative ($p \leq 0.05$).

** = significantly different than Pre-Operative ($p \leq 0.01$).

^ = significantly different than Six-Weeks ($p \leq 0.05$).

^^ = significantly different than Six-Weeks ($p \leq 0.01$).

Table 2.10 UKA Biomechanical Variables Over Time During Stair Negotiation (n = 9)

	Pre-Operative			Six-Weeks			6W to PO	Three-Months			3M to PO	3M to 6W
Stair Ascent	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	10.33	±	2.61	10.50	±	1.67	0.835	11.03	±	0.69	0.432	0.284
PKFA (º)	63.91	±	6.57	63.54	±	5.67	0.820	67.97	±	6.21	0.112	0.007
PKFM (Nm/kg)	0.76	±	0.52	0.59	±	0.17	0.195	0.61	±	0.18	0.290	0.534
Peak Trunk Flexion (º)	25.62	±	7.70	23.50	±	3.87	0.450	22.11	±	4.59	0.250	0.175
Peak Trunk Side Bending (º)	8.02	±	3.74	4.89	±	2.29	**0.010**	4.23	±	1.69	**0.031**	0.480
Time (s)	3.49	±	1.29	2.59	±	0.42	**0.019**	2.28	±	0.13	**0.018**	0.042
Stair Descent												
Max vGRF (N/kg)	9.80	±	5.63	11.18	±	7.03	0.584	14.29	±	4.01	**0.022**	0.063
PKFA first 25% of stance (º)	20.18	±	10.34	22.18	±	8.87	0.365	27.63	±	5.82	**0.054**	0.058
PKFM first 25% of stance (Nm/kg)	0.68	±	0.53	0.75	±	0.52	0.624	1.05	±	0.36	**0.020**	**0.010**
PKFA first 50% of stance (º)	27.28	±	17.26	27.14	±	8.45	0.997	32.45	±	5.80	0.414	0.066
PKFM first 50% of stance (Nm/kg)	0.81	±	0.57	0.82	±	0.52	0.901	1.13	±	0.29	**0.050**	**0.021**
Peak Trunk Flexion (º)	23.54	±	16.85	19.16	±	11.79	0.409	13.12	±	2.39	0.096	0.163
Peak Trunk Side Bending (º)	12.90	±	12.52	7.28	±	7.04	0.216	3.14	±	1.84	**0.044**	0.131
Time on Force Plate (s)	2.22	±	1.25	1.47	±	0.87	0.152	0.96	±	0.96	**0.012**	0.088
Time to Max vGRF (s)	0.75	±	0.68	0.47	±	0.53	0.349	0.22	±	0.22	**0.026**	0.115
Total Time on Stairs (s)	3.32	±	1.94	2.03	±	0.88	0.095	1.47	±	0.25	**0.016**	0.071
Knee Extensor Strength (lbs)	78.70	±	39.14	70.25	±	28.66	0.257	76.84	±	26.29	0.738	0.240

TKA = total knee arthroplasty, 6W = six weeks, PO = pre-operative, 3M = three months, SD = standard deviation,
Max = maximum, vGRF = vertical ground reaction force, N/kg = newtons per kilogram, PKFA = peak knee flexion angle,
˚ = degrees, PKFM = peak knee flexion moment, Nm/kg = newton meters per kilogram, s = seconds, lbs = pounds

* = significantly different than Pre-Operative ($p \leq 0.05$).

** = significantly different than Pre-Operative ($p \leq 0.01$).

^ = significantly different than Six-Weeks ($p \leq 0.05$).

^^ = significantly different than Six-Weeks ($p \leq 0.01$).

Discussion

Results of this study indicate that during the first three months of the post-operative recovery process, the UKA group demonstrated significant improvements in the ability to negotiate stairs to levels comparable to healthy controls, whereas the TKA group continued to display biomechanical deficits. Pre-operative function between both TKA and UKA groups was similar and both demonstrated deficits in biomechanical variables compared to controls. Knee extensor strength favored UKA patients as TKA patients' strength remained significantly decreased out to three months post-operatively. Deficits in PKFM were observed in both TKA and UKA groups at six-weeks post-operatively during both stair ascent and descent, but at three months after surgery UKA group exhibited no differences in PKFM values compared to healthy controls, which indicates that three months following surgery, UKA patients were willing to load the limb during a highly demanding functional task. Compensatory motions involving the trunk are common in OA patients and were present in the TKA group, but were not present in the UKA group at either time period.

The perception that improved performance following surgery among UKA patients compared to TKA patients was related to improved pre-operative function in UKA patients was not supported in this research study. Even though patients in the current study underwent different surgical procedures, both groups exhibited similar levels of functional deficits prior to surgery. In fact, during stair descent, the UKA group demonstrated more biomechanical deficits compared to controls that the TKA group. Additionally, they took longer to descend the stairs compared to the TKA group. This outcome was not expected and provides insight into the limitations of function that are present during a the highly challenging task of stair negotiation in both TKA and UKA patients prior to surgery.

Post-operatively, patients that were unable to perform the stair negotiation tasks at six-weeks following surgery, were initially removed in comparing both the TKA and UKA group to controls. This resulted in a biomechanical profile for surgical patients that was similar to the control group at that time period. However, this analysis was potentially misleading as only 63% (12/19) of the TKA group could descend the stairs compared to 88% (8/9) in the UKA group. Subsequent analysis including all surgical patients, including those unable to perform the task, through the use of "ceiling" values, produced a biomechanical profile that was more representative of the arthroplasty groups and provided a more accurate assessment of the deficits that remain in post-operative arthroplasty patients at six-weeks following surgery.

Stair negotiation has been shown to be a strong predictor of functional decline in older adults[22] and is a more difficult task which places a high demand on knee extensor musculature[23,28]. Chung et al[24], reported no differences in strength values between TKA and UKA at six-months and one-year following surgery. While no statistical differences were found between TKA and UKA patients at either six-weeks or three-months in the present study, TKA patients were still significantly weaker that controls at three months while UKA were not. Significant knee extensor strength deficits were observed at both six-weeks and three-months following surgery in the TKA group, but not in the UKA group, compared to the healthy controls. This can be attributed to the surgical process during TKA in which the surgeon cuts a portion of the knee extensor musculature. However, during a UKA procedure, the knee extensor musculature remains intact. Although knee extensor strength improved over time in TKA patients, it still remained significantly decreased compared to controls at three months post-operatively. Conversely, knee extensor strength in the UKA group did not differ from controls as early as the six-week gait analysis.

Knee flexion moment is an important indicator of the forces acting on the knee joint, with a larger PKFM demonstrating an increase in joint loading and a willingness to load the knee[29]. In the present study, PKFM was collected at 25% of stance to gain insight into the forces within the knee joint during the initial loading phase of stair descent, whereas PKFM at 50% of stance was collected to gain an understanding of the overall PKFM during stance time before the transitioning down the stairs to toe-off. Knee flexion moments were decreased compared to controls in both TKA and UKA groups at the six-week data analysis during both stair ascent and descent, suggesting that functional deficits remain in both groups out to six-weeks post-operatively. Interestingly, three-months following surgery, the UKA group still demonstrated decreased PFKM during stair ascent compared to controls, but during stair descent the PKFM did not differ from the control group. Changes in PKFM are modulated primarily through changes in the magnitude of vGRF and through adjustments in the length of the lever arm by manipulations in the center of mass through increasing trunk flexion. During stair ascent, the UKA group produced similar Max vGRF and trunk motion as found in the control group, as opposed to those in the TKA group who still demonstrated decreased vGRF and increased trunk flexion compared to controls at this time point. Therefore, the precise mechanism behind the decreased PKFM found during stair ascent in the UKA is unclear. However, the performance of UKA patients during stair descent suggests that UKA patients experience an increased willingness to load the limb as early as three-months following surgery, as evidence by the increase in PKFM values.

In comparing the results PKFM of UKA and TKA groups, the findings of the present study generally disagree with those of Jung et al.[30] who reported no differences in joint moments during stair negotiation in patients undergoing a TKA and a UKA, however, participants had undergone a TKA in one knee and a UKA in the other knee and were evaluated at greater than two years following surgery. Both of these factors likely contributed to the differences in results when compared to the current study. At both six-weeks and three-months post-surgery, PKFM was significantly different or trending toward significantly different ($p<0.10$) between UKA and TKA for all loading measures of PKFM during both stair ascent and descent. The UKA group demonstrated improved PKFM values compared to the TKA group suggesting that they are functioning at a higher level compared to those undergoing a TKA.

During demanding functional tasks, patients may compensate with an increase in trunk flexion in order to decrease overall PKFM. This is accomplished by moving the body's center of mass anteriorly and decreasing the length of the lever arm through which the GRF can act to produce force at the knee[31,32]. The same mechanism is also present when trunk side bending is utilized to decrease frontal plane moments at the knee[33,34]. At six-week following surgery, the TKA group presented with increased compensatory motions in both trunk flexion and trunk side bending compared to controls during stair ascent, whereas the UKA group demonstrated trunk flexion and side bending values that were not significantly different than controls. The presence of compensatory motions, combined with decreased PKFA, PKFM, Max vGRF, and increased time to complete stair negotiation in the TKA group, suggest that six-weeks post-surgery may be too early in the recovery process to biomechanically assess stair negotiation in this group of patients. However, at three-months following surgery most arthroplasty patients have successfully completed rehabilitation programs and a biomechanical assessment at this point in the recovery process may provide insight into patient function. At three-months following surgery, the TKA group still presented with increased trunk compensatory motion compared to controls, however both trunk flexion and side bending had improved over time. The persistence of compensatory trunk flexion and side bending in the TKA at three-months in the attempt to decrease PKFM during stair negation is likely related to the persistence of knee extensor weakness seen in this group compared to controls[32]. However, in the UKA group, knee extensor strength returned to the levels of controls by three months reducing the need for compensatory trunk motions during stair negotiation.

Decreased physical activity levels prior to surgery have been reported in both TKA and UKA patients[15]. Therefore, the significantly lower UCLA Activity scores pre-operatively in both arthroplasty groups was an expected outcome in the present study. At six-weeks following surgery, the same trend continued with both TKA and UKA groups reporting decreased physically activity levels compared to controls. Arthroplasty patients are typically still enrolled in rehabilitation programs six-weeks following surgery so this outcome was expected. However, at three-months following surgery, TKA patients still reported physical activity levels significantly below both UKA and controls groups, which agrees with previous research[15,17]. The knee extensor strength of TKA patients remained decreased up to three-months post-surgery when compared to controls, suggesting that physical activity can be effected by this decrease in strength. These findings agree with previous research indicating that functional performance is highly correlated to knee extensor strength[35].

Several limitations were present in the current study. The small sample size of the UKA group, compared to TKA and controls groups could have impacted the biomechanical averages and statistical outcomes. In addition, the statistical procedure using ceiling variables in order to include those participants who could not preform the activity for data analysis, may serve as a limitation. However, at six-weeks post-operatively, only 66% (12/18) of the TKA group could descend the stairs compared to 78% (7/9) of the UKA group. Excluding these participants from statistical analysis yielded results that were not representative of the true differences between controls and arthroplasty groups.

Conclusion

In summary, results of this study indicate that short-term post-operative stair negotiation function favored the UKA patients as they were similar to healthy controls as soon as three-months following knee replacement surgery. Total knee arthroplasty patients displayed more compensatory motions including decreased knee extensor strength, decreased PKFM as well as an increased in both trunk compensatory motions and increased time to negotiate stairs suggesting that important deficits remain during stair negotiation tasks in TKA patients three-months post-operatively. Regaining knee extensor strength is recommended as the focus of rehabilitation programs in post-operative TKA patients. Improvements in post-operative function in UKA patients may be attributed to the minimally invasive procedure which decreases recovery time[10-12] and the surgical process which retains cruciate ligaments[13] which may contribute to an increase in function during highly demanding tasks. Unicompartmental knee arthroplasty may be used in some cases as an alternative to TKA and favorable functional short-term post-operative outcomes may be expected.

CHAPTER 3:

A LONG TERM POST-OPERATIVE BIOMECHANICAL ANALYSIS DURING STAIR NEGOTIATION IN PATIENTS UNDEROING TOTAL KNEE ARTHROPLASTY COMPARED TO HEALHY CONTROLS

Abstract

Context: The presence of knee osteoarthritis makes stair negotiation more difficult. In addition, in the subsequent years following total knee arthroplasty (TKA), deficits in stair negotiation remain. Common implant designs used in TKA are the multi-radius (MR) and single-radius (SR) designs. Clinical results including greater knee flexion angles, improved self-reported clinical score and decreased compensatory motions, support the benefits of SR implants for improving function during stair negotiation following TKA. However, research evaluating biomechanical differences during stair negotiation following TKA between implant designs compared to age-matched controls are limited. **Objective:** To compare the stair negotiation in TKA patients to age-matched controls as well as to analyze implant designs to determine if one is more similar to healthy controls. **Design:** Longitudinal gait analysis. **Setting:** Biomechanics laboratory. **Patients:** Fourteen TKA patients, nine MR (seven males,11 knees) implant and five SR implant (four males, eight knees) were compared to 22 controls (15 males, 15 knees). **Intervention:** Patients randomly received either SR (GetAroundKnee™, Stryker Orthopedics, Mahwah, NJ) or MR (Balanced Knee® System, Ortho Development Corporation, Draper, UT) implants. All TKA patients underwent 3D motion gait analysis during a three-step staircase pre-TKA and post-TKA at six-months and one-year. Control data was collected at a one-time data collection. Multiple multivariate general linear model tests were used to compare variables of interest, with an alpha level at $p<0.05$, to determine differences between implants at each time period. **Main Outcome Measures:** Knee flexion angle (PKFA), knee flexion moment (PKFM), vertical ground reaction force, trunk forward flexion, trunk side bending, strength and time measurements. **Results:** At six-months post-TKA during stair ascent, compared to healthy controls, PKFM were significantly decreased in both MR ($p = 0.000$) and SR ($p = 0.000$) groups. The MR group took longer to descend the stairs ($p=0.000$), had an increased trunk flexion ($p = 0.012$) and an increased trunk side bending when compared to controls. During stair descent at six-months, PKFM were decreased in both MR ($p = 0.000$) and SR ($p=0.00$) groups compared to healthy controls. In addition, the MR implant had increased trunk side bending ($p = 0.000$) and had increased time on force plate ($p=0.000$), time to max vGRF ($p = 0.001$) and total time to descend the stairs (0=0.000) compared to healthy controls. Knee extensor strength was decreased in SR implant ($p=0.008$) six-months post-TKA compared to healthy controls. At one-year post-TKA during stair ascent, compared to healthy controls a decreased PKFM in was

present both MR (p = 0.000) and SR implants (p = 0.000). The MR implant had decreased vGRF (p = 0.013), and an increased in both trunk flexion (p = 0.004) and time to ascend the stairs (p=0.000) compared to healthy controls. During stair descent at one-year post TKA decreased PKFM was observed in both MR (p = 0.000) and SR (p = 0.00) implant groups compared to healthy controls. The MR implant continued to have increased time on force plate (p = 0.000) and total time to descend stairs (p = 0.000) compared to healthy controls.

Conclusion: Functional deficits in KFM remain in TKA patients and effect their ability to negotiate stairs out to one-year post-TKA. One may surmise from this study that data would favor the SR implant over the MR implant design as certain variables were more similar to controls. However, at both time periods during both stair ascent and descent tasks, KFM were decreased in both the MR and SR implant groups compared to healthy controls and was independent on implant design. Using KFM as an indication of patient function, drawing a conclusion regarding which implant is more superior was not inferred since both MR and SR implant groups had decreased KFM compared to healthy controls. In the presence of these deficits, TKA patients increase trunk motions during stair negotiation to manipulate knee joint loading. Using the more challenging task of stair negotiation, results of this study provide a better understanding the functional deficits that remain in TKA patients following surgery and how post-operative deficits can be addressed through more challenging rehabilitation.

Introduction

To maintain independence as we age, the ability to negotiate stairs is essential. Not only is stair negotiation more challenging than level walking[22], more falls occur during stair descent than during ascent in older adults[36]. The presence of knee osteoarthritis (OA) makes stair negotiation more challenging[37] due to increased pain and restrictions in range of motion, and is often the primary functional limitation of OA patients[38]. In subsequent years following TKA procedures, deficits during stair negotiation activities remain, which is attributed to the increased physical demand of stairs[39].

During stair negotiation tasks following TKA, patients compensate for knee extensor weakness by manipulating their external knee flexion moments, a biomechanical variable which is commonly used to assess and measure overall extensor function[40]. This quadriceps avoidance gait during stair negotiation results in decreases in knee flexion angle at foot contact in TKA patients when compared to controls[39,41]. Manipulations in ground reaction forces (GRF) have also been reported following TKA[42]. Functional deficits remain following TKA, but the extent of these deficits, specific to implant design, is not clear[42].

Single-radius (SR) and Multi-radius (MR) femoral implants are commonly used during total knee arthroplasty (TKA), the gold standard of treatment for end stage knee OA. The MR implant is similar to the natural knee with two sagittal axes of rotation, moving anteriorly during terminal extension[43-45]. The MR design has been reported to increase the demand on knee extensors, therefore increasing the compensatory motions in patients during activities of daily living[46]. In contrast, the SR implant has one sagittal axis throughout knee range of motion[45,47,48]. The more posterior axis of rotation in the SR implant creates a mechanical advantage, allowing the quadriceps to work more efficiently throughout extension activities[47,49]. Previous research has evaluated stair negotiation between MR and SR implants and it has been reported that post-TKA MR implant patients displayed compensatory adaptations and also displayed increased muscle activation of their quadriceps muscles, reflecting a need for greater force generation for knee extension[46]. Conversely, the SR implant design allowed for adequate knee extensor moments and required less quadriceps force, providing functional benefits to patients[45,46]. Research supports the theory that MR implant require greater quadriceps force to generate knee extension[45,49].

Clinical results including greater knee flexion angles, improved self-reported clinical score and decreased compensatory motions, support the benefits of SR implants for improving function during stair negotiation following TKA[22,47,49,50]. However, research evaluating biomechanical differences during stair negotiation following TKA between implant designs compared to age-matched controls are limited. Therefore, the purpose of this study is to compare the stair negotiation function between implant designs of OA patients undergoing TKA to age-matched controls.

Methodology

Participants

A randomized, longitudinal design was conducted consisting of 14 osteoarthritis patients (19 knees) undergoing gait assessment within two weeks prior to total knee arthroplasty (TKA) and post-TKA at six weeks, three months, six months and one-year. Inclusion criteria for TKA patients included: under 75 years of age, no previous history of lower extremity fracture, osteotomy, or joint replacement, undergoing an unilateral or bilateral UKA or TKA for the treatment of osteoarthritis, and physically able to walk without an aid. Total Knee Arthroplasty patients were screened for inclusion for this study and were randomly assigned to receive either a single radius (SR) (GetAroundKnee™, Stryker Orthopedics, Mahwah,NJ) or a multi-radius (MR) implant (Balanced Knee® System, Ortho Development Corporation, Draper, UT) design. All surgeries were performed by the same board certified orthopedic surgeon. Biomechanical assessment of enrolled arthroplasty patients occurred within two-weeks prior to TKA and post-TKA at six-weeks, three-months, six-months and one-year. Additionally, data were collected on 30 healthy controls subject (15 male) at a one-time data collection. Inclusionary criteria for the controls included: 1) no history of lower extremity joint surgery, 2) no history of arthritis diagnosis, 3) no diagnosed neurological disorders and 4) no physical activity restrictions from their physician.

Prior to enrollment in the study, all participants signed informed consent forms approved by the Institution's Institutional Review Board. Once consent was gained, participants were assigned an ID number that was used for all data collection sessions and paperwork. All participant data was kept in a filing cabinet in a locked office within the Biomechanics Human Performance Lab at the University of Hawaiʻi at Mānoa. All adverse events, such as injury during testing sessions, were monitored and reported to the Institutional Review Board in accordance to the reporting criteria.

Procedures

All biomechanical analyses were conducted at the University of Hawai‘i Gait Laboratory. Upon arrival at each visit participants completed The University of California at Los Angeles (UCLA) activity questionnaire which asks the participant “to circle a number from 1-10 that best describes their current activity level” (1-being wholly inactive, dependent on others, and cannot leave residence, 10-regularly participates in impact sports). Control participants, in addition to completing the UCLA activity questionnaire, completed a health history questionnaire. The purpose of this survey is to determine if individuals were eligible to participate as a control subject in this study. Following completion of the surveys, participant’s height was collected using a wall-mounted stadiometer (Model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA) and body mass was collected using a Detecto certifier scale (Webb City Mo, USA). Shank lengths were determined and measured from the lateral knee joint line to the distal lateral malleolus and 80% of shank lengths will be calculated and marked. These markings served as location points for placement of the hand-held dynamometer during knee extensor strength testing, to allow for consistent placement of the dynamometer relative to each patient.

Stair negotiation biomechanics data were collected using a three-dimensional motion capture system (Vicon, Inc. Centennial, CO) and one force plate (Advanced Mechanical Technology Incorporated Boston, MA). Twenty-nine reflective markers were placed bilaterally on the following landmarks: first metatarsophalangeal joint, second metatarsophalangeal joint, fifth metatarsophalangeal joint, base of fifth metatarsal, medial and lateral malleolus, posterior calcaneus, medial and lateral epicondyles of the femur, posterior superior iliac spine, anterior superior iliac spine, and acromioclavicular joint. Unilateral markers were placed on the following structures: jugular notch, xiphoid process, spinous process of the seventh cervical vertebrae, spinous process of the tenth thoracic vertebrae, and on the inferior portion of the right scapula. Four arrays consisting of four markers (Vicon, Inc. Centennial, CO) were secured laterally on the shaft of each femur and shank. Markers on the medial femoral epicondyle, medial malleolus and head of the first metatarsal were used for calibration purposes during a static trial only and were removed for stair trials.

Three laboratory steps with dimensions of an 18 cm step rise, a 46 cm width, and a 28 cm tread, were used during stair negotiation[27]. Ground reaction forces of the involved limb were measured using one force plate (Advanced Mechanical Technology Incorporated, Boston, MA) which was embedded on the second step. Kinematic data were collected at 240 Hz and time synchronized with kinetic data collected at 960 Hz[5]. All kinematic and kinetic data were smoothed using a Butterworth filter with a 10 Hz cut-off and ground reaction force was filtered using a 50 Hz cut-off frequency. External joint moments were calculated using inverse dynamics based on marker trajectories and kinetic data which was also filtered using a 10 Hz cut-off frequency[5]. All data was processed using Visual 3D (C-Motion, Inc., Germantown, MD). Following stair descent trials, strength tests were conducted using a Microfet2 hand held dynamometer (Hoggan Health Industries, West Jordan, UT). All data collections were conducted at the University Gait Laboratory.

Stair negotiation trials followed a similar protocol performed by Vallabhajosula et al.[27]. Patients were instructed to walk up the stairs "as quickly and as safely as possible." Each patient was asked to take two additional steps on the stair platform to ensure a natural gait is continued through the last step and deceleration did not occur. For stair descent, patients took a step on the stair platform prior to stepping down with the involved limb. An additional three steps were taken after completion of the stair descent trials. Handrails were provided for safety but patients were instructed not to use them unless balance was compromised. If the handrails were used, the trial was discarded and not included for data analysis. Due to high intra-subject variability previously reported during stair climbing in the osteoarthritis population, five successful trials were averaged.

Following stair negotiation trials, bilateral knee extensor muscle strength was then assessed using a hand held dynamometer (HHD) (Hoggan Health Industries, West Jordan, UT), performed in a gravity dependent position the muscular testing. Knee extensor strength was measured from the previously mentioned marked shank length, while the patient was seated with the knee placed at 60° of knee flexion and their trunk extended 130° from the surface of the table. The HHD was placed on the anterior shank, just proximal to the medial malleolus and secured with a strap. Participants were instructed to build a force over three seconds, holding the maximal force contraction for two seconds. Two trials of a three-second maximal effort isometric knee extension contraction were completed. A third trial was completed if the second trial did not measure within 10% force output of the first trial. Verbal encouragement was provided to help elicit maximal force production by the participant during strength testing.

Statistical Analysis

Normality was assessed using Shapiro-wilk test. To test where the variances between standard deviations the Levene's test was used. Multiple Multivariate General Linear Model tests tested for significance. Post-hoc Tukey tests determined where significant differences existed among groups and the dependent variables. The relationship between knee extensor strength and knee flexion moment was evaluated using Pearson's Correlation Coefficient will be used. All data were analyzed using SPSS Version 22.0. All knee joint moments are reported as external moments. Knee and trunk flexion values are reported as a positive number, additionally, during side bending, a positive value indicates trunk motion towards the stance leg.

Results

A total of 14 TKA patients were included for data analysis and compared to 22 healthy controls. Of the patients undergoing TKA, nine patients (11 knees) received the MR implant design and five patients (8 knees) received the SR implant design. All TKA patients were present for the six-month data collection time period. However, two MR participants dropped out of the study and were not included in the one-year analysis. Therefore, the one-year data collection included seven MR patients (8 knees). All TKA biomechanical variables were compared to 22 (15 males) healthy controls (CON). Demographics were not statistically different between the TKA and healthy control groups and are listed in Table 3.1.

When compared to controls during stair ascent six-months post-TKA, the TKA group had a statistically significantly lower PKFM (CON = 1.06 Nm/kg, TKA = 0.57 Nm/kg, $p \leq 0.01$). The TKA group also had statistically significantly lower Max GRF (CON = 11.57 N, TKA = 10.92 N, $p \leq 0.05$). Statistically significant increases in both trunk flexion (CON = 18.51°, TKA = 24.93°, $p \leq 0.01$) and trunk side bending (CON = 2.84°, TKA = 5.13°, $p \leq 0.01$) were observed in the TKA group. It took the TKA group statistically significant longer to ascend the stairs as well (CON = 1.91 s, TKA = 2.36 s, $p \leq 0.01$). During stair descent trials at six-months post-operatively, the TKA group had statistically significantly lower PKFM at 25% of stance (CON = 1.32 Nm/kg, TKA = 0.81 Nm/kg, $p \leq 0.01$) and PKFM at 50% of stance (CON = 1.39 Nm/kg, TKA = 0.91 Nm/kg, $p \leq 0.01$). The TKA group also had statistically significant increased trunk side bending (CON = 1.82°, TKA = 5.44°, $p \leq 0.01$). Total knee arthroplasty patients had a statistically significant increased time on force plate (CON = 0.72 s, TKA = 1.11 s, $p \leq 0.01$), time to max vGRF (CON = 0.12 s, TKA = 0.20 s, $p \leq 0.05$) and total time on the stairs (CON = 1.13 s, TKA = 1.61 s, $p \leq 0.01$). Additionally, knee extensor strength was significantly decreased in TKA patients (CON = 85.41 lbs, TKA = 62.84 pound, $p \leq 0.01$). Descriptive information for the six-month stair ascent and decent biomechanical variables can be found in Table 3.2 and Table 3.3 respectively.

Compared to controls during stair ascent one-year post-TKA, the TKA group had a statistically significantly lower PKFM (CON = 1.06 Nm/kg, TKA = 0.63 Nm/kg, $p \leq 0.01$). The TKA group also had statistically significantly lower Max vGRF (CON = 11.57 N, TKA = 10.55 N, $p \leq 0.01$). Statistically significant increase in trunk flexion (CON = 18.51°, TKA = 24.40°, $p \leq 0.01$) were observed in the TKA group. It took the TKA group statistically significant longer to ascend the stairs as well (CON = 1.91 s, TKA = 2.38 s, $p \leq 0.01$). At one year post-operatively, during stair descent trials, the TKA group had statistically significantly lower PKFM at 25% of stance (CON = 1.32 Nm/kg, TKA = 0.86 Nm/kg, $p \leq 0.01$) and PKFM at 50% of stance (CON = 1.39 Nm/kg, TKA = 0.94 Nm/kg, $p \leq 0.01$). The TKA group also had statistically significant increased trunk forward flexion (CON = 10.91°, TKA = 16.46°, $p \leq 0.01$), time on force plate (CON = 0.72 s, TKA = 1.09 s, $p \leq 0.01$) and total time to descend the stairs (CON = 1.13 s, TKA = 1.60 s, $p \leq 0.01$). There was also a decreased Max vGRF in TKA group (CON = 15.41 N/kg, TKA = 13.78 N/kg, $p \leq 0.05$). Additionally, knee extensor strength remained significantly decreased in TKA patients (CON = 85.41 lbs, TKA = 63.46 lbs, $p \leq 0.01$). Descriptive information for the six-month stair ascent and decent biomechanical variables can be found in Table 3.4 and Table 3.5 respectively.

Table 3.1 Patient Demographics

	Controls (n=22, 22 knees)			Multi-Radius (n=9, 11 knees)			MR to CON	Single-Radius (n=5, 8 knees)			SR to CON	SR to MR
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Age	67.4	±	4.7	64.8	±	6.3	0.280	64.5	±	3.5	0.360	0.999
Height (mm)	1.7	±	0.1	1.6	±	0.7	0.251	1.7	±	0.8	0.919	0.607
Body Mass (kg)	75.1	±	15.3	72.1	±	13.4	0.982	80.4	±	11.5	0.653	0.634

MR = Multi-Radius, CON = Control, SR = Single-Radius, n = number; SD = standard deviation; mm = millimeters;
Kg = kilograms

Table 3.2 Six-Month Biomechanical Variables During Stair Ascent

	Controls (n=22 knees)			TKA (n=18 knees)			P-value
	Mean	±	SD	Mean	±	SD	
Max vGRF (N/kg)	11.57	±	1.13	10.92	±	0.63*	**0.033**
PKFA (º)	67.10	±	5.85	66.40	±	6.28	0.719
PKFM (Nm/kg)	1.06	±	0.20	0.57	±	0.21**	**0.000**
Peak Trunk Flexion (º)	18.51	±	5.81	24.93	±	6.57**	**0.003**
Peak Trunk Side Bending (º)	2.84	±	2.27	5.13	±	3.46**	**0.004**
Time (s)	1.91	±	0.21	2.36	±	0.40**	**0.000**

TKA = total knee arthroplasty; n = number; SD = standard deviation; Max = maximum
vGRF = vertical Ground Reaction Force; N/kg = newtons per kilogram;
PKFA= peak knee flexion angle; ˚ = degrees;
PKFM = peak knee flexion moment; Nm/kg = newton meter per kilogram; s = second;
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).

Table 3.3 Six-Month Biomechanical Variables Stair Descent

	Controls (n=22 knees)			TKA (n=18 knees)			P-value
	Mean	±	SD	Mean	±	SD	
Max vGRF (N/kg)	15.41	±	2.33	14.22	±	2.37	0.119
PKFA first 25% of stance (º)	27.45	±	6.94	25.86	±	6.93	0.474
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.81	±	0.30**	**0.000**
PKFA first 50% of stance (º)	32.30	±	5.60	31.74	±	6.49	0.771
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.91	±	0.27**	**0.000**
Peak Trunk Flexion (º)	10.91	±	6.27	14.73	±	8.49	0.110
Peak Trunk Side Bending (º)	1.82	±	2.39	5.44	±	4.83**	**0.004**
Time on Force Plate (s)	0.72	±	0.10	1.11	±	0.62**	**0.007**
Time to Max vGRF (s)	0.12	±	0.03	0.20	±	0.18*	**0.045**
Total Time on Stairs (s)	1.13	±	0.16	1.61	±	0.70**	**0.004**
Knee Extensor Strength (pounds)	85.41	±	25.97	62.84	±	25.04**	**0.005**

TKA = total knee arthroplasty; n = number; SD = standard deviation;
Max = maximum; vGRF = vertical Ground Reaction Force;
N/kg = newton per kilogram; PKFA = peak knee flexion angle;
% = percentage; ˚ = degrees; PKFM = peak knee flexion moment;
Nm/kg = newton meter per kilogram; s = seconds;
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).

Table 3.4 One-Year Biomechanical Variables During Stair Ascent

	Controls (n=22 knees)			TKA (n=14 knees)			P-value
	Mean	±	SD	Mean	±	SD	
Max vGRF (N/kg)	11.57	±	1.13	10.55	±	0.67*	**0.004**
PKFA (º)	67.10	±	5.85	63.04	±	6.50	0.408
PKFM (Nm/kg)	1.06	±	0.20	0.63	±	0.19*	**0.000**
Peak Trunk Flexion (º)	18.51	±	5.81	24.40	±	5.15*	**0.004**
Peak Trunk Side Bending (º)	2.84	±	2.27	4.13	±	3.65	0.198
Time (s)	1.91	±	0.21	2.38		0.41*	**0.000**

TKA = total knee arthroplasty; n = number; SD = standard deviation; Max = maximum
vGRF = vertical Ground Reaction Force; N/kg = newtons per kilogram;
PKFA = peak knee flexion angle; ˚ = degrees;
PKFM = peak knee flexion moment; Nm/kg = newton meter per kilogram; s = second;
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).

Table 3.5 One-Year Biomechanical Variables During Stair Descent

	Controls (n=22 knees)			TKA (n=14 knees)			P-value
	Mean	±	SD	Mean	±	SD	
Max vGRF (N/kg)	15.41	±	2.33	13.78	±	1.76*	**0.033**
PKFA first 25% of stance (º)	27.45	±	6.94	27.09	±	5.27	0.871
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.86	±	0.24**	**0.000**
PKFA first 50% of stance (º)	32.30	±	5.60	32.17	±	6.22	0.871
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.94	±	0.20**	**0.000**
Peak Trunk Flexion (º)	10.91	±	6.27	16.56	±	6.10**	**0.012**
Peak Trunk Side Bending (º)	1.82	±	2.39	3.06	±	3.01	0.180
Time on Force Plate (s)	0.72	±	0.10	1.09	±	0.52**	**0.002**
Time to Max vGRF (s)	0.12	±	0.03	0.19	±	0.24	0.222
Total Time on Stairs (s)	1.13	±	0.16	1.60	±	0.60**	**0.002**
Knee Extensor Strength (pounds)	85.41	±	25.97	63.46	±	21.82**	**0.008**

TKA = total knee arthroplasty; n = number; SD = standard deviation;
Max = maximum; vGRF = vertical Ground Reaction Force;
N/kg = newton per kilogram; PKFA = peak knee flexion angle;
% = percentage; ˚ = degrees; PKFM = peak knee flexion moment;
Nm/kg = newton meter per kilogram; s=seconds;
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).

Six-month analysis of specific implant design and healthy controls revealed a decreased PKFM during stair ascent in MR (CON = 1.06 Nm/kg, MR = 0.57 Nm/kg, $p \leq 0.01$) and SR (CON = 1.06 Nm/kg, SR = 0.57 Nm/kg, $p \leq 0.01$) implant designs. Additionally, the MR implant had a statistically increased time to complete the stair ascent task (CON = 1.91 s, MR = 2.49 s, $p \leq 0.01$), peak trunk flexion (CON = 18.51°, MR = 25.95°, $p \leq 0.01$) and peak trunk side bending (CON = 2.84°, TKA = 4.96°, $p \leq 0.05$). The MR group also had a decreased Max vGRF (CON = 11.57 N/kg, MR = 10.91 N/kg, $p \leq 0.05$). When compared to the MR group, the SR group had statistically increased PKFA (SR = 70.37°, MR = 63.08°, $p \leq 0.05$) implant design. Six-month stair ascent implant design descriptive statistics are found in Table 3.6.

In the six-month analysis of implant designs to controls during stair descent, PFKM at 25% of stance was statistically decreased in both MR (CON = 1.32 Nm/kg, MR = 0.70 Nm/kg, $p \leq 0.01$) and SR (CON = 1.32 Nm/kg, SR = 0.95 Nm/kg, $p \leq 0.01$) implant groups. Peak knee flexion moment at 50% of stance also was statistically decreased in both MR (CON = 1.39 Nm/kg, MR = 0.86 Nm/kg, $p \leq 0.01$) and SR (CON = 1.39 Nm/kg, SR = 0.99 Nm/kg, $p \leq 0.01$). The MR implant group also had an increased trunk side bending (CON = 1.82°, MR = 7.15°, $p \leq 0.01$) compared to controls. Multi-radius group also had statistically increased time on force plate (CON = 0.72 s, MR = 1.29 s, $p = 0.000$), time to Max vGRF (CON = 0.12 s, MR = 0.26 s, $p \leq 0.01$) and total time to descend the stairs (CON = 1.13 s, MR = 1.81 s, $p \leq 0.01$). When compared to the MR group, the SR group had a decreased peak trunk side bending (SR = 3.31°, MR = 7.15°, $p \leq 0.05$) as well as a decreased time to Max vGRF (SR = 0.12 s, MR = 0.26 s, $p \leq 0.05$). Knee extensor strength remained decreased in the SR implant group (CON = 85.41 lbs, SR = 54.95 lbs, $p \leq 0.01$). Six-month stair descent implant design descriptive statistics are found in Table 3.7.

One-year analysis of implant design to healthy controls during stair ascent revealed decreased PKFM during stair ascent in both the MR (CON = 1.06 Nm/kg, MR = 0.56 Nm/kg, $p \leq 0.01$) and SR (CON = 1.06 Nm/kg, SR = 0.68 Nm/kg, $p \leq 0.01$) implant groups. The MR implant had statistically decreased Max vGRF (CON = 11.57 N/kg, MR = 10.21 N/kg, $p \leq 0.01$) when compared to controls. Additionally, the MR implant had a statistically increased trunk flexion (CON = 18.51°, MR = 27.13°, $p \leq 0.01$) and an increased time to ascend the stairs (CON= 1.91 s, MR = 2.66 s, $p \leq 0.01$) compared to controls. The SR implant had statistically decreased time to ascend the stairs (SR = 2.18 s, MR = 2.66, $p \leq 0.01$) compared to the MR group. One-year stair ascent implant design descriptive statistics are found in Table 3.8.

In the one-year analysis of implant designs to during stair descent, PFKM at 25% of stance remained statistically decreased in both MR (CON = 1.32 Nm/kg, MR = 0.73 Nm/kg, $p \leq 0.01$) and SR (CON = 1.32 Nm/kg, SR = 0.95 Nm/kg, $p \leq 0.01$) implant groups. Peak knee flexion moment at 50% of stance also remained statistically decreased in both MR (CON = 1.39 Nm/kg, MR = 0.84 Nm/kg, $p \leq 0.01$) and SR (CON = 1.39 Nm/kg, SR = 0.99 Nm/kg, $p \leq 0.01$). The MR implant group continued to have an increased time on force plate (CON = 0.72 s, MR = 1.46 s, $p \leq 0.01$) and total time to descend the stairs (CON = 1.13 s, MR = 2.01 s, $p \leq 0.01$). Compared to the MR implant group, SR implant group had significantly decreased time on the force plate (SR= 0.82 s, MR = 1.46 s, $p \leq 0.01$) and time to complete stair descent (SR = 1.29 s, MR = 2.01 s, $p \leq 0.01$). One-year stair descent implant design descriptive statistics are found in Table 3.9.

Table 3.6 Six-Month Biomechanical Variables During Stair Ascent, by Implant Design

	Controls (n=22 knees)			Multi-Radius (n=10 knees)			MR to CON	Single-Radius (n=8 knees)			MR to CON	SR to MR
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	11.57	±	1.13	10.68	±	0.79*	**0.045**	11.19	±	0.24	0.602	0.486
PKFA (º)	67.10	±	5.85	63.23	±	5.52	0.185	70.37	±	5.16^	0.352	**0.030**
PKFM (Nm/kg)	1.06	±	0.20	0.57	±	0.21**	**0.000**	0.57	±	0.24**	**0.000**	0.998
Peak Trunk Flexion (º)	18.51	±	5.81	25.84	±	5.93**	**0.012**	23.52	±	8.06	0.147	0.722
Peak Trunk Side Bending (º)	2.84	±	2.27	5.60	±	3.15*	**0.030**	5.36	±	3.27	0.076	0.982
Time (s)	1.91	±	0.21	2.49	±	0.51**	**0.000**	2.17	±	0.09	0.103	0.076

MR = Multi-Radius; CON = Control; SR = Single-Radius; n = number; SD = standard deviation; Max = maximum
vGRF = vertical ground reaction force; Nm/kg = newtons per kilogram; PKFA = peak knee flexion angle;
˚ = degrees; PKFM = peak knee flexion moment; Nm/kg = newton meters per kilogram, s = seconds
* = significantly different than Controls (p ≤ 0.05).
** = significantly different than Controls (p ≤ 0.01).
^ = significantly different than Multi-Radius (p ≤ 0.05)

Table 3.7 Six-Month Biomechanical Variables During Stair Descent, by Implant Design

	Controls (n=22 knees)			Multi-Radius (n=10 knees)			MR to CON	Single-Radius (n=8 knees)			SR to CON	SR to MR
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	15.41	±	2.33	13.67	±	2.58	0.139	14.91	±	2.04	0.862	0.511
PKFA first 25% of stance (º)	27.45	±	6.94	23.56	±	7.23	0.302	28.73	±	5.69	0.892	0.256
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.70	±	0.33**	**0.000**	0.95	±	0.18**	**0.009**	0.192
PKFA first 50% of stance (º)	32.30	±	5.60	30.04	±	7.07	0.583	33.87	±	5.36	0.799	0.372
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.84	±	0.30**	**0.000**	0.99	±	0.20**	**0.003**	0.482
Peak Trunk Flexion (º)	10.91	±	6.27	13.93	±	10.81	0.539	15.73	±	4.76	0.269	0.866
Peak Trunk Side Bending (º)	1.82	±	2.39	7.15	±	5.81**	**0.001**	3.31	±	1.96^	0.562	0.065
Time on Force Plate (s)	0.72	±	0.10	1.29	±	0.80**	**0.002**	0.88	±	0.13	0.617	0.098
Time to Max vGRF (s)	0.12	±	0.03	0.26	±	0.22**	**0.006**	0.12	±	0.03^	0.998	**0.006**
Total Time on Stairs (s)	1.13	±	0.16	1.81	±	0.91**	**0.002**	1.36	±	0.15	0.495	0.118
Knee Extensor Strength (pounds)	85.41	±	25.97	69.84	±	24.16	0.220	54.95	±	8.70**	**0.008**	0.393

MR = Multi-Radius; CON = Control; SR = Single-Radius; n = number; SD = standard deviation; Max = maximum; vGRF = vertical Ground Reaction Force; N/kg = newton per kilogram; PKFA = peak knee flexion angle; % = percentage; ˚ = degrees; PKFM = peak knee flexion moment; Nm/kg = newton meters per kilogram; s = seconds;
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).
^ = significantly different than Multi-Radius ($p \leq 0.05$).

Table 3.8 One-Year Biomechanical Variables During Stair Ascent, by Implant Design

	Controls (n=22 knees)			Multi-Radius (n=6 knees)			MR to CON	Single-Radius (n=8 knees)			MR to CON	SR to MR
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	11.57	±	1.13	10.21	±	0.27**	**0.013**	10.80	±	0.78	0.150	0.513
PKFA (º)	67.10	±	5.85	58.58	±	7.28	0.224	67.88	±	5.34	0.984	0.272
PKFM (Nm/kg)	1.06	±	0.20	0.56	±	0.19**	**0.000**	0.68	±	0.19**	**0.000**	0.495
Peak Trunk Flexion (º)	18.51	±	5.81	27.13	±	4.15**	**0.004**	22.36	±	5.08	0.215	0.249
Peak Trunk Side Bending (º)	2.84	±	2.27	3.49	±	3.01	0.880	4.62	±	4.21	0.310	0.751
Time (s)	1.91	±	0.21	2.66	±	0.43**	**0.000**	2.18	±	0.27^	0.052	**0.006**

MR = Multi-Radius; CON = Control; SR = Single-Radius; n = number; SD = standard deviation; Max = maximum
vGRF = vertical ground reaction force; Nm/kg = newtons per kilogram; PKFA = peak knee flexion angle;
˚ = degrees; PKFM = peak knee flexion moment; Nm/kg = newton meters per kilogram, s = seconds
* = significantly different than Controls (p ≤ 0.05).
** = significantly different than Controls (p ≤ 0.01).
^ = significantly different than Multi-Radius (p ≤ 0.05)

Table 3.9 One-Year Stair Descent Biomechanical Variables, Mean ± SD

	Controls (n=22 knees)			Multi-Radius (n=6 knees)			MR to CON	Single-Radius (8 knees)			SR to CON	SR to MR
	Mean	±	SD	Mean	±	SD	P-value	Mean	±	SD	P-value	P-value
Max vGRF (N/kg)	15.41	±	2.33	13.59	±	2.55	0.178	13.99	±	1.05	0.237	0.956
PKFA first 25% of stance (º)	27.45	±	6.94	27.13	±	6.32	0.994	27.07	±	4.80	0.989	0.999
PKFM first 25% of stance (Nm/kg)	1.32	±	0.30	0.73	±	0.23**	**0.000**	0.95	±	0.20**	**0.006**	0.285
PKFA first 50% of stance (º)	32.30	±	5.60	32.45	±	7.12	0.998	31.96	±	5.90	0.990	0.987
PKFM first 50% of stance (Nm/kg)	1.39	±	0.28	0.86	±	0.18**	**0.000**	0.99	±	0.20**	**0.002**	0.609
Peak Trunk Flexion (º)	10.91	±	6.27	20.52	±	5.68**	**0.003**	13.59	±	4.76	0.518	0.090
Peak Trunk Side Bending (º)	1.82	±	2.39	4.19	±	3.61	0.134	2.21	±	2.37	0.931	0.348
Time on Force Plate (s)	0.72	±	0.10	1.46	±	0.63**	**0.000**	0.82	±	0.13^^	0.633	**0.000**
Time to Max vGRF (s)	0.12	±	0.03	0.27	±	0.36	0.071	0.12	±	0.02	0.999	0.131
Total Time on Stairs (s)	1.13	±	0.16	2.01	±	0.75**	**0.000**	1.29	±	0.16^^	0.489	**0.001**
Knee Extensor Strength (pounds)	85.41	±	25.97	61.53	±	26.54*	**0.049**	65.63	±	16.53	0.139	0.937

MR = Multi-Radius; CON = Control; SR = Single-Radius; n = number; SD = standard deviation; Max = maximum;
vGRF = vertical Ground Reaction Force; N/kg = newton per kilogram; PKFA = peak knee flexion angle;
% = percentage; ˚ = degrees; PKFM = peak knee flexion moment; Nm/kg = newton meters per kilogram; s = seconds;
* = significantly different than Controls ($p \leq 0.05$).
** = significantly different than Controls ($p \leq 0.01$).
^ = significantly different than Multi-Radius ($p \leq 0.05$).
^^ = significantly different than Multi-Radius ($p \leq 0.01$).

Discussion

The most important finding in this study was that biomechanical and strength deficits remain during stair negotiation one-year post-TKA when compared to healthy controls. Though level ground walking performance has previously been reported to return to normal based on performance in the Timed Up and Go (TUG) test[39], this assessment may not accurately represent the range of post-operative TKA function needed in daily life. In the present study, during the more demanding task of stair negotiation, patients presented decreased knee flexion moments, along with increased trunk compensatory motion at one-year post-TKA, providing evidence of functional deficits that remain in this patient population. Additionally, subsequent investigation of TKA implant designs determined these deficits were independent of implant design.

The Timed-up-and-go (TUG) test is commonly administered to assess TKA function following surgery[51-54]. The outcome variable of the TUG test is total time to complete the activity, therefore, patients may show improvements in completion time but may rely on compensatory motions or a limp during walking gait to accomplish the task[52]. However, activities more challenging than level ground walking, such as stair negotiation, are also important components of daily living[22]. Even for those who do not have stairs in their home, the ability to step up or down confidently and without assistance on a curb or change in walking surface level is important for high quality of life following TKA.

Stair negotiation was used in this biomechanical analysis to assess post-operative function in TKA patients because it has been shown to be a strong predictor of functional decline in older adults[22] and is a more difficult task which places a high demand on knee extensor musculature[23]. Results of this study suggest that stair negotiation analyses adequately identified functional deficits that may not be evident with level ground walking assessments following TKA. Though generally discharged from rehabilitation around three months following surgery, the ability to negotiate a three step staircase remained a difficult task for patients in this cohort. All patients were able to complete the stair negotiation at both six months and one-year following TKA. However, TKA patients required greater time to accomplish both stair ascent and stair descent tasks compared to controls, which speaks to the difficulty of stairs and suggests that the stairs may have been a better test of TKA patients' functional limits. Additionally, though not quantified in the present study, the majority of patients indicated that stair descent remained a task that was associated with a fear of falling, anxiety and negative feelings.

Older individuals require a higher percentage of maximum knee extensor strength during activity than younger adults which may lead to difficulty performing stair negotiation tasks[28]. It is clear that following TKA, recovering knee extensor strength is of importance since muscle weakness can greatly influence joint kinematics[55] and has a direct influence on clinical outcomes[56]. Patients undergoing TKA in the current study displayed a decrease in knee extensor strength at six-months and one-year compared to healthy controls. Although improvements in strength occurred in the one year following TKA, these values did not return to that of aged match controls. Weaknesses in knee extensor strength among TKA patients appears to have most greatly affected KFM during stair negotiation in the present study. Previous research suggests that knee extensor strength is highly correlated to functional performance and that improved post-TKA knee extensor strength could improve functional performance[35].

Knee flexion moment is an important indicator of the forces acting on the knee joint, with a larger KFM demonstrating an increase in joint loading and a willingness to load the knee[29]. Knee flexion moments were decreased during both stair ascent and descent at six months and one year following TKA compared to healthy controls. Changes in KFM are modulated primarily through changes in the magnitude of GRF and through adjustments in the length of the lever arm through which the GRF can act to produce force at the knee, typically via changes in KFA. However, neither GRF or KFA differed between the TKA patients and healthy controls during stair descent in the current study.

Trunk flexion is one common compensatory motion employed due to quadriceps weakness. Increased trunk flexion serves to decrease KFM by moving the body's center of mass anteriorly and decreasing the length of the lever arm through which the GRF can act to produce force at the knee[31]. Compared to healthy controls, TKA patients in the present study exhibited an increase in trunk flexion, during both stair ascent and descent which provides insight to the observed decrease in KFM. By adopting this increase in trunk flexion as a compensatory motion, TKA patients reduce the demand on the quadriceps and reduced their overall FKM during stair negotiation[32]. It is important to note, however, during stair ascent that, in addition to an increased trunk forward flexion, TKA patients also exhibited a decreased vertical GRF which likely contributed to the observed decrease in KFM. It is possible that TKA patients utilized increased trunk flexion to generate momentum to propel themselves up the stairs due to an unwillingness to load the knee as indicated by the decreases in KFM and vertical GRF. Overall, functional deficits in knee extensor strength effected stair negotiation biomechanics by decreasing KFM and increasing compensatory motion during stairs in this TKA patient group.

In addition to comparing all TKA patients' stair negotiation to healthy controls, a further analysis was performed analyzing MR and SR implant designs. Previous research has suggested that SR implants produce a more efficient extensor mechanism post-TKA[49,57,58]. For instance, during stair ascent, the MR implant group demonstrated decreased vGRF and increased trunk forward flexion and total stair ascent time at both six-months and one-year compared to controls. Conversely, the SR group, at six-months post-TKA, displayed an increased knee flexion angle and decreased trunk forward flexion compared to controls. The SR implant group demonstrated stair descent times more similar to controls at six-months but were significantly weaker in knee extensor strength. At one-year post-TKA the MR implant group, in addition to the deficits observed at six-months, also presented significantly increased trunk flexion compared to controls while the SR implant group displayed trunk flexion values that were not different compared to one-year post-TKA. Taken as whole, these results may suggest SR implant designs produced favorable results compared to MR implants as certain variables were more similar to controls over time. However, during stair negotiation at both time periods, KFM was decreased in both the MR and SR implant groups compared to healthy controls, independent of implant design. Therefore, it is difficult to conclude that either SR or MR implant design produce a superior outcome following TKA based on KFM values at one-year post-surgery.

The results of this study demonstrate that functional deficits remain in those individuals undergoing TKA but limitations were present, particularly when comparing implant types. Sample size was limited once the TKA group was subdivided by implant type. Additionally, differences in the number of bilateral patients varied for each implant type. Two of nine patients in the MR implant group (22%) underwent bilateral TKA compared to three of six patients in the SR implant group (50%). These differences may have affected biomechanical variable averages when examining TKA patients and implant design. Further, non-standardization of rehabilitation programs may have constituted a limitation in the present study. However, the rehabilitation protocols for all TKA patients were based on direction from the same board certified physician with the aim of the rehabilitation being to recover patient range of motion. Finally, the healthy controls used for comparison in this study negotiated stairs at a self-selected velocity that was faster than TKA participants, which may have influenced the kinematics and kinetics used in this biomechanical comparison.

Conclusion

Deficits in KFM during stair negotiation and decreased knee extensor strength remained in TKA patients at one-year post-surgery when compared to healthy controls. In compensation for strength deficits, TKA patients increased trunk motions during stair negotiation to manipulate knee joint loading. As the number of TKA's performed annually continues to rise, the ability to be active following TKA is of utmost importance. Though it is known that stair negotiation is a more difficult task than level walking, analyzing stair mechanics in this group of TKA patient identified functional limitations. Total knee arthroplasty patients may be successfully completing rehabilitation programs that are not physically challenging enough or fail to develop restore confidence in loading the knee during more challenging ADL's like stair negotiation. An evaluation of stair ability by rehabilitation specialists may serve to further identify functional weaknesses to be addressed to not only improve patient function but also to restore patient confidence in completing challenging tasks. Results of this study provide a better understanding of the functional deficits that remain in TKA patients following surgery.

CHAPTER 4:

A BIOMECHANICAL ANALYSIS ON THE EFFECT OF PATELLAR THICKNESS FOLLOWING TOTAL KNEE ARTHROPLASTY

Abstract

Context: During total knee arthroplasty (TKA), the patella often undergoes resurfacing during which surgeons replicate the original patellar thickness. Surgical outcomes are favorable when assessed using patient reported questionnaires, but limited biomechanical research has been performed. **Objective:** To evaluate the relationship between the post-patellar thickness (PPT) following TKA, knee extensor strength and sagittal plane biomechanical variables during walking gait and stair negotiation **Design:** Longitudinal. **Setting:** Biomechanics laboratory **Patients or Other Participants:** This study included 15 patients (21 knees) osteoarthritis patients undergoing TKA. **Intervention:** Three dimensional gait kinematics (240 Hz) and kinetics (960 Hz) were collected on participants as they performed walking and stair negotiation tasks at self-selected velocity. Gait was analyzed prior to TKA and again at six weeks, three-months, six-months and one-year post-TKA. **Main Outcome Measures:** The effect of post-TKA patellar thickness on knee kinematics and kinetics during walking and stair negotiation. **Results:** During walking gait, no significant correlations are present between PPT and peak knee flexion angle (PKFA), peak knee flexion moment (PKFM) or vertical ground reaction force (vGRF) at any of the post-TKA data collection time periods. During stair ascent, patellar thickness and PKFA had a strong positive correlation ($r=0.589$, $p=0.027$) at one-year post-TKA. A weak negative correlation was present during stair ascent between patellar thickness and PKFM at six-months ($r=-0.254$) and at one-year ($r=-0.253$). Additionally, a moderate correlation was present at six-months between vGRF ($r=-0.307$), patellar thickness and stair ascent. During stair descent, a strong negative correlation existed between patellar thickness and vGRF ($r=-0.658$, $p=0.014$) at one-year post-TKA. There was also a weak negative correlation between PKFA_25 ($r=-0.225$) and vGRF ($r=-0.244$) and a weak positive correlation between patellar thickness and PKFM_25 ($r=0.278$) at the six-month stair descent data collection time period. Knee extensor strength was positively correlated to post-TKA patellar thickness at both three-months ($r=0.491$, $p=0.053$) and at one-year ($r=0.526$, $p=0.044$). **Conclusions:** Lack of correlation between PPT and walking gait biomechanics may indicate that despite some decrease in knee extension strength related to decreased PPT, these changes were not great enough to prevent normal function during walking gait. As the demands of functional activities increase,

such as in stair negotiation, PPT may become a more important consideration relative to overall function though the precise effect is unclear since increased compensatory motions are present.

Introduction

Degenerative osteoarthritis (OA) commonly affects the articular surface of the patella and is the reason resurfacing is common during total knee arthroplasty (TKA). When the posterior aspect of the patella is resurfaced, the articular surface is removed and a polyethylene button is inserted which is intended to provide a durable articulating surface as well as recreate normal patellofemoral mechanics by restoring the patella to its original thickness[59]. Previous studies have reported no differences in functional Knee Society Scores[60-62], postoperative range of motion measurements[60,62], stair climbing ability[62] or anterior knee pain[62] between patients whose patellae have been resurfaced and those that have not.

Currently, as there is no-gold standard recommendation regarding appropriate patellar thickness to guide surgeons during patellar resurfacing procedures, replicating the original patellar thickness is recommended[59]. The overall rate of patellofemoral complications has been reported to be as low as 7%[63] and good clinical results have been reported with bony patellar remnants of 12 to 13 mm[64]. In vivo analysis has indicated that a patellar thickness of 11-15 mm or less may increase the risk for patellar fracture[65,66]. In addition to fracture, a patella that is too thin may also cause patellar maltracking and anterior knee pain[59,60]. However, a patella that is too thick may lead to post-surgical complications including subluxation[59,67], abnormal patellar tracking[59,60,67,68] greater post-TKA patellar tilt[67] and decreased knee flexion[69].

Limited research has assessed differences in gait parameters related to patellar resurfacing. Smith et al. reported improvements in all spatial-temporal parameters but no differences between patellar resurfaced or non-resurfaced patients following TKA, although patients with patellar resurfacing demonstrated a trend toward increased knee flexion at initial contact[70]. Decreases in knee flexion angle and moment may be characteristic of a “quadriceps avoidance gait” which may develop as a habitual compensatory gait pattern due to pain in the knee joint or muscle weakness[71]. Similarly, from a biomechanical perspective, decreases in patellar thickness following TKA may also lead to a decreased knee flexion angle and moment associated with decreased knee extensor strength from decreased patellar leverage. Therefore, the purpose of this study is to evaluate the relationship between patellar thickness following TKA, knee extensor strength and sagittal plane biomechanical variables during walking gait and stair negotiation. It is hypothesized that patellar thickness will be positively correlated to knee extensor strength and to increased knee function as displayed by increases in knee flexion angle and moment during both walking and stair negotiation.

Methodology

Participants

This study included 15 patients (21 knees) that were recruited for a longitudinal study of osteoarthritis patients undergoing gait analysis prior to TKA and at six weeks, three months, six-months and one year post-TKA. Inclusionary criteria included: 1) under 75 years of age, 2) no previous history of lower extremity fracture, osteotomy, or joint replacement, 3) undergoing an unilateral or bilateral TKA for the treatment of osteoarthritis and 4) able to walk without an aid. The same board certified orthopedic surgeon performed all TKA procedures for the study and all patients signed informed consent forms approved by the Institution's Committee on Human Studies. Patellar thickness measurements were acquired from patient charts as measured by the surgeon from radiographs.

Procedures

All biomechanical analyses were conducted the University Gait Laboratory. Walking gait and stair negotiation biomechanics were collected using 29 retroreflective markers placed on bony landmarks throughout the thorax, pelvis and lower extremities and four marker arrays secured on the thigh and shank segments. Data were collecting using Vicon motion capture system and Vicon Nexus software (Vicon, Inc., Centennial, CO), kinematic data were collected with at 240 Hz and time synchronized with kinetic data collected at 960 Hz[5] collected from two force plates (Advanced Mechanical Technology Incorporated, Boston, MA), one embedded flush with the floor and one instrumented within the second step of the stairs. All kinematic data were smoothed using a low-pass Butterworth filter with a 10 Hz cut-off and ground reaction force was filtered using a 50 Hz cut-off frequency[5]. External joint moments were calculated using inverse dynamics based on marker trajectories and kinetic data which was also filtered using a 10 Hz cut-off frequency[5]. All data was processed using Visual 3D (C-Motion, Inc., Germantown, MD).

Biomechanical variables evaluated in this study included peak knee flexion angle (PKFA), peak knee flexion moment (PKFM) and maximum vertical ground reaction force (vGRF) in walking and stair ascent trials. For the biomechanical analysis of stair descent, in addition to vGRF, knee angle and moment during loading were the variables of interest. These variables were defined as the peak knee flexion angle (PKFA25) and peak knee flexion moment

(PKFM25) during the first 25% of the stance phase while stepping down onto the force plate.

Walking gait was collected barefoot and at self-selected velocity. A successful walking trial required placement of the entire foot on the force plate without a visible change in gait in an attempt to target the force plate with the appropriate foot. Participants performed the minimum number of trials necessary to obtain three acceptable trials for the involved limb undergoing TKA. The stair ascent and decent test included three steps with the following dimensions: step rise, 18 cm; step width, 46 cm; step tread, 28 cm. Handrails were provided for safety but patients were instructed not to use them unless balance was compromised. If the handrails were used, the trial was discarded. Participants were instructed to walk up the stairs at a comfortable speed. After a brief break the participant was then asked to walk down the stairs in the same manner. Patients were instructed to continue walking on the level ground for an additional two steps to ensure a natural gait was continued through the last step and deceleration did not occur before completion of the test. Due to high intra-subject variability previously reported during stair climbing in the OA population, five successful trials were averaged[27].

Bilateral muscle torque was then assessed using a hand held dynamometer (HHD) (Hoggan Health Industries, West Jordan, UT), performed in a gravity dependent position for knee extension. Knee extension torque was measured with the patient seated in a recumbent position (approximately 115° of trunk extension) and the knee placed at 65° of flexion. The HHD was placed on the anterior shank at 80% of the tibial length and secured with a strap. The participant was instructed to extend their knee without extending their trunk. For each strength measure the patient was asked to build a maximum force over a three second time period. Two trials were performed. If the two strength measures were not within ± 10%, a third trial was collected. In addition to the biomechanical variables, peak values from knee extension strength trials was used for statistical analysis.

Statistical Analysis

Two patients were dropped for data analysis, one due to a crouched walking gait and the other for missing two of the three post-TKA data collections, therefore 13 patients (19 knees) were used for data analysis. Data underwent log transformation, for the comparison of post-TKA patellar thickness on post-TKA biomechanical variables of interest. This log transformation will be referred to as post-patellar thickness procedure (PPT). Pearson correlation coefficients were performed to evaluate the relationship between patellar thickness and variables of interest after undergoing log transformations. Statistical analyses were conducting using SPSS version 23.0 (IMB, Armonk, NY, USA). All moments reported are external. A power analysis was performed for the group size of 19 knees in this study and for a power of 0.80, correlations are statistically significant ($p<0.05$) with a $r = 0.456$ value.

Results

Descriptive statistics for all participant demographics and knee extensor strength measurements can be found in Table 4.1. Biomechanical variables descriptive statistics are reported in Table 4.2 (walking), Table 4.3 (stair ascent) and in Table 4.4 (stair descent). The pre- to post-TKA patellar thickness was strongly, positively correlated ($r = 0.818$, $p \leq 0.01$).

Table 4.1 Participant Demographics (n=13, 19 knees)

	Mean	±	SD
Age	66.2	±	5.2
Body Mass (kg)	77.3	±	10.2
Height (m)	1.7	±	0.7
BMI	28.1	±	3.4
Pre-Patellar Thickness (mm)	22.5	±	1.8
Post-Patellar Thickness (mm)	21.1	±	1.8
Delta Thickness	-1.4	±	1.2
Knee Extensor Strength (pounds)			
Pre-TKA	66.8	±	27.5
3 Months Post-TKA	56.6	±	14.3
6 Months Post-TKA	64.0	±	21.0
1 Year Post-TKA	65.0	±	24.6

n = number, SD = standard deviation, kg = kilograms, m = meters, BMI = body mass index, mm = millimeters, TKA=Total Knee Arthroplasty

Table 4.2 Walking Biomechanical Variables

	Pre-TKA (n=19 knees)			3 Months Post-TKA (n=16 knees)			6 Months Post-TKA (n=19 knees)			1 Year Post-TKA (n=16 knees)		
	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
PKFA (°)	22.02	±	9.46	20.69	±	6.23	20.19	±	5.56	19.48	±	4.29
PKFM (Nm/kg)	0.59	±	0.31	0.62	±	0.23	0.68	±	0.23	0.64	±	0.26
vGRF (N/kg)	10.15	±	0.67	10.15	±	0.45	10.36	±	0.80	10.10	±	0.59

TKA = total knee arthroplasty, n = number, SD = standard deviation, PKFA = peak knee flexion angle, ° = degrees, PKFM = peak knee flexion moment, Nm/kg = Newton meters per kilogram, vGRF = vertical ground reaction force, N/kg = newton per kilogram

Table 4.3. Stair Ascent Biomechanical Variables

	Pre-TKA (n=19 knees)			3 Months Post-TKA (n = 18 knees)			6 Months Post-TKA (n = 18 knees)			1 Year Post-TKA (n = 15 knees)		
	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
PKFA (°)	70.69	±	5.95	67.18	±	4.34	66.79	±	16.24	63.72	±	22.78
PKFM (Nm/kg)	0.61	±	0.31	0.64	±	0.66	0.57	±	0.24	0.62	±	0.23
vGRF (N/kg)	10.60	±	0.72	10.53	±	0.72	11.03	±	2.59	10.61	±	2.82

TKA = total knee arthroplasty, n = number, SD = standard deviation, PKFA = peak knee flexion angle, ° = degrees, PKFM = peak knee flexion moment, Nm/kg = Newton meters per kilogram, vGRF = vertical ground reaction force, N/kg = newton per kilogram

Table 4.4. Stair Descent Biomechanical Variables

	Pre-TKA (n=19 knees)			3 Month (n = 16 knees)			6 Month (n = 19 knees)			1 Year (n= 14 knees)		
	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
PKFA_25 (°)	27.35	±	10.33	25.27	±	5.84	26.84	±	7.92	27.46	±	4.42
PKFM_25 (Nm/kg)	0.80	±	0.41	0.82	±	0.26	0.93	±	0.33	0.95	±	0.27
vGRF (N/kg)	13.57	±	4.27	13.56	±	2.32	14.72	±	3.91	13.95	±	1.41

TKA = total knee arthroplasty, n = number, SD = standard deviation,

PKFA_25 = peak knee flexion angle during first 25 percent stance, ° = degrees,

PKFM_25 = peak knee flexion moment during first 25 percent stance, Nm/kg = newton meters per kilogram, vGRF = vertical ground reaction force, N/kg = newton per kilogram

During walking gait, no significant correlations are present between PPT and PKFA, PKFM or vGRF at any of the post-TKA data collection time periods (Table 4.5). During stair ascent (Table 4.6), patellar thickness and PKFA had a strong positive correlation ($r = 0.589$, $p \leq 0.05$) at one year post-TKA. A weak negative correlation was present during stair ascent between patellar thickness and PKFM at six months ($r = -0.254$) and at one year ($r = -0.253$). Additionally, a moderate correlation was present at six months between vGRF ($r = -0.307$) and patellar thickness and stair ascent.

During stair descent (Table 4.7), a strong negative correlation existed between patellar thickness and vGRF ($r = -0.658$, $p \leq 0.01$) at one year post-TKA. There was also a weak negative correlation between PKFA_25 ($r = -0.225$) and vGRF ($r = -0.244$) and a weak positive correlation between patellar thickness and PKFM_25 ($r = 0.278$) at the six-month stair descent data collection time period. Knee extensor strength (Table 4.8) was positively correlated to post-TKA patellar thickness at both three months ($r = 0.491$, $p \leq 0.05$) and at one year ($r = 0.526$, $p \leq 0.05$).

Table 4.5 Walking and Strength Correlations

	PKFA	P-value	PKFM	P-value	vGRF	P-value	Strength	P-value
3 Month	-0.064	0.814	-0.144	0.595	-0.189	0.483	0.547*	**0.019**
6 Month	-0.001	0.998	0.061	0.803	-0.240	0.323	0.336	0.090
1 Year	-0.011	0.970	-0.016	0.956	-0.212	0.447	0.526*	**0.034**

PKFA = peak knee flexion angle, PKFM = peak knee flexion moment,
vGRF = vertical ground reaction force,
* = significantly correlated at p<0.05.

Table 4.6 Stair Ascent Correlations

	PKFA	P-value	PKFM	P-value	vGRF	P-value
3 Month	-0.061	0.822	-0.108	0.692	0.105	0.711
6 Month	-0.011	0.964	-0.254	0.309	-0.307	0.884
1 Year	0.589*	**0.027**	-0.253	0.383	0.091	0.756

PKFA = peak knee flexion angle, PKFM = peak knee flexion moment,
vGRF = vertical ground reaction force,
* = significantly correlated at p<0.05

Table 4.7. Stair Descent Correlations

	PKFA_25	P-value	PKFM_25	P-value	vGRF	P-value
3 Month	-0.169	0.564	0.172	0.556	-0.200	0.492
6 Month	-0.225	0.301	0.278	0.508	-0.244	0.355
1 Year	-0.193	0.527	0.238	0.435	-0.658*	**0.014**

PKFA_25 = peak knee flexion angle first 25 percent stance, PKFM_25 = peak knee flexion moment first 25 percent stance, vGRF = vertical ground reaction force, * = significantly correlated at p<0.05

Discussion

The most important finding of this study was resulting PPT from TKA was not associated with changes in biomechanical variables of interest during walking at all post-TKA time points despite the presence of significant positive correlations between PPT and knee extensor strength at three months and one year post-TKA. During the more demanding task of stair negotiation, PPT may effect some biomechanical variables as indicated by the presence of significant correlations. However, due to the increase in compensatory motions during stair negotiation and the lack of consistency in these correlations across time points, interpreting the biomechanical effects indicated by these correlations was difficult.

The relationship between decreased patellar thickness and increased risk of spontaneous patellar fracture following TKA is well understood within orthopedic research[65,66]. Even though none of the patients in the present study suffered a patellar fracture during the year following TKA, the relationship between the patellar thickness and the forces acting on the knee joint is important to consider relative to the effect on function. Knee flexion angle and the associated knee flexion moment were utilized as indicators of the forces acting on the patella during gait and stair negotiation since higher knee flexion angles have been associated with higher knee flexion moments and increased joint loading[29]. There were however, no associated moderate to strong correlations in this study between PKFM and PPT during walking and only one strong correlation during stair descent.

In the current study, a decreased pre- to post-TKA patellar thickness was reported in 18 of 21(86%) of the knees evaluated. This general decrease in PPT was significantly positively correlated with a decrease in knee extension strength at the three-month and one-year data collection, suggesting that a decrease in patellar thickness following surgery may effect patient function following surgery since changes in the extensor moment arm has been shown to effect quadriceps efficiency[72]. However, no significant correlations were found between PPT and biomechanical variables at any time period during walking and only two significant correlations were present during stair negotiation.

The ability to walk on a level surface following TKA is a reasonably simple task requiring minimal compensatory motions compared to the much more demanding task of negotiating stairs[23]. With this increase in functional demand, patients may increase their whole body compensatory motions, reducing the extent to which differences in function and changes in biomechanical variables during stair ascent and descent may be attributable to the degree of change in patellar thickness. One year post-TKA, functional deficits and compensatory motions still remain during the stair descent[31,73] thus decreasing the extent to which changes in patellar thickness effect biomechanical variables when descending stairs.

Conclusion

Previous research has indicated that in order to limit post-operative complications following TKA, surgeons should avoid overstuffing the patellar leading to decreased range of motion and increased patellar subluxation risk[59,60,67,68], or leaving the patella too thin, due to the risk of patellar fracturing[65,66]. However, the effect of PPT within the ranges normally produced from TKA on walking and stair negotiation biomechanics has not been adequately examined. The results of the present study indicate that maintenance of patellar thickness serves to improve knee extensor strength which may improve patient function post-TKA. However, the lack of correlation between PPT and walking gait biomechanics may indicate that despite some decrease in knee extension strength related to decreased PPT, these changes were not great enough to prevent normal function during walking gait. As the demands of functional activities increase, such as in stair negotiation, PPT may become a more important consideration relative to overall function though the precise effect is unclear since increased compensatory motions are present. Therefore, when viewing the relationship between PPT and biomechanical variables following TKA, it seems appropriate to conclude that as long as patellar thickness remains above the threshold of 11-15 mm[65,6665,66,68,70] for fracture risk, the effect of PPT on function following TKA is likely to be limited.

CHAPTER 5:

EVALUATION OF CHANGES IN WALKING GAIT BIOMECHANICS THAT OCCUR ACROSS AGE-GROUPS

Abstract

Context: The extent to which biomechanical gait variables change due to aging, apart from pathologies is unclear. Deficits in the presence of pathology are identified in the literature. **Objective:** To compare lower extremity biomechanical variables in healthy controls across age-groups. **Design:** Experimental. **Setting:** Biomechanics laboratory **Participants:** 147 community dwelling individuals were placed into four age-groups based on the age of participant. **Intervention:** Three dimensional gait kinematics (240 Hz) and kinetics (960 Hz) were collected on participants as they performed self-selected walking trials. **Main Outcome Measures:** The effect of aging on lower extremity strength, spatiotemporal parameters, joint kinematics, joint kinetics as well as joint powers and joint work. **Results:** Sagittal plane lower extremity variables were not different across the age-groups ($p>0.05$). There were some changes in frontal plane variable of interest varus velocity, which trended to increase statistically in group three ($p=0.065$) and increased statistically in groups four ($p<0.05$). Knee extensor ($p<0.01$) and hip abductor ($p<0.01$) strength decreased statistically across all age-groups. Additionally, step wide decreased ($p<0.01$), step length decreased ($p<0.01$) and cycle time decreased ($p<0.01$) across age-groups. Leg-stiffness increased statistically ($p<0.05$) from age-group one to age-group two. Total work across age-groups did not differ statistically ($p>0.05$). However, sagittal plane ankle work significantly decreased ($p<0.05$) in group four. Additional in the frontal plane a statistically decreased in ankle work was observed ($p<0.05$) and an increased contribution in knee work ($p<0.05$) was observed. **Conclusions:** Biomechanical variables, power and work remained stable in the absence of pathology in a wide range of age-groups. Gait adaptations, deficits and compensatory motions present in individuals with a pathology, appear to be present due to the pathology itself, as data from the current study suggests that the majority of biomechanical parameters remain relatively stable during walking gait across age-groups. Health care professionals observing subtle changes in spatiotemporal parameters in older patients should strongly encourage initiation of a strengthening program and an active lifestyle to prevent further declines in muscular strength and neuromuscular control and subsequent changes to frontal plane gait biomechanics.

Introduction

Functional declines in activities of daily living have been reported to be evident in individuals around the age of 50 years old[74]. Along with functional declines, decreases in muscular strength occur that effect walking gait characteristics during the natural aging process[75]. These changes can include shorter stride lengths[76,77], increased stance times[77] a higher stride frequency[76] and less joint range of motion[78] during gait. Resistance training programs in aging individuals is associated with an improvement in balance[79-81]and proprioceptive abilities[81,82] which deteriorate during the aging process.

Compared to young, elderly individuals experience a decreased muscle mass which leads to decreases strength and changes to power used propel humans forward during locomotion[75]. Research studies have identified similar walking velocities among young and older healthy participants, but how these velocities were achieved were different between age-groups[76,83]. Young individuals, relied more heavily on power from the plantarflexor musculature (73%) to propel forward and very minimal hip power (16%)[76]. However, to account for age-related decreases in plantarflexor strength, elderly individuals heavily rely up hip joint musculature (44%) for propulsion, and decreased plantarflexor power (51%)[76,77,84].

In addition to changes to gait, strength and power, alterations in knee kinematics also occur with aging[85]. For example, in the sagittal plane, decreased ROM during walking gait is associated with aging[78]. In healthy elderly people, the frontal plane variable peak knee adduction moment (KAM) can be used as a tool to measure medial knee contact forces[86]. The presence of certain pathologies can compromise an individuals' mobility to a greater extent than aging alone [87,88]. Osteoarthritis is a highly prevalent disease[37] and accounts for much of the cause of chronic disability in elderly individuals[89]. In healthy individuals an in increased KAM is present which has been highly correlated to OA progression[9,87,90-98]. Additionally, increases in varus velocity (VV) in the knee has been attributed to a decreased neuromuscular control[99]. It has been reported that OA patients experience deficits in proprioceptive acuity and muscular strength[99]. The extent to which gait changes directly due to age apart from pathologies is unclear.

Deficits in OA patients have been identified in the literature, however, having an understanding of the changes in biomechanical variables in healthy subjects as we age is essential. Therefore, the purpose of this study is to compare the function of healthy control age-groups across a lifespan. The main objective of this study is to compare spatiotemporal parameters, frontal and sagittal plane moments and angles, joint powers, leg stiffness and lower extremity strength across age-groups.

Methodology

Subjects

A one-time data collection was performed on 147 community dwelling individuals. All participants were assigned to one of four groups based on their age. Forty-five participants (22 males) were in Group 1, 39 participants (18 males) were in Group 2, 32 participants (11 males) were in Group 3 and 31 participants (15 males) were in Group 4. Subjects were excluded from this study if they were not cleared for physical activity from a physician, or been diagnosed with Parkinson's, a neurological disorder, rheumatoid arthritis, osteoarthritis, experienced a previous lower extremity joint surgery, or lower extremity joint injury within the past six months, or dizziness, fainting or chest pain previously with exercise. Prior to enrollment in the study, all patients signed informed consent forms approved by the Institution's Institutional Review Board.

Procedures

Upon arrival to the laboratory, informed consent was given and participants then completed the Health Questionnaire. In addition to the Health Questionnaire, the UCLA Activity Score was also completed. The UCLA Activity Score, is an ordinal survey from 1-10 that patients use to describe activity level and was used to assess self-reported overall functional and physical activity[26]. Higher UCLA scores indicate a higher amount of rigorous activity level, with choice #10 stating: “Regularly participates in impact sports.” The participants answered the UCLA Activity Score for their current physical activity level, as well as a response was collected for every decade of their life to assess their physical activity across their lifespan. Anthropometric data including height, using a wall mounted stadiometer (Model 67032, Seca Telescopic Stadiometer, Country Technology, Inc., Gays Mills, WI, USA) and body mass was determined using a Detecto certifer scale (Webb City Mo, USA). In addition, age was recorded and the participant was placed in the proper age-group. Femur length was measured from the head of the trochanter to lateral knee joint line and shank lengths were measured from the lateral knee joint line to distal end of the lateral malleolus and were then determined and 80% of femur and shank lengths were calculated and marked. These markings served as location points for the hand held dynamometer during strength testing. This allowed for consistent placement of the handheld dynamometer, relative to each individual.

Walking gait biomechanics were collected using 29 retroreflective markers placed on the thorax, pelvis and lower extremities and four marker arrays on thigh and shank segments, with a Vicon motion capture system and Vicon Nexus software (Vicon, Inc., Centennial, CO). Markers on the medial femoral epicondyle, medial malleolus and head of the first metatarsal were used for calibration purposes during a static trial only and will be removed for stair trials. Kinematic data will be collected at 240 Hz and time synchronized with kinematic data collected at 960 Hz. A low-pass Butterworth filter will be used to filter kinematic data and kinetic data used for calculation of external joint moments at a 10 Hz cut-off frequency and ground reaction force data will be filtered using a 50 Hz cut-off frequency. All data was processed using Visual 3D (C-Motion, Inc., Germantown, MD) and Matlab (Mathworks, Natick MA). Joint moments will be calculated using inverse dynamics based on filtered marker trajectories and kinetic data. All joint moments will be reported as external moments and knee flexion values will be reported as a positive number. Knee and trunk flexion values are reported as a positive number. Additionally, during trunk side bending, a positive value indicates trunk side bending towards the stance leg. Total work was calculated by each plane as the integral of the joint power curve during the stance phase of gait[100].

Participants were asked to walk barefoot across the four-meter data collection field at a self-selected velocity recorded by infrared timers (Speed Trap II, Brower Timing Systems, Draper, UT, USA). A successful walking trial included placement of the entire foot on the force plate without a visible change in gait in an attempt to target the force plate with the appropriate foot. Participants performed the minimum number of trials necessary to obtain three acceptable trials for each leg.

Upon completion of walking, bilateral muscle torque was then assessed using a hand held dynamometer (HHD) (Hoggan Health Industries, West Jordan, UT), performed in a gravity dependent position. Hip abductor strength was tested while the patient was side-lying, with the non-test limb in contact with the table. A pillow was placed between the patients knees for support and to ensure a starting position of 0° hip abduction. The HHD was placed on the mark indicating 80% of the femur length and was secured in place with a strap. The patient was instructed to abduct the hip while maintaining an extended hip and knee. Knee extension torque was measured while the patient is seated with the knee placed at 65° of knee flexion and the trunk was place in 130° of extension with the patients' hands placed behind them on the table to support the trunk extension during knee extension trials. The HHD was placed on the anterior shank, at 80% of the shank length and secured in place with a strap. The patient was instructed to extend their knee, without extending their trunk. For each strength measure, the patient was asked to build force over a two second time period and then maximal contraction for three seconds (a total of five seconds). Two trials were performed unless strength measures did not fall within ± 10%, a third trial was collected. Only the peak measurement recorded was used for data analysis.

Statistical Analysis

Normality was assessed using Shapiro-wilk test. To test for homogeneity of variance between standard deviations the Levene's test was used. Multiple, general linear models were performed to determine significance differences in each age-group for frontal and sagittal knee angles, moments, spatiotemporal parameters, power, work and leg stiffness during gait between groups. Post-hoc Tukey tests were performed to determine where significance differences exist when significant main effects by group were found. All data were analyzed using SPSS Version 22.0 with an alpha level of $p<0.05$ defined as statistical significance and $p<0.10$ defined as a trend toward significance.

Results

Forty-five participants (22 males) were in Group 1 (20-39 years old, mean age 25.5 ± 5.0), 39 participants (18 males) were in Group 2 (40-54 years old, mean age 48.0 ± 3.8), 32 participants (11 males) were in Group 3 individuals (55-64 years old, mean age 59.4 ± 3.1) and 31 participants (15 males) were in Group 4 (65-75 years old, mean age 69.7 ± 2.4). Descriptive statistics for patient demographics are in Table 5.1.

In the analysis of walking biomechanical variables, there were no statistically significantly differences ($p>0.05$) in Group 2(G2) when compared to Group 1 (G1). Compared to G1, there was a significant increase observed in Group 3 (G3) in both knee adduction moment (G1 = 0.40 Nm/kg, G3 = 0.29 Nm/kg, $p<0.01$) and peak hip adduction moment (G1 = 0.90 Nm/kg, G3 = 0.98 Nm/kg, $p<0.05$). Group 3 also had a significantly increased peak ankle inversion moment compared to G2 (G2 = 0.88 Nm/kg, G3 = 0.98 Nm/kg, $p<0.05$). And finally, Group 4 had significantly increased knee varus velocity (G1 = 45.05 ˚/sec, G4 = $p<0.05$) compared to G1. By gender males demonstrated a statistically increased knee varus velocity in G3 when compared to G1 (G1 = 43.77 ˚/sec, G3 = 73.49 ˚/sec, $p<0.05$). When compared to G1, females demonstrated an increased varus velocity (G1 = 46.28 ˚/sec, G4 = 70.32 ˚/sec, $p<0.05$), an increased peak ankle inversion angle (G1 = 6.96˚, G4 = 9.44˚, $p<0.05$) and an decreased trunk forward flexion (G1 = 2.73˚, G4 = 1.38˚, $p<0.05$). Walking biomechanical variables descriptive statistics can be found in Table 5.2.

Table 5.1 Participant Demographics

	Group 1 (n=45)			Group 2 (n=39)			Group 3 (n=32)			Group 4 (n=31)		
	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
Age (years)	25.49	±	4.98	47.97	±	3.84**†	59.46	±	3.16**‡	69.71	±	2.36*^
Height (m)	1.71	±	0.08	1.68	±	0.09	1.67	±	0.11	1.64	±	0.11*
Body Mass (kg)	76.73	±	18.59	77.75	±	14.97	73.35	±	17.43	70.39	±	15.67
UCLA Score	8.33	±	1.31	8.05	±	1.35	7.44	±	1.37*	7.43	±	1.45*

n = number; SD = standard deviation; m = meters; kg = kilograms;
BMI = body mass index; kg/m = kilogram per meter squared;
UCLA = University of California Los Angeles; m/s = meters per second
* = statistically different than Group 1 ($p \leq 0.01$)
^ = significantly different than Age-Group 2 ($p \leq 0.01$).
† = significantly different than Age-Group 3 ($p \leq 0.01$).
‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

Table 5.2 Walking Biomechanical Variables Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)			G4: Age 65-75 (n=31)		
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
Max vGRF (N/kg)	11.24	±	1.22	11.21	±	1.06	10.87	±	0.82	11.00	±	0.82
Peak Knee Flexion Angle (˚)	18.75	±	4.84	18.41	±	6.29	18.93	±	4.11	17.75	±	4.19
Peak Knee Flexion Moment (Nm/kg)	0.82	±	0.26	0.82	±	0.26	0.82	±	0.21	0.78	±	0.24
Peak Knee Adduction Angle (˚)	1.34	±	2.80	2.44	±	3.17	2.89	±	3.64	1.89	±	2.51

Peak Knee Adduction Moment (Nm/kg)	0.40	±	0.11	0.46	±	0.15	0.49	±	0.18*	0.45	±	0.12
Knee Varus Velocity (˚/sec)	45.05	±	22.21	54.21	±	24.17	60.95	±	29.17*	62.31	±	35.79*
Peak Ankle Flexion Angle (˚)	10.09	±	2.70	10.11	±	2.54	9.63	±	2.44	9.78	±	3.22
Peak Ankle Flexion Moment (Nm/kg)	0.19	±	0.06	0.19	±	0.07	0.20	±	0.08	0.21	±	0.06
Peak Ankle Inversion Angle (˚)	7.34	±	2.43	8.31	±	4.59	8.74	±	2.53	9.00	±	2.65
Peak Ankle Inversion Moment (Nm/kg)	0.08	±	0.04	0.02	±	0.21*	0.07	±	0.04	0.07	±	0.04
Peak Hip Flexion Angle (˚)	30.70	±	7.17	28.09	±	7.19	27.69	±	7.35	27.91	±	6.31
Peak Hip Flexion Moment (Nm/kg)	0.62	±	0.18	0.62	±	0.18	0.60	±	0.19	0.58	±	0.22
Hip Adduction Angle (˚)	8.24	±	3.54	6.42	±	2.88	6.77	±	3.36	7.42	±	3.60
Hip Adduction Moment (Nm/kg)	0.90	±	0.14	0.88	±	0.15	0.98	±	0.13*^	0.96	±	0.16
Trunk Forward Flexion (˚)	5.39	±	2.65	5.25	±	3.96	6.92	±	3.35	5.87	±	4.27
Trunk Side Bending (˚)	2.51	±	1.69	2.25	±	1.40	1.96	±	2.48	1.57	±	1.90
Loading Rate	6156.33	±	1512.81	6596.27	±	1748.50	6599.99	±	1696.01	5838.64	±	1742.37
Males	Group 1 (n=22)			Group 2 (n=18)			Group 3 (n=11)			Group 4 (n=15)		
Max vGRF (N/kg)	11.39	±	1.56	11.24	±	1.44	10.81	±	0.95	10.84	±	0.69
Peak Knee Flexion Angle (˚)	20.28	±	5.25	21.40	±	4.98	18.58	±	5.24	18.88	±	2.93
Peak Knee Flexion Moment (Nm/kg)	0.91	±	0.31	0.88	±	0.25	0.85	±	0.23	0.81	±	0.19
Peak Knee Adduction Angle (˚)	1.82	±	3.04	3.25	±	3.05	3.15	±	2.35	2.75	±	2.53
Peak Knee Adduction Moment (Nm/kg)	0.39	±	0.11	0.48	±	0.15	0.48	±	0.13	0.44	±	0.11
Knee Varus Velocity (˚/sec)	43.77	±	24.44	54.89	±	27.83	73.49	±	40.81*	65.57	±	27.48
Peak Ankle Flexion Angle (˚)	9.76	±	2.75	8.87	±	2.05	9.26	±	2.70	9.70	±	3.54
Peak Ankle Flexion Moment (Nm/kg)	0.18	±	0.06	0.19	±	0.08	0.24	±	0.09	0.20	±	0.06
Peak Ankle Inversion Angle (˚)	7.74	±	2.22	8.08	±	5.16	7.11	±	2.34	8.36	±	2.45
Peak Ankle Inversion Moment (Nm/kg)	0.09	±	0.04	0.02	±	0.21	0.06	±	0.04	0.07	±	0.03
Peak Hip Flexion Angle (˚)	31.42	±	5.68	29.51	±	8.15	27.87	±	7.82	28.49	±	7.22
Peak Hip Flexion Moment (Nm/kg)	0.57	±	0.18	0.65	±	0.19	0.67	±	0.15	0.55	±	0.20
Hip Adduction Angle (˚)	6.92	±	3.30	5.61	±	2.12	5.43	±	2.70	5.12	±	2.37
Hip Adduction Moment (Nm/kg)	0.87	±	0.15	0.89	±	0.17	0.95	±	0.10	0.91	±	0.11

Trunk Forward Flexion (˚)	5.04 ± 2.60	5.55 ± 4.44	7.41 ± 3.87	7.32 ± 3.61
Trunk Side Bending (˚)	2.28 ± 1.41	2.09 ± 1.22	3.51 ± 2.06	1.85 ± 1.42†
Loading Rate	6536.27 ± 1516.44	7391.73 ± 1758.44	6884.68 ± 1025.20	6163.58 ± 2020.73
Females	Group 1 (n=23)	Group 2 (n=22)	Group 3 (n=21)	Group 4 (n=16)
Max vGRF (N/kg)	11.11 ± 0.79	11.18 ± 0.61	10.93 ± 0.78	11.14 ± 1.01
Peak Knee Flexion Angle (˚)	17.27 ± 3.98	15.85 ± 6.26	19.37 ± 3.41	17.60 ± 5.26
Peak Knee Flexion Moment (Nm/kg)	0.72 ± 0.16	0.77 ± 0.26	0.82 ± 0.20	0.79 ± 0.31
Peak Knee Adduction Angle (˚)	0.90 ± 2.55	1.74 ± 3.17	2.85 ± 4.30	1.07 ± 2.26
Peak Knee Adduction Moment (Nm/kg)	0.42 ± 0.11	0.43 ± 0.14	0.50 ± 0.16	0.46 ± 0.15
Knee Varus Velocity (˚/sec)	46.28 ± 20.31	53.63 ± 21.22	55.44 ± 18.68	70.32 ± 44.80*
Peak Ankle Flexion Angle (˚)	10.40 ± 2.68	11.17 ± 2.47	9.96 ± 2.31	10.17 ± 3.03
Peak Ankle Flexion Moment (Nm/kg)	0.19 ± 0.06	0.20 ± 0.06	0.19 ± 0.07	0.21 ± 0.07
Peak Ankle Inversion Angle (˚)	6.96 ± 2.61	8.52 ± 4.16	9.68 ± 2.25*	9.87 ± 3.01*
Peak Ankle Inversion Moment (Nm/kg)	0.08 ± 0.04	0.02 ± 0.22	0.07 ± 0.04	0.08 ± 0.05
Peak Hip Flexion Angle (˚)	30.01 ± 8.43	26.87 ± 6.18	28.12 ± 7.06	28.78 ± 5.59
Peak Hip Flexion Moment (Nm/kg)	0.67 ± 0.17	0.59 ± 0.18	0.56 ± 0.21	0.54 ± 0.22
Hip Adduction Angle (˚)	9.5 ± 3.36	7.12 ± 3.29	7.54 ± 3.59	9.57 ± 3.18
Hip Adduction Moment (Nm/kg)	0.92 ± 0.13	0.88 ± 0.14	1.00 ± 0.14	0.99 ± 0.19
Trunk Forward Flexion (˚)	5.73 ± 2.70	4.50 ± 3.59	6.64 ± 3.20	5.06 ± 4.97
Trunk Side Bending (˚)	2.73 ± 1.93	2.38 ± 1.56	1.14 ± 2.37*	0.91 ± 2.29*
Loading Rate	5792.92 ± 1449.02	5914.45 ± 1456.99	5602.40 ± 1831.33	5682.23 ± 1769.64

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; Max = maximum;
vGRF = vertical ground reaction force; N/kg = newtons per kilogram; ˚ = degrees; Nm/kg = newton meters per kilogram;
˚/sec = degrees per second

* = significantly different than Age-Group 1 ($p \leq 0.05$).

** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$).

† = significantly different than Age-Group 3 ($p \leq 0.05$).

In the analysis of spatiotemporal variables, compared to G1, G2 demonstrated a decreased knee extensor strength (G1= 109.92 lbs, G2 = 87.57 lbs, $p<0.01$), decreased hip abductor strength (G1 = 77.69 lbs, G2 = 64.01 lbs, $p<0.01$), decreased stride width (G1 = 0.16 m, G2 = 0.13 m, $p<0.01$), decreased stride length (G1 = 2.24 m, G2 = 2.01 m, $p<0.01$) and a decreased cycle time (G1 = 2.12 s, G2 = 1.62 s, $p<0.01$). A significant increase in hip abductor strength was observed when G2 was compared to G4 (G2 = 64.01 lbs, G4 = 54.06 lbs, $p<0.05$). When G3 was compared to G1 a significantly decreased knee extensor strength (G1= 109.92 lbs, G3 = 81.59 lbs, $p<0.01$), decreased hip abductor strength (G1 = 77.69 lbs, G2 = 53.85 lbs, $p<0.01$), decreased stride width (G1 = 0.16 m, G2 = 0.12 m, $p<0.01$), decreased stride length (G1 = 2.24 m, G2 = 1.95 m, $p<0.01$) and a decreased cycle time (G1 = 2.12 s, G2 = 1.12 s, $p<0.01$). Compared to G1, G4 demonstrated a decreased knee extensor strength (G1= 109.92 lbs, G4 = 75.87 lbs, $p<0.01$), decreased hip abductor strength (G1 = 77.69 lbs, G4 = 54.06 lbs, $p<0.01$), decreased stride width (G1 = 0.16 m, G4 = 0.12 m, $p<0.01$), decreased stride length (G1 = 2.24 m, G2 = 1.85 m, $p<0.01$) and a decreased cycle time (G1 = 2.12 s, G4 = 1.62 s, $p<0.01$). In addition, G4 demonstrated a significantly decreased hip abductor strength compared to G2 (G2= 64.01 lbs, G4 = 54.06 lbs, $p<0.05$). All spatiotemporal descriptive statistics for all genders combined, as well as by males and females are located in Table 5.3.

Table 5.3 Strength and Spatiotemporal Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)			G4: Age 65-75 (n=31)		
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
Knee Extensor Strength (lbs)	109.92	±	36.68	87.57	±	31.16**	81.59	±	25.37**	75.87	±	27.26**
Hip Abductor Strength (lbs)	77.69	±	21.85	64.01	±	20.33**‡	53.85	±	16.29**	54.06	±	17.40**^
Velocity (m/sec)	1.25	±	0.16	1.26	±	0.15	1.23	±	0.16	1.21	±	0.16
Stride Width (m)	0.16	±	0.04	0.13	±	0.03**	0.12	±	0.03**	0.12	±	0.03**
Stride Length (m)	2.24	±	0.33	2.01	±	0.29**	1.95	±	0.27**	1.85	±	0.34**
Cycle Time (s)	2.12	±	0.34	1.62	±	0.21**	1.61	±	0.16**	1.62	±	0.23**
Males	Group 1 (n=22)			Group 2 (n=18)			Group 3 (n=11)			Group 4 (n=15)		
Knee Extensor Strength (lbs)	126.88	±	31.58	105.91	±	34.61*	103.86	±	27.32*	91.93	±	23.48**
Hip Abductor Strength (lbs)	89.13	±	18.65	76.94	±	20.17*	69.12	±	16.16*	63.88	±	16.58**^
Velocity (m/s)	1.23	±	0.14	1.27	±	0.15	1.30	±	0.10	1.18	±	0.17
Stride Width (m)	2.29	±	0.33	2.08	±	0.25	1.93	±	0.26*	1.94	±	0.38**
Stride Length (m)	0.17	±	0.05	0.14	±	0.02	0.13	±	0.03**	0.12	±	0.03**
Cycle Time (s)	2.17	±	0.28	1.63	±	0.20**	1.53	±	0.11**	1.68	±	0.25**
Females	Age 20-39 (n=23)			Age 40-54 (n=22)			Age 55-64 (n=21)			Age 65-75 (n=16)		
Knee Extensor Strength (lbs)	109.92	±	36.68	87.57	±	31.16**	81.59	±	25.37**	75.87	±	27.26**
Hip Abductor Strength (lbs)	77.69	±	21.85	64.01	±	20.33**	53.85	±	16.29**^^	54.06	±	17.40**^
Velocity	1.27	±	0.17	1.26	±	0.15	1.19	±	0.18	1.21	±	0.18
Stride Width (m)	0.65	±	0.15	0.75	±	0.13**	0.73	±	0.10**	0.70	±	0.13**
Stride Length (m)	0.15	±	0.03	0.12	±	0.03*	0.12	±	0.02	0.11	±	0.03**
Cycle Time (s)	2.06	±	0.38	1.60	±	0.22**	1.66	±	0.17**	1.55	±	0.19**

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; lbs = pounds; m/sec = meters per second; m = meters; s = seconds

* = significantly different than Age-Group 1 ($p \leq 0.05$).

** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$).
‡ = significantly different than Age-Group 4 ($p \leq 0.05$).

In an analysis of leg stiffness and power, G2 compared to G1 demonstrated an increased Kvert (G1 = 0.97, G2 = 1.39, $p<0.05$) and a decreased sagittal plane knee power min value (G1 = -.156 Watts, G2 = -1.84 Watts, $p<0.05$). Additionally, compared to G4, G2 demonstrated an increased knee stiffness (G2 + 6.12, G4 = 4.84, $p<0.05$). Males in G2 had a statistically increased Kvert (G1 = 0.96, G2 = 1.80, $p<0.05$) compared to G1. Males in G4 had a statistically increased sagittal plane knee power min value (G1 = -1.46, G4 = -1.88, $p<0.05$) compared to G1. Females on the other hand demonstrated no statistically significant differences in leg stiffness and power ($p>0.05$). All leg stiffness and power descriptive statistics are located in Table 5.4.

Total joint work for the hip, knee and the ankle was not significantly different in an analysis of all genders combined across age-groups ($p>0.05$), males across age-groups ($p>0.05$) or in females across age-groups ($p>0.05$). Total ankle joint work decreased in all genders in G4 when compared to G1 (G1 = 0.80 J/kg, G4 = 0.73 J/kg, $p<0.05$). In males there was a statistically significant decrease in total ankle joint work when G4 was compared to G1 (G1 = 0.78 J/kg, G4 = 0.72 J/kg, $p<0.05$, $p<0.05$). In females, when G4 was compared to G2 there was a significantly decrease in total ankle joint work (G2 = 0.88 J/kg, G4 = 0.74 J/kg, $p<0.05$). Total work descriptive statistics can be found in Table 5.5.

Table 5.4 Stiffness and Power Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)			G4:	
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±
Ankle Stiffness	3.33	±	1.27	3.45	±	1.63	3.18	±	1.49	2.90	±
Knee Stiffness	5.50	±	1.92	6.12	±	2.30‡	5.06	±	1.67	4.84	±
Kvert	0.97	±	0.53	1.39	±	1.01*	1.38	±	0.79	1.26	±
Sagittal Plane Hip Power Max	1.11	±	0.30	1.15	±	0.31	1.17	±	0.39	1.24	±
Sagittal Plane Hip Power Min	-0.90	±	0.32	-0.98	±	0.42	-0.87	±	0.26	-0.96	±
Sagittal Plane Knee Power Max	0.77	±	0.42	0.79	±	0.44	0.70	±	0.42	0.77	±
Sagittal Plane Knee Power Min	-1.56	±	0.36	-1.84	±	0.53*	-1.82	±	0.49	-1.84	±
Sagittal Plane Ankle Power Max	3.18	±	0.81	3.00	±	0.73	3.08	±	0.70	2.75	±
Sagittal Plane Ankle Power Min	-0.71	±	0.24	-0.73	±	0.23	-0.78	±	0.35	-0.71	±
Males	Group 1 (n=22)			Group 2 (n=18)			Group 3 (n=11)			Grou	
Ankle Stiffness	3.59	±	1.04	4.67	±	1.56	4.16	±	1.55	3.08	±
Knee Stiffness	5.77	±	1.61	6.05	±	1.97	5.68	±	1.86	4.88	±
Kvert	0.96	±	0.71	1.80	±	1.32*	1.41	±	0.97	1.34	±
Sagittal Plane Hip Power Max	1.02	±	0.27	1.11	±	0.35	1.22	±	0.40	1.15	±
Sagittal Plane Hip Power Min	-0.80	±	0.27	-0.78	±	0.39	-0.86	±	0.20	-0.85	±
Sagittal Plane Knee Power Max	0.76	±	0.45	0.72	±	0.30	0.68	±	0.37	0.69	±
Sagittal Plane Knee Power Min	-1.46	±	0.34	-1.73	±	0.45	-1.81	±	0.39	-1.88	±
Sagittal Plane Ankle Power Max	3.03	±	0.78	3.05	±	0.73	3.09	±	0.49	2.57	±
Sagittal Plane Ankle Power Min	-0.70	±	0.24	-0.65	±	0.25	-0.85	±	0.32	-0.66	±
Females	Group 1 (n=23)			Group 2 (n=22)			Group 3 (n=21)			Grou	
Ankle Stiffness	3.08	±	1.44	2.40	±	0.70	2.60	±	1.20	2.51	
Knee Stiffness	5.23	±	2.18	6.17	±	2.60	4.74	±	1.54	4.88	
Kvert	0.98	±	0.28	1.03	±	0.43	1.38	±	0.71	1.30	
Sagittal Plane Hip Power Max	1.19	±	0.30	1.19	±	0.27	1.15	±	0.40	1.26	
Sagittal Plane Hip Power Min	-0.99	±	0.33	-1.13	±	0.38	-0.87	±	0.31	1.06	
Sagittal Plane Knee Power Max	0.77	±	0.40	0.84	±	0.53	0.72	±	0.46	0.85	
Sagittal Plane Knee Power Min	-1.65	±	0.36	-1.93	±	0.59	-1.87	±	0.52	1.84	
Sagittal Plane Ankle Power Max	3.33	±	0.82	2.96	±	0.75	3.12	±	0.80	2.82	
Sagittal Plane Ankle Power Min	-0.72	±	0.23	-0.80	±	0.17	-0.75	±	0.36	0.75	

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation;
Kvert = vertical leg stiffness; Max = maximum; Min = minimum

* = significantly different than Age-Group 1 ($p \le 0.05$).

^ = significantly different than Age-Group 2 ($p \le 0.05$).

‡ = significantly different than Age-Group 4 ($p \le 0.05$).

Table 5.5 Total Work Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)		
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD
Total Joint Work (J/kg)	1.69	±	0.30	1.72	±	0.34	1.74	±	0.28
Total Ankle Joint Work (J/kg)	0.80	±	0.16	0.81	±	0.20	0.81	±	0.15‡
Total Knee Joint Work (J/kg)	0.37	±	0.10	0.40	±	0.11	0.41	±	0.09
Total Hip Joint Work (J/kg)	0.52	±	0.11	0.52	±	0.11	0.52	±	0.10
Sagittal Plane Total Joint Work (J/kg)	0.96	±	0.18	1.00	±	0.21	0.98	±	0.17
Frontal Plane Total Joint Work (J/kg)	0.61	±	0.12	0.61	±	0.12	0.63	±	0.12
Transverse Plane Total Joint Work (J/kg)	0.12	±	0.04	0.12	±	0.04	0.12	±	0.04
Male									
Total Joint Work (J/kg)	1.63	±	0.32	1.57	±	0.32	1.73	±	0.1
Total Ankle Joint Work (J/kg)	0.78	±	0.17	0.71	±	0.16	0.84	±	0.12^
Total Knee Joint Work (J/kg)	0.37	±	0.12	0.38	±	0.09	0.39	±	0.07
Total Hip Joint Work (J/kg)	0.48	±	0.10	0.48	±	0.11	0.51	±	0.09
Sagittal Plane Total Joint Work (J/kg)	0.95	±	0.20	0.90	±	0.20	1.00	±	0.12
Frontal Plane Total Joint Work (J/kg)	0.58	±	0.12	0.57	±	0.13	0.62	±	0.11
Transverse Plane Total Joint Work (J/kg)	0.10	±	0.03	0.10	±	0.034	0.12	±	0.04
Female									
Total Joint Work (J/kg)	1.74	±	0.27	1.84	±	0.31	1.74	±	0.31
Total Ankle Joint Work (J/kg)	0.82	±	0.16	0.88	±	0.19	0.79	±	0.16
Total Knee Joint Work (J/kg)	0.37	±	0.08	0.41	±	0.13	0.42	±	0.09
Total Hip Joint Work (J/kg)	0.55	±	0.11	0.545	±	0.11	0.53	±	0.11
Sagittal Plane Total Joint Work (J/kg)	0.98	±	0.16	1.08	±	0.19	0.98	±	0.19
Frontal Plane Total Joint Work (J/kg)	0.63	±	0.12	0.64	±	0.12	0.64	±	0.13
Transverse Plane Total Joint Work (J/kg)	0.14	±	0.04	0.12	±	0.04	0.12	±	0.04

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation, J/kg = joules per kilogram

* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 (p ≤

^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 (p ≤

† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 (p ≤

‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 (p ≤

The percentage of work performed by the ankle decreased in G4 compared to G1 (G1 = 45%, G2 = 45%, $p<0.05$) and the percentage of work performed by the knee increased in G4 when compared to G1 (G1 = 22%, G4 = 24%, $p<0.05$). In males the total work performed at the ankle was statistically decreased when G4 was compared to G1 (G1 = 48%, G4 = 45%, $p<0.05$). In females there was an observed increase in total work at the knee when G3 was compared to G1 (G1 = 21%, G4 = 24%, $p<0.05$). All percentages of work done by each joint are located in Table 5.6.

The percentage of all work done in each plane, by each joint stayed relatively stable in both the sagittal and transverse plane. However, there was an observed decreased sagittal plane ankle work performed in all genders in G1 and G4 (G1 = 41%, G2 = 38%, $p<0.05$). And, in females the sagittal plane knee work increased when G1 was compared to G3 (G1 = 28%, G3 = 31%, $p<0.05$). In the transverse plane in all genders, there was an increased in knee work between G2 and G3 (G2 = 20%, G3 = 35%, $p<0.01$). Additionally, the females had a decreased transverse palne knee work when G2 was compared to G1 (G1 = 32%, G2 = 27%, $p<0.05$) and in G3 (G3 = 35%, G2 = 27%, $p<0.05$). In all genders in the frontal plane, the ankle work decreased between G1 and G4 (G1 = 64%, G4 = 61%, $p<0.05$). Compared to G1, the knee frontal plane work in all genders increased in G2 (G1 = 8%, G2 = 9%, $p<0.05$), G3 (G1 = 8%, G3 = 10%, $p<0.05$) and G4 (G1 = 8%, G4 = 10%, $p<0.05$). In the male only comparison, compared to G1, frontal plane ankle work decreased in G2 (G1 = 64%, G2 = 60%, $p<0.01$) and G4 (G1 = 64%, G2 = 60%, $p<0.01$). However, frontal plane ankle work increased when G3 was compared to G2 (G2 = 60%, G3 = 66%, $p<0.01$) and G4 (G4 = 60%, G3 = 66%, $p<0.01$). Frontal plane knee work in the males increased when G1 was compared to G2 (G1 = 8%, G2 = 10%, $p<0.01$) and G4 (G1 =

8%, G4 = 10%, $p<0.01$). Frontal plane hip work decreased in G3 when compared to G2 (G2 = 30%, G3 = 25%, $p<0.05$) and G4 (G4 = 30%, G3 = 25%, $p<0.05$). In females there was an observed decreased frontal plane ankle work when G2 was compared to G3 (G2 = 67%, G3 = 60%) and G4 (G2 = 67%, G4 = 61%). Additionally, in the females there was an observed decrease in frontal plane hip work when G1 was compared to G2 (G1 = 28%, G2 = 25%, $p<0.05$). There was a increase in frontal plane hip joint work in females when G2 was compared to G3 (G2 = 25%, G3 = 31%, $p<0.01$) and G4 (G2 = 25%, G4 = 30%, $p<0.01$). All percentages of work done in each plane by each joint can be found in Table 5.7.

5.6 Percentage of Total Work Done By Each Joint

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)			G4: Age 65-75 (n=31)		
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
Total Work at Ankle	48%	±	5%	47%	±	5%	46%	±	4%	45%	±	6%*
Total Work at Knee	22%	±	4%	23%	±	4%	24%	±	4%	24%	±	4%*
Total Work at Hip	31%	±	4%	30%	±	4%	30%	±	3%	32%	±	4%
Males												
Total Work at Ankle	48%	±	6%	45%	±	3%	48%	±	3%	45%	±	4%*
Total Work at Knee	22%	±	4%	24%	±	4%	23%	±	4%	24%	±	4%
Total Work at Hip	30%	±	4%	30%	±	2%	29%	±	3%	31%	±	3%
Females												
Total Work at Ankle	47%	±	5%	48%	±	6%	45%	±	4%	45%	±	7%
Total Work at Knee	21%	±	4%	22%	±	4%	24%	±	3%*	23%	±	4%
Total Work at Hip	32%	±	3%	30%	±	5%	31%	±	3%	32%	±	5%

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation,
% = percentage
* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).
^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).
† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).
‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

5.7 Percentage of All Work Done In Each Plane By Each Joint

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)			G4: Age 65-75 (n=31)		
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
Sagittal Plane Ankle Work	41%	±	6%	39%	±	5%	40%	±	5%	38%	±	6%*
Sagittal Plane Knee Work	29%	±	5%	31%	±	6%	31%	±	5%	31%	±	5%
Sagittal Plane Hip Work	31%	±	5%	30%	±	5%	29%	±	4%	31%	±	4%
Frontal Plane Ankle Work	64%	±	7%	64%	±	8%	62%	±	6%	61%	±	7%*
Frontal Plane Knee Work	8%	±	3%	9%	±	4%*	10%	±	4%*	10%	±	3%*
Frontal Plane Hip Work	28%	±	6%	27%	±	6%	28%	±	5%	30%	±	6%
Transverse Plane Ankle Work	26%	±	8%	28%	±	7%	24%	±	7%	27%	±	8%
Transverse Plane Knee Work	33%	±	8%	30%	±	7%	35%	±	9%^^	31%	±	9%
Transverse Plane Hip Work	41%	±	11%	42%	±	9%	40%	±	13%	41%	±	13%
Males												
Sagittal Plane Ankle Work	40%	±	6%	38%	±	5%	41%	±	5%	38%	±	5%
Sagittal Plane Knee Work	30%	±	5%	32%	±	6%	30%	±	5%	31%	±	5%
Sagittal Plane Hip Work	30%	±	5%	29%	±	3%	30%	±	5%	31%	±	4%
Frontal Plane Ankle Work	64%	±	6%	60%	±	5%**	66%	±	4%^^	60%	±	5%*††
Frontal Plane Knee Work	8%	±	2%	10%	±	3%**	9%	±	2%	10%	±	3%**
Frontal Plane Hip Work	28%	±	7%	30%	±	4%	25%	±	4%^	30%	±	5%†
Transverse Plane Ankle Work	28%	±	7%	29%	±	8%	25%	±	7%	28%	±	7%
Transverse Plane Knee Work	35%	±	8%	34%	±	8%	36%	±	11%	33%	±	10%
Transverse Plane Hip Work	36%	±	10%	37%	±	9%	39%	±	14%	39%	±	13%
Females												
Sagittal Plane Ankle Work	41%	±	6%	39%	±	5%	39%	±	5%	38%	±	7%
Sagittal Plane Knee Work	28%	±	5%	30%	±	6%	31%	±	4%*	31%	±	5%
Sagittal Plane Hip Work	31%	±	5%	31%	±	6%	29%	±	4%	32%	±	4%

Frontal Plane Ankle Work	63%	±	7%	67%	±	8%	60%	±	6%^^	61%	±	8%^
Frontal Plane Knee Work	8%	±	3%	9%	±	4%	10%	±	4%	9%	±	3%
Frontal Plane Hip Work	28%	±	6%	25%	±	6%*	31%	±	5%^^	30%	±	6%^^
Transverse Plane Ankle Work	24%	±	8%	27%	±	7%	24%	±	8%	26%	±	8%
Transverse Plane Knee Work	32%	±	8%	27%	±	5%*	35%	±	9%^^	30%	±	8%
Transverse Plane Hip Work	45%	±	12%	45%	±	8%	41%	±	12%	44%	±	13%

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation,
% = percentage

* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).

† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$)

‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

The work contributions from each plane in the hip joint with all genders combined was not statistically significantly different ($p>0.05$). In males, there was an observed decreased in frontal plane contribution to work when G2 was compared to G3 (G2 = 36%, G3 = 31%, $p<0.05$). However, in females there was an observed decrease in the sagittal plane hip contribution when G2 was compared to G1 (G2 = 61%, G1 = 56%, $p<0.05$) and G3 (G2 = 61%, G3 = 54%, $p<0.05$). Additionally, there was an increase in female frontal plane hip contribution when G2 was compared to G3 (G2 = 28%, G3 = 36%, $p<0.01$). In the knee joint in all genders, there was an observed decrease in transverse plane contribution when G1 was compared to G2 (G1 = 11%, G2 = 9%, $p<0.01$) and when G2 was compared to G3 (G2 = 9%, G3 = 11%, $p<0.05$). There were no statistically significant differences in knee contributions by plane in the males ($p>0.05$). However, in the females there was an observed increase in percentage of knee joint sagittal plane work when G2 was compared to G1 (G1 = 74%, G2 = 78%, $p<0.05$) and in G3 (G3 = 74%, G2 = 78%, $p<0.05$). There was a decrease in female knee joint contribution from the transverse plane when G1 was compared to G2 (G1 = 12%, G2 = 8%, $p<0.01$) and G4 (G1 = 12%, G4 = 9%, $p<0.01$). Additionally, there was an observed decrease in female transverse plane knee joint contribution when G2 was compared to G3 (G2 = 8%, G3 = 10%, $p<0.05$). All work contributions from each plane by joint across age-groups is located in Table 5.8.

5.8 Work Contributions From Each Plane By Joint Across Age-Groups

	G1: Age 20-39 (n=45)			G2: Age 40-54 (n=39)			G3: Age 55-64 (n=32)			G4: Age 65-75 (n=31)		
All Genders	Mean	±	SD	Mean	±	SD	Mean	±	SD	Mean	±	SD
Hip Joint:												
% Sagittal Plane	57%	±	8%	59%	±	8%	56%	±	9%	57%	±	1%
% Frontal Plane	33%	±	7%	32%	±	7%	34%	±	7%	33%	±	6%
% Transverse Plane	10%	±	5%	9%	±	3%	10%	±	6%	10%	±	4%
Knee Joint:												
% Sagittal Plane	76%	±	6%	77%	±	7%	74%	±	6%	76%	±	5%
% Frontal Plane	13%	±	5%	14%	±	6%	15%	±	5%	15%	±	4%
% Transverse Plane	10%	±	3%	9%	±	3%**	11%	±	2%^	9%	±	3%
Ankle Joint:												
% Sagittal Plane	48%	±	1%	48%	±	1%*	48%	±	1%	48%	±	1%*
% Frontal Plane	48%	±	1%	48%	±	1%	48%	±	1%	48%	±	1%
% Transverse Plane	4%	±	1%	4%	±	1%	4%	±	1%^	4%	±	1%*†
Males												
Hip Joint:												
% Sagittal Plane	59%	±	9%	55%	±	8%	59%	±	11%	57%	±	9%
% Frontal Plane	34%	±	7%	36%	±	7%	31%	±	6%^	34%	±	6%
% Transverse Plane	8%	±	4%	8%	±	3%	10%	±	7%	9%	±	4%
Knee Joint:												
% Sagittal Plane	78%	±	6%	75%	±	7%	75%	±	3%	75%	±	5%
% Frontal Plane	13%	±	5%	15%	±	5%	14%	±	4%	15%	±	4%
% Transverse Plane	10%	±	3%	9%	±	3%	11%	±	2%	10%	±	3%
Ankle Joint:												
% Sagittal Plane	49%	±	2%	48%	±	1%*	48%	±	0%	48%	±	1%
% Frontal Plane	48%	±	2%	48%	±	1%	48%	±	0%	48%	±	1%
% Transverse Plane	3%	±	1%	4%	±	2%*	3%	±	1%^	4%	±	1%*†

Females				
Hip Joint:				
% Sagittal Plane	56% ± 8%	61% ± 7%*	54% ± 8%^^	57% ± 6%**
% Frontal Plane	32% ± 8%	28% ± 6%	36% ± 8%^^	33% ± 7%**
% Transverse Plane	12% ± 6%	10% ± 3%	10% ± 5%	10% ± 4%
Knee Joint:				
% Sagittal Plane	74% ± 5%	78% ± 7%	74% ± 7%^	77% ± 6%
% Frontal Plane	14% ± 5%	14% ± 6%	16% ± 6%	14% ± 4%
% Transverse Plane	12% ± 2%	8% ± 3%**	10% ± 3%^	9% ± 3%*
Ankle Joint:				
% Sagittal Plane	48% ± 1%	48% ± 1%	48% ± 1%	48% ± 1%
% Frontal Plane	48% ± 1%	48% ± 1%	48% ± 1%	48% ± 1%
% Transverse Plane	4% ± 1%	4% ± 1%	4% ± 1%	4% ± 1%

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation,
% = percentage
* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).
^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).
† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).
‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

Discussion

The most important finding of the present study was that walking gait remained relatively consistent across a wide range of ages in the absence of pathology. Functional and biomechanical gait variables, including walking velocity, loading characteristics, sagittal plane joint moments, powers and planar joint work changed very little between groups representing a stable gait pattern across the lifespan. However, despite the consistency of gait variables in the sagittal plane, important differences existed in strength and frontal plane mechanics between age-groups that provide important insights into age related adaptions in gait not owing to pathology.

Decreased knee extensor and hip abductor muscular strength occurred across the four age-groups which were expected outcomes of this study (Table 5.3). Additionally, significant differences in step length and width were present between age-groups which was also anticipated based on previous research and the expected age related changes in gait. However, while step length predictably shortened, as age increased, step width became narrower which was contrary to expectations based on gait changes common to age-related pathologies and an attempt to stabilize walking gait[101]. These spatiotemporal changes in females occurred across every age-group, which coincided with knee extensor strength decreases of 22%, 25% and 33% in groups two, three and four respectively. Though changes in spatiotemporal parameters in males occurred only between age-groups three and four, these also coincided with knee extensor strength decreases of 19% and 27% in groups three and four, respectively. However, despite decreases in lower extremity strength and changes to spatiotemporal parameters, self-selected walking velocity did not differ across age-groups. These findings contradict those of previous research suggesting that walking velocity decreases with aging[77]. However, walking velocity in the present study was similar to the 1.17 m/sec self-selected walking velocity reported in subjects ages 55-75 by Kirkwood et al.[100]. The maintenance of walking velocity has important clinical implications as preferred walking speed has been associated with increased independence as well as decreased number of hospitalizations and overall health care costs[102].

Human locomotion depends on movements that occur primarily in the sagittal plane[100]. Older participants in this study demonstrated little difference in sagittal plane trunk, ankle, knee or hip joint biomechanics compared to those in the younger age-groups (Table 5.2). However, changes in frontal plane biomechanics found in this study may provide insight into ways in which modulations are made in the frontal plane to accommodate the aging process. Increases in varus velocity (VV) and peak knee adduction moment (KAM) values are highly correlated and are typically associated with OA disease progression[95,103,104]. In OA patients, VV is also attributed to decreased neuromuscular control[99]. Although participants in the current study were non-pathologic, both males and female demonstrated an increase in VV across age-groups, which may be attributed to a decrease in neuromuscular control with aging. Males in the present study significantly increased VV between age-groups two and three (ages of 55-64), and females demonstrated increases in VV later in life, between age-group three and four (ages 65-75). The observed decreases in step-width associated with aging in the present study for both males and females may represent gait compensations aimed at overcoming decreases in frontal plane neuromuscular control that occur with aging. These decreases in step width may serve to improve frontal plane control by decreasing the lever arm through which GRF acts on the knee in the presence of compromised neuromuscular control. This is supported by the small but non-significant increases in KAM observed in this study across groups, despite the significant increases in VV in the older participants.

Common gait changes that occur in the presence of OA gait include: a decreased walking velocity[87,91], increased stride width[105], decreased stride length[87,93], toe out gait[106] and an increased trunk compensatory motions[34,107,108]. These changes in gait occur to manipulate the center of mass in an attempt to make gait more stable and decrease pain in the knee joint[31]. Across the age-groups in the present study, although not significantly significant, trunk forward flexion values did increase. An increased trunk forward flexion in this healthy population, suggests that this adaptation may occur in attempt to make gait more stable and to reduce forces passing through the knee joint, similar to compensations observed in OA patients[31,32].

Despite the presence of pathology related sagittal plane compensatory trunk motions being present in the non-pathologic participants in the present study across age-groups, the same was not true for frontal plane trunk motion. Though an increase in trunk side bending is a common compensatory motion in patients with lower extremity joint pain[33,34], in the present study trunk side bending was significantly decreased in the oldest age-group, when compared to the youngest age-group indicating that in the increased trunk side bending associating with pathology may be attributable only to the pathology and not related to the aging process as well.

It is well understood in literature that propulsive power contributions from the ankle decline whereas contributions from the hip increase in elderly individuals[76]. In the present study, a statistically significant decrease in ankle joint work was observed when group three was compared to group four (Table 5.5). Additionally, the percentage of all work done by the ankle joint decreased in the sagittal and frontal plane across the lifespan (Table 5.7). Previous research has demonstrated that when age-groups walked with similar self-selected velocities, power generation strategies were significantly between age-groups[76,83]. Young individuals relied more heavily on power from the plantarflexor musculature (73%) and very minimally from the hip

(16%) for propulsion[76]. However, to account for age-related decreases in plantarflexor strength, elderly individuals heavily rely on hip joint musculature (44%) for propulsion in the presence of decreased plantarflexor power (51%)[76,77,84]. Despite the significant decreases in ankle power observed across age-groups in the present study, a concomitant increase in hip power was not observed.

In the present study, another notable finding was that the total amount of work performed by the lower extremity during gait, that is, the sum of all of the work performed by each joint in each plane, stayed remarkably stable across all age-groups. Total work for all three planes and all three joints were not statistically significantly different between groups (Table 5.5). Therefore, regardless of age, the amount of total work performed during gait was similar among the age-groups which was not a surprising result because walking velocity and body mass was not statistically different between age-groups. More importantly, the percentage of total work performed by each joint and in each plane also was exceptionally stable across age-groups (Table 5.6).

Since there were no differences across the age-groups, groups three and four were combined to allow comparison to previous research examining gait in older individuals aged 55-75 years. The total amount of work generated at the hip during gait for all participants over 55 years old in the present study was 0.52 J/kg, with 57% contribution from the sagittal plane, 34% in the frontal plane and 10% in the transverse plane. The total amount of work generated at the knee joint in this group during gait was .40 J/kg, with 75% contribution from the sagittal plane, 15% in the frontal plane and 10% in the transverse plane. The total amount of work generated from the ankle joint during gait in this group was 0.77 J/kg, with 48% contribution in the sagittal plane, 48% in the frontal plane and 4% in the transverse plane. These percent work values by joint are markedly higher for the frontal and transverse plane than those reported by Kirkwood et al in individuals 55-75 years old.[100] Kirkwood et al.[100] collected on a total of 30 participants (17 male) compared to a total of 63 (32 male) participants in the current study which could explain the differences in values between studies.

Although there were no differences in the in the total amount of work performed by the lower extremity, a decreased ankle joint work occurred in age-group four (Table 5.6). In male participants, a trend toward decreased ankle joint work (p=0.059) occurred after 40 years old (between groups one and two) and reached significant decrease by the age of 65 years old compared to female participants, in whom a trend toward decreased ankle joint work (p=0.095) began around 65. The overall percentage of work performed at the ankle was also significantly decreased in age-group four when compared to group one. Interestingly, the percentage of total work performed by the knee significantly increased at 65 years old, suggesting that to make up for the decrease in work performed by the ankle as we age, there is an increase in knee work. There was also a statistically significant decrease in both sagittal and frontal plane ankle work observed in age-group four, whereas in the knee in there was an increase in both sagittal and frontal plane work but starting at a younger age (40-55 years old). The sagittal plane decrease in ankle work is likely related to the decrease in ankle power that occurs during that aging process[76,77,84] resulting in a compensatory increase in percentage of knee work observed in the current study. Additionally, at the knee joint there was a statistically decreased percentage of knee transverse plane work performed in the oldest participants (>65 years old). When analyzed by gender, the males experienced a decreased frontal plane joint work in age-group two (40-54 year olds), while the women experience this decrease later in life as observed in the age-group three (55-64). The changes in frontal and transverse work observed at the knee are likely attributable to an increased demand for work at the knee in the presence of decreased ankle power and work as well as decreased neuromuscular control at the knee.

Conclusion

Results of this study indicate that walking gait biomechanical variables, spatiotemporal values, power and work remained stable in the absence of pathology across a wide range of age-groups. Decreases in step-width observed in older participants was contrary to expectations but consistent with observed increases in VV wherein decreased step width may serve as a mechanism by which to control KAM and maintain frontal plane stability related to decreased strength and neuromuscular control associated with aging. Gait adaptations, deficits and compensatory motions present in individuals with a pathology, appear to be present due to the pathology itself, as data from the current study suggests that the majority of biomechanical parameters remain relatively stable during walking gait across age-groups. In elderly individuals, improvements to neuromuscular control and balance have been demonstrated with the initiation of exercise programs[81,82,109]. Therefore, health care professionals observing subtle changes in spatiotemporal parameters in older patients should strongly encourage initiation of a strengthening program and an active lifestyle to prevent further declines in muscular strength and neuromuscular control and subsequent changes to frontal plane gait biomechanics.

REVIEW OF LITERATURE

Methodological Considerations of Research in the Field of Biomechanics

Introduction

Participants in gait research studies are often given instructions to “walk with your head and eyes forward at a comfortable pace” and to “walk as naturally as possible”[1]. Targeting the force plate can be defined as identifying the outline of the force plate to the participant, prior to collecting data, so that they can visually guide their steps to deliberately contact within the force plate boundary, potentially altering their natural gait in attempt to accomplish this task[2,3]. Targeting the force plate has been reported to have no significant impact on spatiotemporal parameters during walking[1,4]. However, in running, significant differences in spatiotemporal parameters were found between the short and long strides when compared to the normal stride lengths, demonstrating that subjects made adjustments to their stride to strike the force plate[3]. No differences have been reported between normal and targeting conditions for ground reaction force (GRF) vectors and the timings of the forces in walking or running gait studies[1-4].

Review of Literature

Changes in step length have been shown to affect ground reaction forces which can be a major limiting factor of gait studies. The study by Wearing et al.[4] investigated the effect of visual targeting the force plate on Spatiotemporal and kinetic measurements in 11 healthy volunteers. Walking gait data were collected at a self-selected speed under two gait conditions; targeting and non-targeting. The 10-meter walkway was covered with paper and the modified foot printing method was used to collect Spatiotemporal data including: step length, heel-to-target distance and step width. Data was analyzed using repeated measures analyses of variance (ANOVA), paired t-tests, Levene’s median test and Kolmogorov-Smirnov test with Lilliefor’s correction. Step length variability significantly increased during the targeting condition. This reflects that the subject’s used visual control strategies while approaching the force plate and adjusted their step length to hit the target. In terms of ground reaction force, there was no difference in magnitude, timing and variability over the five walking trials. They concluded that variability in walking gait step length to hit the force plate, had no effect on the ground reaction force parameters when gait protocols having a defined starting point, specific for each individual’s preferred step length are used.

Effects of targeting the force plate measures and its effect on lower limb joint motion variability are scarce in the current literature. Therefore, Verniba et al.[1] investigated the effect of visual targeting on spatiotemporal, kinematic and kinetic measures during barefoot walking gait in ten (males n=6) healthy individuals. The participants walked on an instrumented carpet which provided the spatiotemporal variables of interest. Participants walked at a self-selected speed for all trials. Visual 3D software was used and data was analyzed using a mixed effects repeated measures of analysis of variance (rmANOVA), and a Tukey correction test was performed. Evidence of targeting was reported because the mean heel-target distance variability for targeting trials decreased progressively for the steps approaching the targeting step, and the post-target steps as well. However, no significant differences between targeting and natural trials were detected in spatiotemporal, kinematic and kinetic gait measures. If adjustments are made to tailored to the individual's gait step and stride length, visual targeting the force plate has no effect on the magnitude or variability of any gait measures.

The purpose of the study by Grabiner et al.[2] was to evaluated the influence of force plate targeting on the variability of ground reaction forces (GRF) in 15 healthy subjects. The subjects were tested under two conditions in which the distance to the force plate was reached when the subject took one step, or multiple steps. An analysis of variance (ANOVA) and post hoc test was used for data analysis. They reported that targeting the force plate did not significantly affect GRF variability in either stepping condition. The number of steps required to reach the force plate on anterior/posterior GRF variability was found to be significant. In conclusion, GRF variability was not significantly affected by targeting the force plate.

The purpose of the study by Challis et al.[3] was to examine the influence of force plate targeting on the magnitude and consistency of the ground reaction force profiles, peak forces and segment angles of the support leg during the stance phase of running in seven male experienced runners. During the practice trials, the starting position of the run was adjusted so that their foot hit the force plate by the fourth footfall, and they were asked to run at a velocity of 3.2 m•s^{-1}±5%. Four trials were recorded for this condition and it was referred to as the "normal" condition. Then the subject's starting position was moved up 50 cm ("short" condition) and then 50 cm ("long" condition) back from their original starting position. Kinematics were determined using a video-based motion analysis system. A repeated analysis of variance (ANOVA), a Bartlett's test and Fisher post hoc test was used for data analysis. They reported no differences between the coefficients of variation of GRF and segment angle profiles among the three conditions. Significant differences between conditions were reported for peak vertical impact forces and their timings, and for the three lower limb segment angles at the start of force plate contact. In summary targeting the force plate may be acceptable depending on the variables being analyzed.

Conclusion

The effect of targeting in walking and running gait parameters has been previously researched with limited consensus[1-4]. There tends to be more agreement that targeting the force plate does not affect spatiotemporal parameters during walking[1,4]. But, that during running trials it is important for participants not to target as it may influence results[4]. However, more research needs to be performed.

Knee Anatomy

Introduction

Having knowledge of the anatomical structures found within the knee joint is of extreme importance. The use of magnetic resonance imaging (MRI) is a common tool used to evaluate knee anatomy[110,111]. This tool can be used on cadaveric knees[110] as well as in those patients under-going total knee arthroplasty[111].

Review of Literature

Iwaki et al.[110] used MRI of six male cadaveric knees to determine the shapes and relative movements of the femur and the tibia. In the sagittal plane, the medial compartment is composed of the arc of two circles and that of the tibia of two angled flats. The anterior facets articulate in extension. At about 20° the femur "rocks" to articulate through the posterior facets. The medial femoral condyle does not move anteroposteriorly with flexion to 110°. Laterally, the femoral condyle is composed entirely, or a single circular facet similar in radius and arc to the posterior medial facet. The tibia is mainly flat. The femur tends to roll backwards with flexion creating a more posterior axis of rotation. The tibia internally rotates with extension due the fact that no anteroposterior motion occurs medially and the backwards rolling that is occurring laterally.

Approximately 20% of TKA patients have pain after implantation and it may be due to impingement of soft tissue around the knee and could be due to imprecise geometry of the tibial implant. The purpose of this study by Hartel et al.[111] was to describe and analyze the anatomy of the tibial plateau at the arthroplasty resection level. They also wanted to determine if there were differences in the shape when compared within genders and different age groups. A total of 237 knee MRI's were evaluated in this study and comprised 107 left and 130 right. For the extraction of the tibial bone silhouette, and active contour detection algorithm (snake) was employed. Comparison among male and female tibial plateaus shows high similarity. The results of this study were that the tibial plateaus were asymmetric and that there was no statistical significance difference between tibial plateaus among gender or age. Indicating that developing total knee arthroplasty implants for specific age groups or specific gender is of minor relevance. However, the subjects used in this study were healthy individuals with no incidence of knee osteoarthritis (OA) so the application of the results may not be applicable for the OA population.

Conclusion

The tibial plateaus are very flat and remain similar among males and females[111]. Using cadaveric knees, it has been identified that the medial compartment is composed of the arc of two circles and that of the tibia of two angled flats[110]. And, the anterior facets articulate in extension[110]. In healthy individuals with no history of OA, developing implants for specific age groups or specific genders is of minor relevance[111].

Aging Characteristics on Functional Ability

Introduction

As we age decreases in strength[28] which is attributed to sarcopenia[75] and functional limitations[74]. Additional a greater total effort is generated at the hip with aging[100]. Elderly work at a higher effort relative to their maximal capability and therefore experience an increased effort during stair negotiation[23]. The functional demand on the knee extensor is the highest during stair descent[28]. Changes in knee kinematics[85] and ankle joint moments[112] is evidence of the effects of aging on functional ability.

Review of Literature

With aging comes functional decline which is usually been measured by self-reported measurements of activities of daily living, or by instrumental activities of daily living (shopping, doing laundry). The purpose of the study by Freedman et al.[74] was to examine recent trends in functional limitations using the US Bureau of the Census's Survey of Income and Program Participation in 25,993 individuals. The functions evaluated and assessed if they had any difficulty in the following: seeing the words and letters in ordinary newspaper print, even when wearing glasses or contacts, lifting or carrying something over 10 pounds, climbing a flight of stairs without resting and walking a quarter of a mile. Data were examined using logistic regression models. Large declines in the prevalence of functional limitations were reported in all age groups but especially in those 80 years or older. In those individuals 50 years and older, walking three block posed the greatest challenge and seeing the words in the newspaper posed the least.

There are certain medical diseases (arthritis, hip fracture, low back pain, diabetes, shortness of breath) that are associated with physical disability. The relationship between arthritis and physical function has been shown to be limited to those activities that require the individual to use the affected joint. The purpose of the study by Guccione et al.[37] was to identify associations between specific medical conditions in the elderly and limitations in seven functional tasks and to compare risks of disability across medical conditions in 1,769 individuals (1060 female) mean age of 73.7 years. The seven functional tasks measured were: walking up and down stairs to the second floor, walking a mile, housekeeping, heavy home chores including: shoveling snow and washing windows, cooking, grocery shopping and carrying bundles weighing 10 pounds. Logistic regression analysis was performed for data analysis. The most prevalent diseases reported were heart disease and knee osteoarthritis, the least prevalent disease were hip fracture and congestive heart failure. Stroke was associated with functional limitations in all seven tasks. Knee osteoarthritis was associated with limitations in four tasks, carrying bundles weighing 10 pounds, walking a mile, housekeeping and stair climbing. In general, stroke, depressive symptoms, hip fracture, knee OA and heart disease account for more physical disability in this group of elderly men and women.

Older adults perform activities of daily living (ADLs) at a substantially greater effort when compared to young adults and an older adults' ability to perform ADLs declines with age. Therefore, the purpose of the study by Hortobágyi et al.[23] was to determine the relative effort to necessary for older adults to execute ADLs and to assess the magnitude of muscle coactivity while negotiating stairs as well as while rising from a chair in 14 (7 women) participants, mean age of 74 years old when compared to 13 young adults (mean age of 22 years). Muscle activity was recorded using electromyography. The laboratory stairs were embedded with a force plate in the second step, step height was 0.19 m and step depth was 0.37 m and participants were not allowed to use the handrails. Five trials of stairs and rising from the chair were averaged for data analysis. Data was analyzed using intraclass correlation coefficient, analysis of variance and a Tukey post hoc test. Compared to young adults, older adults had an increased effort in both ascent (54% of maximal effort in young compared to 78% in older) as well as during stair descent (42% of maximal effort in young compared to 88% in older). Similar results were reported during the chair rise activity, young participants used 42% effort compared to 80% effort in older individuals. The difficulty in ADLs that is observed and reported in elderly individuals may be due to the elderly working at a higher effort relative to their maximum capability than to the absolute functional demands imposed by these two tasks.

Stair descent is a challenging and demanding task for aging adults and the elderly may be operating closer to their maximal joint range of motion limits when compared to young adults. The purpose of the study by Reeves et al.[113] was to identify joint moment and ROM demands using 3D gait analysis in 17 young and 15 elderly adults and to establish the demands relative to maximal capacities using electromyography. Participants walked down the three step staircase, each step was mounted with a force plate. The rise of the stairs was 170 mm, tread depth was 280 mm with a width of 900 mm. Peak strength measurements were collected using an isokinetic dynamometer. Ground reaction force in all three directions were similar between young and elderly participants. The elderly worked at a higher relative capacity (42%) compared to the young adults (30%) as evidence when knee joint moments were normalized to maximal eccentric knee extensor moment. At the beginning and ending of single support phase ankle joint moments were lower in the older adults. The elderly generate lower absolute joint moments at the ankle compared to the young adults which allows them to operate at the same relative proportion of their maximal capacity during stair descent. All four muscles tested by EMG displayed similar patterns of EMG activity between the elderly and young adults. No differences in displacement of center of mass in frontal or sagittal planes. In conclusion the ankle joint is of critical importance during stair descent movements and that exercise-based interventions should target the ankle specifically to improve the safety of stair descent in older adults.

Due to decreases in physical capacity, increases in the demand on lower extremity joints may occur with everyday activities. The purpose of the study by Samuel et al.[28] was to characterize functional demand (FD) at the knee and hip joints during everyday activities in 84 healthy participants ages 60-88 years old. Participants were divided into three age groups: 60-69 years old, 70-79 years old and 80 years old and over. A biomechanical analysis was performed while the participants performed chair rising, sit-down, stair ascent and descent activities. Muscle strength measurements were performed using a dynamometer that was attached to a plinth to measure isometric muscle strength. Data was analyzed using Shapiro-Wilks test and analysis of variance tests. Functional demand was defined as “the muscle moment generated during a task, divided by the maximum isometric strength (expressed as a percentage)." It was reported that participants in their 80's had 76-84% of the strength of those who were in their 60's. Functional demand was higher in the group of participants over the age of 80 and FD was greatest on hip and knee joint extensors than flexors across all activities. The knee extensor demand during gait (101%), stair ascent (103%) and stair descent (120%) and these three activities were reported to be the most demanding to the participants due to the FD associated with these activities. Stair descent places a higher FD on the knee extensors and extensors as well when compared to stair ascent. The high demands of these activities could result in the older adult losing their ability to perform these everyday tasks safely.

Sarcopenia is a progressive withdrawal of anabolism and an increased catabolism combined with a reduced muscle regeneration capacity leads to a decreased muscle mass as we age. The purpose of the paper by Narici et al.[75] was to review the mechanism leading to muscle wasting in old age and its functional consequences. From the second to the eighth decade of life, total lean body mass declines by about 18% in men and 27% in women and is affects the lower extremity more than in the upper extremity musculature which may be due to the detraining effect and decrease in physical activity as we age. During sarcopenia both the muscle fiber size and the actual number leads to the decrease in muscle mass. Functionally, a greater decline in muscle power than in force is problematic for the quality of life of older people since most daily actions, such as raising for a chair or climbing a flight of stairs require the development of muscle power. It is reported that there is a "decline in force per unit of muscle cross-sectional area and in peak power per unit volume" due to sarcopenia. Aging not only leads to changes in skeletal muscles but also in tendon elasticity as well. Strong evidence exists that regular exercise slows down age related decreases in muscle mass and regular exercise should be encouraged as we age.

Limitations in mobility impairs activities of daily living and overall quality of life in older adults. The purpose of the study by Boyer et al.[85] was to test for a significant effect of age in seventy-four healthy participants on knee function during the stance phase of walking. The participants were divided into three groups, younger (mean age of 24 years), middle-age (mean age of 48, and older (mean age of 64). Principal component analysis was performed to characterize and statistically compare the patterns of knee joint movement and their relationships in walking. No differences were reported in walking speeds between the age groups. The magnitude and pattern of deviation from the overall means in knee frontal plane angle and coupling knee flexion kinematics were reported when comparing the younger to the middle and older age groups. Changes in knee kinematics occur with aging. The middle aged and older adults tended to be more abducted, internally rotated and had a posteriorly positioned tibia relative to the femur. These changes occurred in the middle age and older age groups suggesting that midlife changes in neuromuscular physiology or decreases in physical activity may have important consequences. These difference in kinematic measures offer the potential to identify early markers for the assessing the risk of developing knee OA with aging.

Gait changes are common as we age, however, few studies describe hip and knee joint power and mechanical work during gait. Therefore, Kirkwood et al.[100] quantified ROM, joint moments, power and mechanical work using 3D gait analysis during walking gait in 30 subjects aged 55-75 years old. Students t-test were used for data analysis. The average speed for the females were 1.13 m/sec and for the male was 1.35 m/sec for a total speed average of 1.17 m/sec for both genders combined. The total effort generated by the hip joint during gait was greater than the one generated at the knee joint. For example, the hip joint generated a total effort of 0.40 joules/kg; with 22% contribution from the frontal plane, 76 % from the sagittal plane and only 2% from the transverse plane. The total effort generated at the knee joint during gait was 0.30 joules/kg; with 7% occurring in the frontal plane, 90% from the sagittal plane and 3% from the transverse plane. Reduced work at the hip was thought to be due to hip abductor weakness that occurs in elderly. It was reported that the strategy employed by elderly subjects was to reduce knee work in the frontal plane and increase it at the sagittal plane in attempt to use a larger portion of knee flexor/extensor muscles and contributing to an increased balance during gait.

Conclusion

Through the use of survey's[74], functional assessments[23,28,37,112] and 3D gait analysis[28,100,112] are common tools used assess functional ability as we age. Decreases in strength[28] and functional limitations[74] occur as we age. To accommodate for these declines that occur with aging, the human body changes knee kinematics[85], ankle kinetics[112] and relies on the hip in the sagittal plane[100] in order to maintain function during the aging process.

Healthy Controls and Stair Negotiation

Introduction

Prior to gaining the knowledge of osteoarthritis sufferer's ability to negotiate stairs, it is important to evaluate this task in healthy individuals. Even in those individuals whom are free from pathology, middle and older individual's negotiated the stairs more slowly[114] and produced smaller vertical ground reaction force than younger individuals[114]. In the frontal plane, the varus-valgus (VV) moment of the knee were increased when compared to level walking[115]. In terms of kinetics and stair ascent, vertical ground reaction force (vGRF) was reported as having a double hump, with the second peak maximum more dominant and mediolateral GRF was a major contributor to the variation in the knee moments[114]. Alternatively, the ground reaction force (GRF) waveforms during stair descent were reported to have variation, with or without and second peak[114] and that vGRF was a major contribution to variation in the frontal plane knee moments[114]. The demands on our lower extremities are increased with stair descent, and it was reported that GRF's of 1.49-1.6 higher when compared to stair ascent[114].

Review of Literature

During level walking the vGRF curve is generally known to be highly repeatable within individuals. However, during stair ascent and descent, little is known about the shape of the vGRF curve and its characteristics. Therefore, Stacoff et al.[114] compared the vGRF parameters during level walking, stair ascent and descent on three different age groups. There were 20 subjects each, which were established based on age (young 33.7 years, middle 63.6 years, old 76.5 years). Subject wore shoes for their stair negotiation trials. Each subject was asked to repeat seven test conditions 8-10 times at his or her comfortable speed. Analysis of variance (ANOVA), Bonferroni post hoc test and two-tailed t-test was used for data analysis. The vGRF curves showed considerable variations between the test conditions, the subjects, and within each subject. During stair ascent, a double waveform is present, but the second maximum peak is more dominant then the fist and the force values were just above one body weight. During stair descent, the typical waveform is no longer present and is replaced by a peak first curve and a large variation with and without a second peak, and values were between 1.49 and 1.6 times body weight. Age was found to be a factor which should be considered because the young group walked fasted and produced larger vGRF during level walking and on stair ascent than the middle and old age groups. Differences between the middle and old age group were found to be very small.

It is essential to have basic knowledge and an understanding of the kinematics and kinetics of the VV motion of the knee during stair negotiation. Therefore, the purpose of the study by Yu et al.[115] compared the maximum VV moment of the knee and the VV moment of the lower leg of ten healthy adult participants during level walking and during stair climbing. Participants were instructed to walk at a self-selected speed and gait analysis was performed for level walking, stair ascent and stair descent. For data analysis of variance and regression analysis were performed. All ten subjects demonstrated a valgus moment at the knee in the stance phases of stair climbing and level walking. They reported a significant increase in the VV moment of the knee during stair ascent when compared to level walking. There was a coupling that occurred between the VV moment and the flexion-extension motions at the knee. The vGRF was a major contributor to the within subject variation in the VV moment of the knee during stair descent and level walking, and the medial lateral GRF was the major contributor to the variation in the moment during stair ascent. And finally the knee VV angle was a major contributor to the between subject variation in the valgus moment of the knee during stair climbing and level walking. The results of this study suggest that the valgus moment of the knee may be an important dynamic factor for differentiating subject with knee VV deformities, especially in stair climbing.

Conclusion

Through gait analysis during stair negotiation, we are able to gain valuable information regarding GRF[114] and frontal plane moments[115] in healthy, non-pathologic individuals. Variability in frontal plane moments were reported during stair ascent and were attributed to mediolateral GRF. However, vGRF was major contributor to the variation of frontal plane moments during stair descent[115]. Further evaluation of GRF waveforms reveals changes reported during both stair ascent and descent[114].

Elderly and Walking Gait

Introduction

Walking gait characteristics change as we age. Elderly individuals in general walk with a decreased walking velocity, which is attributed to a short stride length as well as a decrease in power during push-off [77]. Additionally, an increased walking velocity in elderly has been highly correlated improved health status[102]. Knee adduction moment (KAM) is an important biomechanical knee variable and has been linked to individuals with knee osteoarthritis (OA). Even in healthy, elderly individuals, a positive correlation exists between knee adduction moment and maximal medial knee contact forces[86]. Elderly individuals make adaptations to gait variables lead to a safer, more steady gait[77].

Review of Literature

In the elderly a lower cadence, shorter and more variable step length, increased head and torso flexion, and an increased knee flexion have been identified in the literature. Fifteen fit and healthy elderly individuals underwent a gait analysis and were used in a study by Winter et al.[77] to determine the normal biological degeneration that takes place with aging and to examine if major or subtle changes would point to the degeneration or to the compensations that reduce the change of the person stumbling or losing their balance. Identical walking gait variables were taken from 12 young adults with a mean age of 24.6 years for a comparison. Modified t-tests were used for data analysis. The elderly in the study had a decrease walking velocity attributed to a short stride length that was associated with an increase in stance time and a change in total double-support (from 24.6% in young to 31% in the elderly). The researchers attributed these changes due to the decrease power during push-off in elderly because of the decrease in strength the plantarflexors an adaption that leads to a safer, less destabilizing gait stride. Toe clearance did not change among groups. In research it is important to understand and recognize the biomechanical gait changes that occur due to aging in elderly subjects when conducting research on elderly individuals with balance disorders.

Some studies use a musculoskeletal model-based simulation analysis and have demonstrated a correlation between the external KAM and the medial knee contact force. The purpose of the study by Ogaya et al.[86] was to investigate the correlation between the external KAM and the medial knee contact force during gait in 122 healthy older people using the musculoskeletal model-based simulation analysis. Three dimensional gait analysis was performed and participants walked at a self-selected speed and a musculoskeletal model was created for each of the participants. Inverse dynamics were used to calculate muscle force and joint reaction forces. Pearson's correlation coefficient test was used for data analysis. It was reported that the first peak of medial knee contact force had a strong correlation with the first peak of KAM as well as the maximum extension moment. The second peak value of the medial knee contact force had significant moderate positive correlations with both the second peak values of KAM and the maximal external extension moment. Through analyzing the gait and medial knee contact forces of healthy elderly people, a significant positive correlation between maximum medial knee contact force and maximum KAM and external knee extension moment and that KAM value can be used as a measure of the medial knee contact force.

Healthcare providers can evaluate walking gait velocity which is an important clinical assessment and is easy to conduct. The purpose of the study by Purser et al.[102] was examine the relationship between walking velocity and hospitalization related health services sought out in 1,388 (mean age 74.2) males. It was reported that for every 0.10 m/s reduction in baseline walking velocity, that it was associated with decreased physical functioning, more disabilities, additional rehabilitation visits, longer hospital stays and higher costs. Additionally, an increase in 0.10 m/s in walking velocity, an improved physical function, fewer basic disabilities, a decrease in hospital stays and an overall reduction of $1,188 a year were reported. Walking speed is a useful assessment tool for elderly individuals and may help predict those who will need and use more health-related services.

Conclusion

As previously mentioned, walking gait velocity decreases, and elderly individuals may adopt compensatory strategies in order to walk with more confidence[77,86]. Assessment of walking gait may help predict elderly individuals need for an increase in health-related services[102]. Three dimensional gait analyses is the preferred method to evaluating gait characteristics and to gain insight into elderly gait[77,86].

Elderly and Stair Negotiation

Introduction

The ability to climb stairs is a key activity that allows for functional independence, and stair negotiation is a great way to assess functional status in elderly individuals[22]. Stair descent is the leading cause of falls during stair negotiation[36]. Handrail usage was increased in elderly individuals with poor vision, lower leg strength and balance[116]. By increasing lower limb strength and balance, the elderly may increase their ability to negotiate stairs[116]. Stair ascent times was greatest in older men, and were associated with older individuals, regardless of gender[22]. Stair descent has been reported to be more difficult than ascent[22,36].

Review of Literature

The ability to negotiate stairs is an important activity of daily living and independency in the elderly. The purpose of the study by Tiedemann et al.[116] was to investigate the contributions of physical and psychological factors with stair negotiation in 644 community dwelling individual’s aged 75-98. Participants were timed while they walked up and down a set of eight stairs, in which they could use the handrail and a mobility aid if preferred. Visual acuity and visual contrast sensitivity was assessed using a log Minimum Angle Resolvable letter chart and the Melbourne Edge Test. Ankle dorsiflexion, knee extension and knee flexion strength was assessed and normalized for body weight, balance and reaction time was also measured. The Short-Form 12 Health Status Questionnaire (SF-12) assessed pain, depression, anxiety and vitality. For data analysis Pearson correlation coefficients, hierarchical multiple regression, independent t-tests were used. Knee extension and flexion strength were reported to be predictors of both stair ascent and descent speed. Balance deficits also resulted in slower, more tentative stair negotiation. Individuals who were more likely to perform poorly in vision, strength testing, and balance tests, had a fear of falling and reduced vitality were reported to use the handrails during stair negotiation. The results of this study imply that exercise training, to increase lower limb strength and balance, in addition to visual interventions for older individuals, may result in an increased ability to negotiate stairs in a safe and efficient manner.

For older adults, the ability to climb stairs is considered a key marker of functional independence. However, there are no current standard stair negotiation performance tests. Therefore, the purpose of the study by Oh Park et al.[22] was to establish reference values for stair ascend and descent times in community dwelling older adults and to examine their predictive validity for functional decline. Participants climbed three stairs at their preferred pace and could use the handrails. Each step measured 18 cm in height, 26 cm in depth and 110 cm in width. Stair ascent trial times were recorded independently of stair descent times. In addition, seven activities of daily living (ADL's) were assessed every two to three months and functional decline was defined as an increment of 1-point or more on the disability score. Data was analyzed using the Kaplan-Meier method, Cox proportional-hazards regression analysis and the Goodness-of-fit test. Stair descent time was greater with older age in both genders and stair ascent time was greater with the older aged men. In addition, stair descent time was a strong predictor for functional decline in this group of older adults and was a more difficult ADL than stair ascent. The stair negotiation time is a quick and simple measure that can be used in clinical settings to assess functional status.

The ability to negotiate stairs is essential as we age to maintain independent living. The purpose of the study by Karamanidis et al.[117] was to examine the external knee adduction moments in 27 older (mean 64 years) and 16 young (mean 28 years) individuals during stair the descent activity. Stair measurements were: 17 cm height, 29 cm tread depth and 39 cm width of a two-step staircase. Subjects' were not allowed to use the handrails and three valid trials were recorded using Vicon cameras. An analysis of variance was used for data analysis. The older adults had higher external flexion moments at the end of the single support phase (trailing leg) demonstrating that the older adults made a greater use of the trailing leg for preparing the initiation of the double support phase when compared to the younger participants. In the frontal plane, KAM values as well as knee adduction angular impulse were 50% higher during the single support phase for the older adults when compared to the young which increases the mechanical load at the medial compartment of the knee in these elderly participants. The magnitude of the ground reaction force vector was lower during the double support phase (leading leg) and during the single support phase (trailing leg) for the elderly. In the frontal plane, the older adults had a significantly more medial position of the line of action of the GRF relative to the ankle and the knee joint when compared to the healthy. These observed changes between the leading and trailing leg, which may be affected by the age-related decline in muscle strength in the elderly cause a redistribution of the mechanical load at the knee joint affecting the initiation and progression of knee OA in the elderly.

Understanding the loading forces of the knee joint is required for various investigations in total knee replacements. Previously musculo-skeletal models in addition to gait analysis were used and algorithms calculate the joint forces and moments acting on the knee joint. The purpose of this study by Heinlein et al.[118] was to provide in vivo loading data of the knee joint using a telemetric tibial tray to measure the forces and moments, during level walking and stair negotiation trials at one week, six and ten-months post-total knee replacement. The highest mean values of the peak load components were 276% body weight (BW) in the axial direction, 21% BW medio-laterally and 29% BW in the antero-posterior direction. During stair negotiation, stair descent produced the highest forces of 352% BW (axial), 35% BW medio-lateral and 36% BW antero-posterior. The sagittal and frontal plane moments increase to 2.8% BW*mass and 4.6% BW*mass respectively. This study is one of the first to evaluate mechanical load tibial baseplates during functional activities. These simulators can validate musculo-skeletal models.

Age-related decline in strength and physiological characteristics are some of the reasons that elderly have difficulty with stair negotiation. The impact of physical fitness on stair negotiation in elderly has not been well established. Therefore, the purpose of the study by Mian et al.[119] was to compare lower extremity motion using motion capture in 23 young and in 34 healthy older adults and to determine the effect of a 12-month exercise training program had on lower extremity stair descent kinematics in healthy older adults. The stair dimensions included a three step staircase with a rise of 17 cm, tread of 28 cm and a width of 50 cm and force plates were mounted into each step and participants performed six trials of stair descent. After the initial data collection participants in the older group were randomly assigned to either a control group or a fitness group (n=14 for both groups). The fitness group attended two supervised exercise program and completed one home based session, all session lasting an hour in length for the following 12 months. The control group carried on with their normal daily activities. Student t-tests were used to compare the young and older values, while an analysis of variance was used to determine the effect of the exercise program. Total descent times, stride cycle and single support times were longer in the older group when compared to the young. Peak knee motion flexion was also lower in the old which was surmised to be associated to the frontal plane hip and pelvis motion. While frontal and transverse plane pelvis and hip motion were higher in the older group which was attributed to insufficient neuromuscular control. It was reported that the exercise training program did not reduce the age-related differences in the stair descent activity.

Older adults and those suffering from pathology may be forced to adjust their stair gait pattern due to decreases in muscular strength, proprioception, and balance associated with aging and disease. These individuals may adopt an increased handrail usage or a step-by-step pattern for stair descent negotiation. The purpose of the study by Ried et al.[120] was to compare the kinematics and kinetics of the knee joint during traditional step-over-step (SOS) and compensatory step-by-step lead-leg (SBSL) and step-by-step trail-leg (SBST) stair negotiation in 17 healthy adults (mean age = 23 years). Each participant performed five stair ascent and descent trials using three different stepping patterns. Each step height was 15 cm, depth was 26 cm and step tread was 56 cm. All data was analyzed using independent t-tests. During stair ascent different peak anteroposterior forces were observed across all three stepping patterns. During stair descent, the initial AP peak force for SOS was larger than the SBSL and SBST. However, the second peak force for SOS and SBST were larger than SBSL. In terms of joint moments, during stair ascent the flexion moments of SOS and SBSL patterns were similar and much larger than the SBST moments. During stair descent the initial peak flexion moment for the SOS pattern was larger than SBSL and SBST, whereas during the second peak, SOS and SBST were no different and larger than SBSL. Overall, the SBSL during stair ascent and the SBST during stair descent had the highest loads. During SOS descent, the data suggests that the lowering of the body is controlled by the trial leg while the lead leg reaches for the lower step. Results from this study suggest that in the presence of arthritis, pain may be reduced during stair ascent by using the more painful leg as the support or trail leg. This is of importance because it increases our knowledge of alternative stepping patterns during stair negotiation and have important clinical implications.

Falls on stairs occur in elderly people causing serious and sometimes fatal injuries. The purpose of the study by Svanstrom et al.[36] was to investigate, provide data and insight into falls associated with stair negotiation. Stair descent was reported to be the most dangerous and accounted for 76% of the accidents occurring on stairs. Injuries to the head accounted for 36% of injuries, 19% and 15% were of the upper and lower extremity respectively and 8% had an injury to the abdominal area.

The purpose of the study by Fisher et al [71] was to determine whether 90 subjects suffering from knee OA had a reduced muscle strength when compared to 104 healthy controls. The OA subjects had increased difficult and pain for reported activities of daily living and significantly lower strength for knee extension and flexion. Additionally, they had significantly lower quadriceps and hamstring endurance and velocity when compared to healthy controls. This data demonstrates that patients suffering from knee OA have reduced muscle function and a decreased muscle capacity when compared to a healthy control group.

Conclusion

Elderly individuals tend to ascend the stairs more slowing[22], and are more apt to use handrails with decreases in vision, balance and lower leg strength[116]. Clinically, stair negotiation can be used to assess functional status in elderly individuals[22]. And it is important for elderly to increase lower limb strength and balance to improve function and ability to negotiate stairs[116].

Aging Characteristics and Biomechanical Changes

Introduction

As we age we are able to maintain walking velocity but to maintain that velocity we decrease spatiotemporal variables[84]. A walking velocity of 1.7 meters per second (m/sec) has been identified in elderly individuals[100]. Elderly individuals while walking at a higher speed comes with higher metabolic cost of walking[121]. More specifically they experience more knee flexion at initial contact (IC)[76] but less range of motion over walking gait cycle[122] and stepped down with less ROM which resulted in a stiffer leg[78,83] to accommodate for a reduced neuromuscular system. They also really on larger contributions from the hip and less contribution from knee and ankle[76,84,123,124]. The knee mostly functions in the sagittal plane[100].

Review of Literature

Gait kinematics are different between healthy young and elderly adults and much of the discrepancies can be attributed to differences in self-selected walking velocity. The purpose of the study by DeVita et al.[76] was to compare the joint torques and power of 12 elder adults and 14 young adults while walking at 1.48 m/s over a force plate while being videotaped and all subjects had shoes on. No differences were reported in walking velocity. The elderly participants spent less time in swing and walked with a shorter stride length and at a higher stride frequency compared to the young. At the hip joint, the elderly group were more flexed throughout the entire gait cycle, which was attributed to a more forward trunk lean. At the knee they were in more knee flexion at initial contact but flexed the knee less throughout the remainder of stance compared to the young group. And, at the ankle joint they maintained a more neutral ankle position through the stance phase of walking. The joint torques were significantly different between the groups, but they overall combined to produce nearly identical torque curves. The elderly experienced larger contributions of power from the hip extensors and smaller contributions from the knee extensors and ankle plantarflexors. These results support the concepts that biomechanical and physiological consequences of aging are not solely a reduction in motor abilities but are a qualitative change in underlying neuromuscular components of a motor performance.

It remains unclear weather changes in walking mechanics that occur with aging are natural consequences and to what extent these changes are attributed to a reduction of physical activity and fitness. Therefore the purpose of the study by Boyer et al.[123]was to determine if the walking mechanics in 123 older (>50 years old) highly active individuals and compared the variables to 33 younger (<40 years old) individuals. When compared to the young group, the older group walked at a similar walking velocity however, how this was achieved varied due to the significant differences were reported in all spatiotemporal variables between the two groups.

Walking speed was maintained by increasing cadence while reducing stride length in the older group. During heel contact, the ankle in the older group was in a more plantar-flexed position compared to the young adults. Additionally, the ankle in the older group was less plantarflexed at push-off compared to young adults. Ankle dorsiflexion and plantarflexion moments as well as hip extension moments were different between the age groups. These results indicate that there is a small effect of age on walking gait mechanics in a population of highly active older walkers and therefore increased activity with aging can mitigate declines in walking performance and mechanics with age. The physical activity in the older participants may have minimized the magnitude of age-related changes that occur in gait mechanics.

The purpose of the study by Peterson et al.[121] was to determine how age and walking speed affect metabolic cost of walking in 14 young (mean age of 25 years) and 14 older (mean age of 71) while participants walked on a treadmill at four differing speeds while electromyography and oxygen consumption were measured. The net cost of walking was higher in older adults at each walking speed, and was 23% higher in older adults across all walking speeds. Similar spatiotemporal parameters were reported. Older adults had higher coactivation in the thigh musculature. Total coactivation showed a significant positive relationship to the metabolic cost of walking at all walking speeds. Higher metabolic cost of walking and higher coactivation in older adults, along with the positive relationship between metabolic cost of walking and coactivation implies that coactivation contributes to a higher metabolic cost of walking in older adults.

A reduction in ankle power generation leads to gait changes in a healthy older adult population. The purpose of the study by Cofre et al.[124] was to investigate lower joint power and work in eight older adults (66.8 years old) to 12 young adults (age 26.6 year old) when walking at matched speeds of 1.0, 1.3 and 1.6 m/s^{-1} and self-selected speeds. Data were analyzed using multiple MANOVA's. Speed did not differ between groups. The older group generated 17% less ankle power and 21% less work compared to the young group. Additionally, the older group generated 47% more hip work, 30% more hip peak power, 30% more knee peak power and 19% more peak knee power. The action by the older group were associated with less ankle plantarflexion (44% less peak ankle plantarflexion), more hip flexion and anterior pelvic tilt. Additionally, the older group adopted a different gait pattern at the faster speeds by generating more hip work than ankle work, meaning that this group relied more on hip flexors to propel the leg into swing with a reduced reliance on ankle plantarflexion function.

In order to descend stair, sufficient lower limb strength is required to control and support the entire body mass on the single limb while moving down the stairs. A lack of joint stiffness could result instability, therefore, the purpose of the study by Lark et al.[122] was to determine whether the stance limb ankle and knee joint torques and dynamic joint stiffness differed between six young men (23.6 years old) and six elderly men (67.7 years old) during a step down tasks at three steps heights of 200, 250 and 300 mm. Repeated measures ANOVA were used for data analysis. The elderly had a 15% less passive range of motion for knee flexion and 41% less ankle dorsiflexion compared to the young men. Total time to complete the movement did not differ, however, the elderly spent more time in foot-flat compared to the young. This allowed for the elderly to maintain foot flat position longer and was attributed to maintain a larger base of support for longer and thereby increasing stability. Maximum ankle torque values were lower in the elderly and occurred at a larger dorsiflexion angle. Ankle stiffness was significantly less in the elderly group at all step heights compared to the young group. In both groups ankle stiffness (from heel-off of the supporting limb to contra-limb touch down) increased with step height, while knee joint stiffness decreased. During the initial phase of the task (initiation of movement until heel-off of the supporting limb), the elderly had significantly less ankle stiffness at all step heights compared to the young group. The differing torque pattern and lower dynamic ankle stiffness suggests that the elderly have an altered control strategy and highlight the importance of dynamic joint stiffness during a stepping down task.

Aging is related to a reduction in walking gait velocity which can in large part be attributed to a shortened step length. The purpose of the study by Judge et al.[84] was to compare the relationship between joint kinetics and step length in 26 older subjects (79 years old) and 32 young subjects (26 years old) and to determine if hip extension or ankle plantarflexion power was primarily responsible for the shorter step length. Additionally, they wanted to determine if quadriceps strength, measured with a dynamometer, and knee kinetics were responsible for the shortened step length in older individuals. The older subjects had a 10% shorter step length compared to the young subjects. In addition, they had decreased ankle plantarflexion and power during late stance. Older subjects were unable to increase ankle plantarflexion power at maximal walking pace, but increased hip flexor power 72%. Therefore, older subjects were unable to generate power with their ankle plantarflexors and therefore compensated by increasing hip flexor power. Appropriate training of ankle plantarflexor musculature may be important in maintaining step length as we age.

As we age the ability to climb stairs becomes increasingly difficult. Therefore the purpose of the study by Vallabhajosula et al[27] investigated the frontal plane joint dynamics on ten healthy subjects during two ascent conditions. The first was to ascend the stairs from a walk and using momentum while initiating stairs compared to initiating stairs ascent from a stand. Repeated analysis of variance tests were used for data analysis. Subjects generated greater peak hip abductor moments to counteract pelvic drop on the contralateral side. Greater peak knee abductor moments were generated when initiating stair ascent from a walk. This is important result and emphasizes the importance of using stair climbing as a testing to evaluate hip strength in individuals with documented frontal plane abnormalities, or individuals suffering from knee osteoarthritis. Those patients suffering from pathologies my not be able to generate sufficient moments to counteract this pelvis drop with may result in mechanically inefficient stair ascent.

The purpose of the study by Keller et al.[8] was to determine the differences in vGRF during walking and running at different velocities from 13 male and 10 female athletes. Analysis of covariance was used for data analysis. Vertical ground reaction force increased with an increase in walking and running velocity. Maximum ground reaction force was linearly correlated to loading rate. However, at speeds greater than 60% of the subjects' maximum speed, the vGRF forces remained constant at 2.5 times body weight. An interesting finding reported was that slow jogging was associated with a greater than 50% higher vGRF and loading rate when compared to walking or fast running due to the higher center of gravity and bouncier running style. Running style therefore appears to be particularly important in the determinacy of vGRF.

For patients with early to moderate osteoarthritis difficulty during stair negotiation is often an early complaint. The purpose of the study by Costigan et al.[38] was to investigate knee kinetics in 35 healthy volunteers using optoelectric motion tracking system while participants negotiated stairs (rise 20 cm, run 30 cm). The contact forces occurred at high degree of knee flexion where there is a smaller joint contact area resulting in high contact stresses. The peak knee abduction moment was 0.42 N m/kg while the flexion moment was 1.16 N m/kg. The knee flexion moments were higher during stair climbing that during level walking. An interesting finding from this research was the patellofemoral contact force was 8 times higher during stair ascent than during walking. Compared to level walking, stair climbing produces greater joint forces and moments. Results of this study can be used as baseline measures in pathological studies.

During physical activity the muscles, tendons and ligaments all work together and act like a spring. The stiffness of the leg spring represents the average stiffness of the overall musculoskeletal system during contact with the ground with a greater stiffness leading to a shorter ground contact time. Additionally, stiffness can be adjusted to allow changes in stride frequency or surface stiffness during hopping. The purpose of the study by Farley et al.[125] was to determine the mechanisms by which humans adjust leg stiffness during hopping in place in five subjects under two, two-legged hopping conditions. The first hopping condition which they hopped to whatever height they preferred ("preferred hopping height") and they second condition in which they hopped as high as possible ("maximum height hopping"). Leg stiffness was twice as great for maximum height hopping when compared to preferred height hopping. Ankle stiffness was 1.9 times great while knee stiffness was 1.7 times greater during maximum height hopping when compared to preferred hopping height. Through the use of a computer simulator and reported that ankle stiffness increased 1.9 fold which caused overall leg stiffness to increase by 2.0 fold. Increasing the knee stiffness by 1.7 fold had no effect on leg stiffness. Therefore, the primary mechanism for leg stiffness adjustments is through increasing the stiffness of the ankle.

As we age, the human neuromuscular system negotiates challenges to ambulation by stiffening the muscle and joints of the lower extremity. Additionally, there is a reduction of joint range of muscle a loss of muscle strength with aging. The purpose of the study by Hortobagyi at al.[78] was to compare the joint stiffness of 14 older women (mean age 70.1) to 16 young women (mean age 20.8) during a step-down task at 10% and 20% of their individual heights. The older women had 50% greater lower extremity stiffness and 28% less linear shortening of the limb when compared to the young group. In the ankle the older group stepped down with 92% less dorsiflexion and an overall 42% less ankle range of motion. In the knee they experienced 28% less knee flexion and 57% less knee joint range of motion. With a more demanding task (20% of height step-down), the older group experienced a reduced adaptability of the aged neuromuscular system with greater motor challenges. Therefore, this group of older women stepped downward with a more erect lower extremity alignment which resulted in a stiffer leg which would allow for a safer movement strategy.

Elderly women step downwards with a substantial greater leg stiffness and may be attributed by muscle coactivity. Therefore, the purpose of the study by Hortobagyi et al.[83] was to investigate whether an increased leg stiffness was associated with muscle pre and coactivity in aging in 11 young (20.8 years old) and 12 elderly (69 years old) participants. Participants stepped down from a platform (20% of their height) and also performed strength trials with electromyography recording. Data were analyzed using two-tailed *t*-tests and linear regression. When compared to the young, the elderly group had a 9% more maximal force applied to the leg, 27% less displacement and 64% greater leg stiffness while stepping downward. Additionally, the elders muscle activity during the loading phase of stepping downwards was 96% greater when compared to young. The biceps femoris and tibialis anterior coactivity during ground contact was 120% larger in the elderly group which increases joint stiffness and stability. The muscle pre and coactivity accounted for 50% of the variance in leg stiffness. To compensate for neuromuscular deficits, elderly increased their muscle pre- and coactivity to stiffen the leg during downward stepping motions.

The purpose of the study by Crowinshield et al.[6] was to determine the effect of walking velocity and age on hip kinematics and kinetics. They concluded that kinematic and kinetic parameters of gait are dependent upon subject age and walking velocity. An increased walking velocity was associated with an increase in hip flexion/extension, stride length, hip resultant force, hip moment and hip contact force. In addition, all variables were affected by the age of the subject, except for the variable hip resultant force. When doing gait research, it is important to have an understanding of the effects of walking velocity on gait biomechanical variables.

Few studies address the hip and knee joint powers and mechanical work during walking in elderly individuals. Therefore, Kirkwood[100] quantified hip and knee; range of motion, joint moments, power and mechanical work during walking gait using 3D analysis on 17 males and 13 females aged 55-75 years old. Data were analyzed using t-tests. The average walking velocity for this age-group was 1.17 m/sec. The total effort generated by the hip joint during gait was greater than in the knee joint. The hip generated 0.40 J/kg with 22% from the frontal plane, 76% from the sagittal plane and only 2% from the transverse plane. Compared to the total effort generated at the knee joint was 0.30 J/kg with 7% contribution from the frontal plane, 90% from the sagittal plane and 3% in the transverse plane. In the sagittal plane, the knee ROM was 60.4° and 4.9° of movement occurred in the frontal plane. Walking at a decreased walking velocity caused a reduced ROM in all joints, especially in the sagittal plane. This study provided insight into the role of each joint during gait with the knee working solely in the sagittal plane.

Conclusion

To assess how biomechanical variables change, 3D gait analysis during walking[84,100,76,123,124,6] and running trials[8], was commonly used. It was additionally used during more functional movements such as; during a step down task[122,78,83], stair negotiation[27,38] and during hopping movements[125]. A major change in biomechanics is a shift of reliance on ankle for propulsion to the hip[84, 76,123,124].

Osteoarthritis and Gait

Introduction

By evaluating gait using motion analysis systems[34,90,91,126-128], researchers can gain valuable insight into the effect of osteoarthritis (OA) on walking gait. Individuals suffering from OA walk with slower velocities[87,88] and have an increased stance time[87,90]. In the frontal plane, varus thrust[92,103,129] and knee adduction moment (KAM) have been correlated to and is higher in OA patients[9,87,90-98], increasing the walking forces medially in the knee. These patients with severe OA, present clinically with greater knee varus alignment[91,97,106,130]. To compensate for increased in KAM, OA sufferers can reduce KAM by walking with a toe-out gait[106,126], decreasing walking velocity[87,91] or with a lateral trunk lean[34]. In the sagittal plane, higher knee flexion moments are associated with higher knee loading[29]. OA patients walk with smaller knee flexion and knee extension moments, or a “quadriceps avoidance gait.” decreasing the overall load and with decreased range of motion[87,88,90,93,94,98] decreasing pain and knee joint loading.

Individuals with medial knee OA can often have frontal plane laxity at the knee. Lewek et al.[97] quantified frontal plane knee joint laxity in patients with medial knee OA and genu varum to determine the effect of both on joint laxity and gait. Participants for the study totaled 12 patients (six females and six males) who were all diagnosed with OA and presented with a genu varum stance. Radiographic changes were observed and measured from the standing postero-anterior radiographs with the knee flexed to 30°. Measurements of joint laxity were made from stress radiographs taken with the subject lying supine on the radiograph table with the knee flexed to 20°. A TELOS stress device was used to apply a 15 daN (33lbs) force in both a varus and valgus direction. All subjects completed a knee out-come survey-activity of daily living scale (KOS-ADLS) and underwent a gait analysis using six camera VICON motion analysis system. Walking electromyography (EMG) data was collected using a 16-channel system interfaced with the VICON simultaneous recording. Participants walked a 10m walkway at a self-selected velocity. All data was analyzed using SPSS, ANOVA and linear regression. The OA group had statistically significant greater laxity in the medial compartment, when compared to the control group. Subjects in the OA group recorded higher KOS-ADLS knee impairment function scores than the control group. After accounting for differences in walking velocity, the peak knee adduction moment was significantly higher in the OA group compared to the control group. When examining the EMG the OA subjects had significantly great VMMG co-contraction when compared to the controls. The OA patients, who presented with medial knee OA and genu varum, did indeed have greater frontal plane laxity and instability than an age-matched control group. Excessive laxity was accompanied by greater medial muscle co-contraction. The medial location of the excessive frontal plane laxity is likely contributing to the cycle of articular cartilage degeneration, joint malalignment and altered joint loading. Therefore, the researchers concluded that the medial laxity in OA patients is likely contributing to the altered gait patters observed and that medial joint laxity should be addressed to slow the progression on OA.

Knee adduction moment may be increased in people with knee OA. The purpose of this study by Baliunas et al.[93] is to evaluate subjects with knee OA of varying radiographic severity to determine if the peak KAM moment during gait is increased. This retrospective study selected 31 subjects (18 females, 13 males) on the basis of definite medial joint space narrowing and no definite lateral joint space narrowing from a pool of 64 subjects who were tested in the gait laboratory in a double blind study which investigated NSAID's on knee joint loading during gait in patients with knee OA. Pain levels were assessed using the pain subsection of the Rush modified Hospital for Special Surgery functional knee evaluation. Thirty-one asymptomatic control subjects underwent the gait protocol as well and were comparable in age, height, weight and gender distribution. Standard anterior-posterior weight bearing x-rays were obtained and the KL grade was determined for each subject. Reflective markers were applied and subjects were asked to walk at self-selected speeds of "slow," "normal," and "fast." Primary analysis consisted of testing for a significant difference in the peak external knee adduction moment between the knee OA and normal groups using Student's *t*-test. Pearson correlation coefficients were used to test for significant correlations between the knee angles and external moments. Subjects with knee OA and radiographic evidence of medial compartment cartilage damage, the peak external knee adduction moment was significantly greater than normal and also tended to walk with a decreased external extension moment. This external extension moment gait pattern has previously been referred to as a "quadriceps avoidance gait" and was observed in subjects with knee OA as well as some normal subjects. Minimum knee angle of the knee OA groups was significantly greater than normal and knee range of motion (ROM) was significantly less than normal. Osteoarthritis subjects walked at a significantly decreased stride length (stride length divided by height) and an increased cadence. In conclusion, the peak external knee adduction moment in higher than normal in subjects with radiographic signs of knee OA with medial compartment cartilage damage who are being managed by conservative medical therapy. This implies that higher medial compartment knee joint loads are present in this population and that an increased knee adduction moment during gait is associated with patients with knee OA.

Increased mechanical loading has been consistently linked with medial OA. Furthermore, evidence from healthy adults indicates that knee flexion kinematics may also influence knee joint loading. The purpose of the study by Creaby et al.[29] was to investigate the association between knee flexion kinematics and indicators of knee joint loading during walking in 89 patients with medial OA. Walking gait kinematics were collected to measure stance phase vertical ground reaction force (vGRF), knee joint moments and knee flexion kinematics. Linear regression was used for data analysis. A greater knee flexion excursion was associated with higher peak vGRF and accounted for 10% of its variance. Additionally, greater peak knee flexion was associated with a higher flexion moment and accounted for 44% of its variance. No association was reported between knee adduction moment and knee flexion kinematics during walking. This data suggests that greater knee flexion is associated with higher joint loading in the sagittal plane. However, knee flexion kinematics were not associated with KAM or increased medial joint loading.

The purpose of the study by Favre et al.[55] is to compare age-related sagittal plane patterns during walking in 81(27 asymptomatic, 28 with moderate medial OA and 26 with severe medial OA) older patients and 29 younger asymptomatic subjects. Data were analyzed using ANOVA tests. During initial contact, the knee was less extended and the shank less inclined in the three older groups compared to the young group. Both OA groups also had the femur less posterior relative to the tibia and small extension moment compared to the young. This study showed that this difference was due to the shank being less tilted with increasing age and disease severity and is consistent with age-related decline in quadriceps strength. Muscle function is closely related to joint kinematics and may lead to an increase in incidence of knee OA with again or disease progression. The severe OA group also had a less extended knee and smaller knee extension moment than the younger and older moderate OA group at push-off. These differences at initial contact and push-off are associated with both age and disease severity and could form a basis for looking at mechanical risk factors for initiation and progression of knee OA.

The purpose of the study by Kuroyanagi et al.[92] was to quantitatively measure varus thrust and the relationship it has with other static and dynamic parameters. The sample size comprised of 44 knees in 32 patients and radiographic severity of OA was at least grade 2 (25 knees), grade 3 (13 knees) and grade 4 (6 knees) according to the Kellengren-Lawrence (KL) scale. Ten knees from 10 healthy elderly subjects were included in the control group. Hospital for Special Surgery (HSS) scores were used as the clinical scores. Femorotibial Angle (FTA) was measured from a standard full-length anteroposterior weight-bearing radiograph. All subjects performed 10 m level trials at a comfortable walking speed. Pearson's coefficient was used to analyze the correlational relationship between FTA, peak knee adduction moment and the amount of varus thrust. Kruskal-Wallis test was conducted to determine difference in knee mechanics between K-L grade groups. Patient's HSS scores and FTA were significantly related to KL grade. The first peaks appeared at 32.2% of the stance phase. Therefore, the amount of varus thrust, and knee adduction moment all increased and were significantly related to K-L grade. The moment increased in severe OA knees and KL grade also exhibited a significant relationship with the moment. The amount of thrust was clearly more closely correlated to knee adduction moment than FTA. The correlation between FTA and knee adduction moment was also significant. Evaluation of the varus thrust, which is a potent risk factor for knee OA progression was correlated to knee adduction moment, coronal limb alignment and x-ray joint degeneration and appears to offer an important index for knee OA disease severity. Therefore, the amount of thrust correlated to static and dynamic parameters and may offer an important clinical index for knee OA.

An association exists between knee alignment and OA progression. However, there are few longitudinal studies examining the effect of alignment and the risk of incident of knee OA. Therefore, Sharma et al.[95] examined whether alignment influences the risk of incident and progression in radiographic knee OA. At baseline a total of 2,958 knees were without osteoarthritis and 1,307 knee had OA. Alignment was from a full-limb radiograph, which also included hip and tibiotalor joints and was assessed at baseline and after 30 months. Varus alignment was defined as $\leq 178°$ and any measurement $\geq 182°$ was valgus alignment. Multiple logistic regression with generalized estimating equations was used for data analysis. Varus but not valgus alignment increased the risk of and incidence of tibiofemoral OA which is not too surprising of a result due to the KAM being greater when varus alignment increases. In knees with reported OA, varus and valgus alignment each increased the risk of OA progression in the biomechanically stressed compartment.

There are many mechanical factors that can be measured with gait analysis. The purpose of this study by Astephen et al.[90] was to describe the biomechanical factors that have been investigated in patients with OA. This study included 60 asymptomatic subjects, 60 with moderate knee OA and 61 with severe knee OA. Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) self-reported pain, function, stiffness and total scores were significantly greater in the severe group then the moderate and greater moderate group. Kellgren-Lawrence (K-L) radiographic scores were significant between the moderate and severe groups. Participants walked at a self-selected pace. Analysis of variance, Bonferroni correction, and Tukey post-hoc test were performed for data analysis. Stance percentage, stride time and stance time all increased in OA patients. All OA differences at the knee joint included early stance flexion moments and higher mid-stance adduction moments. Both OA groups also had reduced peak and first peak hip adduction moments and late stance hip extension moments. Progressive gait changes included two sagittal plane changes at the knee: successively smaller knee flexion angles during stance phase and successively smaller knee extension moments in early stance phase. This study is the first to associate changes in both peak knee flexion angle and peak extension moments just after foot contact with increasing levels of knee OA severity. Results also support the suggestion that mid-stance knee adduction moment is a more important parameter than the peak value for distinguishing between asymptomatic and OA gait patterns, and the mid-stance knee adduction moment is a speed-independent measure, unlike the peak knee flexion moment.

It is not clear whether OA patients walk slower due to increasing in age or due to disease severity to reduce loading in the medial compartment of the knee and is the purpose for this study by Mundermann et al.[91]. Forty-four subjects (24 women and 20 men) were had OA and a control group was matched for size, age, height, weight or BMI. Participants were instructed to walk at 3 speeds: slow, self-selected normal, and fast. Linear regression analysis, Bonferroni correction, and repeated *t*-tests were used to relate the maximum knee adduction moment at the different walking speeds. Differences in maximum KAM between control, less severe OA and more severe OA were determined using repeated-measures ANOVA. The maximum knee adduction moment at self-selected normal walking speed was linearly correlated with self-selected normal walking speed for patients with knee OA when data from patients with all disease severities were combined. Maximum KAM at self-selected normal walking speed was not significantly different between all OA knees and asymptomatic control knees. However, when the data was stratified on the basis of disease severity, the maximum KAM was significantly higher in knees with more-severe OA than in asymptomatic matched control knees and in knees with less severe OA. These participants also had a greater varus alignment of the knee. Changes in the maximum KAM may not be readily predicated from walking speed from all patient and control groups because they are subject to large individual variability. The differences in magnitude and slope of the theoretical relationship between maximum KAM and walking speed for the 3 groups of subjects suggests that patients with less-severe knee OA walk with unique gait mechanics that are different from the gait mechanics of asymptomatic control subjects and patients with more-severe knee OA despite similar age and sex distributions in all 3 subject groups. This unique walking pattern may lead to reduced loading of the medial compartment of the knee when walking at slower speeds. Patients with less-severe knee OA can reduce the maximum knee adduction moment by reducing walking speed. Knees with more-severe OA had significantly greater maximum knee adduction moments than did knees with less-severe and asymptomatic control knees, and were in more varus alignment than were knees with less-severe OA. These results suggest that increased maximum knee adduction moment may not be the initial cause of OA but rather the effect of morphologic changes in the pathologic joint.

Dynamic alignment in association with dynamic loads during gait in patients with medial OA has not been compared to a heathy control. Therefore, the purpose of the study by Foroughi et al.[128] was to determine whether dynamic alignment is altered in medial knee OA and how dynamic alignment is related to KAM in 17 women with medial knee OA and 15 sedentary body mass index-matched controls with no diagnosis or symptoms of OA. Gait analysis was captured while participants walked barefoot at two speeds; self-selected and maximal speeds. Muscle strength was assessed using a one repetition maximum on a resistance machine, static balance and body sway were measured via the force plate, and the WOMAC was used to assess pain, stiffness and physical function. Analysis of co-variance and linear regression were used for data analysis. Shank adduction angle and shank mean angular velocities were reported to have reached its peak around 30% of stance, corresponding with first peak KAM and were higher in the medial knee OA group when compared to controls and were the best predictors of KAM. Knee adduction moments were not different between groups. This research suggests that the greater shank adduction angle puts the medial compartment under more pressure and it therefore supports higher loads and forces during gait.

Individuals with a high KAM during gait are more likely to have medial compartment OA. During gait modifications can be made to reduce KAM while walking, therefore, Shull et al.[127] was to evaluate the effectiveness of a six-week gait retraining program on first peak KAM and self-reported pain in ten subjects with medial knee OA. The participants came to the laboratory eight times for data collections, the first time to establish a baseline, once weekly at the conclusion of gait retraining sessions and one month after the post-training session. Gait retraining was accomplished through real-time sensing and feedback while participants walked on a treadmill. To assess knee pain and function the WOMAC questionnaire was at each data collection as well. Repeated measure, one-way analysis of variance (ANOVA) was used for data analysis. After six weeks of a gait retraining program, peak KAM was reduced by 20%, foot progression angle also decreased by an average of 7° while trunk sway angle did not change. It also resulted in a 29% improvement in WOMAC pain scores and a 32% improvement in WOMAC function scores. The reported reduction in KAM could be used as a non-surgical intervention to slow the progression of OA. This study demonstrated that gait retraining can reduce KAM, pain and improve function in patients with medial OA. These results demonstrated the potential to use a gait retraining program as a non-surgical intervention to improve symptoms and slow OA progression for individuals with medial OA and knee pain.

Knee adduction moment has been proposed to be an indirect measure of medial compartment knee joint load during walking. KAM is calculated using the resultant frontal plane ground reaction force (GRF), and the perpendicular distance from the GRF to the knee joint center. Hunt et al.[94] examined the frontal plane ground reaction forces, frontal plane lever arm, and knee adduction moment in 100 subjects with knee OA. Paired t-tests, two-factor repeated measures analysis of variance and Tukey post hoc analysis were used on data. Peak KAM and peak frontal plane lever arm magnitudes were significantly greater in the affected limbs. Peak GRF magnitudes were significantly less in the affected limbs. This was the first study to report that the lever arm was predominately located medial to the knee joint center of rotation and varied little in magnitude throughout the stance phase of walking. The data suggests that there was a higher association between peak KAM and peak frontal plane lever arm than between peak KAM and peak frontal plane GRF, particularly in knees with OA, suggesting that the frontal plane lever arm assessed during walking is an important variable in the examination of knee OA.

Adverse mechanical loading, more specifically, peak KAM places high forces on the medial aspect of the knee and leads the development and progression of knee OA. However, a direct link between KAM and medial knee compartment loading has never been demonstrated in vivo. Therefore, Zhao et al.[96] used a single subject with an instrumented knee implant to evaluate medial compartment load. Video motion and ground force reaction were collected for five patterns of gait: normal, fast, slow, wide stance and toe-out, due to their influence on KAM. The implant contained four single axis load cells that provided a total axial load between medial and lateral compartments. Data was analyzed using multivariable linear regression. Statistically significant correlations were found between KAM and both medial contact force and the medial total contact force ratio. This strong correlation of KAM with medial compartment contact force could lead clinicians to screen for KAM in patients with previous medial compartment injury and they should also consider patient interventions (strengthening or gait retraining) to lower KAM.

One risk factor for medial knee OA is having a large KAM and can be used as a predictor of radiographic disease severity, rate of disease progression and the development of chronic knee pain. One gait modification by OA patients to reduce KAM is to increase their trunk lean which manipulates the center of mass location and thereby reduced the ground force reaction lever arm at the knee, thus reducing KAM. Simic et al.[34] evaluated the effect of increased lateral trunk lean on knee load in 22 patients with medial knee OA. All participants under-went 3-dimensional gait analysis under four conditions; normal gait, and with a 6°, 9° and 12° lateral trunk lean. Participants were trained to lean their trunk toward the symptomatic leg during ipsilateral stance phase of the gait cycle. Standardized anteroposterior knee radiographs were obtained and disease severity was assessed using the Kellgren/Lawrence grading scale. The WOMAC was used to assess self-reported pain and physical function. Repeated measures ANOVA was performed for data analysis. They reported that by increasing lateral trunk lean toward the symptomatic knee significantly reduced knee load throughout stance in participants with knee OA and the larger the lean the greater reductions in KAM was observed. However, it did not immediately affect symptoms at the knees, hip or back in this sample. Results of this study support that by using the compensatory motion of increasing trunk lean, OA patients reduce KAM and the medial compartment load on the knee.

The purpose of Kaufman et al.[88] was to analyze gait characteristics in patients with knee OA. Participants in this study included 139 (47 males and 92 females) adults who were diagnosed with knee OA. Kinematic and kinetic data was collected using six video cameras (Expertvision-Motion Analysis Corporation, Santa Rosa, CA) and 3D ADTECH motion analysis software system (AMASS). Gait cycle was defined as the time from foot contact to ipsilateral foot contact. For stair ascent/decent the gait cycle was defined for the foot strike beginning on the first stair through foot strike on the third stair. Subjects walked along a 12 m walkway at a self-selected pace. The stairs were a flight of four, 18 cm high with a 25 cm run. SAS Statistically Analysis System and a repeated measures Analysis of Variance were used for data analysis. Knee kinematic patterns for the patients with OA were similar to the normal subjects. Osteoarthritis patients did have a slower walking velocity, had a decreased peak knee moment, as well as a decrease in peak knee extension moment. The varus moment significantly increased in OA patients and in OA subjects the knee was slightly more extended when the peak varus moment occurred. In attempt to reduce pain, subjects with OA reduced their knee extensor moment and knee joint loading. Gait adaption's made by patients suffering from OA provide pain relief from the dynamic joint loading that occurs with walking. This study therefore is an example that objective gait analysis can be used and beneficial to document gait adaptations in patients with OA.

Toe-out gait is a frequent compensation for patients suffering from knee OA. This toe-out gait allows for a more lateral center of pressure and reduces the knee adduction lever arm. Jenkyn et al.[126] examined the mechanism of reducing the adduction moments and frontal plane lever arms in 180 patients with medial OA. The patients under went gait analysis and were examined using two frames of reference. The first frame was attached to the tibia, which reported actual toe-out data, and the second frame was attached to the laboratory to simulate no-to-out gait. Paired t-tests were performed for statistical analysis. The KAM lever arm was shorter in the toe-out condition throughout the gait cycle whereas the flexion and extension lever peak lever arm magnitudes were longer in the toe-out condition. Toe-out gait in patients with medial knee OA causes a decrease in KAM by increasing their flexion moment in early stance phase, shifting the load away from the medial compartment and thereby decreasing pain.

Greater KAM can be decreased in walking gait by increasing toe out angle. In the study by Hurwitz et al.[106] tested weather the peak external KAM during walking in 62 subject with knee OA were correlated with the mechanical axis of the leg, radiographic measures of OA severity, toe out angle or clinical assessments of pain, stiffness or function. Radiographic x-rays were obtained to determine radiographic severity and the mechanical axis was obtained. Pain and function were assessed using the WOMAC, and a gait analysis was performed to determine biomechanical factors. For the gait analysis all subjects walked at three selected speeds; slow, normal and fast walking. Data was analyzed using t-tests, Person correlations and multiple linear regression models. Subjects with varus knees presented with the greater peak external KAM, therefore, the mechanical axis was the single best predictor of peak KAM in this group of individuals with mild to moderate OA. Radiographic measures of OA severity in the medial compartment were also predictive of both peak KAM and the sum of the WOMAC scores. Toe out angle was reported to be predictive of peak KAM only during late stance. An interesting reported finding was that the WOMAC scores negatively correlated with the peak KAM moments which was attributed to the high variability between subjects in how pain or functional abilities are perceived. Having an understanding of which clinical measures of OA are most closely associated with dynamic knee joint loads is important for the understanding of OA progression.

The purpose of this study by Al-Zahrani et al.[87] was to identify and define gait abnormalities and compensatory strategies in patients with OA. Fifty-eight subjects (14 males and 44 females) with severe OA were referred to the study from an orthopedic clinic. Twenty-five age and sex-matched healthy subjects were recruited as a control group. Kinematic and kinetic gait parameters were collected and the systems were fully integrated with elctromyographic (EMG) telemetry. Subjects walked barefoot, the length of the laboratory (15 m long and 4 m wide) at a self-selected speed. The Visual Analog Scale (VAS) was used to determine the presence of knee pain throughout the data collection. Data was analyzed using Statistical Package for Social Sciences for Windows. Between group comparisons were made with Mann-Whitney U test. The researchers observed that patients with OA walked more slowly and had a shorter stride length than healthy control subjects. Subjects with OA also had a shorter stride length and a delayed onset of mid-stance and mid-swing phases of gait cycle. Subjects with OA also had reduced range of motion (ROM) at the hip, knee and ankle joints compared with the control group. Peak moments generated at the knee in stance were higher in the OA group. In 40 out of the 58 patients with OA the rectus femoris was active throughout the stance phase of the gait cycle. By contrast, the activation of this muscle was observed in early to mid-stance in the control subjects. The onset of contraction of the gastrocnemius muscle was delayed in the patient group compared with that in the healthy subjects. No consistent differences between the groups in the pattern of activation of the other muscles were seen. The findings of this study confirm that OA patients walked with shorter stride lengths, reduced walking velocity, and had a longer duration of the stance phase of gait.

Varus thrust is defined as the visualized bowing-out of the knee laterally. The impact of varus thrust on the progression of knee OA has not been previously reported. Therefore, Chang et al.[129] evaluated the presence of varus thrust at baseline and its risk of progression of medial knee OA in 237 individuals' with knee OA. To assess alignment and OA progression, anteroposterior full-limb radiographs were obtained, pain intensity was assessed using the visual analog scale, physical function was assessed by the chair stand performance test and the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) physical function scale. All participants under-went a visual gait analysis to assess varus thrust and 64 participants underwent kinetic and kinematic gait analysis. Data was collected to establish a baseline and again 18 months later. Data was analyzed using Odds Ratio and multiple logistic regression.

They reported that a varus thrust was associated with a 4-fold increase in the likelihood of medial OA progression over the next 18 months due to the added stress to the medial compartment. In the knees that had varus thrust also had a greater knee adduction moment (KAM). Varus thrust is a risk factor for the progression of medial knee OA which relates at least in part to the severity of static varus. Considering knees separately, a varus thrust increased the odds of progression among varus-aligned knees, suggesting that knees with varus thrust are particularly high risk for OA progression.

Increases in knee adduction moment may be a risk factor for the development of knee OA. With walking, forces acting the leg produce KAM at the knee, positioning the knee in a varus alignment. Amin et al.[9] used kinetic and kinematic analysis on 80 participants to establish baseline data and then reevaluated the participants three to four years later for another analysis, at that time seven of the 80 subjects developed knee pain. Participants performed four locomotor tasks barefoot; chair rise, a self-selected 10-m walking trial, stair descent and a standing trial to access balance. Data was analyzed using multivariate analysis of variance. The data from this study suggest that among those whom developed knee pain, higher peak KAM occurred with the standing and stair descent tasks and are associated with the development of future chronic knee pain. The knee loading biomechanical characteristics reported during chair rising and stair descent may influence the development of future knee pain. Participants that developed chronic knee pain had higher KAM, when compared to baseline, with locomotor activities 3-4 years prior, which suggests that KAM may play an important role in the development of knee pain.

In patients with low varus-valgus motion it is assumed they have a more efficient use of of muscle strength during walking. This implies that those patients presented with muscle weakness would lead to more severe functional disability and a higher varus-valgus motion. To evaluate this Van der Esch et al.[131] assessed the relationship between knee varus-valgus motion and functional ability and the impact this motion on the relationship between muscle strength and functional ability on 63 subjects with osteoarthritis (OA). At each visit the subjects completed a questionnaire, muscle strength testing of quadriceps and hamstrings using a isokinetic dynamometer, functional ability was assessed using the Western Ontario and McMaster Universities Osteoarthritis Index, a 100 m walking test and a get-up-and-go test, as well as a three-dimensional gait analysis. Multilevel (linear mixed model) analysis and Pearson correlation coefficients for data analysis. Subjects with muscle weakness had a higher varus-valgus range of motion that was associated with a stronger reduction in functional ability than in patients with low varus-valgus motion. A pronounced varus position and a difference between the left and right knees in varus-valgus position were related with reduced functional ability. The results of this study suggest that subjects with high varus-velocity range of motion is associated with inefficient use of muscle strength in the loading response phase and a reduction of functional ability.

Stability of the knee is defined as the ability of the joint to maintain a position or to control movement under differing external loads. A possible measure of knee stability is the varus-valgus motion during walking. The purpose of this study by Van der Esch et al.[130] is to determine the validity of varus-valgus motion as a measure of knee joint stability in 63 subjects with knee OA. Muscle strength was measured isokinetically, joint proprioception was measured as a detection of joint movement, knee alignment was measured using a goniometer and three-dimensional gait analysis was performed on each subject. During midstance the difference between the peak excursion in varus direction and the peak excursion in the valgus direction defined the varus-valgus range of motion (VV-ROM) and was the measure for joint laxity. Pearson correlation coefficients and regression analysis were used for data analysis. The VV-ROM was found to be not correlated with muscle strength, joint proprioception or with skeletal alignment. Since the midstance VV-ROM was not correlated it suggests that the varus-valgus motion is not a valid measure of joint stability.

The purpose of the study by Landry et al.[98] was to identify biomechanical variables during self-selected and fast walking velocities in 41 patients with knee OA and compare to 43 control subjects. Student *t*-tests, repeated-measures ANOVA were used for data analysis. The faster walking speed resulted in overall increases in stride length and decreased stride time, stance time and stance percentage compared to self-selected in both groups. In both walking conditions, OA patients had larger knee adduction moments. Additionally, these knee adduction moments were higher during stance phase and this magnitude was sustained for a longer portion of the gait cycle. The OA group walked with a reduced flexion moment and a decreased external rotation moment during early stance. Increasing the speed of walking was associated with an increase in magnitude of all joint moments. The increased walking velocity did not increase or bring out biomechanical differences between the OA and control group that did not exist during the self-selected walking velocity.

Conclusion

In addition to motion analysis[34,88,90,91,98,126-128] to gain insight into OA patients gait characteristics, researchers used EMG[9,87], radiographs[92,93,95,97,106,129] and self-reported questionnaire scores[90-92,97,106,127-130]. Osteoarthritis patients adapt walking gait characteristics[87,88,90,91,126,127] to decrease frontal and sagittal plane moments, and overall reported pain. In the frontal plane varus thrust[92,103,129], KAM[9,87,90-98] and varus alignment[91,106,130] present as risk factors for knee OA and could possible lead to progression of knee OA. In the sagittal plane, most observations report a "quad avoidance gait"[87,88,90,93,94].

Osteoarthritis and Stair Negotiation

Introduction

When an individual suffers from osteoarthritis (OA) changes can occur with their ability to negotiate stairs. When compared to healthy controls, 80% of total knee arthroplasty (TKA) patients required assistance to negotiate stairs[40]. Osteoarthritis patients may compensate for their strength deficits by reducing their speed while performing stair tasks[132-134] by increasing contributions from the hip[32,135] or ankle[41]. They may also adopt certain characteristics, which can include; forward trunk lean[32], greater dependence on the uninvolved limb[132] or may attempt avoid knee flexion as much as possible[40,41]. Due to the demands of stair ascent, OA individuals may adopt changes in the way they ascend the stairs[41,134,136,137]. Stair descent is a greater challenge for OA suffers[40] and knee adduction moments were reported to be highest with this task[136]. Increases in loading, almost two times the body weight was reported due to muscular weakness in TKA patients. Although handrail usage changes biomechanical variables in the laboratory and usage should be limited during biomechanical studies, the safety of the individual and balance are both improved with the use of handrails[113].

Review of Literature

With a decrease in quadriceps strength, osteoarthritis patients will adopt patterns of movement during stair ascent to compensate, most likely with an anterior trunk lean, for the loss of quadriceps function. Therefore, Asay et al.[32] determined if 23 patients with knee OA of varying severity adopt an altered pattern of movement to reduce the forces acting on the quadriceps by leaning their trunk forward during stair ascent and compared them to a control group. Each subject underwent gait analysis by performing three stair-ascending trials at a self-selected speed. Data were analyzed using analysis of variance (ANOVA), correlations were evaluated using linear regression and Student's t-test. Patients with more severe OA had greater peak trunk flexion angles, as well as lower peak knee flexion moments and greater peak hip flexion moments when compared to control subjects. In conclusion, patients with more severe OA adopt an anterior trunk lean during stair ascent to reduce the demand on the quadriceps by reducing the net quadriceps moment. They were also able to conclude that this stair ascent adoption comes at later stages in the disease progression and could be a useful objective functional marker of the disease.

Compensatory strategies for TKA patients with weak quadriceps and/or hamstring strength post-surgery on level ground includes walking at a slower velocity as well as presenting with an anterior trunk lean Compensatory strategies are well understood for level ground walking, therefore Bjerke et al.[133] evaluated electromyography (EMG) of the vastus lateralis and semitendinousus activity in 23 unilateral, 19 month post-TKA subjects while ascending stairs. Participants climbed the stairs in a step-over-step pattern at a self-selected speed for a total of six trials; three trials were collected on each limb. Isokinetic measurements of the quadriceps and hamstring strength were collected with an isokinetic dynamometer prior to the stair ascent, the EMG electrodes were placed on the previously mentioned muscles and whole body kinematics were collected with an eight-camera system. For data analysis ANOVA was performed. In this study, the EMG activity was found to be positively correlated with gait velocity. The TKA group had significantly decreased quadriceps and hamstring strength when compared to the control group, however, kinematic analysis did not reveal an increased forward lean of the trunk in the TKA group. Therefore, they concluded that in order to compensate for the muscular weakness in the lower extremities the TKA subjects in the current study reduced gait velocity and did not managed the muscular weakness with a forward trunk lean.

Descending stairs requires more knee flexion than ascending stairs and there is evidence of reduced knee flexion in stair ascent post-TKA the underlying mechanism for this reduction is not clear. The purpose of the study by Bjerke et al.[135] was to investigate peak knee flexion during stair descent (PKSD) in TKA-side compared to the contralateral side, and compared to age matched controls without knee problems. The stair descent protocol followed the same process as described in their previous article[133]. Following the stair descent peak passive knee flexion (PPKF) was measured and a joint position sense test was performed. All whole body kinematics were recorded at 100 Hz with eight-camera system. In addition, leg length, current pain, anterior knee laxity, fear of movement, and current pain were also assessed. For data analysis, paired t-tests and one-way ANOVA's were run. Pearson's Correlation Coefficient were run as well. Only the PPKF, although sufficient to allow uncompensated stair descent, explained the smaller PKSD in the TKA-group. Reduced quadriceps peak torque may contribute to PKSD. A combination of reduced PPKF and quadriceps peak torque may affect the length-tension relationship in the muscle which may explain why TKA subjects did not use the sufficient PPKF to descend stairs without compensation. Increased hip adduction in the TKA-group indicated a compensation for reduced PKSD or reduced hip abductor strength or quite possibly both.

There is a need to examine how TKA affects the knee varus angle and moment under conditions of dynamic loading. Therefore, the purpose of this study by Mandeville et al.[134] was to determine the effect of TKA on knee varus angle and moments during level walking and stair ascent compared on 21 TKA patients. Level walking and stair ascent kinematic and kinetics data were collected on all participants within two weeks of surgery and post-TKA at six months and WOMAC/VAS questionnaires were also completed at that time. For data analysis purposes two-way mixed ANOVAs, Bonferroni correction and Pearson's correlation coefficients were used. In level walking they reported a decrease in frontal plane knee moment in participants receiving TKA, so much so that the values between TKA group and control group were similar. Frontal knee angle was significantly related to the frontal plane knee moment at both time periods. During stair ascent no significant difference in frontal plane knee angle were found at either data collection. As a result of the TKA, the mean patient frontal knee angle and moments were significantly altered to approach the reported control values. With stair ascent the was a significant reduction in varus frontal knee moments from pre to post-surgery suggesting that the subjects accommodate to the demands of stair ascent as well as they do to level walking by reducing medical compartment loading due to the realignment of the knee joint. In terms of WOMAC/VAS scores, they were not found to be significantly correlated to frontal plane knee angles or moments suggesting that there is not a correlation between the patients' perception of pain and dysfunction differ from their objective knee function.

Most biomechanical studies evaluating stair negotiation in older adults do not use handrails, but it is known that to improve the safety of older people while on stairs, they use handrails. Therefore, Reeve et al.[113] examined the influence of light handrail use on the kinetics and kinematics of stair negotiation in 13 older adults whose average age was 74.9 years old. Patients were asked to negotiate four stairs at their self-selected speed and in a step-over manner. In one condition participants were asked to negotiate the stairs unaided; in the second condition, they were asked to negotiate the stairs with use of the handrails, as a guide only, and not to perform a large proportion of the work with their arms. They were asked not to pull during ascents, or to accept their body mass during descent. Student's t-test were used for data analysis. During stair ascent, there were no significant differences in the vertical or anterior-posterior ground reaction forces, in the peak hip, knee or ankle extension and flexion angles or in the center of mass-center of pressure separation in sagittal or frontal planes between the two conditions. During stair ascent light handrail use caused a redistribution of joint moments between the knee and the ankle and an altered strategy causing lower ankle joint moments in the trailing leg and higher knee joint moments in the lead leg. This redistribution can be considered a safe strategy for stair ascent. During stair descent there were no difference in anterior-posterior or medial-lateral GRFs. The second knee joint moment peak and the peak hip flexion angle were both lower when using the handrails compared to the unaided stair descent. The ankle joint moment increased with handrail usage and was associated with more effective control of balance as shown by a reduced COM-COP separation during stair descent. These results indicate that although the biomechanical mechanisms are different for stair ascent and descent, the safety of stair negotiation and balance is improved in older adults with light use of the handrails.

It has been suggested that individuals with knee OA increase their step-width (SW) while negotiating stairs to reduced peak knee abduction moments, thereby reduce pain. The purpose of this study by Paquette et al.[138] is to examine the effects of increased step width on knee biomechanics in 13 patients with medial knee OA during stair descent. At the time of data collection, participants completed a Knee injury and Osteoarthritis Outcome Score (KOOS) and underwent gait analysis while they descended five stairs under three conditions, preferred SW, wide SW (26% of SW) and wider SW (39% of SW). Repeated measures ANOVA and Mauchly's Test of Sphericity was used for data analysis. The first peak knee adduction angle and the first and second peak knee abduction moments were not changed with increased SW. Their results show that both the moment arm and the frontal plane GRF vector at the time of first peak abduction moment were unchanged between SW conditions. Knee pain also remained unchanged between SW conditions, but could be attributed the fact that the pain levels for these participants fell in the lower range of visual analog pain score values during walking and stair negotiation. In summary, these findings indicate that increased SW during stair descent did not reduce peak internal knee abduction moments or knee pain in patients with medial compartment OA.

In walking gait, there appears to be a relationship between foot progression angle (FPA) and first KAM moment, but the results are confounding. Therefore, the purpose of the study by Guo et al.[136] examined the effect of increasing FPA on KAM during walking and stair negotiation in ten participants with pain free, mild to moderate medial compartment knee OA. Participants performed five walking trials, stair assent and stair descent at self-selected speeds and with shoes on. During stair negotiation data was collected under two FPA conditions, self-selected FPA and with an additional 15° of toe-out relative to their self-selected FPA. Dependent t-test and one-tailed tests were used for data analysis. Knee adduction moments for the first and second peaks and for both FPA conditions were largest for stair descent, followed by stair ascent and then walking. Increasing FPA by 15° resulted in a 1% increase in magnitude of walking first peak KAM, an increase of 11% in stair ascent and a 2% decrease was reported during stair descent. Increasing the FPA had no effect on the magnitude of the first peak KAM but it did significantly decrease the second peak KAM during walking. Walking with the greater FPA brings the ground reaction force vector closer to the knee joint center during the second half of the stance phase, therefore, reducing the second peak KAM. For stair ascent the first peak KAM was significantly greater for the increased FPA condition and this same condition significantly reduced second peak KAM. There were no differences noted during stair descent. These results suggest that walking with a toe out strategy may benefit persons with early stages of medial knee OA.

The external flexion angle, which is correlated with the quadriceps ability to avoid the collapsing of the body, is 2.3 times greater during stair ascent then when compared to level walking. The purpose of the study by Pozzi et al.[132] is to compare kinematic and kinetic variables in 20 patients six months after TKA and compare them to controls during a step up and over task as well as to evaluate the relationship between quadriceps strength in the operated limb during step up and over task as well as stair ascent. Participants completed the Knee Outcome Survey-Activity of Daily Living (KOS-ADL) scale, the Timed up and Go (TUG) test, Stair Climbing Test (SCT) and Six Minute Walk (6MW) test, performed quadriceps maximal voluntary isometric using an electromechanical dynamometer and underwent gait analysis during which the step up and over task was completed. Analysis of variance, bivariate correlations were performed for data analysis. Individuals in the TKA group took 33% longer to complete the TUG and 41% longer to complete the SCT when compared to the controls. Quadriceps strength was 18% higher in the non-operated limb when compared to the operated limb. The TKA group had lower peak moments, power, and sagittal plane excursion in the operated knee when compared to the contralateral limb while the hip on the operated side had greater power generation. Compared to the control group, all symmetry ratios were significantly lower in the surgical group. Stair climbing time was correlated with quadriceps strength of the operated limb. Patients six months after TKA completed the step up and over task with biomechanical asymmetries that reduce the demand on the operated knee and increase reliance on the contralateral limb and ipsilateral hip. When compared to controls, TKA patients had abnormal movement patterns during the step up and over task, considerable impairments in the quadriceps strength, worse performance-based test results and lower self-reported questionnaire scores.

Conclusion

By using motion analysis[32,40,41,113,134,135,138], electromyography[133] as well as other functional assessment tests[132], we can gain a greater understanding of the increased demands negotiating stairs places on an individual with OA. These patients make adaptations to both stair ascent[40,113,134,136] and stair descent[40,113,136,137] to accomplish this task. Although stair descent is more difficult for TKA patients[113], increasing the strength of the lower leg musculature in this population may provide improvements in their ability to negotiate stairs[132-135]. Handrail usage is not advised when conducting biomechanical studies due to its effect on variables, it can however, promote safety of stair negotiation and balance improvements in this population[113].

Total Knee Arthroplasty versus Unicompartmental Knee Arthroplasty

Introduction

Total knee arthroplasty (TKA) is gold standard for end-stage knee osteoarthritis (OA). However, for individuals with OA present only in the medial knee compartment, unicompartmental knee arthroplasty (UKA) can be a viable surgical option. In studies comparing TKA and UKA, patient records were used to identify short-term risk factors[139] and post-operative long-term pain and function was assessed using questionnaires[11,140]. The UKA preserves knee anatomy so increases in range of motion favors this procedure[11,139], and these patients return to function faster when compared to TKA patients[11]. A downside of UKA is the risk of having a higher revision rate when compared to TKA[140]. Therefore, it should be left to the discretion of the physician to determine if UKA or TKA is the better option for patients with isolated medial knee OA.

Review of Literature

The purpose of the study by Duchman et al.[139] was to use the American College of Surgeons National Surgical Quality Improvement Program database to identify the incidence of and risk factor for patients undergoing TKA or UKA, as well as to determine complication rates in 29,333 patients (TKA group=27,745, UKA group=1,588). Data were analyzed using multivariate logistic regression, Student two-tailed *t*-test, chi-square analysis, Pearson correlation, and McNemar analysis. There were no reported differences in the overall short-term, thirty-day, complication rate between the TKA and UKA patients. However, individual outcome measures, including the rate of deep venous thrombosis, operative time and duration of hospital stay were greater in TKA group. With that being said, morbidity remained low, with no mortality in this cohort. They did point out however, that these differences are likely inconsequential and were only significant due to the large sample size in this study. These findings suggest that there is no difference in thirty-day morbidity and mortality between TKA and UKA surgical procedures.

Revision rates are higher for UKA than that of TKA. The purpose of the study by Lygre et al.[140] was to compare the two-year post-operative pain levels, using the Visual Analog Scale, and function, using the Knee Injury and Osteoarthritis Outcome Index (KOOS) score on the EuroQol-5D health-related quality-of-life instrument, among TKA (n=972) and UKA (n=372) patients. Independent-samples Student *t*-test, Pearson chi-squared test, multiple linear regression and Bonferroni correction method were used for data analysis. They reported differences in favor of the UKA implants for the KOOS subscales of “symptoms”, “function of daily living” and “function in sport and recreation.” In the patients undergoing UKA, men scored better than women in regards to “pain”, “activities of daily living” and “function in sport and recreation.” They observed only small or no differences in pain and function between UKA and TKA at least two years following surgical intervention. The main advantage that favors UKA over TKA, is the preserved knee anatomy that allows for better range of motion in activities that involve bending of the knee for these patients.

With the introduction of a mobile bearing UKA, there is limited data evaluating individuals implanted with UKA compared to those undergoing TKA and the length of the recovery process. Therefore, Lombardi et al.[11] compared Knee Society (KS) clinical outcome scores, Lower Extremity Activity Scale scores, Oxford knee scores, range of motion and return to work or sport in 115 knees undergoing UKA and 115 knees undergoing TKA. All patients underwent the same post-operative recovery protocol. Data were analyzed using two-tailed Student *t* tests and Pearson's chi-square tests. Following surgery UKA patients had shorter hospital stays, greater knee ROM and could walk a greater distance prior to being discharged from the hospital. At eight-weeks post-surgery, UKA ROM remained better in UKA patients and lower extremity activity scale scores favored the UKA group. An important finding reported was that the UKA patients returned to work and/or sports at an average of eight weeks, whereas TKA patients didn't return to work/sport until 11 weeks after surgery although not statistically significant. Seven TKA patients underwent manipulation to regain motion and no UKA patients underwent this procedure. Clinical and functional KS scores and Oxford scores were reported to be similar in both groups. This data favors the minimally invasive UKA allows for patients to gain ROM and function faster than those undergoing TKA.

Unicompartmental knee arthroplasty has several advantages over TKA including a minimally invasive procedure. The purpose of the study by Kim et al.[141] was to compare the patient reported outcome scores using the WOMAC, Forgotten Joint Score (FJS), High Flexion Knee Score (HFKS) and the patients' satisfaction two-years after surgery in 100 UKA and 100 TKA patients. Data were analyzed using the chi-square test and the Wilcoxon signed-rank test. There were no differences in WOMAC scores between UKA and TKA patients. The UKA group scored higher on the FJS and the HFKS questionnaires. Eighty-six percent of the UKA patients were satisfied compared to only 71% of the TKA patients were satisfied with their operation. With higher FJS and HKFS scores, UKA patients facilitated less knee awareness during activities and better function. Therefore, overall patient satisfaction favored the UKA patients.

Conclusion

Despite very different surgical procedures, UKA and TKA patients experience similar post-operative surgical risks[139] pain and function[140] two-years post-operatively. Immediate recovery prior to discharge from the hospital, favors UKA[11] which can be attributed to the less invasive surgical procedure. Range of motion improvements and self-reported functional scores also favor the UKA procedure[11]. Overall patient satisfaction favored the UKA procedure[141]. When determining which implant to use for the diseased joint, both TKA and UKA can be viable options for these patients.

Unicompartmental Knee Arthroplasty (Oxford) Implant Design

Introduction

Mobile-bearing and fixed-bearing implants are used for unicompartmental knee arthroplasty (UKA) in patients with medial knee osteoarthritis (OA). In studies comparing these implant designs both implants have been reported to provide excellent pain relief and improvements in patient function[142,143]. No differences have been reported in patient outcome scores between the two UKA designs[142-144]. The mobile-bearing implant has been reported to more mimicked the normal knee[142]. Patients undergoing UKA with an absent anterior cruciate ligament, have altered sagittal plane knee kinematics[144]. Regardless of implant design, progression of the OA into the lateral compartment was the main reason for undergoing a revision surgery[143]. And in ALC deficient knees, loosening of the tibial component was large reason for implant failure[144].

Review of Literature

Unicompartmental knee arthroplasty implants have improved significantly due to changes in implant designs and material used over the years. However, whether a fix or mobile meniscal bearing UKA should be used remains controversial. Therefore, Li et al.[142] evaluated knee kinematics, clinical outcome scores using Knee Society scores (KSS), Western and McMaster Osteoarthritis index (WOMAC) and Short-from 36 scores (SF-36), and radiographic findings, between fixed (Miller/Galante implant) and mobile bearing (Oxford implant) UKAs in 48 patients. Patients undergoing UKA were randomly assigned either a fixed or mobile bearing implant and were evaluated pre-operatively and again at two-years post-operatively. The Mann-Whitney U-test or Chi-square test, analysis of variance was performed for data analysis. No differences in clinical scores were determined between the two implants. The mobile-bearing UKA demonstrated knee kinematics similar to that of the normal knee by displaying larger and a more consistent tibial internal rotation movement, a more stationary medial femoral condyle and a lateral femoral condyle roll back during knee flexion. They also reported a higher incidence of radiolucent lines at the implant interface in the fixed bearing UKA which may imply that the fixation quality may be compromised in this implant group. The mobile bearing UKA demonstrated more normal knee kinematics and a lower incidence of radiolucency was reported, however, it did not translate into any improved clinical outcomes at two years-post-operatively.

Unicompartmental knee arthroplasty implants are available with fixed- and mobile-bearing designs, with no advantages in using one design over another. Therefore, Whittaker et al.[143] examined in 179 patients, whether an Oxford mobile-bearing (n=79) or a Miller-Galante fixed-bearing (n=150) UKA design differed in clinical outcome, using both KSS and WOMAC scores, survivorship, revisions and timings of failures via chart review. Data were analyzed using Students *t*-test, Mann-Whitney U test, Kolmogorov-Smirnov Z test, Kaplan-Meier analysis, Chi square test and the Breslow statistical test. They reported no differences in clinical outcome scores between the two implant groups, both designs provided excellent pain relief and improved patient function. The five-year survival rates were 96% for the fixed-bearing and 89% for the mobile bearing design. Progression of the arthritis and aseptic loosening were the dominant reasons for revisions in both groups. The mobile-bearing underwent revisions around 2.6 year, post-UKA and the fixed-bearing underwent revision at a mean of 6.9 years. No differences were noted in the indications or complexity of revision surgery, or in the midterm survivorship between the two groups. They concluded that both implant designs were successful in relieving pain and restoring function and had similar reasons for revision and no differences between implant designs were identified in this study.

Patients undergoing medial compartment UKA in knees with a deficient anterior cruciate ligament (ACL), have reported higher rates of implant failure. However, there is some evidence that UKA can be successful in ACL deficient patients. Therefore, the purpose of the study by Pegg et al.[144] was to determine there was a difference in sagittal plane kinematics, using fluoroscopy and evaluating the patellar tendon angle (PTA), in 16 people undergoing UKA without an ACL and in 16 individuals undergoing UKA with an intact ACL. Fluoroscopy was recorded during step-up and forward lunge activities, which were chosen because of the strain exerted on the ACL and the high flexion moment within the knee. Data were analyzed using non-parametric Wilcoxon signed-rank test, Mann-Whitney U tests and intra-observer correlation coefficient tests. The ACL deficient group was reported to take longer to perform both movement tasks. Reductions in the PTA in the ACL deficient group were observed between 40°-60° of flexion during step-up activity and may be attributed to muscle imbalance and/or a loss of proprioception. During the forward lung activity, the PTA was also reduced at 100° and

110° however, the contribution of the ACL at that extreme range of motion would be small and at that flexion angle responsibility of the anterior translation of the tibia heavily relies on the hamstring musculature. When compared to total knee arthroplasty, overall sagittal plane kinematics in UKA ALC deficient knees more resembled healthy knees. However, were not as similar to UKA patients with intact ACLs'. Based on these results, more long-term outcome data is required before UKA can be recommended for ACL deficient patients.

Conclusion

Evaluation of clinical outcome scores reported from KSS[142,143], WOMAC[142,143], Oxford Knee Scores[144]and the SF-36[142], no differences were reported between UKA implant designs[142,143] or with or without the presence of the ACL[144]. In terms of knee kinematics, the mobile bearing implant functioned was similar to the normal knee[142] and the presence of an ACL is important[144]. Researchers could not draw a conclusion as to which implant design was better, they both provided excellent pain relief and improved patient function[142,143].

Unicompartmental Knee Arthroplasty Oxford Implant Survivorship

Introduction

An important concern for patients undergoing unicompartmental knee arthroplasty (UKA) has to do with the Oxford knee implant longevity. In studies evaluating UKA revisions, very low revision rates have been reported[19,145-147]. The average years post-UKA when a revision procedure is performed is 2.8 years[145]. The most common reason for the revision was due to progression of the osteoarthritis (OA) into the lateral compartment, which is unrelated to UKA surgical procedures or implant design, with those patients undergoing a total knee arthroplasty[146,147]. Another reason for revision was due to the dislocation of the mobile bearing[145], however thickness of the polyethylene bearing had no impact on survival of the implant[19]. Emerson et al.[146], reported no revision due to tibial component failure. Mid-and long-term survival of the Oxford UKA implant design was concluded to be excellent[14,19,145,146].

Review of Literature

Improvements unicompartmental implant design, surgical equipment and the introduction of minimally invasive surgical techniques has led to a rise of UKA. A prospective study by Lisowski et al.[145] evaluated the functional and radiological outcomes of 244 UKA patients receiving the Oxford Phase 3 UKA. Pain, function and health related quality of life were evaluated pre- and post-operatively using the Western Ontario and McMaster Osteoarthritis (WOMAC) Questionnaire, the Knee Society Score (KSS), Oxford Knee Score and the visual analog scale. Post-operative implant alignment, and progression of OA was determined using standing anteroposterior and lateral radiographs. Data were analyzed using non-parametric Wilcoxon test, general linear modeling, post hoc Bonferroni test and the Kaplan-Meier survival analysis. The medium-term survival results of the Oxford Phase 3 UKA at a seven-year follow-up were reported as 94.4%. Revisions for all causes (progression of OA, dislocation, aseptic loosening and instability) occurred at a mean follow-up of 2.8 years. The most frequent cause for revision in this study, at 1.2%, was early dislocation of the meniscal bearing which was attributed to a technical error during surgery. An interesting finding reported was the outcome scores after one year post-operatively do not significantly change so a follow-up of one year is justified. Major complication rate of the Oxford Phase 3 UKA was low and this study showed high survival rate with this implant.

Literature regarding the survivorship of UKA have mixed reviews. The purpose of the study by Emerson et al.[146] was to determine post-operative limb alignment, survivorship and identify the modes of failure in 55 knees implemented with the completely unrestrained polyethylene mobile bearing Oxford partial implant. Knee Society scores were used to assess function. Radiographs were used to assess for osteolysis, progressive joint degeneration and knee alignment. Student t-tests, Pearson coefficient and Kaplan-Meier tests were used for data analysis. The overall alignment of the knee was restored to neutral and was reported to average 5.6° of valgus. Seven of the knees underwent revision surgery, six of those seven ended up receiving a total knee replacement due to the natural progression of arthritis in the lateral compart which was not related to the initial post-operative alignment. None of the patients in the study experienced tibial component failure due to polyethylene wear or osteolysis. The rate of survival at ten years post-operatively was reported to be 85% with failure for any reason, 90% with progression of lateral compartment arthritis and 96.3% with component loosening as an end point.

The Oxford UKA uses metal components, for the bone ends and contains a fully congruent unconstrained mobile polyethylene bearing that ranges in thickness from 3.5-11.5 mm. The purpose of the study by Price et al.[148] was to evaluate the thickness of the polyethylene bearing and the 15-year survival of three different Oxford UKA implant in 439 knees. One-hundred and fourteen knees were reviewed clinically at a minimum of 10-years post-operatively and used the Hospital for Special Surgery knee score (HSS). The Oxford Knee Phase I, the

Oxford Knee Phase II, and finally the Oxford knee Phase III were used for the purpose of this study. The core features of the implant remained essentially unchanged, however, the non-articulating surfaces of the femoral implant between the Phase I and II implant, and in the Phase III implant four femoral implant sizes were introduced. An analysis of variance test was used for data analysis. At 10-years post-operatively 91% of the 114 knees had good or excellent results, and 82% reported being pain free. The 10-year survival rate of the polyethylene bearings less than 6 mm thick was 95%, compared to 94% with bearings greater than 6 mm. The results of this study demonstrate the excellent long-term clinical and survival rates of patients implanted with the Oxford knee implant. The thickness of the polyethylene bearing did not impact the survival of the implant.

In the younger population with knee osteoarthritis, an on-going debate exists regarding the best surgical treatment intervention for this patient group. Therefore, Price et al.[19] compared the clinical outcome, using HSS, in patients undergoing medial UKA who were <60 years old (n=44) at the time of operation, to the results of those patients $\geq$60 years old (n=403). Survival analysis was performed and survival rates were calculated from a life-table and a statistical comparison was performed using a log rank test. The ten-year results in the <60 age group was 91%, suggesting that this implant functions well and is durable when used in younger patients. Not surprisingly, the HSS scores of the younger UKA group were seven points greater which is attributed to an increased physical activity post-operatively in this age-group. The results of this study suggest that the Oxford UKA can benefit from the reduced morbidity and improved function of the knee after UKA and that age is not a contraindication for using this implant to treat patients with medial knee OA.

Berger et al.[14] did a ten-year follow-up on 49 knees in patients undergoing UKA. Thirty-nine knee had excellent result (80%) and six (12%) had good result at this minimum of 10-year follow-up. Two patients underwent revision to TKA due to the progression of OA into the patellofemoral joint. Thirty-nine knees (80%) were able to obtained 120° of knee flexion. A Kaplan-Meier analysis revealed a 10-year survival rate of 98%, a 13-year survival rate of 95.7% with revision or radiographic loosening at the end point. Tibial aseptic loosening and accelerated polyethylene wear are two of the most common reason of failure in UKA, however these complications were not seen in the present study. Although UKA was associated with excellent clinical and radiographic results in this 10-year follow-up, it is encouraged that the implant be used only in properly selected patients.

The purpose of the study by Price et al.[147] was to investigate the twenty-year survival rate of the Oxford mobile bearing medial UKA in 682 patients. Participants were monitored for complications and surgical revisions and were withdrawn from the analysis due to death. A survival analysis was performed for data analysis. In total, only 29 (4%) revision procedures were performed on this cohort, and of these participants, total knee arthroplasty was the surgical intervention in 93% of the cases due to the progression of osteoarthritis to the lateral compartment which may be due to the overcorrection of the varus deformity during surgery. However, with the mean age of 70 in this cohort, they report that the device need not be considered a pre-TKA procedure. In terms of revisions, if the patient remains unrevised at 10 years, then survival of the implant to 20 years is to be expected according to this analysis. They reported the 10- and 20-year survival rates to be 94% and 91% respectively, meaning the Oxford mobile bearing UKA can have a low revision rate through the second decade of life after implantation.

Conclusion

Through retrospective chart review to evaluate revision rates and explanations for revisions, various questionnaires are used to evaluate implant survival including: KSS[145,146], WOMAC[145], VAS[145], Oxford Knee Score[145] and the HSS[14,19]. Radiographs were also used to assess knee alignment and joint degeneration [145,146]. The Oxford UKA implant's survival rate is excellent with seven-year rates of 94.4%[145], ten-year rates of 85-95% [146,14,148] and 20 year long-term rates were reported 91%[147].

Unicompartmental Knee Arthroplasty and Function

Introduction

Post-surgical functional assessment is a very important surgical outcome following unicompartmental knee arthroplasty (UKA). Prior to UKA surgery, the patients' physical function and mobility is low. Using Knee Society Scores, Lim et al.[15] reported no difference in pain or total Knee Society Scores (KSS) between UKA and total knee arthroplasty (TKA) patients. However, after surgery, improvements in kneeling[149], management of stairs[149] and physical activity involvement [18] is observed and is proof that a greater function is achieved postoperatively in UKA patients. Most UKA patients participate in hiking, cycling and swimming activities after surgery, an example that Oxford UKA patients can continue to be active, or become physically active after UKA[18].

Review of Literature

Patient-based performance outcome scores typically favor the UKA procedure over TKA which is attributed to the ability to bend the knee and the preservation of the cruciate knee ligaments. However, there is little difference in reported pain and function between the two groups, with revision rates higher in UKA patient. The purpose of the retrospective study by Lim et al.[15] was to compared the medium-term outcomes, using KSS which assessed pain and function, of 608 UKA patients and in 608 TKA age and gender matched patients. The Komogrorov-Smirnov test and Mann-Whitney U test were used for data analysis. They reported no differences in pain or total KSS for both UKA and TKA groups. The function scores of the UKA group were better than the TKA group and the TKA groups reported better pain control scores although neither of these values were statistically significant. Higher rates of medical complications were noted in the TKA group, which may have implications for physicians when deciding on which procedure to perform on frail, elderly individuals. The UKA group experienced higher revision rates at 6.3% when compared to the revision rate of TKA at 2.99%, aseptic loosening was the main cause for the UKA revision procedure. Despite differing surgical procedures, differences in pain and function were not reported, but this study suggests that UKA is associated with fewer post-operative complications, however, TKA provide better initial pain relief and is less likely to require a revision.

Post-operative ability to walk down stairs and to knee are two important functional demands of those individuals under-going knee arthroplasty. The purpose of the study by Hassaballa et al.[149] was to assess the stair descent ability versus kneeling ability in 231 patients following a TKA, UKA or a patellofemoral replacement (PFR). The TKA group had 113 knees, the UKA had 70 and the PFR consisted of 58 knees. Data were analyzed using Kruskal Wallis, Bonferroni correction and ordinal logistical regression. Prior to surgery only 3% of patients could kneel and 20% could manage stair ascent with ease, and no difference among the different types of arthroplasty was reported. Post-operatively, 41% of TKA patients could kneel while 53% of the UKA group could. In terms of managing the stairs with little to no difficulty, 20% could manage stairs with ease, post-operatively that number increased to 75%. One-year post-operatively, the UKA group performed better than both the TKA and PFR groups during both stair climbing and kneeling activities. Although the patients' ability to kneel and ascent stairs after surgery improved, disparities among the two activities remain suggesting that there could be other factors that affect kneeling ability other than preoperative arthritic pain, range of motion, or patellofemoral joint involvement.

Unicompartmental knee arthroplasty is minimally invasive and preserves the cruciate ligaments in the knee allowing patients to recover faster and return to sport. The purpose of the study by Jahnke et al.[18] was to investigate, using the Heidelberg Sports Activity Score (HAS), Joint Discomfort questionnaire, Oxford 12-score, Tegner, UCLA and the change of sports activities before and after medial UKA in 135 patients. Wilcoxon signed-rank test and Spearman's correlation coefficient were used for data analysis. In this study 74% of the patients practiced sports one a week prior to UKA, post-operatively that number increased to 84%. They identified that hiking, cycling and swimming as the three most practiced sports in this population post-UKA, all are low-impact sports, which is recommended after UKA. Similarly, the most impaired sports due to UKA were jogging, tennis and ball games, which would all be considered high impact sports. Sporting activities were lower in women when compared to men both pre- and post-UKA. And, in terms of age, younger patient (<65) were sportier than those patients above age 65. After surgery, the increase in the HAS significantly increased in the older age group, whereas that HAS of younger patients only increased slightly. This study demonstrated that Oxford UKA patients continue to be active, or become more active after UKA.

Conclusion

Although undergoing a knee joint arthroplasty, UKA patients can improve function and physical activity after surgery[18,149]. Activity scores and questionnaires are commonly used to assess UKA function prior to and after surgery[15,18,149]. When compared to other knee arthroplasty patients, (TKA and patellofemoral) UKA patients perform better in both kneeling and stair negotiation activities[149]. Proof, that Oxford UKA patients can continue to live physically active lives after their procedure[18].

Unicompartmental Knee Arthroplasty and Biomechanics

Introduction

To investigate kinematic variables in unicompartmental knee arthroplasty (UKA) patients, cadaveric studies[150] and studies using fluoroscopy[10,20,151,152] are most commonly used to evaluate the patellar tendon angle[20,151] which is used for a biomechanical analysis of patient function. Following UKA, sagittal plane kinematics return to normal[10,20,150,151]. The importance of anterior cruciate ligament retention during the UKA procedures has been attributed to normal knee kinematic function[150] and it has been suggested to contribute to UKA longevity[10].

Review of Literature

Unicompartmental knee replacement can be a good alternative for young or middle aged individuals with medial knee osteoarthritis. The purpose of the study by Patil et al.[150] was to evaluate the knee kinematics of a unicompartmental knee replacement during stimulated stair-climbing in six fresh frozen cadaver models. The cadaveric knees were mounted in a dynamic, quadriceps driven, closed kinetic chain knee simulator based on the Oxford knee rig design. Electromagnetic tracking sensors were attached to the knees and measure three-dimensional motion during simulated stair-climbing. Each knee was tested under multiple conditions: first baseline knee kinematics were recorded with an intact joint capsule, in the second condition the bicruciate-retaining unicompartmental implant was implemented, and in the final condition the capsule of the knee was incised and a routine posterior cruciate-retaining tricompartmental replacement procedure took place. A repeated measures multifactorial analysis of variance and Bonferroni correction was used to adjust for the three post hoc pairwise comparisons were used for data analysis. Knee kinematics of the fixed-bearing unicompartmental implant design were similar to intact knee. This suggests that near normal function may be an expected outcome of a UKA. This study also shed light into the importance of an intact anterior cruciate ligament in maintaining knee kinematics and quadriceps force. The tricompartmental arthroplasty significantly affected femoral rollback and changed knee kinematics. This in vitro cadaver study suggests that the unicompartmental implant design has the potential to restore normal knee kinematic function better than tricompartmental implants. With the restoration of normal knee function, this may have a positive effect on patient rehabilitation, extensor function, implant survival and wearing of the implant design.

In vitro studies suggest that the Oxford medial UKA implant can display more physiologic sagittal plane kinematics, but this has not confirmed in vivo. Therefore, Price et al.[151] developed a fluoroscopic tool for the dynamic measurement of patella femoral angle and to compare the in vivo sagittal plane kinematics at one and 10 years post-operatively. Sagittal plane video fluoroscopy was obtained from five normal, non-arthritic knees, five post-operative total knee arthroplasty (TKA) knees (12-24 post-TKA) and 10 knees following UKA (five at one year, and five at ten years post-operatively). The fluoroscopy images were collected during three exercises. Linear interpolation, repeated measures analysis of variance, post-hoc tukey tests and a separate 1-way ANOVA were used for data analysis. The Oxford UKA patellar tendon angle with flexion did not change out to ten years post-operatively when compared with healthy controls. This suggest that a normal pattern of sagittal plane knee kinematics exists. It also implies that the anterior cruciate ligament function is maintained in the long term as well. However, an abnormal patellar tendon angle was reported in TKA patients when compared to both controls and UKA groups and was attributed to the anterior cruciate ligament not being present due to the surgical procedures of the TKA.

The purpose of the study by Pandit et al.[10] was to compare mid-sagittal plane kinematics, using a standard fluoroscopic technique during a step-up activity with a deep knee bend, and mobile bearing movement of knees in ten patients with a combined anterior cruciate ligament (ACL) reconstruction and Oxford UKA with a matched group of Oxford UKA patients with an intact ACL. In addition to the fluoroscopy, both groups were evaluated clinically using the Oxford Knee Score, the American Knee Society Score (KSS) and the Tegner activity assessment. Data were analyzed using non-parametric tests, Mann-Whitney U tests and the Kruskal Wallace test. Both groups displayed a patellar tendon angle similar to that of the normal knee in both the step up and deep knee bend activities which require knee extension and flexion movements. In terms of the mobile bearing, both groups' bearings moved posteriorly as the knee extended, whereas in the unloaded knee the bearing tends to move anteriorly during extension. This study confirmed that the kinematics of the knee is normal after Oxford UKA and is due the patients return to a high level of function after both UKA and ACL reconstruction.

Previous cadaveric studies have reported that healthy, intact ACL is critical for the success of UKA. There are limited in vivo, weight bearing UKA analysis in the literature. Therefore, Argenson et al.[152] used fluoroscopy in 20 patients to analyze the kinematics of UKA patients, with an intact ACL during deep knee bend maneuvers to maximal flexion. The contact position of the components was used for sagittal plane analysis and the angle between the longitudinal axis that passes through the femoral component and the fixed axis through the tibial component were used to assess axial rotation. Normally, anterior femorotibial contact in full extension and 14.2 mm of posterior rollback on the lateral femoral condyle with progressive flexion. Subjects that underwent medial UKA experienced a normal anteroposterior kinematic pattern, but less rotation than in a normal knee. The medial condyle remained in a similar contact position throughout the deep-knee bend. Eight subjects in this study experienced an anterior contact position in full extension, mimicking that of a posterior cruciate retaining TKA in which the ACL is no longer intact suggesting that the the ACL was unable to thrust the femur anteriorly at full extension. This anterior translation observed could have a number of potential negative consequences including, less maximal knee flexion, decreased quadriceps efficiency and polyethylene wear. These finding support the case that the ACL may contribute to UKA longevity and plays a significant role in knee kinematics of UKA patients.

The mechanism of failure of the UKA implant is not well understood, but abnormal knee kinematics may contribute to failure. Therefore, the purpose of the study by Hollinghurst et al.[20] assessed knee kinematics determine the cruciate ligament function in 24 patients whom had undergone a St Georg Sled fixed bearing UKA. The patients were divided into two groups: early (2-5 years post-operatively) and late (>9 years post-operatively). Video fluoroscopy was used while the participants performed three exercises and were used to determine patella tendon angle (PTA) which was the primary kinematic variable. Data were analyzed using independent *t*-tests and one-way analysis of variance. They reported that the average sagittal plane kinematics of the knee following UKA (in both the early and late groups) are similar to the normal knee in both the short-term and it is preserved into the medium to longer term post-operatively. This suggests that the ligaments and soft tissues around the knee continue to function normally following UKA, and that perhaps more importantly, the cruciate function is retained 10 years after the operation.

Conclusion

Sagittal plane kinematics are often evaluated through cadaveric studies, or in vivo through fluoroscopy[10,20,150-152]. In these studies, the Oxford Knee Society and Knee Society Scores and the Tegner activity assessment questionnaire have also been used to assess self-reported patient functional outcomes[10]. In assessment of the sagittal plane kinematics via the patellar tendon angle following UKA, normal sagittal plane kinematics are reported[10,20,150,151]. However, when evaluating patellar tendon angle in a study evaluating total knee arthroplasty patients and UKA patients, the TKA patients had an abnormal patellar tendon angle, which was attributed to those patients not having an ACL. Retention of the ACL has been reported to maintain kinematic knee function [150] and contributes to UKA longevity[10].

Unicompartmental Knee Arthroplasty and Stair Negotiation

Introduction

Gait analysis can lend insight into post-operative unicompartmental knee arthroplasty (UKA) function. In assessment of gait, UKA patients walked with increased walking velocity[21,153] and increased both stride length cadence[21,153] meaning they are confident and able to accept weight normally during gait[154] which is an important clinical outcome[153]. During the surgical procedure of UKA, the anterior cruciate ligament is retained and is attributed to a normal biphasic flexion/extension knee moment observed during gait as well as an ability to maintain a normal quadriceps contraction as evidence by the normal flexion/extension range of motion[154].

Review of Literature

Following UKA questions remain weather these patients increase their walking velocity or either increasing their cadence and/or step length. Therefore, the purpose of the study by Webster et al.[153] was to characterize footstep patterns and knee kinematics in 12 patients undergoing UKA. Kinematic data was obtained using Vicon 3D motion analysis and spatial and temporal parameters were measured using an electronic mat. All participants, wore shoes and four walking trials each were performed at self-selected and at fast speeds. For data analysis paired *t*-tests, and correlation matrix was used. They reported an increases in maximum walking velocity of 28% and they increased both stride length and cadence to achieve this velocity. The ability of post-operative velocity is an important functional outcome. The UKA patients are reported to preserve the biphasic knee flexion-extension pattern post-operatively and was comparative to controls. An interesting finding that warrants further investigation was that eight of 12 patients displayed significant increases in knee flexion in the operated limb when compared to the contralateral limb. Results provide evidence that patients undergoing UKA are able to achieve good clinical outcomes as determined by gait analysis.

The intactness of the anterior cruciate ligament (ACL) depends on the type of knee arthroplasty procedure performed. Deficiency of the ACL is known to cause downhill walking difficulty. The purpose of the study by Wiik et al.[21] was to examine whether downhill walking gait pattern was different between different types of knee arthroplasty implant types. Fifty-two subjects walked an on instrumented treadmill and were assigned to one of three groups: 1) total knee arthroplasty (TKA) implant design (Genesis II cemented cruciate retaining TKA, Smith & Nephew, Warwick, UK), 2) UKA implant design (Oxford UKA, Biomet, Swindon UK) or 3) young healthy controls. The patients' downhill preferred walking speed (PWS) was chosen by the subject. After determining the PWS, the speed was increased incrementally until downhill top walking speed (TWS) was attained. One-way analysis of variance with Tukey post hoc test and independent *t*-tests were used for data analysis. They reported the the gait patterns of the UKA group resembled the normal controls more than the TKA group. The UKA also walked 15% faster than the TKA group which was attributed to the longer stride length and more normal weight acceptance of the UKA group. The ability of the UKA to walk downhill in a near normal physiological gait pattern served as the functional difference between these two arthroplasty implant designs.

There is limited data suggestion a functional advantage of the UKA over TKA. Therefore, Chassin et at.[154] performed a gait analysis on ten UKA patients. Results were compared to a similar cohort of patients whom had undergone TKA and to a group of controls. Patients walked a 10-m walkway at a self-selected, slow and fast velocities. Fisher's exact test and linear regression were performed for data analysis. The UKA group exhibited a normal biphasic flexion/extension moment during stance which may be attributed to the intact ACL. Knee adduction moment (KAM) was significantly larger in the UKA group when compared to the TKA group. The post-operative limb alignment correlated with KAM recorded during the gait analysis. Post-operative alignment of UKA patients tended to be in varus alignment and may explain the increase in KAM in this group. By having an intact ACL, UKA patients maintain normal quadriceps contraction and knee flexion/extension range of motion during gait.

Conclusion

Walking gait analysis post-UKA reveals that sagittal plane kinematics[154] return to normal and these patients walk with an increased walking velocity[21,153] a positive functional outcome. Unicompartmental surgery has a less invasive surgical procedure, and retains the ACL[154] which can be attributed to the return of normal gait parameters in these patients.

Unicompartmental Knee Arthroplasty and Stair Negotiation

Introduction

With aging, stair negotiation can become challenging due to decreases in muscle strength, loss of balance and confidence. Biomechanical changes can be observed during stair negotiation in patients undergoing UKA in the sagittal plane[155], frontal plane[155] as well as the transverse plane[30]. These changes can be attributed to knee extensor strength deficits and less compressive loading on the knee joint surfaces with stair negotiation[155]. However, following UKA, the motion in the transverse plane resembles that of the normal knee and kinematics can return to normal after surgery[30].

Review of Literature

Compressive loading of the knee joint could exacerbate joint osteoarthritis (OA), and little is known regarding UKA moments during stair ascent. Therefore, Fu et al.[155] investigated knee kinetics in 26 patients with medial and lateral UKA's during stair ascent. Mechanical limb alignments were measured using uniplanar radiographs. Patients walked barefoot up the stairs at a self-selected speed, and force plates were embedded flush with the floor and in the first step of the stairs. Data were analyzed using paired *t*-tests and Pearson's correlations. The medial knee UKA exhibited significantly less peak knee extensor moments as well as greater late stance peak abductor moments for the UKA limb then the non-UKA limb. The lower peak knee extensor moment may be due to the deficits in knee extensor strength or shifting of the body weight more toward the non-UKA limb. The greater peak abductor moments may be attributed to a shifting of the body more toward the non-UKA side, or speculated to be due to the persistent deficits in knee extensor strength due to the chronic arthritis (strength was not performed in this research). In terms of knee moment patterns, the UKA limbs were similar to the non-UKA limb, but two distinct patters emerged which was attributed to patient age; pattern one have been observed in stair ascent studies of healthy, younger individuals, whereas, pattern two has been attributed to the lack of muscle strength in older populations. Limb dominance and post-operative time were correlated with peak knee abductor moment, indicating that loading increased on the implant component of the UKA limb with increased post-operative time. They concluded that reduced knee extensor moments of the UKA limbs indicate less compressive loading on the knee joint surfaces.

When compared to total knee arthroplasty (TKA), UKA possesses several advantages, however objective means comparing the two implant are limited. Therefore, Jung et al.[30] compared knee kinematics of TKA and UKA implant designs in six patients during stair negotiation. All patients in this study had received a TKA in one knee and undergone a UKA in the other knee. All participants negotiated a four step staircase with force plates embedded in the second and third step, barefoot, and did not use handrails. Nonparametric Friedman test was used for data analysis. The UKA knees exhibited significantly greater degrees of rotation in the transverse plane which resembles normal knee kinematics during stair negotiation, but there were no differences when comparing other parameters. In terms of stair ascent versus stair descent, overall greater knee angle, vertical ground reaction force (GRF), joint reaction force and moments were observed during stair descent. For example, maximum GRF was reported to be about 31.2%BW larger with stair descent than stair ascent. In conclusion, both TKA and UKA overall demonstrate similar knee kinematics, although UKA resembles the normal knee kinematics during stair negotiation.

Conclusion

By evaluating stair negotiation biomechanically, we can gain insight into UKA patients function post-operatively. To compensate during stair negotiation, UKA patients may compensate by shifting the body weight more toward the uninvolved limb due to muscle weakness, or experience a reduction of extensor moments demonstrating less compressive loading on the knee [155]. In the transverse plane, movements resemble the normal knee [30] demonstrating that even after surgery biomechanical variables can mimic the normal knee.

Unicompartmental Knee Arthroplasty and Strength

Introduction

Following unicompartmental knee arthroplasty (UKA), flexion and extension knee strengths deficits are to be expected and are at their weakest one month post-UKA[156]. Improvements in strength and function occur throughout the first year post-UKA[24,156,157]. Flexion and extension remains weaker in the operated leg when compared to the uninvolved limb [156] and this trend continues throughout the second post-operative year[157]. The improvements of strength and function stabilize during the first year after surgery[157].

Review of Literature

A strength asymmetry exists prior to UKA between the involved and uninvolved limbs. There are limited studies evaluating limb strength in both the knee flexors and extensors prior to UKA and up to one year post-UKA. Therefore, the purpose of the study by Rossi et al.[156] was to compare inter-limb torque production, using a cybex isokinetic system, and strength recovery of the knee extensors and flexors prior to UKA and post-UKA at one month, two months and one year in 13 UKA patients. During the isokinetic testing, patients completed full range of motion at speeds of 1.047 radians/second and 3.142 radians/second, three repetitions were completed at each speed. For data analysis analyses of variance, effect size and standard response mean were used. For both knee flexion and extension, the involved side was weaker than the uninvolved side, at all data collection time points. One-month post-UKA, both knee extensors and flexors are at their weakest. One year after surgery, the patients were able to generate 83 and 84% of the average knee extension peak torque and generated 78 and 83% knee flexion peak torque produced on the uninvolved limb at 1.047 and 3.142 radians/sec, respectively. Therefore, limb asymmetry is present even one year post-UKA which is similarly found in total knee arthroplasty patients. An emphasis on rehabilitation and continued longitudinal studies of lower limb muscle weakness after UKA is necessary.

A bi-compartmental knee arthroplasty is a partial knee replacement of medial knee osteoarthritis, with the use of the Oxford knee implant, in addition, the patella is also replaced. The purpose of the study by Chung et al.[24] was to compare isokinetic knee flexor and extensor strength, using an isokinetic dynamometer, and physical performance tasks, including the 6-minute walk test, Timed-up-and-go test as well as a 12 step stair climbing test, in 24 patients undergoing partial knee implant (n=11) or total knee arthroplasty (TKA), n=13. All patients were evaluated pre-operatively and again post-operatively at six-months and one-year. Independent *t*-test, analysis of variance, and the Scheffe and Bonferroni methods were used for data analysis. There were no reported differences between groups in knee extensor and flexor torque, or physical performance outcomes when evaluated collectively, preoperatively or at either post-operative data collection time point. When evaluating them individually, both groups showed improvements in the 6-minute walk test, only the TKA group demonstrated improvements in the stair climbing test. In theory, the partial knee replacement should favor post-operative knee kinematics and function due to the preservation of knee cruciate ligaments. In this study however, it was not superior in recovery of knee muscle strength or physical performance tasks when compared to TKA patients.

Recovery and return to function have been reported to be quicker in patients undergoing UKA. However, studies have shown that muscle asymmetry remains one-year post-UKA and most of these studies focus on strength, rather than power. The purpose of the study by Barker et al.[157] was to examined leg extensor power (LEP), using the Leg Extensor Rig which measures explosive power, and function, using the Oxford Knee Score and Tegner Activity Score in 44 patients undergoing UKA. Data were collected six weeks prior to UKA and at one and two years post-UKA. At one year post-UKA all patients had made significant improvements on all functional measures. There were significant increases in leg extensor power in both operated and non-operated limbs. Between years one and two, there were very slight improvements in strength in both legs. When compared to healthy aged-matched controls, LEP measures at two years post-UKA were decreased. The recovery of function and strength following UKA stabilized by one year, and further improvements the following year were minimal.

Conclusion

Lower leg muscle strength can be assessed using isokinetic cybex systems[156], with dynamometers[24] or by leg extensor power rig systems[157]. In addition, strength can be assessed by functional and activity scores which can shed light into patients' reported muscle strength [157]. An ability to improve upon or participate in physical performance tasks can also be directly correlated to muscle strength[24]. Improvements in function and strength have been reported[24,156,157] and stabilize within the first year after UKA[157].

Total Knee Arthroplasty and Implant Design

Introduction

When conservative treatment of osteoarthritis fail, the patient often undergoes a total knee arthroplasty (TKA). Implant designs used for TKA include a single radius (SR) and a multi-radius (MR) design. The SR implant has one radius, or axis, which is located more posteriorly in the knee[45,47] and has a more anatomical alignment[49]. Whereas, the MR implant has two or more axis varying the joint center throughout range of motion (ROM), which mimics a normal knee[45]. In studies, single radius implants are reported to have a greater mechanical advantage by lengthening extensor mechanism moment arm with decreases quad muscle force making them more efficient[45,47,49,50] and have a high survival rate out to ten years post-operatively[158]. In post-operative TKA patients those with a SR implant gained more flexion rapidly and were able to rise from chair without using their arms and with less anterior knee pain[47]. In contrast, the MR implant is reported to require greater eccentric quad force to generate knee extension[45,49], and patients adopt compensatory motions by having significantly longer sit to stand (STS) times, greater trunk flexion angles during the (STS) and an increased muscle activation of quadriceps due to the need for greater force generation for knee extension to occur[46]. However, with posterior cruciate ligament (PCL) retaining implant, no difference in anterior knee pain and unassisted chair rise test was detected which was attributed to intact PCL lengthening the extensor moment arm in both groups[57].

There are numerous reported advantages to using SR implant design in TKA surgery. The purpose of the study by Gomez et al.[50] was to assess the difference in functional recovery from TKA in 60 patients receiving either a SR or MR implant. Functional recovery was assessed numerous ways including Knee Society Scores (KSS), number of required days of physical therapy, number of postoperative days with crutch use, isokinetic evaluation using a dynamometer of the quadriceps and hamstrings, balance was assessed using a dynamometric balance platform and gait analyzation. Data were analyzed with a Kolmogorov-Smirnov test, Student's t-test, and a Chi square test. Patients with a SR implant had higher functional KSS, attended fewer physical therapy sessions and spent less time on crutches when compared to MR patients. In terms of strength, the SR group had a decreased flexion peak torque, increased extension peak torque and an overall lower flexion/extension ratio demonstrating a greater mechanical advantage when compared the MR group. Therefore, SR implant patients obtained better functional outcomes and had improved extensor performance when compared to MR patients.

Multi-radius implants may not restore the extensor mechanism in patients undergoing TKA for the treatment of osteoarthritis (OA). The purpose of the study by Mahoney et al.[47] was to compare the extensor mechanism in 184 participants undergoing TKA with either a MR or SR knee implant. The single radius knee implant had a more posteriorly located center of rotation. Participants were evaluated preoperatively, and postoperatively at six-weeks, 3-months, 6-months, one year and two years using the Knee Society scoring system and the chair rise test to assess extensor mechanism function. Unconditional logistical regression was used for statistical analysis. The SR knees gained flexion more rapidly than the MR group. At six weeks, more patients with the SR implant were able to rise from a chair without using their arms, this trend continued throughout the two years. Participants with the SR knee reported significantly lets anterior knee pain when rising from the chair throughout the study. In conclusion, a more posterior flexion-extension axis, lengthens the extensor mechanism moment arm which decreases the quadriceps muscle force, making the quadriceps more efficient and reduces the knee joint reaction force, thereby reducing anterior knee pain.

With varying axes and therefore, different length extensor moment arms and the amount of force required to perform knee extension should vary among SR and MR TKA implant designs. Therefore, the purpose of the study by Hall et al.[57] was to determine if the SR implant has an advantage in obtaining earlier knee ROM and function when compared to a MR implant in 100 TKA patients. Active knee ROM, Knee Society scores, and the patients' ability to independently rise from a chair were assessed prior to surgery and at four to six weeks, three months and one year postoperatively. Students t-tests and Pearson X^2 tests were used for data analysis. Flexion values, Knee Society, ability to rise from a chair unassisted and anterior knee pain did not differ between these two groups and implants. They attributed the results of the study to using two implants that were PCL retaining. The presence of an intact PCL may have served to lengthen the extensor moment arm in both groups and thus allowed for equal performance in the chair-rise test, so results are limited to PCL-retaining TKA implant designs.

Total knee replacement designs are based on the most common theories and knowledge of the location and orientation of the flexion/extension axis of normal knees. Most MR knee joint implants have two or more joint centers within a function range of motion, however, a SR has one axis. The purpose of the study be Wang et al.[45] is to investigate the SR and MR TKA implant design and the effects of the design on knee kinematic and activation of knee joint muscles on 16 unilateral participants during stand to sit activity. Kinematic and electromyography (EMG) was collected while the subject sat down in a "natural" manner. One-way ANOVA's were used for data analysis. The SR group exhibited less quadriceps and hamstring co-activation than the MR group during knee flexion. Therefore, the MR group had to use a greater quadriceps eccentric force to generate the knee extension torque. The SR group also demonstrated less abduction angular displacement and reached peak abduction earlier than the MR group which would lead to an increase in medial and lateral knee stability. With the more posterior axis, the SR TKA design reduced the amount of knee extensor muscle force necessary to safely lower the body while performing the stand to sit motion.

Even after TKA patients improve their quadriceps strength, but their strength does not reach the level of healthy controls. Ostermeier, et al.[49] investigated the amount of quadriceps force required to extend the knee in 12 human knee specimens, before and after TKA with either a SR or a MR implant design in vitro. The test design simulated an isokinetic extension cycle of the knee, reproducing the physiological forces and moments of the knee. Data were analyzed using a non-parametric paired Wilcoxon signed-rank test. The SR knee resulted in a lower maximum extension force than the MR design which is thought to increase the efficacy of the extensor mechanism. The maximum quadriceps load with the MR implant occurred at a lower knee flexion angle and the forces remained higher in further extension when compared to the SR implant. In addition, the SR design had a more anatomical alignment, leading to a more physiological kinematics of the patella which may also be the reason for the reduced quadriceps force required to achieve knee extension. It can be theorized that TKA patients whom receive a SR implant will have a mechanical advantage in knee extension compared to those receiving a MR implant.

The SR TKA implant contains one axis of rotation that has been shown to decrease the quadriceps muscle force to produce 40 N m-1 knee extension. However, little is known about the SR implant design and its effect on physical function. Therefore, in a follow-up to the previous research study, Wang et al.[46] compared unilateral SR and MR implants during STS activates in 16 participants. All participants underwent 3D sit-to-stand testing while kinematic data was collected as well as EMG for knee flexor and extensor muscle groups. Data was analyzed using one-way analysis of variance, one-way analysis of covariance and Fisher exact probability tests. The MR group displayed compensatory adaptations by have significantly greater STS time, a greater trunk flexion angle, and a tendency of greater trunk flexion velocities, especially reported during the forward-thrust phase. The MR group also displayed increased muscle activation of their quadriceps muscles, reflecting a need for greater force generation for knee extension. Electromyography revealed greater co-activation of the hamstrings in the MR group and it was theorized to have occurred to increase joint stability. The SR implant design allows for adequate knee extensor moments with less quadriceps force required, which provides more functional benefits to patients.

For the purpose of TKA, there are two main types of implant designs; multi-radius and single radius. The purpose of the study by Ji et al.[158] was to determine the survival rates (minimum of ten years) of the single radius TKA implant and to document patellofemoral complication rates in 80 TKA knees (n=54). The Knee Society score, radiographs were assessed using the Knee Society evaluation system, range of motion was measured using a standard goniometer, and anterior knee pain was defined as persistent knee complaint upon standing or stair climbing after TKA and was assessed using the visual analog scale. The Mann-Whitney *U*-test and Kaplan-Meier survival analysis were used for data analysis. Two revisions were performed due to tibial loosening and joint infection. Survivorship was reported to be 96.7% out to ten years. Anterior knee pain was present in six patients, 7.5% of this population, however all six patients were able to climb up and down stairs slowly. It was important to note that no patellar revision surgeries were performed in this sub-group. The single radius TKA implant demonstrated excellent 10-year survivorship rates. The lower rates of implant loosening and anterior knee pain are comparable with results reported in the literature.

The purpose of the study by Stoddard et al.[159] was to evaluate the stability of a single radius versus multi-radius implant design for mid-range instability in cadaveric knees. Repeated-measures two-way ANOVA's were used for data analysis. Both TKA's has limits of laxity that did not differ from each other, from 30-60°. Both implant designs were different from the natural knee allowing for greater anterior drawer laxity near extension and was attributed to excision of the anterior cruciate ligament. Mid-range instability was not attributed to the specific design of the implants and may be related to unrecognized ligament laxity during surgery.

Improvement in knee extensor strength following TKA using a SR implant design has been attributed to the ration between forces in the quadriceps and the patellar tendon, or attributed to the patellar tendon moment arm. The purpose of the study by Ward et al[58] was to investigate and explain the mechanical advantage using three different SR implant designs in six cadaveric knees. The increase in patellar tendon moment arm did not explain the reduced knee extensor strength in the TKA. No significant differences between implant designs were noted and implant designs did not affect the outcome variables. A possible explanation for this is that trochlea on the femur is shaved down which thereby reduced knee flexion angle as the knee flexed.

Conclusion

With advances in technology and new implant designs on the market, the question of the best knee implant design remains. Implant designs are assessed many different ways; biomechanically through gait analysis and electromyography[45], functionally through the sit to stand time test[46] and subjectively through Knee Society Scores[47]. In studies, single radius implants require less quadriceps muscle force making them more efficient[45,47,49,50] and have a high survival rate out to ten years post-operatively[158]. Those patients receiving MR implants require greater eccentric quad force to generate knee extension[45,49], and patients may adopt compensatory motions[46]. Implant design research must continue in order to continue to improve TKA patient outcomes and to provide them with the most function and efficient knee implants.

Total Knee Arthroplasty and Patient Satisfaction

Introduction

The rates of total knee arthroplasties (TKA) and revision surgeries related to TKA is expected to rise 673% and 601% respectively[160]. With the rise in this number of surgeries patient satisfaction and life post-TKA is an important surgical outcome. Patients who reported to be satisfied or very satisfied after TKA ranged from 68-81%[161-164]. Of those individuals who were satisfied, a higher proportion of them were under the age of 60 years old[163]. Of those patients who reported being dissatisfied, most were older in age, (60-75 years old)[162,163], lived alone[163], experienced knee stiffness or swelling once a week[162], had a decreased range of motion (ROM)[163,165] and pain in their knee[162,163]. Revision surgery was performed on those patients with a lack of progress, therefore require a joint manipulation[161] or due to aseptic loosening[166]. Within the first two post-operative years following TKA infection and instability are the main reasons for revision and are surgeon dependent outcomes that can be controlled[166]. Long-term failure, greater than 15 years, was due to implant component wearing[166]. Most TKA patients, even post-operatively may still experience pain in their knee[161]. It has been reported that implant type[163], patellar resurfacing[163] or higher post-operative range of motion[165], did not influence the fulfillment of expectations.

Review of Literature

The incidences of total knee replacement surgeries have been on the rise. The purpose of the study by Kurtz et al.[160] was to formulate projections for the number of primary and revision total hip and TKA that will be performed in the United States through the year 2030. Data was collected using the Nationwide Inpatient Sample database (from 1990 to 2003) was used in conjunction with the United States Census Bureau data and were used for the projections. The prevalence of arthroplasty surgery was modeled with the use of Poisson regression model with age, gender, race and/or ethnicity. For data analysis Pearson chi square test was also used. In the year 2003, a total of 402,100 primary and 32,700 revisionary total knee arthroplasties were performed. Total knee arthroplasties are projected to grow by 673% to 2.95-4.14 million performed annually between 2005 and 2030. Revision for TKA are also projected to grow by 601% during the same time frame. This study provides a quantitative basis for the number of orthopedic surgeons needed to perform these procedures and hospitals should begin to have the appropriate resources in place to serve this need. With the rise in number of people undergoing TKA in the future, it is important for patients to have the best clinical satisfaction as an outcome.

The purpose of this study by Noble et al.[162] was to determine which factors contribute to patient satisfaction with TKA. Researchers developed a self-administered survey called the Total Knee Function Questionnaire (TKFQ), which consists of 55 multiple choice questions relating to the patient's symptoms and functional ability that was administered to 253 patients one-year post-TKA. The patients were asked for responses relating to three different categories of activities involving the knee: 1) baseline activities defined as activities of daily living (ADL); 2) advanced activities which required greater ROM, greater strength and control; and 3) recreational activities. Any finally, patients answered questions regarding current symptoms of their knee, ability to walk, pain medications consumed due to knee pain as well as demographic data. Statistical significance of the differences in TKFQ scores was assessed using an analysis of variance (ANOVA) test. Seventy-five percent of the patients reported that they were either satisfied (18%) or very satisfied (57%) with their knee replacement. Factors associated with a higher proportion of patient satisfaction were age 60 or under at follow-up. Factors that were associated with a lower proportion of satisfied patients included an age of 60-75 years old, knee stiffness at least once per week, swelling of the affected knee at least once per week and use of analgesics at least once per day to treat pain associated with the affect knee. There was a correlation between patient satisfaction and limitations in performing activities involving the replaced knee. One half of the dissatisfied patients reported that they were not as active as they expected they would be before the operation compared with 15% for the satisfied patients. Results of this study suggest that satisfaction with the outcome of TKA has more to do with each patient's subjective perception of their knee function than the biomechanical performance of their knee. Dissatisfaction was generally due to disability and inability to perform functional activities that they consider important to the extent that they want without difficulty or a recurrence of pain and symptoms. Suggesting that real improvements of post-TKA patient's satisfaction will be realized one physician's address patients' preoperative concept of satisfactory outcome as much as the functional performance of the knee implant.

Numerous studies indicate that only 82%-85% of patients were satisfied with their primary TKA. Therefore, Bourne et al.[163] performed a cross-sectional study of patient satisfaction after 1,703 primary TKAs. Western Ontario and McMillan Osteoarthritis Index (WOMAC) questionnaires were given at the time of the primary TKA and one year later for the purpose of this study. Data was analyzed using Kolmogorov-Wilcoxon Test and SPSS. Overall satisfaction revealed that 81% were satisfied or very satisfied while 19% were very dissatisfied, dissatisfied or neutral. Satisfaction did not vary by type of prosthesis nor whether the patella was resurfaced during the surgical procedure. Only 72% were satisfied with their ability to go up or down starts as compared to 85% with walking on a flat surface. Dissatisfied TKA patients were older, lived alone, were less likely to have 90° of flexion preoperatively and have extreme pain on the WOMAC pain score while lying or sitting. Satisfaction with pain relief varied from 72%-86% and with function from 70%-84% for specific activities of daily living. The most significant factors associated with primary TKA patient dissatisfaction were expectations not being met, a low one year WOMAC score, a low preoperative WOMAC score and a complication requiring hospital readmission.

Symptoms of OA do not always correlate well with radiographic changes; however radiographic changes play a key role a patient's need for a TKA. The purpose of this study by Peck et al.[161] was to access weather patients with only mild radiographic changes, whom underwent TKA, had as good as outcome as others undergoing TKA. Zimmer NexGen Cruciate Retaining prosthesis was the implant of choice for all participants. Oxford Knee Scores (OKS) were collected on patients at both three and 12 month post-operative appointments. Unpaired *t*-test and paired sample *t*-test were used for continuous data and for non-continuous data; chi-squared test. In this study, patients who presented with early radiographic changes of OA had statistically significant final post-operative OKS scores. Overall patients who were "satisfied" or "very satisfied" with surgical outcomes was 68% of the population. Eight of the 44 TKA's had to under-go further surgery (3 for a lack of progress, 3 knees underwent revision surgery, one due to infection and one for loosening of the un-cemented femoral component). All 44 patients complained of significant knee pain but only mild OA changes were observed radiographically supporting the case of severity of OA symptoms does not lead to more progressive form of OA. Despite significant symptomatic complaints of OA pain, backed up by arthroscopic evidence, absence of significant radiological changes can lead to poor postoperative TKA outcomes and should be considered with caution.

The purpose of the study by Schroer et al.[166] was to attempt to identify the reasons why TKA patients need revision surgery and why the failure mechanism has changed over the past 10-15 years. This was a retrospective study with 844 failed knee TKA's from six different orthopedic institutions who were in need of revision surgery. Mean age was 65.0 years and body mass index (BMI) was 33.8. Men made up 37% (313) knee failures and women 63% (531). All institutions used a standardized spreadsheet to record data and categorize the mechanism of the failure. Mean time to revision was 5.9 years (range from 10 days to 31 years). More than one-third (35.3%) of knee revisions occurred in the first two years, 24.9% from 2-5 years, 29.5% from 5-15 years and 10.3% after 15 years. Aseptic loosening was the predominant mechanism of the knee implant failure, followed by: instability, infection, polyethylene wear, arthofibrosis, and malalignment. These six mechanisms of failure represented 89.7% of all the failures. Failure mechanisms vary over time. Instability (25.2%) and infection (22.8%) were the most common failure mechanisms needing revision within two years of their initial surgery, but were rare after 15 years. Both of these causes are in large part under the surgeon's control. Arthrofibrosis represented 10% of all revisions less than five years, but was uncommon after five years. Again, arthrofibrosis also surround the surgical procedures and therefore are under the surgeon's control. Polyethylene wear represented 1% of revisions under five years, but was the leading failure mechanism after 15 years. This can be attributed to the improved implant design changes that have occurred over the last 15 years that have decreased the wear of the implants. Aseptic loosening was the only failure mechanism that was consistent across time, representing more than 19% of failures in each time period. Of all the failure mechanisms, Aseptic loosening is not well understood and is often times the "catch all" diagnosis phrase in which an alternative diagnosis could not be made. In conclusion, implant performance does not seem to be the predominant factor in knee failure. Early failure mechanisms are primarily surgeon dependent and over time (15 years) the implant may wear enough to warrant a revision surgery.

Post-operatively, most TKA patients rarely beyond 120° of knee flexion and are able to complete most activities of daily living within that ROM. However, 140° of knee flexion are required for kneeling activates and for getting up and down from the floor and few studies have examined whether post-operative flexion is correlated with patients' perception of surgical outcome. Therefore, Devers et al.[165] determined whether high knee flexion lead to improved benefits in patient satisfaction, perception and function, using the Total Knee Function Questionnaire (TKFQ) in 122 post-operative TKA patients. Patients for this study were evaluated pre-operatively as well as one year post-operatively. In addition to the TKFQ knee ROM was assessed and Knee Society scores were completed and patients were categorized into three groups based on maximum knee flexion ROM: low flexion ($\leq$ 110°), midflexion (111°-130°) or high flexion (> 130°). Analysis of variance and chi-squared tests were used for data analysis. The individuals within the high flexion group reported to that; the surgery achieved their expectations, their knee "felt normal", they had no limitations to what they wanted to do, and their post-operative activity level was more than it was prior to TKA. No correlation was discovered with knee society scores and knee ROM however, the TKFQ demonstrated that high knee flexion was significantly associated with achievement of pre-operative expectations and with the elimination of functional limitations post-operatively. Although the degree of post-operative knee flexion did not affect overall patient satisfaction, it did influence the fulfillment of expectations, functional ability and knee perceptions. This study suggests that increased knee flexion, of greater than 130°, may lead to improved outcomes post-TKA.

The goal of total knee arthroplasty is to reduce pain and improve the patient's mobility. However, the ultimate goal of the treatment should be for patients to have long-term satisfaction. Therefore, Robertsson et al.[164] used the Swedish Knee Arthroplasty Register (SKAR) questionnaire in 27,372 knees. Student *t*-tests, Mann-Whitney U-test, the Krustkal-Wallis H-test and the Spearman analysis were used for data analysis. The patients with the highest reported satisfaction were the TKA patients and the medial unicompartmental knee arthroplasty (UKA) patients. In this study, 81% of the patients overall were satisfied, 8% were dissatisfied while 11% remained uncertain. Of those patients undergoing TKA, those undergoing patellar resurfacing with a button were more satisfied when compared to those not undergoing patellar resurfacing. In those with osteoarthritis, patients undergoing revision surgery of the primary TKA were more dissatisfied than after revision of primary UKA. Patient satisfaction has been significantly correlated to pain and physical function after TKA, this study supports that satisfaction after TKA is long-lasting and even after revision, most patients remain satisfied with the surgical outcomes.

Conclusion

Total knee arthroplasty post-operative patient satisfaction is mostly assessed by using self-reported questionnaires[161-165]. But can also be evaluated using patient chart review[166] or computer databases[160]. With the predicted rise of TKA surgeries[160], patient satisfaction is an important clinical outcome. Satisfaction after TKA ranges from 68-81%[161-164] and has been associated with those younger individuals undergoing the operation[162]. Patient dissatisfaction was typically due to disability and an inability to perform functional activities[162,163]. Revision rates are fairly low for this surgical procedure[161,166], but are attributed to lack of patient rehabilitation progress[161] or aseptic loosening[166] and anterior knee pain was still a post-surgical complaint[161]. It is important for surgeons to explain surgical outcomes so that the patients' expectations, and therefore, satisfaction can be met by undergoing TKA.

Total Knee Arthroplasty and Function

Introduction

Following total knee arthroplasty (TKA) balance and mobility testing allows for the assessment of patient function. In addition to patient questionnaire to assess function[26,51,52,54,167-169], clinical tests include the Berg Balance Test (BBT)[54], the Timed up and go (TUG)[51-54] and the sit to stand test[31,52]. The most common clinical test is the TUG test[51-54] which is used for a quick clinical assessment of the patients overall function[52]. This test has been reported to be correlated with the Western Ontario and McMaster Osteoarthritis Index (WOMAC) scoring questionnaire[51] and TUG times even one- to two-years postoperatively are slower when compared to controls[51]. Questionnaires have identified patients, even as far out as two years postoperatively, reported improvements in quality of life, health and activities of daily living[167-169]. When evaluating greater function through stair negation, as identified through patient questionnaires, moderate pain was reported[51] and women were more likely than men to use the handrails for assistance[54]. Those with ischemic heart disease or had low preoperative function[168], reported more residual knee pain and poorer functional outcomes post-TKA. Preoperative characteristics that predict function at six months post-TKA include: greater joint function, lower comorbid conditions, and ability to walk a greater distance[168]. Recommendations for those undergoing TKA would be to improve patient pre-TKA function and activity, for improved post-TKA function[167,168].

Review of Literature

Assessing physical activity following total joint replacement (TJA) is important given the negative consequences of activity in patients undergoing these surgeries. The purpose of the study by Naal et al.[26] was to determine which is the best activity rating scale using the University of California, Los Angeles (UCLA), the Tegner score and the Activity Rating Scale for use in 105 total hip arthroplasty patients (THA) and 100 TKA patients. These scales will be correlated with the International Physical Activity Questionnaire (IPAQ) which has validity and is widespread in scientific studies. The 95% confidence intervals were used for reliability values. In patients undergoing TJA the UCLA activated scale correlated better with the other measures, provided better reliability and completion rate than the Tegner scale and the ARS. Therefore, of the three scales evaluated in this study, the UCLA seems most appropriate for assessing activity levels in patients undergoing TJA.

After a TKA is performed, residual knee pain post-TKA can adversely affect patient satisfaction and functional outcome. A study by Nashi et al.[167] identified the incidence, progression of knee pain and the functional outcome post-TKA in a retrospective review of 357 patients. Patients pain was assessed using Knee Society Scores (KSS) and WOMAC scores assessed functional outcome and each patients scores were reviewed at three months, six months, one year and two years post-operatively. Chi-square test and Spearman's correlation test were used for data analysis. The main finding demonstrated that a significant proportion of the patients (28.9%) reported residual knee pain at two years post-operatively, though their functional scores continued improving. Patients suffering from ischemic heart disease were more likely to have residual knee pain and reported poorer functional outcome scores. And finally, it was reported that males and patients with a posterior-stabilizing implants were found to report better functional outcomes and both the one and two year post-TKA mark. It is important to understand that factors such as gender, the presence of ischemic heart disease and the implant design may have an effect on the development of post-TKA residual knee pain and functional outcomes.

The purpose of the study by Jones et al.[168] was to identify preoperative determinants of functional status after TKA in 276 patients. This prospective, longitudinal study used the WOMAC scores to measure pain, functional outcomes and stiffness as well as the Medical Outcomes Study 36-Item Short-Form Health Survey (SF-36) which assessed quality of life. Univariate linear regression and multiple linear regression were used for data analysis. They reported that patients with greater dysfunction prior to surgery functioned at a lower level post-operatively at six months than those with a higher preoperative functional status. Furthermore, preoperative joint function, comorbid conditions, preoperative walking distance and walking devices were more predictive of function at six months than preoperative knee flexion. At six months post-TKA 60% of patients reported moderate to extreme difficulty descending stairs. Therefore, patients with low preoperative function may require pre-surgery rehabilitation and/or further rehabilitation post-TKA to improve functional outcomes.

Improvements in quality of life and functional ability are considered the most important outcome in major joint replacements. Few studies have addressed these, therefore, the purpose of the study by Rissanen et al.[169] was to describe and explain changed in 276 total hip and 176 knee arthroplasty patients' quality of life and functional ability using the health-related quality of life (HRQOL), the Nottingham Health Profile (NHP) and the Activities of Daily (ADL) questionnaires. Data were analyzed using t-test, multivariate regression models and ordinal least orders squared techniques. From the NHP major improvements were reported in pain, sleep and physical mobility. Only 9.7% of knee patients had a worse HRQOL score and the oldest patients gained least in terms of the scores. And, in terms of ADL scores, post-surgical scores decreased so these patients had a diminished need for help in everyday activities. Generally speaking, post-surgical improvements to quality of life, health and activities of daily living were reported.

To assess gait, balance and fall risk in elderly individuals the dynamic gait index (DGI) was developed. Low DGI scores are likely to provide a good indication of fall risk. The purpose of the study by Herman et al.[54] was to evaluate the DGI and its association with psychological components and measurements of balance and mobility in 278 healthy elderly individuals. The BBT was used to measure balance and mobility and the TUG test assess functional mobility. The self-reporting activities specific balance confidence (ABC) scale and the geriatric depression scale were used to assess the individual's fear of falling, depression and anxiety. For data analysis Pearson's correlation coefficient, Chi-squared and Student's t-tests were used. The DGI was moderately correlated with the BBT, TUG and the ABC scale, meaning that those with balance problems or were fallers performed worse on the DGI. This study reported the DGI most gender-specific item was related to stair climbing. Women were reported to more often than men hold onto the hand railing, even when comparing healthy, non-fallers. The DGI was able to identify subtle changes in performance and it appears to be an appropriate tool for assessing function in healthy older adults.

Most studies use self-reported questionnaires; few studies have used the TUG test to assess overall function mobility after a total knee arthroplasty. Therefore, Rossi et al.[51] was to explore the relationships between mobility and self-reported function of 11 patients who had undergone TKA approximately 17 months prior. To access mobility, the TUG test was used and the WOMAC was used to evaluate self-reported pain, stiffness and physical function. For data analysis intraclass correlation coefficient, 1-tailed paired t-test, Cronbach α and Spearman correlation coefficients were used. Individual's 10 to 26 months after surgery were 28% slower when compared to an age matched healthy control group when completing the TUG test. An interesting reported finding in terms of the TUG test was that none of the TKA patients took longer than 12 seconds to complete the test and therefore would be considered without risk for a fall. A majority of the TKA subjects reported at least moderate pain during stair climbing and that heavy domestic duties and getting in and out of the bathtub are more challenging for post-TKA patients to complete. In this study, post-TKA patients with greater WOMAC pain scores had higher times to complete the TUG test. And perceived function was moderately correlated with the TUG test for mobility. Simple instruments like the WOMAC and TUG test can provide objective measures of perceived function and mobility in post-TKA individuals.

After TKA knee function can be quantified by patient-based scales, with questionnaires, or by performance based measures. Measurements of quadriceps strength, TUG and stair climbing would be examples of performance based measures. The purpose of the study by Boonstra et al.[52] was to assess which functional knee test are most selective and functionally content valid for quantification of knee function in 28 TKA patients. Western Ontario and McMaster Universities Osteoarthritis Index and KSS were used to ass self-reported knee function. For performance-based measures the sit-to-stand test (STS) with the use of a force plate and a bi-axial accelerometer and one gyroscope was used to determine kinetics and joint kinematics. Maximal isometric contraction of the quadriceps and hamstrings were performed to assess lower leg strength. And finally the TUG test was also used to assess function due to it ability to be used in the clinical setting. Student t-test and Pearson correlation coefficients were used for data analysis. The WOMAC, KSS, STS and TUG were all able to discriminate between TKA patients and healthy controls. However, these patient-based scales are largely influenced by pain and this should be cautioned when used to quantify knee function. The STS and TUG were both selective and had functional content validity with the maximal knee angular velocity and loading asymmetry. An interesting finding of this study applies to the TUG test which uses time as a measurement, which is a very global measure. A person can have an asymmetric limping gait pattern when compared to healthy control, and can have a faster time. They recommend that the TUG be used as a part of a more detailed evaluation. In summary, patient-based scales are heavily affected by pain and therefore should not be used to measure knee function. The TUG can be used for a quick clinical initial assessment of global function and the STS is a more biomechanical instrument identifying knee function.

The purpose of the study by Su et al.[31] was to determine the biomechanics of chair rising in 12 patients after TKA and compare to 12 healthy elderly subjects and 14 OA sufferers prior to TKA. Repeated measures ANOVA was used for statistical analysis. The OA group and the

TKA patients had an increased in time to complete the chair rise which was interpreted that they had greater difficulty with this task. Additionally, they both displayed greater forward displacement of the center of mass during than activity when compared to the healthy controls. This compensatory motion helps to decrease the flexion angle, reaction force and flexion moment of the diseased knee where quadriceps strength might be weaker. Both OA and TKA patients had lower maximal knee flexion moments. Patient undergoing TKA develop a compensatory motion during chair rising by increasing the forward body flexion and shifting more weight onto their uninvolved limb.

Despite the high number of total knee arthroplasty surgeries being performed annual, there is limited information on expected outcomes in the early postoperative phase and how these outcomes relate to the prognosis and long-term outcomes. Therefore, Bade et al.[53] assessed the predictive value of functional performance and ROM measures taken preoperatively affect long-term postoperative outcomes after TKA on 64 subjects. Active knee flexion and extension ROM was measured with a long-arm goniometer with the patient lying supine. The TUG test and 6-min walk (6MW) test were used to measure functional performance, the TUG measured acute function whereas the TUG and 6MW tests measured long-term function. Independent t-tests, chi-squared tests, repeated measure linear mixed model, and linear regression were used for data analysis. Preoperative knee flexion and extension measures were found to be a significant predictor of long-term ROM. Preoperative TUG performance was predictive of long-term functional performance on the 6MW test performance. Although clinicians may not have access to preoperative TUG times, the TUG test can be performed acute setting and can predict long-term functional performance.

Following TKA, patients should experience improvements to quality of life and an increased ability to perform activities of daily living. The purpose of the study by Standifird et al.[39] was to compare knee biomechanics during stair ascent between 13 TKA patients and 15 controls. All participants wore standardized running shoes and were unable to use the handrail during the stair trials. All participants completed a physical activity readiness survey and a timed-up-and-go test in addition to the stairs. Mixed model analysis of variance and Bonferroni post-hoc tests were performed for data analysis. Controls had greater ROM than TKA patients and in TKA patients the non-replaced knee had greater passive ROM. There were no differences in velocity or ground reaction force variables between TKA and controls during stair ascent.

Replaced and non-replaced knees of TKA patients were less flexed at contact compared to the control group. The TKA group had reductions in loading response peak knee extension moment compared to control limb and decreased push-off peak knee extension displaying deficits are greater during the first half of the stance phase as well as increased hip ROM. This suggests that the TKA patients rely on their hip joint for stability and propulsion when compared to controls and may be a developed compensatory mechanism for instability and weakness. There were no differences reported in TUG times between controls and TKA patients, however differences were apparent in the functional stair test showing the benefit of more demanding physical clinical tests in order to bring out the differences between these two populations.

There have been few studies that attempt to describe the biomechanics of patients during stair ascent so little is known about the demand for quadriceps control throughout the activity. Therefore, the purpose of a study by McClelland et at.[40] was to investigate the prevalence of abnormal knee flexion-extension patterns during both stair ascent and descent in a group of 40 patients following total knee arthroplasty and were compared to age-matched controls. During the gait analysis, American Knee Society Scores and a Total Knee Function Questionnaire (TFQD) were also collected. Hierarchial cluster analysis, Step-wise discriminant function analysis, and independent t-tests were used for data analysis in this study. Almost half of the TKA patients could not ascend or descend the stairs without assistance compared to 80% of the controls. Stair descent was a greater challenge for patients than stair ascent. Most of the patients that could ascend and descend the stairs did so with a moment that changed direction in a similar patter to all the control participants. Most of the peak knee biomechanics of these patients were also not different from controls. A subgroup of TKA participants adapted an apparent avoidance of generating a knee flexion moment as evidence by a reduction in magnitude of the knee flexion moment and a premature change to the knee extension moment. Rehabilitation strategies that specifically address these characteristics may improve stair climbing ability in elderly patients after TKA.

Understanding the contributions of individual joint moments after TKA may enhance rehabilitation protocols and long term surgical outcomes. Therefore, Mandeville et al.[41] examined individual joint moment patterns during level walking and stair ascent in 21 TKA patients prior to and after surgery and compared them to 21 age matched controls. All participants underwent gait analysis along a 10-meter walkway at their self-selected speed while barefoot. A mixed-model analysis of co-variance (ANCOVA) was used to analyze between and within-group effects, Bonferroni correction and independent t-tests were used for data analysis. Level walking gait stride length and velocity was found to increase post-TKA across the testing time period, but controls walked at a faster velocity than the TKA group. The TKA group ascended stairs significantly slower than the control group. The total support moment and knee joint moment was significantly less in the TKA group during stair ascent. The TKA knee contributions were 12.9% and 22.8% less than controls, appearing to have limited knee extensor moment production. In TKA group, ankle contributions were 2.2% and 6.6% larger. The pre-TKA group was characterized by a slower velocity, shorter stride length and neutral knee extensor moment, with limited knee flexion when compared to controls. Prior to surgery TKA subjects maintain a stiff knee prior to help alleviate pain, and post-TKA the stiff knee angle may represent an attempt to stabilized the knee joint against the external flexion moment generated by ground reaction force. Post-surgical rehabilitation should concentrate on preserving hip and ankle functions which contribute to the total support moment in walking and stair ascent, along with immediate post-surgical rehabilitation that targets knee extensor strength and muscle activation.

Conclusion

In patients undergoing TKA, improvements in quality of life, health and activities of daily living have been reported[168,169]. To assess patient function in the clinic, self-reported questionnaires were used[51,52,54,167-169] as well as balance and mobility testing[51-54]. The TUG test is a popular test[51-54] and can be used for a quick clinical assessment of TKA function. It is important to note that TUG times remain 28% slower when compared to controls following TKA surgery[52] leading for us to believe that although reporting improvements in function, TKA patients do not return to the function and mobility when compared to healthy controls with do not suffer from osteoarthritis.

Total Knee Arthroplasty and the Patella

Introduction

The patella is an important anatomical structure for successful total knee arthroplasty (TKA) surgeries. Within TKA patients, regardless of undergoing a patellar resurfacing, non-surfacing or receiving a patellar implant, similar patellofemoral contact patterns were reported[170,171]. When compared to normal knees however, TKA patients demonstrate varied patellar kinematic patterns of wear[170]. After TKA, complications related to the patella include;

patellar crepitus[172], patellar loosening[173] and patellar fractures[65,173,174]. Researchers[172-174] have reported clinical and radiographic variables that may increase TKA patients' incidences of patellar fractures[173,174] or patellar clunk syndrome[172]. The average time to patellar component failure was reported to be seven years[173].

Review of Literature

Total knee arthroplasty implant designs with a more posterior center of flexion, should theoretically, require lower extensor forces for the same external load applied. Browne et al.[72] investigated two different TKA implant designs; the LMA (with long extensor moment arm) and a control design (that had a changing center of rotation) in six cadaveric knees. Knees were mounted in a dynamic 188 quadriceps driven closed kinetic chain knee simulator and under-went TKA with the two different types of knee implants. Patellar components with a load cell recorded patellofemoral compressive forces in addition to superoinferior and mediolateral shear forces during knee extension for both the LMA and control implants. Repeated measures multifactorial analysis of variance (ANOVA) was used for data analysis. This study reported that the LMA implant, with the longer moment arm, significantly reduced quadriceps tension. This reduction in quadriceps force should theoretically allow for certain activities of daily living and could accelerate patient rehabilitation after TKA. In addition, the LMA implant also reduced patellofemoral forces, which can help alleviate some of the increase in contact stress and less anterior knee pain after TKA as well as positively impact patellar component wear and loosening. This cadaveric study simulated a controlled closed kinetic change knee extension, the LMA design, has the possibility to enhance function after TKA due to the longer extensor moment arm directly having an effect on quadriceps efficiency.

After TKA, the patella may be vulnerable to fracture especially in individuals with small knees, or after patellar resurfacing surgical procedure. The purpose of the study by Lie et al.[65] was to investigate, using a stationary load frame and strain gauge, the likelihood of an increased risk of patellar facture following TKA using eight cadaveric knees. Patellar strain was measured at 0°, 30°, 60° and 90° of knee flexion in four conditions: intact patella, patella intact with TKA, patella resurfaced and patellar component added and with the patella thickness continuously reduced by 2 mm, to 16, 13 and 11 mm thick, until it was too thin to accommodate the patellar component. Linear regression, two-way ANOVA was performed for data analysis. The major reported finding of the study was that the patellofemoral strain readings increased significantly with knee flexion in all conditions of the experiment and the patellar resection magnified the strain. The patellar bone strain decreased into compressive strain due to the resection in the extended knee, and increased, with greater tensile strains in the flexed knee. During flexion, the anterior surface of the patella became more convex, with high tensile bone strains. In extension, resection cause negative anterior strains, represented by a bending in the opposite direction with large tensile strains on the cut posterior. The bending of the patella was attributed to the cancellous bone that remains after a resurfacing, which deforms easily. This study suggests that the resected patella is safe against fracture if the excision is kept as shallow as possible, leaving 16 mm patellar thickness as long as normal isometric strength is not regained post-TKA. The reduction of strength post-operatively in this population widens the safety margin. However, a 11 mm thickness patellar is vulnerable to fracture during activities as simple as standing up from a chair.

With a posterior-stabilized TKA implant, a unique complication of this implant is patellar crepitus (PC) or clunk syndrome. The purpose of the study by Dennis et al.[172] was to conduct a retrospective analysis was to determine when PC occurs post-operatively as well determine the patient clinical, radiographic and surgical variables that increase the risk of developing PC after TKA in 60 patients with PC (n=44) or patellar clunk (n=16) and compared them to 60 well-functioning, aged-matched TKA patients. Data were analyzed using multivariate logistic regression, two-tailed Fisher's exact test, Shapiro-Wilk W test, Students *t*-test and the Hosmer and Lemeshow test. The average time for PC symptoms to develop was 10.9 months. They reported that radiographic variables found to increase the development PC or clunk included a reduced preoperative or post-operative patellar tendon length, a thinner post-operative composite patellar component thickness and an increase in posterior femoral condylar offset. In addition, the incidence of PC is correlated with a greater number of previous knee surgeries, use of a smaller femoral component, a thicker tibial polyethylene inserts and placement of the femoral component in a flexed position. This study provides insight into surgical variables that could lead to the development of PC or patellar clunk after TKA.

The most common cause of TKA failure is due to patellar complications. The purpose of the study by Meding et al.[173] was to identify patient and surgeon factors associated with patellar component failure in 5,620 patients using the same posterior cruciate retaining ligament TKA using a retrospective review of patient's pre-operative and intraoperative records to identify these factors. The average follow-up time averaged 7 years in this study. Cox hazard ration was used for data analysis. Intraoperatively, TKAs performed with a lateral release had the greatest risk of patellar loosening. And patients with a body mass index of greater than 30 kg/m^2 had the greatest risk of patellar fracture. As far as factors that predict patellar fracturing, male gender, preoperative varus alignment of greater than 5° and a large patellar component size predicted a higher risk of patellar fracture was reported. The greatest predictors for patellar loosening were medial patellar component position, tibial component thickness of greater than 12 mm, preoperative alignment of 10° or more and a preoperative flexion of 100° or more. Recognition of these risk factors for patellar component failure may help determine relative indications for both TKA and patellar resurfacing.

Patellar fractures can be a complication after TKA surgical procedures and mechanisms causing such fractures is not clear. Therefore, Seo et al.[174] performed a retrospective case-control analysis to identify clinical, radiological and surgical factors that increase the risk of developing a spontaneous fracture of the patella after resurfacing in 64 knees (n=60), or 1.1% of the population studied. Information obtained for this study included preoperative and post-operative range of motion, the University of California Los Angeles (UCLA) scoring system, Knee Society scores (KSS), the Hospital for Special Surgery (HSS) knee score, radiographic alignment, and surgical data including; tibial, femoral and patellar component sizes, composite patellar thickness and whether a lateral retinacular release had been performed. The Sharpiro-Wilk test, univariate analysis, two-tailed Fisher's exact test, Students *t*-test or the Mann-Whitney U tests were used for data analysis. More fractures occurred in women (n=53) than men (n=11) and 67.2% of all the fractures were asymptomatic. The clinical variables reported included higher post-operative knee flexion, greater post-operative activity and a lower post-operative KSS score. Radiographic variables reported to increase the risk of patellar fracture included higher pre-operative mechanical malalignment, shorter post-operative length of the patellar tendon, higher patellar tilting angles, greater change in anteroposterior femoral diameter and anterior patellar displacement and lower post-operative patellar thickness. And finally, patients having more than one previous knee surgery have an increased risk for patellar fracture. It is vital to have an understanding of the risk factors associated with a spontaneous patellar fracture following TKA. This study provides insight into these factors, in the hopes that with some preventative measures, this challenging complication can be avoided.

Most previous research studies of the patella involved in vitro analysis, however, Stiehl et al.[170] investigated in vivo patellofemoral sagittal plane kinematics, using fluoroscopic surveillance during weight-bearing deep knee bends in 14 normal knees, 12 anterior cruciate ligament (ACL) deficient knees, and 55 TKA knees. Of the TKAs, 39 had resurfacing with a dome-shaped patella, 8 had resurfacing with an anatomic mobile-bearing patella, and 8 were not resurfaced. All patients were asked to stand with the knee in full extension and perform three weight-bearing deep knee bends, fluoroscopy analysis occurred simultaneously. They reported that the patellofemoral contact patterns were similar for the knee types tested in this study, however, the patellar kinematic patterns post-TKA were more variable when compared to subjects having a normal knee or an ACL deficient knees. Knees implanted with the dome shaped patellar prosthesis displayed the most abnormal results attributed to surgical techniques and placement of the prosthesis, with a more superior patellofemoral contact point and greater patellar tilt angles. The subjects with ACL deficient knees and the TKA patients with the normal patella or resurfaced with an anatomical shaped prosthetic patella were comparable to kinetics of the normal knee.

Patellofemoral forces have been estimated using in vitro cadaveric models, and few studies have explored the in vivo patellofemoral mechanics beyond 90° of flexion. Therefore, Sharma et al.[171], investigated the patellofemoral forces into deep flexion (above 90°) for two high-flexion TKA implants, underweight bearing conditions in 20 patients and compared them with seven healthy, normal knees, using fluoroscopy. The patients flexed their knee from full extension to maximum weight-bearing flexion without an, and without lifting their heels off the ground. Along with the fluoroscopy, ground reaction forces were recorded as well, as patients performed this movement on a force plate. For data collection non-parametric descriptive statistics and Spearman's correlation coefficient were used. In all three groups, the quadriceps force decreased with flexion which was attributed to the moment arm increasing, due to the posterior movement of the femur on the tibia, therefore, decreasing the force in the quadriceps during knee flexion. However, at maximum flexion, the normal knees experienced lower forces compared with the TKA groups. The patellofemoral contact forces do not drastically increase in the TKAs at deep flexion. In terms of implants used for the TKA, the posterior cruciate retaining TKA exhibited greater resemblance of patellofemoral forces to the normal knee than the fixed bearing posterior stabilized TKA although it was not significant. The patellar ligament to quadriceps force ratio decreased with the increase in knee flexion, while the patellofemoral to quadriceps force ratio increased. The implanted knees experienced similar

Conclusion

Through the use of cadaveric studies[65,72], retrospective chart review[172-174] and fluoroscopic[170,171], we have gained insight into the patella and its relationship with TKA surgery. Post-operative TKA patients often experience less patellofemoral forces and strain, therefore, decreases in anterior knee pain due to the increased moment arm of the implant design[171]. Having an understanding of clinical variables[173,174] and radiographic variables[65,173,174] is important to reducing the risk of patellofemoral complications[172-174] after TKA.

Total Knee Arthroplasty and Patellar Resurfacing

Introduction

During total knee arthroplasty (TKA), surgeons can resurface the articular cartilage on the posterior aspect of the patella, or opt to leave it un-resurfaced. There are mixed reviews when evaluating the literature on which procedure produces greater surgical outcomes. No differences were reported between patellar groups in patient reported clinical outcome scores[61-63,175-178], in reported functional outcomes[62,70,179] or in anterior knee pain[62] between resurfaced and un-resurfaced groups. Contrarily, it has also been reported that those patients under-going resurfacing experience more anterior knee pain[63,175] when compared to un-resurfaced. In studies, the resurfaced group had better reported satisfaction, outcomes and reported function[62,176-178] as well as lower complaints of anterior knee pain[61].

Review of Literature

When examining studies of resurfacing of the patella, none of them have involved blinded examiners and randomization. Therefore, Barrack et al.[175], investigated the indication for patellar resurfacing in a randomized, prospective, blinded study in 118 knees (n=58) using the Knee Society Scores (KSS) and a patient satisfaction was assess using a detailed questionnaire. Patients were randomly assigned either the resurfacing of the patella group (n=58) or the non-resurfaced group (n=60). Data were analyzed using analysis of covariance, Kruskal-Wallis Tests, Pearson chi-square tests, repeated measures of variance, Wilconxon signed-rank tests and the Fisher exact test. There were no reported differences between groups with regard to patient satisfaction, KSS for pain or function, or the assessment of patellofemoral function. Pre-operative anterior knee pain was reported to be a logical reason to resurface the patella, and was successful in 92% of the cases in this study. A greater number of the patients in the group that did not undergo resurfacing experienced complaints of anterior knee pain post-operatively (13% compared to 7%), but this was not statistically significant (p=0.38), but it is important to point out that in the non-resurfaced group, there was an associated 10% need for subsequent resurfacing. Those TKA patients not undergoing patellar resurfacing but must be willing to accept the risk that anterior knee pain may persist after surgery and that patellar resurfacing surgery might be necessary. On the other hand, patients undergoing patellar resurfacing in conjunction with TKA who suffer from anterior knee post-operatively, may not have surgical options to resolve this pain.

At the time of TKA, whether or not to resurface the patella remains controversial. Roberts et al.[62], performed a prospective study on patient outcomes with or without patellar resurfacing in 327 knees that had remaining articular cartilage on the patellar articular surface. Patients enrolled in the study undergoing TKA, were randomly assigned to patellar resurfacing group (n= 135) or the non-patellar resurfacing group (n=178). Patient satisfaction, revision, Knee Society score and Knee Society function scores were used to assess patient outcomes and scores were recorded preoperatively and post-operatively at two years. One hundred and fourteen of the knees were followed for greater than 10 years and were analyzed separately. A two-sample *t*-test, Fisher's exact test, analysis of covariance and Kalan Meier survivorship analysis were used for data analysis. In this study population, a vast majority of the patients with remaining patellar articular cartilage did very well at an average follow-up of 7.8 years, patellar resurfacing did not affect this. There was no difference among groups with regard to anterior knee pain, stair climbing ability, Knee Society scores or survivorship. However, patients in the patellar resurfaced group demonstrated improvement in patient satisfaction at the final follow-up appointment (mean 7.8 years). No complications of the patellar resurfacing procedure were reported. It was concluded that TKA patients with patellar articular cartilage do very well regardless of patellar resurfacing, but patient satisfaction may be slightly higher in those patients undergoing patellar resurfacing.

The management of the patella in TKA remains problematic. Therefore, Wood et al.[61] reported clinical outcomes, using the Knee Society clinical rating system, of 220 TKA patients who were randomly assigned to undergo TKA with patellar resurfacing (n=92) or without patellar resurfacing (n=128) and followed their progress for a minimum of three years. Clinical evaluations were performed pre-operatively, and post-operatively at three, six and twelve months, and annually thereafter. In addition to the Knee Society clinical rating system, the patients were observed negotiating five stairs. The presence of knee pain, use of handrails, and step approach/technique (reciprocal, operatively treated limb, or non-operatively treated limb) were recorded. For data analysis chi-square test, Student *t*-tests, Kaplan-Meier and the cox proportional hazards regression analysis were performed. There was no difference in the percentage of patients requiring a revision or another type of operation related to the patellofemoral joint. At the last follow-up, there was a significantly higher incidence of anterior knee pain in those patients that had not had the patellar resurfacing performed (31% vs 16%). Weight was reported as the only preoperative variable associated with the development of anterior knee pain post-operatively in patients who did not have their patella's resurfaced. This finding suggests that total joint loading, not obesity, may be a critical factor in the development of anterior knee pain. Resurfacing of the patella does not guarantee a painless post-surgical patellofemoral joint, however, patients undergoing patellar resurfacing had a lower incidence of anterior knee pain in this study.

Chen et al.[176] performed a meta-analysis of all randomized control trials comparing 1,725 knees that underwent a TKA with and without patellar resurfacing to evaluate the efficacy of this procedure. For data analysis the fixed model and random-effects model were used. In this meta-analysis no difference between the two groups were reported in terms of anterior knee pain. They reported that the rate of reoperation was lower following TKA with patellar resurfacing compared to without patellar resurfacing. During long term follow-up (>5 years), the patellar resurfacing group may achieve a higher Knee Society Score than the non-resurfaced group, which would infer a greater clinical outcome in this group of TKA patients. Other benefits to patellar resurfacing may be limited. More follow-up research must be performed to gain insight into functional outcome of patellar resurfacing versus non-resurfacing during TKA.

The leading cause for revision, after infection is due to patellofemoral complications and during a TKA, controversy remains whether or not to resurface the patella. The purpose of the study by Panni et al.[63] was to review the charts of 1,600 TKA's and analyzed the rates of patellar resurfacing. All patients having received patellar resurfacing were asked to complete the Hospital for Special Surgery (HSS) score and undergo radiographic analysis and in select cases a computed tomography scan. They reported an overall patellofemoral complication rate of 7%. Following TKA, anterior knee pain was the greatest complication, followed by symptomatic patellar mal-tracking, due to a greater patellar thickness value, and one patient had patellar component loosening. Advancement in implant geometry and the development of new patellar polyethylene wear, patellar loosening is now less frequent of a complication, and although they are rare, they can be catastrophic events for the patient. Patellar resurfacing must be carried about with a high degree of accuracy to decrease the risk of complications.

Many different tools and questionnaires have been used to identify TKA clinical outcomes. The purpose of the study by Aunan et al. [177] was to compare functional outcomes, using the knee injury and osteoarthritis outcome score (KOOS) in 115 patients undergoing TKA with and without patellar resurfacing at one and three-years post-operatively. Secondary outcome measures were assessed using the Knee Society Score (KSS), the Oxford knee score and patient satisfaction using the visual analog scale (VAS). Data were analyzed using independent *t*-tests, Mann-Whitney U-test, and a mixed model analysis. They reported the mean score for the KOOS outcome measure was in favor of those TKA patients undergoing patellar resurfacing. The KOOS outcome measure was developed for more active patients, and with TKA patients being younger and physically more active, functional assessment after TKA should include measuring tools that take sports activities and other physical activities into account. No differences in patellar groups were observed for the three secondary outcome measures used in this study. The primary outcome measure in this study indicated that patellar resurfacing may be beneficial for knee function after TKA.

The purpose of the study by Berti et al.[178] compared the knee biomechanics in 47 TKA patients with and without patellar resurfacing during stair climbing activities. Clinical evaluation of these patients were performed using the International Knee Society (IKS) and the Hospital for Special Surgery (HSS) scores. Twenty (of the 42 patients) underwent motion analysis during stair ascent using the ELITE six camera system, 10 without patellar resurfacing, 10 without resurfacing and another 10 control subjects were recruited. The staircase had four steps with force plates embedded in the second and third steps. One-way ANOVA, Mann Whitney Test, Levene test, Least Significant Difference non-parametric test and Pearson's chi-square tests were used for data analysis. Differences in IKS, HSS score and passive knee flexion favored the patients whom underwent patella resurfacing and this group had better active knee range of motion during stance phase compared to the non-resurfaced group. Knee adduction moment was reported higher in the non-resurfaced group. When compared to the control group, both patellar groups had a reduced mean velocity, reduction in knee flexion at heel strike, maximum knee flexion in the swing phase and a reduction in the knee flexion moment. Was reduced in the patella resurfaced group. The patella resurfaced group experienced better functional results (passive knee flexion, IKS scores and HSS scores) as well as better kinematic and kinetic results after TKA procedure during stair ascent.

Walking has been reported as the most important activity performed post-TKA for patients. Therefore, Smith et al.[70] evaluated the differences in knee kinematics and kinetics during walking gait in TKA 34 patients (for a total of 41 knees), with (17 patients) and without (24 patients) patellar resurfacing, and compared them to a control group. All participants walked a self-selected speed with footwear on. The control group were asked to walk at slow, medium and fast pace and the data from the speed closest to the patient group was used for comparison. For data analysis Student's *t*-tests, Levene's test and linear regression analysis were used. There were no differences between patellar resurfacing or without resurfacing in any temporal-spatial, kinematic or kinetic parameters between TKA groups. However, when compared to controls used in this study, a number of gait parameters did not improve post-TKA. Both of the patellar groups experienced mild anterior knee pain at similar rates postoperatively (41% in patellar resurfaced and 42% in non-resurfaced). The only gait parameter that demonstrated a trend toward significance was knee flexion at heel-strike, TKA with patellar resurfacing exhibited 3° more knee flexion at heel strike, a difference that was not present prior to TKA ($p=0.023$). This study suggests that pre-TKA gait patterns, rather than patellar surgical procedures are the main determinant of walking function after TKA.

Patellar resurfacing during a TKA surgery is an area of controversy and few investigators have attempted to compare the functional outcome of TKA with and without patella resurfacing. Therefore, Myles et al.[179] measured knee joint motion, using tow flexible electrogoniometers, during functional activities both prior to and after TKA in a randomized group of 42 patients with and without patella resurfacing, and compared these groups to a control group. Patients were tested prior to surgery and post-TKA at four months and between 18-24 months. They performed 11 functional activities at a self-selected speed: level walking, ascending/descending slope 5°, stair ascent/descent, standing/sitting from a low and standard chair and stepping into/out of a bath. Repeated measures analysis of variance, Mauchly's test of Sphericity and Students t-tests were used for data analysis. They reported no differences in the functional electrogoniometery data between the patella resurfaced and patella not resurfaced groups. They did reports changes within both groups performance over the three time periods of nine of the 11 functions. Routine patella resurfacing in TKA does not result in an increase in functional range of movement in this population.

Conclusion

Literature has mixed results regarding whether or not to resurface the patella during TKA. Patient reported outcomes are either in support of patellar resurfacing[62,176-178] or reported no differences in scores[62,175]. Gait analysis was performed between patients undergoing resurfacing or un-resurfaced patella's also reporting no differences between groups[70,178]. Anterior knee pain has been document in both groups post-surgically[61,63,175]. Further evaluation of the patellar resurfacing during TKA is recommended.

Total Knee Arthroplasty and Patellar Thickness

Introduction

Post-operative patellar thickness is one of the most challenging factors facing surgeons performing total knee arthroplasty (TKA) surgeries. Having too thin of a patella places greater strain on the patella[66] and increases the risk of suffering a fracture[59], patellar tracking issues[59] as well as possible anterior knee pain and/or difficulty with stair negotiation[60]. Having a patellar that is too thick, can lead to patellar related post-surgical complications[59,60,67,69], subluxation[59,67] and abnormal patellar tracking[59,60,67,68]. Reproducing the patellar thickness to its original size[59], the thickness which is ½ the patellar width[180] during TKA is the recommendation for surgeons.

Review of Literature

During TKA, the patellofemoral articulation is known to have a significant impact on surgical outcomes. However, remains to be a consensus on the exact relationship between patella-implant thickness and the biomechanical function of the knee after TKA. Therefore, Abolghasemian et al.[69] analyzed the relationship between patellar thickness and range of motion after TKA biomechanical model of the human knee using a computer based biomechanical study and to identify factors influencing this relationship in two cadaveric knees. In the computer based biomechanical study, a virtual 3-dimentional total knee arthroplasty was performed and then the model was used to obtain the maximum possible flexion with differentiating patellar thicknesses of 1 mm. In the experimental investigation portion of this study two cadaveric knees, with anatomically intact joints and full range of motion, under-went a posterior stabilizing TKA. The patellar implants varied in thicknesses ranging from nine to 24 mm with 3 mm increments. With each patellar implant in place the knee was then allowed to bend passively by gravity along without any additional force and knee flexion angle was measured. Paired *t*-tests were used for data analysis. Increasing the thickness of the patella caused an exponential loss of knee flexion in both the biomechanical and cadaveric conditions. This flexion loss followed an exponential pattern with higher patellar thicknesses. From this study, a general recommendation can be made to cut the patella to a depth which restores the native patella's thickness after resurfacing to avoid adverse biomechanical and functional consequences.

Total knee arthroplasty is a common and highly successful surgical procedure, however, patellar thickness is one of the most challenging factors of the surgeon. Therefore, Hsu et al.[59] investigated the effects of patellar thickness on patellar tracking and patellofemoral characteristics after TKA in seven cadaveric knees. The lower leg was secured to a knee loading frame and underwent TKA. Patellar tracking and patellofemoral contact characteristics were analyzed using a magnetic tracking device and force transducer, and through calculations respectively, in the normal knee, with normal patellar thickness after TKA, 2 mm thicker patellar model and a 2 mm thinner patellar model. Repeated measures analysis of variance and Student *t*-test were used for data analysis. There was no difference of patellar thickness on patellar flexion or rotation. The thickness of the patella was reported to increase the effective moment arm significantly only at knee flexion below 35°, even though the actual moment arm exhibited an increase throughout flexion, potentially reducing the range of motion of the knee. With the thicker patellar, the patella was predisposed to subluxation because it remains laterally tilted during most of the knee flexion angles tested. However, a thinner patella reduces contact force, but also poses the potential risks of a stress fracture as well as anteroposterior instability. In both cases, have a thicker or thinner patella had a smaller contact area than the intact and normal thickness patella. Therefore, it is important for the surgeon to reproduce the original patellar thickness while performing resurfacing during TKA.

Over-stuffing the patella of the knee joint occurs during TKA if there is a lack of range of motion due to the thickness of the patellofemoral implant inserted. When over-stuffing occurs, abnormal tensions on the retinacula and other soft tissue structures may contribute to patellar maltracking. The purpose of the study by Ghosh et al.[68] was to determine the effect of overstuffing the patellofemoral joint during TKA on the retinacula using eight cadaveric knees mounted to a test rig and to describe the relationship between the thickness of the patella-plusprosthesis construct and the lengths of the medial and lateral retinaculum. Once in the test rig, a cruciate retaining TKA (Genesis II) implant was inserted and the patellar thickness was tested under four conditions: 2 mm under-stuffed, pre-cut thickness, 2 mm overstuffed and 4 mm overstuffed, knee ROM was measured dynamically using a Polaris optical tracking system and retinacular length changes were measure with a transducer. Two-way analysis of variance and Bonferroni post-tests were used for data analysis. Over- or under-stuffing the patellofemoral joint caused more stretching or slackening of the medial patellofemoral ligament (MPFL), or medial retinacula, than that of the deep transverse iliotibial band (ITB)-patellar band of the lateral retinaculum. The medial retinaculum was stretched or slackened during each of the thickness changes to the patella in this study due to the anatomy of the MPFL being stretched between bony attachments. The lateral retinaculum however, was only stretched during the 4 mm thickness condition which was attributed to the mobile attachment of the ITB and its greater tensile stiffness when compared to the MPFL. As the patellar thickness is increased, the ITB will stretch less than the MPFL, in response to the same load. Abnormal retinacular tensions as the result of an improper patellar thickness may lead to a number of patellofemoral complications which could lead to early implant failure and pain.

In TKA patients correct patellar tracking is critical and plays a vital role in post-operative clinical success and longer-term implant survival. The study by Youm et al.[67] examined weather resurfaced patellar thickness, evaluated using Merchant radiographs, affected postoperative patellar tilting in 272 female knees undergoing TKA. The knees were then categorized into four groups according to the change in patellar thickness: 1) thinker by 1 mm or more, 2) equal or thinner by less than 1 mm, 3) thicker by 1 mm or less and 4) thicker by more than 1 mm. Patellar tilt was determined preoperatively and post-operatively at two weeks and again at six months. For data analysis *t*-tests, analysis of variance and Pearson's correlation tests were used. No differences were reported between groups 1, 2 or 3, however, the post-operative patellar tilt was greater in patients whose patellar resurfacing during TKA, resulted in a patella more than 1 mm thicker than its original thickness (group 4). Most of the patients in this group had preoperative patellar thicknesses of less than 19-20 mm, resulting in an increase in post-operative patellar thickness, which was inevitable. In these patients, the preoperative tilt was also linked to post-operative tilt. It is important for surgeons to not make the patella too thick during the resurfacing, to ensure good post-operative patellar tracking.

Knee range of motion may be impacted due to the thickness of the patellar prosthesis-bone composite after a TK). The patella is measured intraoperatively and its thickness is used to guide the depth of the resection. However, the thickness of the patella is difficult to estimate because of the wearing and osteoarthritis process. Therefore, Iranpour et al.[180] evaluated the relationship between the patellar thickness and various patellar dimensions by three-dimensional computed tomographic scans on 37 normal adult knees. Spearman's rho correlation test was used for data analysis. In non-symptomatic adults, the patellar thickness was reported to be highly correlated to its width, more specifically the thickness was ½ of the maximum width. It is important to point out that this 2:1 ratio does not take into account the thickness of the patellar cartilage, which is estimated to be 4 mm, and it progressively decreases after the age of 50. The width: thickness ratio appears to be anatomically constant and may be a useful guide for estimating patellar thickness.

Due to the pre-surgical patellar thickness in some patients undergoing TKA, achievement of precut thickness is often difficult. Koh et al.[60] retrospectively compared the clinical outcomes of 56 patellae's' resurfaced to less than 12 mm and 56 patellae resurfaced greater than 12 mm using measurements off radiographs. Patients function was assessed objectively using the KSS. Student's *t*-tests and chi-squared tests were used for data analysis. There was no difference in function or clinical outcomes between the two groups in this study. In the group with the patella resurfaced to less than 12 mm, a higher proportion of patients in this group had either anterior knee pain or difficulty climbing stairs, however, it was not statistically significant. With the patella being over-stuffed to greater than 12 mm, more patients in this group suffered from patellofemoral complications when compared with the other group, again although not statistically different. The concluded that an over-stuffed patellar thickness may be associated with patellar related complications and that a thickness of less than 12 mm did not appear to affect the clinical outcome in the patients used for this study.

An area of controversy in TKA, deals with patellar resurfacing. The literature is divided with plenty of evidence for never resurfacing the patella or that resurfacing should be performed but patellar dislocations are very rare in this population. The purpose of a study by Singh et al[181] was to report a rare case of atraumatic spontaneous patellar dislocation in a 63-year-old man who had undergone a TKA for the treatment of osteoarthritis (OA), and eventually a patellar replacement as well due to a consistent complaint of anterior knee pain. He presented clinically with a painful locked knee following a sudden flexion movement, a radiograph confirmed that his patellar implant had displaced into the infrapatellar area and he underwent revision surgery. The original thickness of the patellar prosthesis inserted was 41 mm. An insert of that size, raises the tension in the patellofemoral joint during movements involving high flexion angles and it is speculated to have possible caused the dislocation. The new patellar prosthesis after the revision was, 25 mm, therefore restoring the knee joint forces closer to that of a normal knee. If a patellar replacement surgery is not performed correctly, the patient's condition may be compromised. It is important to have a thorough understanding of the patellofemoral joint during surgery to avoid future complications.

Patellar related complications are the leading cause of failure following total knee arthroplasty surgery. The purpose of the study by Reuben et al[66] was to examine the effect of patellar thickness on the quadriceps strain and the anterior aspect of the patella following TKA in 10 cadaveric knee joints. Tests were conducted in the intact knee, then by either a posterior cruciate ligament retention or sacrifice of TKA without patellar resurfacing. Tests were then performed following patellar resurfacing with an overly thick, optimum and thin patella. Patellar strain was increased as the patella became thinner and was closest to the intact knee when the patellar was not resurfaced. More specifically, a patellar thickness of less and 15 mm resulted in a significantly increased strain on the anterior aspect of the patella. It is recommended that patellar thickness of the patella be at least 15 mm following TKA.

Conclusion

Patellar thickness evaluation through cadaveric studies[59,66,68,69], radiographs[67,180] and evaluation of a clinical outcome questionnaire[60], have identified risks and complications if the patella is left too thick[59,60,67-69] or too thin[59,60] post TKA. The goal of surgeons performing TKA is to reproduce the original patellar thickness while resurfacing the patella[59,69], or at least no less than 15 mm[66] yielding a greatest surgical outcomes.

Total Knee Arthroplasty and Gait

Introduction

One of the main goals for patients undergoing total knee arthroplasty is to restore walking gait function post-operatively. Improvements in walking velocity[182,183], decreases in KAM and knee forces and an improved impact absorption and function[182] have been reported. Spatiotemporal parameters improve following TKA, but when compared to controls post-TKA patients' velocity remained slower and stride length shorter[183]. However, not all literature supports these improvements in walking following TKA. The knee may also be susceptible to higher knee torques, decreased walking velocity, a decrease in walking velocity[184] and remain only slightly improved one-year post-operatively[185].

Review of Literature

It has been previously reported that patients with knee osteoarthritis (OA) present with a greater body sway during standing when compared with age-matched controls due to the result of proprioceptive deficits, muscle weakness and knee joint pain. The purpose of this study by Mandeville et al.[183] was to assess the effect of knee pain and surgery on gait stability in nineteen total knee arthroplasty (TKA) patients during level walking and obstructed walking, when a plastic tube obstacle that was 10% of the participants' height was placed on the walkway. Twenty-one TKA patients and 21 age matched controls underwent motion analysis at two testing periods, pre-surgery and six-months post-surgery. The Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) and Visual Analog Scale (VAS) were used to assess pain, stiffness and activity of daily living difficulty. Two way mixed analyses of variance (ANOVAs), Bonferroni correction, dependent sample t-tests and Pearson correlation coefficients were used for data analysis. Post-TKA subjects significantly increased walking velocity and stride length, and significantly decreased step width and stride time. But when compared to controls post-TKA velocity remained significantly slower and stride length significantly shorter. This trend continued with the post-TKA walking velocity during the obstacle crossing trials. The control and the TKA groups walked and crossed over an obstacle similarly suggesting that TKA subjects have the ability to appropriately manage with center of mass (COM). In conclusion the TKA subjected used a conservative strategy to manage the COM and center of pressure in the sagittal plane, possibly to reduce the kinetic demands on the involved limb

Improvements in objective outcomes (walking velocity, knee range of motion) and functional tests (timed-up-and-go, stair ascent and 6-minute walking tests) post-TKA have been reported. However, they do not provide insight into the effect of the knee implant on the mechanical environment of the knee. Therefore, Hatfield et al.[182] evaluated kinetics and kinematics on 42 patients undergoing TKA. Analysis of gait occurred one week prior to surgery and one year post-TKA. Regression analysis was used to determine the proportion of the postoperative knee adduction moment variance. Minitab statistical software was also used for statistical analysis. Principal component analysis extracted major patterns of variability in the gait waveforms. Walking velocity and WOMAC scores improved post-TKA. Mid-stance knee adduction moment (KAM) magnitude was decreased implying a decrease in medial compartment loading during gait. Overall knee flexion angle magnitude increased due to an increase during swing. In early stance knee flexion moment increased and in late stance knee extension moment was found, indicating improved impact absorption and function. They concluded that TKA changes specific features of the dynamic loading environment and knee motion during gait at one year postoperatively.

There are only a few studies with objective gait assessment being used as a routine functional assessment in the management of patients with knee OA. The purpose of a study by Rahman et al.[185] was to examine the use of inertial measurement units (IMUs) in busy pre- and post-operative outpatient clinics for patients with TKA. Measurements were performed pre-operatively, eight weeks and 52 weeks post-TKA. Motion sensors were attached to each thigh and shank segment while participants walked a 10-meter walkway. Multi-variate analysis of variance (MANOVA) and Bonferroni post-hoc tests were used for data analysis. The gait of the TKA group was only slightly improved one year post-TKA, when compared to the pre-operative data, and both collection periods were significantly less to controls. Knee flexion motion in stance was the most important variable in discriminating between patients and controls. Even after 12 months of surgery, many TKA patients have not improved their gait relative to pre-operative status. Routine gait assessment should be used to guide post-TKA rehabilitation and to develop strategies and ways to improve mobility of these patients.

The purpose of the study by Stan et al.[184] was to assess the changes in human gait and postural control in the early post-operative phases of unilateral TKA in 10 patients, by evaluating the variability of the free moment and postural parameters such as, medio-lateral and antero-posterior displacement and average velocity. All patients received the same posterior cruciate ligament substituting prosthesis. Free moment and postural control were assessed using a force plate. Postural control was measured with eyes open and then with eyes closed. Data collection occurred two days prior to surgery and 12 days post-TKA. Paired samples test, Kilmogorov-Smirnov and Shapiro-Wilk tests were used for data analysis. Their results show that in the early post-TKA phases, free moment is higher on both the operated and the non-operated limbs, meaning that the both knees are subject to higher torques. When compared to controls, the shape of the vertical torque graphs for both knees for the TKA patience demonstrate asymmetries. The stance time was higher post-TKA for both limbs, although the increase for the non-operated limb was greater, attributed to the decrease in walking speed and may reflect a strategy to avoid extra loading, an accommodation for post-TKA patients. In terms of balance, control of balance is weaker in the early post-operative phases of unilateral TKA. The decrease in balance lends to the decrease in gait speed. Therefore, it is important in adopt a well-conducted rehabilitation program to increase walking stability and balance in TKA patients.

It is important for TKA patients to have an understanding that they may experience significant improvements in walking gait parameters post-operatively[182] but that they still may face some walking challenges[184,185]. And when compared to healthy aged-matched controls, TKA individuals remain slower[183].

Total Knee Arthroplasty and Frontal Plane Changes

Introduction

Following total knee arthroplasty (TKA), post-operative self-reported pain and functional scores improved[104,186-188]. In terms of frontal plane changes reported, knee adduction moments initially go down, and return to preoperative values by one-year[186]. It is important to point out that in this group of TKA patients, walking gait velocity increased post-operatively[186]. This may have an impact on implant longevity as knee adduction moment (KAM) has been identified to increase medial joint forces and medial compartment loading. Varus thrust patients had greater knee pain with weight bearing and is again associated with medial joint loading[104]. Compensations for knee motion post-TKA were reported to occur in the ankle[188]. For patients undergoing a unilateral TKA, high KAM values have been reported in the uninvolved knee which may lead to the progression of OA in that limb[187].

Review of Literature

The relationship of static alignment and varus thrust with pain in individuals with established knee OA was the purpose of this research study by Lo et al.[104]. A total of 82 participants were video recorded walking 20 m away from and toward a stationary camera. Varus thrust was determined by two rheumatologists and was classified as being: definitely present, possibly present, or definitely absent. Posteroanterior radiographs of the knee semi-flexed were obtained through a standardized protocol and Kellen-Lawrence Grade (KL) scores were determined from the x-rays. Chi-squared, *t*-test, and ordinal logistic regressions analysis were performed using SAS. In terms of varus thrust; twenty-five of the participants were classified as having definite, 15 possible and 42 had no observed varus thrust. Radiographic OA was more severe in the group with definite varus thrust. In patients with definite varus thrust, 84% of them had static varus corrected anatomical alignment, compared with only 33% of those without varus thrust. This study discovered that in persons with symptomatic knee OA the presence of varus thrust, and possible varus static alignment, are associated with greater overall knee pain, specifically during weight-bearing activities. Treatment of varus thrust with bracing or gait retraining may provide symptomatic relief for patients.

Retrieval studies have shown that the knee adduction moment returns to pre-operative levels as soon as six months postoperatively which is the purpose of this study by Orishimo et al.[186]. A gait analysis was performed on 15 patients preoperatively, six months and one year postoperative TKA. Nine patients presented with a KL three OA and six patients were diagnosed with Grade four OA. Knee Society (KS) scores and KS function scores were collected at each visit as well as static frontal plane alignment using standing anterior/posterior (AP) radiographs. Patients walked 6-m walkway at self-selected pace. For the analysis of data ANOVA, post hoc paired *t*-tests and Pearson correlations were run. Peak knee adduction angles were initially reduced at the six-month data collection but increased 53% of preoperative levels at one year. KS scores and KS function scores improved from preoperative to both the sixth month and one-year gait analysis. Gait velocity increased after TKA. By one year, gait velocity was 11% greater than preoperatively. It was reported that knee adduction moments were initially reduced at the first post-operative data collection but increased to 94% of preoperative levels by year one. Research observations from this study suggest that pre-surgical levels of knee adduction moment might return as early as one year postoperatively.

Although total knee arthroplasty patients experience a reduction of pain post-operatively, many TKA patients do not achieve normal joint function when walking following surgery. Therefore, the purpose of this study by Levinger et al.[188] was to identify biomechanical changes in the lower limb of 32 patients undergoing TKA. Gait analysis was performed prior to TKA and at 12 months post-TKA using 3D motion analysis system. The Western Ontario and McMaster University Osteoarthritis Index (WOMAC) was used to determine physical function, stiffness and pain. Data was analyzed using a mixed-design ANOVA, and Bonferroni post-hoc tests were used. Significant improvements were reported for pain, stiffness, function and overall WOMAC scores following surgery. Peak knee flexion moment, ankle plantarflexion and dorsiflexion moments and peak ankle power generation at push off were significantly increased following surgery, providing insight of the role that the ankle plays in compensating for the impaired functioning of the before and after TKA. Several biomechanical changes in the knee and ankle were identified in the TKA group prior to and after TKA. Rehabilitation strategies may need to focus not only on improving knee function but also on gait retraining to optimize recovery.

The purpose of this study by Alnahdi et al.[187] was to examine frontal plane kinematics and kinetics during walking in patients who underwent TKA. Seventy-five TKA post-surgical participants were enrolled in this study (31 subjects were six-months post-op and 44 subjects were one-year post op). The control group consisted of 20 subjects with no reported knee pain or injury. Gait analysis was performed using 3-D, eight camera motion capture system and seven walking trials were collected (within 5% of the practiced speed) at a self-selected speed. Participants completed the Knee Outcome Survey-Activities of Daily Living Scale (KOS-ADLS), knee flexion and extension ROM was measured, quadriceps strength was assessed, Timed Up and Go Test (TUG), Stair Climbing Test (SCT), and Six-Minute Walk test (6MW) were performed as well. Results from the study revealed that the non-operated knee had larger knee adduction angle and dynamic loading during stance when compared with the operated knee. For the analysis of the persons who underwent TKA, effect of side was significant for the knee angle at peak knee angle, with the non-operated knee being more adducted than the operated knee. Peak knee adduction showed no group by side interaction, no effect of group but the effect of side was significant, the non-operated knee having larger moment. Knee adduction impulse showed no group by side interaction, no group effect, but effect of side was significant with the non-operated knee having larger impulse. Examining stance time revealed no group by side interaction no effect of side, but there was a significant group effect with the six-month group having longer stance time. The presence of the high KAM and dynamic loading may be an underlying reason for OA progression in the non-operated knee.

Conclusion

To evaluate frontal plane knee changes researchers used gait analysis[186-188], questionnaires[104,186-188], functional tests[187], and radiographs[104]. Researchers have identified that KAM returns post-operatively after TKA[186] and during functional tests, KAM is high in the uninvolved knee which could lead to progression of OA in the uninvolved knee[187]. Identifying frontal plane risk factors in knee OA is important to understanding OA and underlying characteristics that may lead to progression of the disease.

Total Knee Arthroplasty and Rehabilitation

Introduction

It is suggested in the previous section that individuals with osteoarthritis (OA) can benefit from improvements in strength. In OA patients undergoing total knee arthroplasty (TKA), the effect of a pre-operative rehabilitation program has been reported to decrease pain[189], higher reported quality of life scores[190], improve function scores[190] and functional tests[189] and strength improved[189,191] when compared to TKA patients not participating in pre-TKA strengthening. A functional post-TKA rehabilitation program was also evaluated and TKA patients were compared to a control group[192]. A further examination into post-TKA strength reveals that throughout the first year, quadriceps strength improved, however, by three-years post-TKA that trend does not continue and is attributed to the worsening of the non-operated limb[73]. The functional rehabilitation group performed better on functional tests and had less pain[192]. Biomechanical improvements in knee flexion excursion, knee flexor moment and vertical ground reaction forces have been observed[191]. Prior to TKA, OA patients may benefit from participating in pre-rehabilitation program[189-191] as well as in a functional rehabilitation program[192].

Following TKA, patients usually walk with asymmetrical movement patterns attributed to weakness of the operated limb especially in the early phases of recovery. This longitudinal cross-sectional study by Yoshida et al.[191] investigated the changes in quadriceps strength and function of both limbs of 14 individuals, for three years post TKA and compared them to age-matched controls. All patients were treated by the same physical therapy clinic with a standardized and progressive rehabilitation program. The Medical Outcomes Study 36-Item Short-Form Health Survey (SF-36), Knee Outcome Survey activity of daily living section (KOS-ADL) were used for clinical assessments. Functional testes included the timed up and go test (TUG), the stair-climbing test (SCT) and the six-minute walk test (6MW). Quadriceps strength was measure with a dynamometer. Data was analyzed using 2-way repeated measures analysis of variance (ANOVA) and paired t-tests. Quadriceps strength was significantly different between groups at three months and one year post-TKA but not at three years post-TKA. As the quadriceps strength equalized between the limbs, there were improvements in symmetry for knee flexion excursion, knee flexor moment, and peak vertical ground reaction force in the TKA group. There was a significant improvement in self-reported function between three months and one year after TKA, but a significant decrease between one and three years. The authors attributed this to the worsening of the non-operated limb, as well as the improvements in the operated limb. Three years post-TKA, differences in kinematic, kinetics and spatiotemporal variables still exist between those with knee pathology and those living without knee pathology. Patients after TKA demonstrate improvements in biomechanical symmetry over time, however, these individuals also demonstrate a progressive loss of strength in the non-operated limb over time. The decreased function during daily activities after TKA is contributed by the quadriceps weakness of the non-operated limb, not due to the TKA.

Prior to TKA measures of strength, functional ability and knee pain have been shown to be significant predictors of post-TKA outcomes. The purpose of this repeated measure design study by Topp et al.[189] was to examine the effect of preoperative exercise program quadriceps strength, pain and functional ability on 26 individuals undergoing TKA. After the TKA was performed all patients participated in the same post-operative functional rehabilitation exercises and after being discharged from the hospital all patients were assigned to nine in-home therapy sessions until they achieved 0-100° of range of motion. During functional tasks knee pain was accessed using the Visual Analog Scale (VAS). To assess functional ability each subject participated in four tasks; distance covered during a 6-minute walk test, number of sit to stand repetitions performed in 30 seconds, length of time required to ascend 22 stairs, rest 30 seconds and finally descend the stairs. Quadriceps strength was measured by a Biodex. Knee pain, functional ability, and quadriceps strength were assessed pre-TKA to establish a baseline and again one month and three months post-TKA. Repeated measure analysis of variance was performed for data analysis. When compared to controls, the exercise group improved their sit-to-stand performance, and had a decrease in pain at both post-TKA data collections. Three months post-TKA the prehabilitation group demonstrated decreases in all measures of pain, improvements in three of the four functional tasks, and improvements in strength in both the surgical and nonsurgical quadriceps. The findings of this study appear to indicate the importance of a prehabilitation program among TKA patients.

Brown et al.[190] also evaluated the effects of prehabilitation on quality of life three months of 17 patients after TKA. Participants were randomly assigned to either the control group who received the usual pre-TKA care or the prehabilitation group that included a warm-up, ten resistance exercises, six stretching exercises, three step exercises and a cool-down for eight weeks prior to TKA. Following TKA all participants received the same standard postoperative care. All participants completed the SF-36 three months post-TKA. Analysis of data included independent t-tests. The data from this study suggest that OA patients who engage in exercise prior to TKA report higher mean general health-related quality of life and well as report higher physical functioning scores three months after surgery. These increases in scores could be the result of the increased pre-TKA strength and functioning of the participant's knee.

Reduction of pain and an improvement in physical function and quality of life are expected outcome of TKA. However the functional benefits of this procedure are not well understood, therefore Moffet et al.[192] evaluated the effectiveness of a new intensive functional rehabilitation (IFR) program on functional ability and quality of life in 77 TKA patients. Two months post-TKA the participants were randomly assigned to two groups, the IFR group received a supervised rehabilitation program between two and four months after TKA, and the control group which were given standard care. Each IFR session included; a warm-up, strengthening exercises, functional task-oriented exercises, and endurance exercises followed by a cool-down. Functional ability and quality of life was accessed post-TKA at two months (to establish a baseline), four months, six months and one year. Subjects in the IFR group walked longer distances in the 6-minute test at the four month, six-month and one-year data collection. At the four and six-month collection they also presented with less pain, stiffness and difficulty performing tasks. Positive changes to quality of life occurred in the IFR group at the six-month evaluation. The IFR was effective in improving the short-term and mid-term functional ability after TKA.

The purpose of the study by Silva et al.[56] was to examine the knee extensor strength in 32 TKA patients compared to 52 normal healthy subjects. Step-wide multivariate regression test was used for data analysis. Knee extensor strength values in patients following TKA were 32.2% lower than healthy control subjects. Women after TKA were weaker than men undergoing the same procedure. Knee society scores were positively correlated to average knee extensor peak torques and a great strength was associated with a better score. Older TKA patients were weaker compared to the younger TKA patients. This data suggests that there is a need for more aggressive rehabilitation after TKA, especially in women, older patients and those more obese patients.

The purpose of the study by Mizner et al.[35] was to evaluate function; by assessing quadriceps strength, knee ROM, TUG test, timed stair-climbing test, pain and knee function questionnaires in 40 unilateral patients at one, two, three and six months post-TKA. One month after surgery it was reported that there was a worsening of knee ROM, quadriceps strength and performance on functional tests. Following the one-month testing, all measurements significantly improved. Quadriceps strength was most highly correlated to functional performance at all testing sessions which suggests that improved post-TKA quadriceps strength could be important to enhance the potential benefits of TKA. Functional measures underwent an expected decline early after TKA, but recovery was more rapid than anticipated.

In TKA patients, the effect of rehabilitation on patient outcomes have been evaluated through clinical outcome surveys[190,191], functional tests[191],[189,192], and through gait analysis[191]. Pre-operative rehabilitation programs improve outcome scores[189,190], function and strength[190]. Functional improvements were reported in TKA patients participating in a functional rehabilitation program[192]. Improvements to sagittal plane walking gait characteristics have also been reported[191]. Patient outcomes may be improved by encouraging TKA patients to participate in strengthening exercises prior to and after surgery.

APPENDIX A

BIOMECHANICAL VARIABLE TABLES FOR ACROSS AGE-GROUPS WITH ALL P-VALUES PRESENT

Participant Demographics

	Group 1 (n=45)	Group 2 (n=39)	G1 to G2	Group 3 (n=32)	G1 to G3	Group 4 (n= 31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Age (years)	25.49 ± 4.98	47.97 ± 3.84**†	0.000	59.46 ± 3.16**‡	0.000	69.71 ± 2.36*^	0.000	0.000	0.000	0.000
Height (m)	1.71 ± 0.08	1.68 ± 0.09	0.474	1.67 ± 0.11	0.239	1.64 ± 0.11*	0.029	0.958	0.497	0.822
Body Mass (kg)	76.73 ± 18.59	77.75 ± 14.97	0.993	73.35 ± 17.43	0.821	70.39 ± 15.67	0.374	0.693	0.270	0.898
UCLA Score	8.33 ± 1.31	8.05 ± 1.35	0.9792	7.44 ± 1.37*	0.032	7.43 ± 1.45*	0.026	0.263	0.225	0.999

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; m = meters; kg = kilograms;
BMI = body mass index; kg/m = kilogram per meter squared; UCLA = University of California Los Angeles;
m/s = meters per second

* = statistically different than Group 1 ($p \leq 0.01$)

^ = significantly different than Age-Group 2 ($p \leq 0.01$).

† = significantly different than Age-Group 3 ($p \leq 0.01$).

‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

Strength and Spatiotemporal Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Knee Extensor Strength (lbs)	109.92 ± 36.68	87.57 ± 31.16**	0.000	81.59 ± 25.37**	0.000	75.87 ± 27.26**	0.000	0.684	0.128	0.742
Hip Abductor Strength (lbs)	77.69 ± 21.85	64.01 ± 20.33**‡	0.000	53.85 ± 16.29**	0.000	54.06 ± 17.40**^	0000	0.014	0.016	0.999
Velocity (m/sec)	1.25 ± 0.16	1.26 ± 0.15	0.984	1.23 ± 0.16**	0.930	1.21 ± 0.16	0.599	0.797	0.421	0.933
Stride Width (m)	0.16 ± 0.04	0.13 ± 0.03**	0.002	0.12 ± 0.03**	0.000	0.12 ± 0.03**	0.000	0.870	0.561	0.954
Stride Length (m)	2.24 ± 0.33	2.01 ± 0.29**	0.005	1.95 ± 0.27**	0.001	1.91 ± 0.35**	0.000	0.848	0.520	0.949
Cycle Time (s)	2.12 ± 0.34	1.62 ± 0.21**	0.000	1.61 ± 0.16**	0.000	1.65 ± 0.23**	0.000	0.999	0.958	0.947
Males	Group 1 (n=22)	Group 2 (n=18)	P-value	Group 3 (n=11)	P-value	Group 4 (n=15)	P-value	P-value	P-value	P-value
Knee Extensor Strength (lbs)	126.88 ± 31.58	105.91 ± 34.61*	0.014	103.86 ± 27.32*	0.026	91.932 ± 23.48**	0.000	0.995	0.255	0.521
Hip Abductor Strength (lbs)	89.13 ± 18.65	76.94 ± 20.17*	0.019	69.12 ± 16.16*	0.000	63.88 ± 16.58**^	0.000	0.428	0.026	0.753
Velocity (m/s)	1.23 ± 0.14	1.27 ± 0.15	0.782	1.30 ± 0.10	0.472	1.19 ± 0.16	0.877	0.925	0.401	0.205
Stride Length (m)	2.29 ± 0.33	2.08 ± 0.25	0.168	1.93 ± 0.26*	0.015	2.00 ± 0.40**	0.040	0.589	0.891	0.935
Stride Width (m)	0.17 ± 0.05	0.14 ± 0.02	0.219	0.13 ± 0.03**	0.052	0.13 ± 0.03**	0.013	0.798	0.593	0.996
Cycle Time (s)	2.17 ± 0.28	1.63 ± 0.20**	0.000	1.53 ± 0.11**	0.000	1.73 ± 0.26**	0.000	0.639	0.640	0.140
Females	Age 20-39 (n=23)	Age 40-54 (n=22)	P-value	Age 55-64 (n=21)	P-value	Age 65-75 (n=16)	P-value	P-value	P-value	P-value
Knee Extensor Strength (lbs)	109.92 ± 36.68	87.57 ± 31.16**	0.000	81.59 ± 25.37**	0.000	75.87 ± 27.26**	0.000	0.684	0.128	0.742
Hip Abductor Strength (lbs)	77.69 ± 21.85	64.01 ± 20.33**‡	0.000	53.85 ± 16.29*'	0.000	54.06 ± 17.40**	0.000	0.014	0.016	0.999
Velocity	1.27 ± 0.17	1.26 ± 0.15	0.999	1.19 ± 0.18	0.376	1.22 ± 0.16	0.756	0.356	0.732	0.959
Stride Length (m)	2.18 ± 0.32	1.94 ± 0.31**	0.048	1.96 ± 0.28**	0.070	1.81 ± 0.28**	0.002	0.999	0.560	0.473
Stride Width (m)	0.15 ± 0.03	0.12 ± 0.03*	0.007	0.12 ± 0.03**	0.015	0.11 ± 0.02**	0.002	0.995	0.908	0.805
Cycle Time (s)	2.06 ± 0.38	1.60 ± 0.22**	0.000	1.66 ± 0.17**	0.000	1.57 ± 0.17**	0.000	0.905	0.980	0.739

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; lbs = pounds; m/sec = meters per second; m = meters; s = seconds

* = significantly different than Age-Group 1 ($p \leq 0.05$).

** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$).

‡ = significantly different than Age-Group 4 ($p \leq 0.05$).

Walking Biomechanical Variables Descriptive Statistics For All Genders and Males Across Age-Groups

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Max vGRF (N/kg)	11.24 ± 1.22	11.21 ± 1.06	0.998	10.87 ± 0.82	0.391	11.00 ± 0.82	0.747	0.509	0.841	0.953
Peak Knee Flexion Angle (°)	18.75 ± 4.84	18.41 ± 6.29	0.973	18.93 ± 4.11	0.991	17.75 ± 4.19	0.997	0.906	0.999	0.922
Peak Knee Flexion Moment (Nm/kg)	0.82 ± 0.26	0.82 ± 0.26	0.999	0.82 ± 0.21	0.999	0.78 ± 0.24	0.952	0.999	0.920	0.931
Peak Knee Adduction Angle (°)	1.34 ± 2.80	2.44 ± 3.17	0.361	2.89 ± 3.64	0.129	1.89 ± 2.51	0.872	0.923	0.876	0.559
Peak Knee Adduction Moment (Nm/kg)	0.40 ± 0.11	0.46 ± 0.15	0.202	0.49 ± 0.18*	0.013	0.45 ± 0.12	0.397	0.705	0.993	0.580
Knee Varus Velocity (°/sec)	45.05 ± 22.21	54.21 ± 24.17	0.429	60.95 ± 29.17*	0.065	62.31 ± 35.79*	0.040	0.735	0.615	0.997
Peak Ankle Flexion Angle (°)	10.09 ± 2.70	10.11 ± 2.54	0.999	9.63 ± 2.44	0.885	9.78 ± 3.22	0.963	0.882	0.960	0.996
Peak Ankle Flexion Moment (Nm/kg)	0.19 ± 0.06	0.19 ± 0.07	0.968	0.20 ± 0.08	0.616	0.21 ± 0.06	0.585	0.871	0.847	0.999
Peak Ankle Inversion Angle (°)	7.34 ± 2.43	8.31 ± 4.59	0.510	8.74 ± 2.53	0.239	9.00 ± 2.65	0.123	0.945	0.808	0.988
Peak Ankle Inversion Moment (Nm/kg)	0.08 ± 0.04	0.02 ± 0.21*	0.074	0.07 ± 0.04	0.948	0.07 ± 0.04	0.989	0.320	0.225	0.997
Peak Hip Flexion Angle (°)	30.70 ± 7.17	28.09 ± 7.19	0.330	27.69 ± 7.35	0.254	27.91 ± 6.31	0.328	0.995	0.999	0.999
Peak Hip Flexion Moment (Nm/kg)	0.62 ± 0.18	0.62 ± 0.18	0.999	0.60 ± 0.19	0.975	0.58 ± 0.22	0.760	0.994	0.849	0.953
Hip Adduction Angle (°)	8.24 ± 3.54	6.42 ± 2.88	0.067	6.77 ± 3.36	0.230	7.42 ± 3.60	0.728	0.973	0.595	0.859
Hip Adduction Moment (Nm/kg)	0.90 ± 0.14	0.88 ± 0.15	0.985	0.98 ± 0.13*^	0.047	0.96 ± 0.16	0.212	0.025	0.126	0.932
Trunk Forward Flexion (°)	5.39 ± 2.65	5.25 ± 3.96	0.998	6.92 ± 3.35	0.249	5.87 ± 4.27	0.939	0.203	0.887	0.644
Trunk Side Bending (°)	2.51 ± 1.69	2.25 ± 1.40	0.918	1.96 ± 2.48	0.577	1.57 ± 1.90	0.140	0.915	0.435	0.843
Loading Rate	6156.33 ± 1512.81	6596.27 ± 1748.50	0.624	6599.99 ± 1696.01	0.997	5838.64 ± 1742.37	0.846	0.440	0.237	0.981
Males	Group 1 (n=22)	Group 2 (n=18)	P-value	Group 3 (n=11)	P-value	Group 4 (n=15)	P-value	P-value	P-value	P-value
Max vGRF (N/kg)	11.39 ± 1.56	11.24 ± 1.44	0.886	10.81 ± 0.95	0.596	10.88 ± 0.67	0.628	0.929	0.960	0.999
Peak Knee Flexion Angle (°)	20.28 ± 5.25	21.40 ± 4.98	0.926	18.58 ± 5.24	0.774	18.30 ± 3.36	0.613	0.487	0.322	0.999
Peak Knee Flexion Moment (Nm/kg)	0.91 ± 0.31	0.83 ± 0.23	0.895	0.85 ± 0.23	0.930	0.80 ± 0.18	0.567	0.999	0.936	0.951
Peak Knee Adducion Angle (°)	1.82 ± 3.04	2.89 ± 2.74	0.623	3.15 ± 2.35	0.561	2.73 ± 2.62	0.754	0.995	0.998	0.981
Peak Knee Adduction Moment (Nm/kg)	0.39 ± 0.11	0.46 ± 0.15	0.197	0.48 ± 0.13	-/110	0.44 ± 0.10	0.499	0.955	0.959	0.799
Knee Varus Velocity (°/sec)	43.77 ± 24.44	54.89 ± 27.83	0.799	73.49 ± 40.81*	0.036	65.57 ± 27.48	0.286	0.243	0.823	0.709
Peak Ankle Flexion Angle (°)	9.76 ± 2.75	8.73 ± 2.02	0.649	9.26 ± 2.70	0.960	9.56 ± 3.41	0.996	0.958	0.829	0.993
Peak Ankle Flexion Moment (Nm/kg)	0.18 ± 0.06	0.19 ± 0.08	0.996	0.24 ± 0.09	0.129	0.20 ± 0.05	0.877	0.227	0.959	0.482
Peak Ankle Inversion Angle (°)	7.74 ± 2.22	8.13 ± 5.31	0.983	7.11 ± 2.34	0.957	8.54 ± 2.32	0.890	0.860	0.986	0.706
Peak Ankle Inversion Moment (Nm/kg)	0.09 ± 0.04	0.01 ± 0.21	0.162	0.06 ± 0.04	0.933	0.07 ± 0.03	0.970	0.635	0.443	0.998
Peak Hip Flexion Angle (°)	31.42 ± 5.68	29.61 ± 8.39	0.865	27.87 ± 7.82	0.546	27.74 ± 7.43	0.431	0.924	0.884	0.999
Peak Hip Flexion Moment (Nm/kg)	0.57 ± 0.18	0.64 ± 0.19	0.591	0.67 ± 0.15	0.420	0.57 ± 0.20	0.999	0.975	0.611	0.442
Hip Adduction Angle (°)	6.92 ± 3.30	5.63 ± 2.18	0.476	5.43 ± 2.70	0.469	5.36 ± 2.49	0.336	0.998	0.992	0.999
Hip Adduction Moment (Nm/kg)	0.87 ± 0.15	0.87 ± 0.17	0.998	0.95 ± 0.10	0.470	0.93 ± 0.11	0.677	0.421	0.613	0.977
Trunk Forward Flexion (°)	5.04 ± 2.60	5.80 ± 4.44	0.917	7.41 ± 3.87	0.298	6.78 ± 3.75	0.487	0.661	0.871	0.971
Trunk Side Bending (°)	2.28 ± 1.41	2.12 ± 1.25	0.987	3.52 ± 2.06	0.125	1.76 ± 1.41†	0.735	0.086	0.909	0.023
Loading Rate	6536.27 ± 1516.44	7222.94 ± 1655.44	0.539	6884.68 ± 1025.20	0.933	6095.18 ± 1893.59	0.839	0.946	0.195	0.594

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; Max = maximum; vGRF = vertical ground reaction force; N/kg = newtons per kilogram; (°) = degrees;
Nm/kg = newton meters per kilogram; °/sec = degress per second

* = significantly different than Age-Group 1 ($p \leq 0.05$).

** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$).

† = significantly different than Age-Group 3 ($p \leq 0.05$).

Walking Biomechanical Variables Descriptive Statistics For Females Across Age-Groups

Females	Group 1 (n=23)			Group 2 (n=22)			P-value	Group 3 (n=21)			P-value	Group 4 (n=16)			P-value	P-value	P-value	P-value
Max vGRF (N/kg)	11.11	±	0.79	11.29	±	0.61	0.879	10.90	±	0.77	0.840	11.12	±	0.95	0.999	0.414	0.919	0.858
Peak Knee Flexion Angle (°)	17.27	±	3.98	16.22	±	6.35	0.882	19.12	±	3.51	0.583	17.23	±	4.89	0.999	0.205	0.920	0.636
Peak Knee Flexion Moment (Nm/kg)	0.72	±	0.16	0.80	±	0.28	0.712	0.81	±	0.21	0.666	0.77	±	0.29	0.933	0.999	0.982	0.969
Peak Knee Adducion Angle (°)	0.90	±	2.55	2.09	±	3.49	0.613	2,76	±	4.21	0.3238	1.09	±	2.20	0.998	0.905	0.791	0.418
Peak Knee Adduction Moment (Nm/kg)	0.42	±	0.11	0.46	±	0.17	0.786	0.50	±	0.15	0.265	0.46	±	0.15	0.805	0.805	0.999	0.861
Knee Varus Velocity (°/sec)	46.28	±	20.31	53.63	±	21.22	0.633	55.44	±	18.68	0.741	70.32	±	44.80*	0.203	0.998	0.814	0.737
Peak Ankle Flexion Angle (°)	10.40	±	2.68	11.18	±	2.41	0.755	9.82	±	2.34	0.883	10.00	±	3.13	0.964	0.333	0.521	0.997
Peak Ankle Flexion Moment (Nm/kg)	0.19	±	0.06	0.19	±	0.06	0.979	0.19	±	0.07	0.999	0.21	±	0.07	0.967	0.904	0.999	0.692
Peak Ankle Inversion Angle (°)	6.96	±	2.61	8.45	±	4.07	0.364	9.59	±	2.22*	0.028	9.44	±	2.94*	0.070	0.364	0.615	0.999
Peak Ankle Inversion Moment (Nm/kg)	0.08	±	0.04	0.03	±	0.22	0.524	0.07	±	0.04	0.997	0.08	±	0.05	0.999	0.663	0.608	0.998
Peak Hip Flexion Angle (°)	30.01	±	8.43	26.91	±	6.04	0.450	27.60	±	7.29	0.663	28.07	±	5.29	0.828	0.998	0.958	0.997
Peak Hip Flexion Moment (Nm/kg)	0.67	±	0.17	0.59	±	0.18	0.539	0.57	±	0.21	0.294	0.59	±	0.24	0.582	0.971	0.999	0.984
Hip Adduction Angle (°)	9.5	±	3.36	7.03	±	3.23	0.075	7.46	±	3.51	0.195	9.38	±	3.44	0.999	0.976	0.158	0.326
Hip Adduction Moment (Nm/kg)	0.92	±	0.13	0.90	±	0.16	0.980	1.00	±	0.14	0.242	1.00	±	0.19	0.377	0.120	0.216	0.999
Trunk Forward Flexion (°)	5.73	±	2.70	4.83	±	3.59	0.823	6.66	±	3.12	0.814	5.02	±	4.67	0.924	0.321	0.998	0.494
Trunk Side Bending (°)	2.73	±	1.93	2.34	±	1.53	0.919	1.14	±	2.31*	0.051	1.38	±	2.20*	0.179	0.211	0.472	0.983
Loading Rate	5792.92	±	1449.02	6112.03	±	1697.21	0.915	5536.58	±	1810.26	0.955	5598.12	±	1611.69	0.983	0.662	0.778	0.999

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; Max = maximum; vGRF = vertical ground reaction force; N/kg = newtons per kilogram; (°) = degrees;
Nm/kg = newton meters per kilogram; °/sec = degress per second

* = significantly different than Age-Group 1 ($p \leq 0.05$).

** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$).

† = significantly different than Age-Group 3 ($p \leq 0.05$).

Stiffness and Power Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Ankle Stiffness	3.33 ± 1.27	3.45 ± 1.63	0.981	3.18 ± 1.49	0.972	2.90 ± 1.43	0.586	0.865	0.396	0.868
Knee Stiffness	5.50 ± 1.92	6.12 ± 2.30‡	0.440	5.06 ± 1.67	0.752	4.84 ± 1.42^	0.453	0.094	**0.029**	0.968
Kvert	0.97 ± 0.53	1.39 ± 1.01*	**0.054**	1.38 ± 0.79	0.089	1.26 ± 0.51	0.358	0.099	0.881	0.917
Sagittal Plane Hip Power Max	1.11 ± 0.30	1.15 ± 0.31	0.935	1.17 ± 0.39	0.935	1.24 ± 0.341	0.876	0.342	0.998	0.698
Sagittal Plane Hip Power Min	-0.90 ± 0.32	-0.98 ± 0.42	0.743	-0.87 ± 0.26	0.980	-0.96 ± 0.36	0.848	0.560	0.999	0.682
Sagittal Plane Knee Power Max	0.77 ± 0.42	0.79 ± 0.44	0.995	0.70 ± 0.42	0.898	0.77 ± 0.41	0.999	0.804	0.999	0.894
Sagittal Plane KneePower Min	-1.56 ± 0.36	-1.84 ± 0.53*	**0.040**	-1.82 ± 0.49	0.085	-1.84 ± 0.50	0.058	0.999	0.999	0.998
Sagittal Plane Ankle Power Max	3.18 ± 0.81	3.00 ± 0.73	0.672	3.08 ± 0.70	0.938	2.75 ± 0.67	0.066	0.965	0.509	0.290
Sagittal PlaneAnkle Power Min	-0.71 ± 0.24	-0.73 ± 0.23	0.977	-0.78 ± 0.35	0.710	-0.71 ± 0.26	0.999	0.911	0.987	0.778
Males	Group 1 (n=22)	Group 2 (n=18)	P-value	Group 3 (n=11)	P-value	Group 4 (n=15)	P-value	P-value	P-value	P-value
Ankle Stiffness	3.59 ± 1.04	4.75 ± 1.04	0.061	4.16 ± 1.55	0.692	3.27 ± 1.56^	0.908	0.698	**0.022**	0.394
Knee Stiffness	5.77 ± 1.61	6.13 ± 2.00	0.921	5.68 ± 1.86	0.999	5.02 ± 1.52	0.579	0.910	0.290	0.781
Kvert	0.96 ± 0.71	1.74 ± 1.34*	**0.053**	1.41 ± 0.97	0.553	1.29 ± 0.54	0.715	0.794	0.516	0.987
Sagittal Plane Hip Power Max	1.02 ± 0.27	1.06 ± 0.31	0.979	1.22 ± 0.40	0.376	1.17 ± 0.38	0.535	0.619	0.795	0.983
Sagittal Plane Hip Power Min	-0.80 ± 0.27	-0.78 ± 0.39	0.991	-0.86 ± 0.20	0.947	-0.88 ± 0.24	0.877	0.869	0.763	0.999
Sagittal Plane Knee Power Max	0.76 ± 0.45	0.72 ± 0.30	0.982	0.68 ± 0.37	0.915	0.65 ± 0.23	0.800	0.989	0.954	0.999
Sagittal Plane KneePower Min	-1.46 ± 0.34	-1.71 ± 0.46	0.306	-1.81 ± 0.39	0.158	-1.83 ± 0.57	0.068	0.943	0.864	0.999
Sagittal Plane Ankle Power Max	3.03 ± 0.78	3.00 ± 0.72	0.999	3.09 ± 0.49	0.997	2.62 ± 0.67	0.309	0.990	0.430	0.353
Sagittal PlaneAnkle Power Min	-0.70 ± 0.24	-0.65 ± 0.25	0.952	-0.85 ± 0.32	0.374	-0.67 ± 0.25	0.986	0.204	0.999	0.285
Females	Group 1 (n=23)	Group 2 (n=22)	P-value	Group 3 (n=21)	P-value	Group 4 (n=16)	P-value	P-value	P-value	P-value
Ankle Stiffness	3.08 ± 1.44	2.44 ± 0.71	0.282	2.67 ± 1.21	0.659	2.54 ± 1.25	0.511	0.926	0.994	0.989
Knee Stiffness	5.23 ± 2.18	6.11 ± 2.55	0.465	4.67 ± 1.50	0.847	4.67 ± 1.35	0.829	0.123	0.1490	0.999
Kvert	0.98 ± 0.28	1.11 ± 0.56	0.846	1.36 ± 0.70	0.098	1.22 ± 0.49	0.506	0.434	0.918	0.874
Sagittal Plane Hip Power Max	1.19 ± 0.30	1.22 ± 0.30	0.991	1.14 ± 0.39	0.970	1.31 ± 0.45	0.724	0.884	0.869	0.484
Sagittal Plane Hip Power Min	-0.99 ± 0.33	-1.13 ± 0.38	0.565	-0.87 ± 0.30	0.686	1.05 ± 0.43	0.962	0.092	0.898	0.451
Sagittal Plane Knee Power Max	0.77 ± 0.40	0.84 ± 0.53	0.957	0.71 ± 0.46	0.973	0.89 ± 0.53	0.881	0.792	0.993	0.679
Sagittal Plane KneePower Min	-1.65 ± 0.36	-1.93 ± 0.57	0.236	-1.82 ± 0.54	0.660	1.84 ± 0.44	0.639	0.889	0.946	0.999
Sagittal Plane Ankle Power Max	3.33 ± 0.82	3.00 ± 0.76	0.492	3.08 ± 0.80	0.725	2.88 ± 0.67	0.287	0.985	0.962	0.852
Sagittal PlaneAnkle Power Min	-0.72 ± 0.23	-0.80 ± 0.17	0.784	-0.74 ± 0.36	0.999	0.76 ± 0.26	0.983	0.862	0.962	0.995

n = number; G1 = Group 1; G2 = Group 2; G3 = Group 3; G4 = Group 4; SD = standard deviation; Max = maximum; Kvert = vertical leg stiffness; Max = maximum; Min = minimum

* = significantly different than Age-Group 1 ($p \leq 0.05$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$).

‡ = significantly different than Age-Group 4 ($p \leq 0.05$).

Table 5.5 Total Work Descriptive Statistics Across Age-Groups

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Total Joint Work (J/kg)	1.69 ± 0.30	1.72 ± 0.34	0.755	1.74 ± 0.28	0.496	1.64 ± 0.35	0.499	0.703	0.349	0.217
Total Ankle Joint Work (J/kg)	0.80 ± 0.16	0.81 ± 0.20	0.854	0.81 ± 0.15‡	0.778	0.73 ± 0.16	0.062	0.662	0.101	**0.052**
Total Knee Joint Work (J/kg)	0.37 ± 0.10	0.40 ± 0.11	0.170	0.41 ± 0.09	0.147	0.39 ± 0.11	0.446	0.857	0.607	0.515
Total Hip Joint Work (J/kg)	0.52 ± 0.11	0.52 ± 0.11	0.894	0.52 ± 0.10	0.891	0.52 ± 0.13	0.872	0.802	0.782	0.984
Sagittal Plane Total Joint Work (J/kg)	0.96 ± 0.18	1.00 ± 0.21	0.528	0.98 ± 0.17	0.586	0.94 ± 0.21	0.624	0.960	0.289	0.346
Fronal Plane Total Joint Work (J/kg)	0.61 ± 0.12	0.61 ± 0.12	0.972	0.63 ± 0.12	0.424	0.58 ± 0.12	0.341	0.420	0.373	0.111
Transverse Plane Total Joint Work (J/kg)	0.12 ± 0.04	0.12 ± 0.04	0.589	0.12 ± 0.04	0.771	0.12 ± 0.04	0.964	0.430	0.636	0.756
Male										
Total Joint Work (J/kg)	1.63 ± 0.32	1.57 ± 0.32	0.576	1.73 ± 0.1	0.357	1.60 ± 0.32	0.757	0.180	0.827	0.265
Total Ankle Joint Work (J/kg)	0.78 ± 0.17	0.71 ± 0.16	0.183	0.84 ± 0.12^	0.350	0.72 ± 0.18	0.268	**0.047**	0.865	0.074
Total Knee Joint Work (J/kg)	0.37 ± 0.12	0.38 ± 0.09	0.600	0.39 ± 0.07	0.529	0.38 ± 0.08	0.711	0.870	0.898	0.784
Total Hip Joint Work (J/kg)	0.48 ± 0.10	0.48 ± 0.11	0.950	0.51 ± 0.09	0.487	0.50 ± 0.11	0.633	0.475	0.612	0.807
Sagittal Plane Total Joint Work (J/kg)	0.95 ± 0.20	0.90 ± 0.20	0.213	1.00 ± 0.12	0.876	0.92 ± 0.19	0.136	0.235	0.781	0.161
Fronal Plane Total Joint Work (J/kg)	0.58 ± 0.12	0.57 ± 0.13	0.788	0.62 ± 0.11	0.431	0.57 ± 0.12	0.734	0.329	0.940	0.309
Transverse Plane Total Joint Work (J/kg)	0.10 ± 0.03	0.10 ± 0.034	0.613	0.12 ± 0.04	0.105	0.11 ± 0.03	0.231	0.256	0.499	0.612
Female										
Total Joint Work (J/kg)	1.74 ± 0.27	1.84 ± 0.31	0.302	1.74 ± 0.31	0.973	1.67 ± 0.38	0.517	0.297	0.116	0.546
Total Ankle Joint Work (J/kg)	0.82 ± 0.16	0.88 ± 0.19	0.261	0.79 ± 0.16	0.505	0.74 ± 0.16^^	0.126	0.081	**0.012**	0.366
Total Knee Joint Work (J/kg)	0.37 ± 0.08	0.41 ± 0.13	0.157	0.42 ± 0.09	0.136	0.39 ± 0.13	0.483	0.927	0.551	0.500
Total Hip Joint Work (J/kg)	0.55 ± 0.11	0.545 ± 0.11	0.900	0.53 ± 0.11	0.606	0.54 ± 0.15	0.843	0.698	0.934	0.783
Sagittal Plane Total Joint Work (J/kg)	0.98 ± 0.16	1.08 ± 0.19	0.072	0.98 ± 0.19	0.990	0.97 ± 0.24	0.851	0.080	0.069	0.845
Fronal Plane Total Joint Work (J/kg)	0.63 ± 0.12	0.64 ± 0.12	0.816	0.64 ± 0.13	0.758	0.59 ± 0.13	0.314	0.938	0.228	0.206
Transverse Plane Total Joint Work (J/kg)	0.14 ± 0.04	0.12 ± 0.04	0.219	0.12 ± 0.04	0.217	0.12 ± 0.05	0.277	0.985	0.968	0.955

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation, J/kg = joules per kilogram

* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).

† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).

‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

5.6 Percentage of Total Work Done By Each Joint

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Total Work at Ankle	48% ± 5%	47% ± 5%	0.310	46% ± 4%	0.532	45% ± 6%*	**0.020**	0.763	0.177	0.122
Total Work at Knee	22% ± 4%	23% ± 4%	0.065	24% ± 4%	0.107	24% ± 4%*	**0.055**	0.929	0.862	0.805
Total Work at Hip	31% ± 4%	30% ± 4%	0.540	30% ± 3%	0.367	32% ± 4%	0.229	0.743	0.086	0.056
Males										
Total Work at Ankle	48% ± 6%	45% ± 3%	0.056	48% ± 3%	0.902	45% ± 4%*	**0.044**	0.086	0.866	0.069
Total Work at Knee	22% ± 4%	24% ± 4%	0.092	23% ± 4%	0.814	24% ± 4%	0.215	0.233	0.707	0.406
Total Work at Hip	30% ± 4%	30% ± 2%	0.523	29% ± 3%	0.618	31% ± 3%	0.160	0.315	0.450	0.101
Females										
Total Work at Ankle	47% ± 5%	48% ± 6%	0.745	45% ± 4%	0.304	45% ± 7%	0.164	0.183	0.095	0.662
Total Work at Knee	21% ± 4%	22% ± 4%	0.377	24% ± 3%*	**0.025**	23% ± 4%	0.142	0.170	0.509	0.539
Total Work at Hip	32% ± 3%	30% ± 5%	0.189	31% ± 3%	0.411	32% ± 5%	0.641	0.634	0.099	0.299

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation, % = percentage

* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).

† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).

‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

5.7 Percentage of All Work Done In Each Plane By Each Joint

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Sagittal Plane Ankle Work	41% ± 6%	39% ± 5%	0.143	40% ± 5%	0.688	38% ± 6%*	**0.030**	0.358	0.434	0.109
Sagittal Plane Knee Work	29% ± 5%	31% ± 6%	0.077	31% ± 5%	0.216	31% ± 5%	0.089	0.702	0.972	0.692
Sagittal Plane Hip Work	31% ± 5%	30% ± 5%	0.827	29% ± 4%	0.370	31% ± 4%	0.472	0.501	0.371	0.141
Frontal Plane Ankle Work	64% ± 7%	64% ± 8%	0.601	62% ± 6%	0.545	61% ± 7%*	**0.031**	0.905	0.107	0.163
Frontal Plane Knee Work	8% ± 3%	9% ± 4%*	**0.055**	10% ± 4%*	**0.042**	10% ± 3%*	**0.034**	0.794	0.751	0.962
Frontal Plane Hip Work	28% ± 6%	27% ± 6%	0.589	28% ± 5%	0.607	30% ± 6%	0.213	0.987	0.091	0.111
Tranverse Plane Ankle Work	26% ± 8%	28% ± 7%	0.154	24% ± 7%	0.385	27% ± 8%	0.508	0.036	0.509	0.163
Tranverse Plane Knee Work	33% ± 8%	30% ± 7%	0.083	35% ± 9%^^	0.291	31% ± 9%	0.298	**0.011**	0.562	0.057
Transvere Plane Hip Work	41% ± 11%	42% ± 9%	0.744	40% ± 13%	0.844	41% ± 13%	0.745	0.630	0.986	0.635
Males										
Sagittal Plane Ankle Work	40% ± 6%	38% ± 5%	0.213	41% ± 5%	0.876	38% ± 5%	0.136	0.235	0.781	0.161
Sagittal Plane Knee Work	30% ± 5%	32% ± 6%	0.136	30% ± 5%	0.899	31% ± 5%	0.360	0.172	0.615	0.373
Sagittal Plane Hip Work	30% ± 5%	29% ± 3%	0.768	30% ± 5%	0.970	31% ± 4%	0.453	0.833	0.329	0.504
Frontal Plane Ankle Work	64% ± 6%	60% ± 5%**	**0.009**	66% ± 4%^^	0.453	60% ± 5%*††	**0.028**	**0.004**	0.746	**0.012**
Frontal Plane Knee Work	8% ± 2%	10% ± 3%**	**0.006**	9% ± 2%	0.191	10% ± 3%**	**0.009**	0.275	0.982	0.296
Frontal Plane Hip Work	28% ± 7%	30% ± 4%	0.212	25% ± 4%^	0.157	30% ± 5%†	0.381	**0.019**	0.754	**0.042**
Tranverse Plane Ankle Work	28% ± 7%	29% ± 8%	0.732	25% ± 7%	0.169	28% ± 7%	0.977	0.112	0.776	0.192
Tranverse Plane Knee Work	35% ± 8%	34% ± 8%	0.579	36% ± 11%	0.752	33% ± 10%	0.433	0.445	0.813	0.344
Transvere Plane Hip Work	36% ± 10%	37% ± 9%	0.815	39% ± 14%	0.507	39% ± 13%	0.526	0.660	0.698	0.934
Females										
Sagittal Plane Ankle Work	41% ± 6%	39% ± 5%	0.405	39% ± 5%	0.431	38% ± 7%	0.111	0.982	0.405	0.391
Sagittal Plane Knee Work	28% ± 5%	30% ± 6%	0.333	31% ± 4%*	**0.035**	31% ± 5%	0.135	0.243	0.543	0.636
Sagittal Plane Hip Work	31% ± 5%	31% ± 6%	0.975	29% ± 4%	0.202	32% ± 4%	0.755	0.218	0.736	0.143
Frontal Plane Ankle Work	63% ± 7%	67% ± 8%	0.144	60% ± 6%^^	0.125	61% ± 8%^	0.286	**0.004**	**0.019**	0.723
Frontal Plane Knee Work	8% ± 3%	9% ± 4%	0.771	10% ± 4%	0.092	9% ± 3%	0.482	0.165	0.665	0.394
Frontal Plane Hip Work	28% ± 6%	25% ± 6%*	**0.043**	31% ± 5%^^	0.400	30% ± 6%^^	0.373	**0.006**	**0.007**	0.913
Tranverse Plane Ankle Work	24% ± 8%	27% ± 7%	0.090	24% ± 8%	0.784	26% ± 8%	0.368	0.164	0.511	0.525
Tranverse Plane Knee Work	32% ± 8%	27% ± 5%*	**0.039**	35% ± 9%^^	0.174	30% ± 8%	0.510	**0.001**	0.214	0.061
Transvere Plane Hip Work	45% ± 12%	45% ± 8%	0.819	41% ± 12%	0.274	44% ± 13%	0.860	0.192	0.702	0.410

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation, % = percentage

* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).

† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).

‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

5.7 Percentage of All Work Done In Each Plane By Each Joint

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Sagittal Plane Ankle Work	41% ± 6%	39% ± 5%	0.143	40% ± 5%	0.688	38% ± 6%*	**0.030**	0.358	0.434	0.109
Sagittal Plane Knee Work	29% ± 5%	31% ± 6%	0.077	31% ± 5%	0.216	31% ± 5%	0.089	0.702	0.972	0.692
Sagittal Plane Hip Work	31% ± 5%	30% ± 5%	0.827	29% ± 4%	0.370	31% ± 4%	0.472	0.501	0.371	0.141
Frontal Plane Ankle Work	64% ± 7%	64% ± 8%	0.601	62% ± 6%	0.545	61% ± 7%*	**0.031**	0.905	0.107	0.163
Frontal Plane Knee Work	8% ± 3%	9% ± 4%*	**0.055**	10% ± 4%*	**0.042**	10% ± 3%*	**0.034**	0.794	0.751	0.962
Frontal Plane Hip Work	28% ± 6%	27% ± 6%	0.589	28% ± 5%	0.607	30% ± 6%	0.213	0.987	0.091	0.111
Tranverse Plane Ankle Work	26% ± 8%	28% ± 7%	0.154	24% ± 7%	0.385	27% ± 8%	0.508	0.036	0.509	0.163
Tranverse Plane Knee Work	33% ± 8%	30% ± 7%	0.083	35% ± 9%^^	0.291	31% ± 9%	0.298	**0.011**	0.562	0.057
Transvere Plane Hip Work	41% ± 11%	42% ± 9%	0.744	40% ± 13%	0.844	41% ± 13%	0.745	0.630	0.986	0.635
Males										
Sagittal Plane Ankle Work	40% ± 6%	38% ± 5%	0.213	41% ± 5%	0.876	38% ± 5%	0.136	0.235	0.781	0.161
Sagittal Plane Knee Work	30% ± 5%	32% ± 6%	0.136	30% ± 5%	0.899	31% ± 5%	0.360	0.172	0.615	0.373
Sagittal Plane Hip Work	30% ± 5%	29% ± 3%	0.768	30% ± 5%	0.970	31% ± 4%	0.453	0.833	0.329	0.504
Frontal Plane Ankle Work	64% ± 6%	60% ± 5%**	**0.009**	66% ± 4%^^	0.453	60% ± 5%*††	**0.028**	**0.004**	0.746	**0.012**
Frontal Plane Knee Work	8% ± 2%	10% ± 3%**	**0.006**	9% ± 2%	0.191	10% ± 3%**	**0.009**	0.275	0.982	0.296
Frontal Plane Hip Work	28% ± 7%	30% ± 4%	0.212	25% ± 4%^	0.157	30% ± 5%†	0.381	**0.019**	0.754	**0.042**
Tranverse Plane Ankle Work	28% ± 7%	29% ± 8%	0.732	25% ± 7%	0.169	28% ± 7%	0.977	0.112	0.776	0.192
Tranverse Plane Knee Work	35% ± 8%	34% ± 8%	0.579	36% ± 11%	0.752	33% ± 10%	0.433	0.445	0.813	0.344
Transvere Plane Hip Work	36% ± 10%	37% ± 9%	0.815	39% ± 14%	0.507	39% ± 13%	0.526	0.660	0.698	0.934
Females										
Sagittal Plane Ankle Work	41% ± 6%	39% ± 5%	0.405	39% ± 5%	0.431	38% ± 7%	0.111	0.982	0.405	0.391
Sagittal Plane Knee Work	28% ± 5%	30% ± 6%	0.333	31% ± 4%*	**0.035**	31% ± 5%	0.135	0.243	0.543	0.636
Sagittal Plane Hip Work	31% ± 5%	31% ± 6%	0.975	29% ± 4%	0.202	32% ± 4%	0.755	0.218	0.736	0.143
Frontal Plane Ankle Work	63% ± 7%	67% ± 8%	0.144	60% ± 6%^^	0.125	61% ± 8%^	0.286	**0.004**	**0.019**	0.723
Frontal Plane Knee Work	8% ± 3%	9% ± 4%	0.771	10% ± 4%	0.092	9% ± 3%	0.482	0.165	0.665	0.394
Frontal Plane Hip Work	28% ± 6%	25% ± 6%*	**0.043**	31% ± 5%^^	0.400	30% ± 6%^^	0.373	**0.006**	**0.007**	0.913
Tranverse Plane Ankle Work	24% ± 8%	27% ± 7%	0.090	24% ± 8%	0.784	26% ± 8%	0.368	0.164	0.511	0.525
Tranverse Plane Knee Work	32% ± 8%	27% ± 5%*	**0.039**	35% ± 9%^^	0.174	30% ± 8%	0.510	**0.001**	0.214	0.061
Transvere Plane Hip Work	45% ± 12%	45% ± 8%	0.819	41% ± 12%	0.274	44% ± 13%	0.860	0.192	0.702	0.410

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation, % = percentage
* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).
^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).
† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).
‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

5.8 Work Contributions From Each Plane By Joint Across Age-Groups

	G1: Age 20-39 (n=45)	G2: Age 40-54 (n=39)	G2 to G1	G3: Age 55-64 (n=32)	G3 to G1	G4: Age 65-75 (n=31)	G1 to G4	G2 to G3	G2 to G4	G3 to G4 to
All Genders	Mean ± SD	Mean ± SD	P-value	Mean ± SD	P-value	Mean ± SD	P-value	P-value	P-value	P-value
Hip Joint:										
% Sagittal Plane	57% ± 8%	59% ± 8%	0.531	56% ± 9%	0.750	57% ± 1%	0.947	0.386	0.526	0.815
% Frontal Plane	33% ± 7%	32% ± 7%	0.684	34% ± 7%	0.876	33% ± 6%	0.824	0.608	0.559	0.954
% Transverse Plane	10% ± 5%	9% ± 3%	0.619	10% ± 6%	0.742	10% ± 4%	0.835	0.447	0.802	0.624
Knee Joint:										
% Sagittal Plane	76% ± 6%	77% ± 7%	0.659	74% ± 6%	0.228	76% ± 5%	0.830	0.119	0.534	0.358
% Frontal Plane	13% ± 5%	14% ± 6%	0.256	15% ± 5%	0.120	15% ± 4%	0.236	0.619	0.911	0.712
% Transverse Plane	10% ± 3%	9% ± 3%**	**0.003**	11% ± 2%^	0.824	9% ± 3%	0.096	**0.016**	0.273	0.194
Ankle Joint:										
% Sagittal Plane	48% ± 1%	48% ± 1%*	**0.049**	48% ± 1%	0.713	48% ± 1%*	**0.020**	0.159	0.645	0.077
% Frontal Plane	48% ± 1%	48% ± 1%	0.774	48% ± 1%	0.227	48% ± 1%	0.452	0.155	0.642	0.074
% Transverse Plane	4% ± 1%	4% ± 1%	0.119	4% ± 1%^	0.567	4% ± 1%*†	**0.035**	**0.053**	0.525	**0.015**
Males										
Hip Joint:										
% Sagittal Plane	59% ± 9%	55% ± 8%	0.272	59% ± 11%	0.923	57% ± 9%	0.605	0.313	0.605	0.599
% Frontal Plane	34% ± 7%	36% ± 7%	0.216	31% ± 6%^	0.227	34% ± 6%	0.919	**0.031**	0.301	0.226
% Transverse Plane	8% ± 4%	8% ± 3%	0.692	10% ± 7%	0.114	9% ± 4%	0.372	0.235	0.628	0.466
Knee Joint:										
% Sagittal Plane	78% ± 6%	75% ± 7%	0.163	75% ± 3%	0.151	75% ± 5%	0.143	0.834	0.909	0.918
% Frontal Plane	13% ± 5%	15% ± 5%	0.063	14% ± 4%	0.292	15% ± 4%	0.103	0.574	0.872	0.685
% Transverse Plane	10% ± 3%	9% ± 3%	0.648	11% ± 2%	0.298	10% ± 3%	0.929	0.171	0.617	0.371
Ankle Joint:										
% Sagittal Plane	49% ± 2%	48% ± 1%*	**0.028**	48% ± 0%	0.479	48% ± 1%	0.025	0.234	0.906	0.206
% Frontal Plane	48% ± 2%	48% ± 1%	0.715	48% ± 0%	0.341	48% ± 1%	0.632	0.226	0.905	0.199
% Transverse Plane	3% ± 1%	4% ± 2%*	**0.016**	3% ± 1%^	0.826	4% ± 1%*†	**0.011**	**0.026**	0.823	**0.018**
Females										
Hip Joint:										
% Sagittal Plane	56% ± 8%	61% ± 7%*	**0.018**	54% ± 8%^^	0.493	57% ± 6%**	0.598	**0.003**	0.097	0.255
% Frontal Plane	32% ± 8%	28% ± 6%	0.073	36% ± 8%^^	0.113	33% ± 7%**	0.833	**0.001**	0.067	0.214
% Transverse Plane	12% ± 6%	10% ± 3%	0.279	10% ± 5%	0.208	10% ± 4%	0.266	0.850	0.905	0.955
Knee Joint:										
% Sagittal Plane	74% ± 5%	78% ± 7%	0.059	74% ± 7%^	0.784	77% ± 6%	0.313	**0.035**	0.463	0.216
% Frontal Plane	14% ± 5%	14% ± 6%	0.812	16% ± 6%	0.235	14% ± 4%	0.921	0.160	0.753	0.325
% Transverse Plane	12% ± 2%	8% ± 3%**	**0.000**	10% ± 3%^	0.114	9% ± 3%*	**0.016**	**0.031**	0.287	0.338
Ankle Joint:										
% Sagittal Plane	48% ± 1%	48% ± 1%	0.941	48% ± 1%	0.539	48% ± 1%	0.546	0.592	0.506	0.251
% Frontal Plane	48% ± 1%	48% ± 1%	0.941	48% ± 1%	0.539	48% ± 1%	0.546	0.592	0.506	0.251
% Transverse Plane	4% ± 1%	4% ± 1%	0.941	4% ± 1%	0.539	4% ± 1%	0.546	0.592	0.506	0.251

G1 = Group 1, G2 = Group 2, G3 = Group 3, G4 = Group 4, n = number, SD = Standard Deviation, % = percentage

* = significantly different than Age-Group 1 ($p \leq 0.05$), ** = significantly different than Age-Group 1 ($p \leq 0.01$).

^ = significantly different than Age-Group 2 ($p \leq 0.05$), ^^ = significantly different than Age-Group 2 ($p \leq 0.01$).

† = significantly different than Age-Group 3 ($p \leq 0.05$), †† = significantly different than Age-Group 3 ($p \leq 0.01$).

‡ = significantly different than Age-Group 4 ($p \leq 0.05$), ‡‡ = significantly different than Age-Group 4 ($p \leq 0.01$).

APPENDIX B

INFORMED CONSENT FORM
TARGETING PROTOCOL

<u>**INFORMED CONSENT**</u>
To Participate in a Research Study

Department of Kinesiology and Rehabilitation Science, University of Hawai‘i at Mānoa
1337 Lower Campus Road, PE/A Complex Rm. 231, Honolulu, HI 96822
Phone: 808-956-7606

I. INVESTIGATORS
Principal Investigators: Cris Stickley, PhD, ATC
Investigators: Elizabeth Parke, MS, ATC

II. TITLE
Biomechanical Analysis of Walking and Running Gait.

III. INTRODUCTION

The following information is being provided to help you decide if you would like to participate in this study. This form may have words that you do not understand. If you have questions, please ask us. The purpose of this study is to evaluate walking and running gait.

IV. DESCRIPTION OF PROCEDURES

You will be asked to report to the University of Hawai‘i at Mānoa Gait Lab (Sherriff 100) for a one-time data collection. When you arrive at the Gait Lab, you will be asked to perform two tasks: (1) walk for four meters at a self-selected speed, 10-16 times; and (2) run for four meters within 4 m/second speed, 10-16 times. The entire procedure will take approximately 45 minutes.

V. RISKS

Due to the low level of physical activity involved, the risk of injury is comparable to your routine activities of daily living. There is a very remote chance of cardiac arrest and/or death.

The investigators are NATABOC certified athletic trainers and First Aid/CPR/AED trained. In the event of any physical injury from the research, only immediate and essential medical treatment is available including an AED. First Aid/CPR and a referral to a medical emergency room will be provided. In the event of any emergency incidence outside the lab as a result of this research, contact your medical doctor and inform the principal investigator, Elizabeth Parke, MS, ATC at 336-402-3816 or Cris Stickley, PhD, ATC, at 513-259-4666. You should understand that if you are injured in the course of this research process that you alone will be responsible for the costs of treating your injuries.

VI. BENEFITS

You may not receive direct/immediate benefits. However, you will obtain information regarding your walking and running gait upon requests.

VII. COMPENSATION

No compensation will be given.

VIII. CONFIDENTIALITY

Your research records will be confidential to the extent permitted by law. Agencies with research oversight, such as The University of Hawaiʻi Committee on Human Studies, have the right to review research records.

An identification number will be used to identify you during the study, which will be known only to you and study personnel. In addition, all data and subject (identity) information will be kept under lock and key in the Department of Kinesiology and Rehabilitation Science at the University of Hawaiʻi at Mānoa. These materials will be permanently disposed of in a period not longer than 5 years. You will not be personally identified in any publication arising from this study. Personal information about your test results will not be given to anyone without your written permission.

IX. CERTIFICATION

I certify that I have read and I understand the foregoing, that I have been given satisfactory answers to my inquiries concerning the project procedures and other matters and that I have been advised that I am free to withdraw my consent participation and to discontinue participation in the project or activity at any time without prejudice.

I herewith consent to participate in this project with the understanding that such consent does not waive any of my legal rights, nor does it release the principal investigator or institution or any employee or agent thereof from liability for negligence.

I attest that I am not currently limited from full participation in my chosen sport due to injury.

I attest that I do not believe that I am currently pregnant.

If you have any questions related to this study, please contact any of the principal investigators: Elizabeth Parke, MS, ATC, at 336-402-3816 or Cris Stickley at 513-259-4666 at any time.

____

Participant's Printed Name

______________________________________ ______________

Signature of Participant Date

______________________________________ ______________

Witness Signature Date

If you cannot obtain satisfactory answers to your questions, or have complaints about your treatment in this study, please contact: Committee on Human Subjects, University of Hawai'i at Mānoa, **1960 East-West Rd., Biomed Bldg. Ste. B-104,** Honolulu, Hawaii 96822, Phone (808) 956-5007.

Anthropometric Data

Subject ID#: _______________ Date_________

Age________________ Gender: F / M

Subject's Dominate Leg: L / R

Vicon/Nexus Measurements

Weight (kg)	
Height (mm)	
Age (yrs)	
Left leg length (mm)	
Left knee width (mm)	
Left ankle width (mm)	
Right leg length (mm)	
Right knee width (mm)	
Right ankle width (mm)	

Data Collection Form

Subject ID#: _______________

Subject Dominant Leg: L / R

Condition: Non-targeting

Total Trials: 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20 21 22 23 24 25 26 27 28

Walking Trials		
Trial	Which foot hit the plate	Walking Pace
1	R / L	
2	R / L	
3	R / L	

Running Trials		
Trial	Which foot hit the plate	Running Pace
1	R / L	
2	R / L	
3	R / L	

Data Collection Form

Subject ID#: _______________

Subject Dominant Leg: L / R

Condition: Targeting

Total Trials: 1 2 3 4 5 6 7 8 9 10 11 12 13 14 15 16 17 18 19 20 21 22 23 24 25 26 27 28

Walking Trials		
Trial	Which foot hit the plate	Walking Pace
1	R / L	
2	R / L	
3	R / L	

Running Trials		
Trial	Which foot hit the plate	Running Pace
1	R / L	
2	R / L	
3	R / L	

APPENDIX C

INFORMED CONSENT FORM AND DATA COLLECTION FORMS FOR TKA STAIR NEGOTIATION PROTOCOL

RESEARCH SUBJECT INFORMATION AND CONSENT FORM

TITLE:	Biomechanical Comparison of Multi- and Single-Radius Implant Designs During Level Walking and Stair Climbing Tasks
PROTOCOL NO.:	2014-018 WIRB® Protocol #20141194
SPONSOR:	Cris Stickley, PhD, ATC
INVESTIGATOR:	Cass Nakasone, MD 888 South King Street Honolulu, Hawaii 96813 United States
SITE(S):	University of Hawaiʻi at Mānoa PE/A Complex Room 231, Lower Campus Road Honolulu, Hawaii 96822 United States Straub Clinic & Hospital 888 S. King Street Honolulu, Hawaii 96813 United States
STUDY-RELATED PHONE NUMBER(S):	Cass Nakasone, M.D. 808-522-4232 Cris Stickley PhD, ATC 808-956-3798

This consent form may contain words that you do not understand. Please ask the study doctor or the study staff to explain any words or information that you do not clearly understand. You may take home an unsigned copy of this consent form to think about or discuss with family or friends before making your decision.

SUMMARY

You are being asked to be in a research study. The purpose of this consent form is to help you decide if you want to be in the research study. Please read this consent form carefully. To be in a research study you must give your informed consent. "Informed consent" includes:

- Reading this consent form

- Having the study doctor or study staff explain the research study to you
- Asking questions about anything that is not clear, and
- Taking home an unsigned copy of this consent form. This gives you time to think about it and to talk to family or friends before you make your decision.

You should not join this research study until all of your questions are answered.

Things to know before deciding to take part in a research study:

- The main goal of a research study is to learn things to help patients in the future.
- The main goal of regular medical care is to help each patient.
- No one can promise that a research study will help you.
- Taking part in a research study is entirely voluntary. No one can make you take part.
- If you decide to take part, you can change your mind later on and withdraw from the research study.
- The decision to join or not join the research study will not cause you to lose any medical benefits. If you decide not to take part in this study, your doctor will continue to treat you.
- Parts of this study may involve standard medical care. Standard care is the treatment normally given for a certain condition or illness.
- After reading the consent form and having a discussion with the research staff, you should know which parts of the study are experimental (investigational) and which are standard medical care.
- Your medical records may become part of the research record. If that happens, your medical records may be looked at and/or copied by the sponsor of this study and government agencies or other groups associated with the study.

After reading and discussing the information in this consent form you should know:

- Why this research study is being done;
- What will happen during the research;
- Any possible benefits to you;
- The possible risks to you;
- How problems will be treated during the study and after the study is over.

If you take part in this research study, you will be given a copy of this signed and dated consent form.

PURPOSE OF THE STUDY

The purpose of this study is to compare the function of patients, implanted with either a multi-radii or a single radius total knee arthroplasty design, during level walking and stair climbing tasks. You are being asked to participate in this study because you are undergoing total knee arthroplasty. About 100 subjects are expected to participate.

PROCEDURES

If you decide to participate in this study, you will be randomly assigned (by chance) to one of four possible groups and receive either a single radius knee implant or one of three multiple radii knee implants. You have an equal chance of being assigned to any one of the four implant groups. The implants that will be used in this study are:

- GetAroundKnee™, Stryker Orthopedics (single radius)
- Balanced Knee® System, Ortho Development (multiple radii),
- Persona ™ Total Knee, Zimmer (multiple radii)
- NexGen®, Zimmer (multiple radii)

These types of implants are approved by the FDA for the type of surgery you are having and will be used according to their approved indication.

You will be asked to report to the University of Hawaiʻi at Mānoa, Kinesiology and Rehabilitation Science Laboratory (Gait Lab) (Sherriff 100) for all testing visits before and after your knee surgery.

Upon arrival to the Gait Lab, you will be asked to fill out one survey in reference to your current pain and activity level. Measurements about your body will be taken and you will be asked to perform the following tasks:
(1) walk for 6 meters at a comfortable speed 6-10 times (Gait Analysis),
(2) walking up and down stairs at a comfortable speed 3-4 times, and
(3) push into stationary objects (fixed dynamometer) with your leg for three seconds for two different leg movements (Isometric Strength).

You will also be asked some questions about your daily activities. The entire visit will take approximately 60 minutes.

You will be asked to go to the Gait Lab for your first study visit before your surgery. You will be asked to return to the Gait Lab 5 more times over the next two years to repeat the procedures listed above (please see Table 1 below for visit schedule). Each visit to the Gait lab will take approximately 60 minutes.

Table 1. Visit Time Line

	Before Surgery	6 Weeks After Surgery	3 Months After Surgery	6 Months After Surgery	1 Year After Surgery	2 Years After Surgery
Gait Analysis (test)	X	X	X	X	X	X
Isometric Strength	X	X	X	X	X	X

Paper/Pencil Survey	X	X	X	X	X	X

RISKS AND DISCOMFORTS
Being randomized to one type of knee implant instead of the others, may lead to greater or lesser stability of the knee post-surgery.

There are risks associated with your knee replacement surgery, whether or not you participate in this study. These include:

- Blood clots that can, in rare cases, be life threatening
- Complications after a blood transfusion
- Allergic reaction to the medications or materials used
- Infection
- Injury to arteries or nerves in your leg
- Surgery may not reduce your pain and stiffness, possibly requiring more treatment
- Surgery may cause more pain
- Risks of anesthesia

You will be asked to review and sign a separate consent form for your knee surgery, and your surgeon will explain the risks of the procedure in more detail.

Gait analysis risks
Due to the level of physical activity involved during the testing procedures, there is a risk of injury. You may have pain in your affected joint during testing. You may also have some discomfort, muscle cramping or soreness during or after test sessions. Although we have people to assist you and handrails in place, there is a chance of falling during the test. There is a very remote chance of cardiac arrest and/or death. These risks are comparable to your routine rehabilitation and activities of daily living, and will not affect your recovery from the surgery.

You cannot participate in this study if you are pregnant because the information collected during the walking test may not accurately represent your normal walking characteristics. If you are unaware that you are pregnant, participation in this study will result in no more danger to the mother or fetus than normal activities of daily living. However, if you become pregnant or think you might be pregnant during the course of this study, you must inform the researchers, and you will be removed from study participation.

NEW INFORMATION

You will be told about anything new that might change your decision to be in this study. You may be asked to sign a revised consent form if this occurs.

BENEFITS

You may not receive direct/immediate benefits from study participation. However, you will obtain information regarding your walking gait, functional activity capacity, hip muscular strength, and behavioral characteristics. Results of this study may assist physicians, physical therapists, and athletic trainers to ensure the optimal clinical outcomes to maintain the beneficial effects of total knee replacement.

PAYMENT FOR PARTICIPATION

You will not be paid for your participation in the study.

You will be given $5 that can be applied towards parking and/or transportation to the University of Hawaiʻi Gait Laboratory each time you come for a visit. The money will be given to you after you arrive at the facility with a receipt, so it is a reimbursement. You will be reimbursed only for the visits that you attend.

COSTS

You are not expected to have additional costs related to the procedures and visits that may result from your participation in this research study.

Any additional costs associated with parking/transportation over and above the $5 provided will be your responsibility. The fee for parking at the University of Hawaiʻi parking structure is $5 during the week and $6 on the weekends.

ALTERNATIVE TREATMENT

If you decide not to participate in this study, you will receive your knee replacement surgery with the type of implant that your doctor feels is best for you. Your follow-up care will be the same whether or not you are in this study.

USE AND DISCLOSURE OF YOUR HEALTH INFORMATION:

By signing this form, you are authorizing the use and disclosure of individually identifiable information. Your information will only be used/disclosed as described in this consent form and as permitted by state and federal laws. If you refuse to give permission, you will not be able to be in this research.

This consent covers all information about you that is used or collected for this study. It includes

- Past and present medical records
- Research records
- Records about your study visits.
- Information gathered for this research about:
    - Physical exams
    - Laboratory, x-ray, and other test results
    - Questionnaires
- Records about the implanted medical device.

Your authorization to use your identifiable health information will not expire even if you terminate your participation in this study or you are removed from this study by the study doctor. However, you may revoke your authorization to use your identifiable information at any time by submitting a written notification to the principal investigator, Cass Nakasone, MD at 888 S. King Street, Honolulu, HI 96813. If you decide to revoke (withdraw or "take back") your authorization, your identifiable health information collected or created for this study shall not be used or disclosed by the study doctor after the date of receipt of the written revocation except to the extent that the law allows us to continue using your information. The investigators in this study are not required to destroy or retrieve any of your health information that was created, used or disclosed for this study prior to receiving your written revocation.

By signing this consent form you authorize the following parties to use and or disclose your identifiable health information collected or created for this study:

- Cass Nakasone, MD and his research staff for the purposes of conducting this research study.
- Straub Clinic & Hospital and Hawai'i Pacific Health

Your medical records may contain information about AIDS or HIV infection, venereal disease, treatment for alcohol and/or drug abuse, or mental health or psychiatric services. By signing this consent form, you authorize access to this information if it is in the records used by members of the research team.

The individuals named above may disclose your medical records, this consent form and the information about you created by this study to:

- The sponsor of this study and their designees (if applicable)
- Federal, state and local agencies having oversight over this research, such as the Office for Human Research Protections in the U.S. Department of Health and Human Services, Food and Drug Administration, the National Institutes of Health, etc.
- The University of Hawai'i
- Hawaii Pacific Health (HPH) Officials, the Western Institutional Review Board, and the HPH Office of Compliance for purposes of overseeing the research study and making sure that your ethical rights are being protected.

Some of the persons or groups that receive your study information may not be required to comply with federal privacy regulations, and your information may lose its federal privacy protection and your information may be disclosed without your permission.

COMPENSATION FOR INJURY

In the event of any physical injury from the research, only immediate and essential medical treatment is available. First Aid/CPR and a referral to a medical emergency room will be provided. In the event of any emergency incidence outside the lab as a result of this research, contact your regular medical doctor and inform the study coordinator: Cris Stickley Ph.D., ATC, at 808-956-3798. You should understand that, if you are injured in the course of this research process, you or your medical insurance will be billed for the costs of treating your injuries.

VOLUNTARY PARTICIPATION AND WITHDRAWAL

Your participation in this study is voluntary. You may decide not to participate or you may leave the study at any time. Your decision will not result in any penalty or loss of benefits to which you are entitled.

Your participation in this study may be stopped at any time by the study doctor or the sponsor without your consent for any of the following reasons:

- it is in your best interest;
- you do not consent to continue in the study after being told of changes in the research that may affect you;
- you become pregnant;
- or for any other reason.

If you leave the study before the planned final visit, you may be asked by the study doctor to have some of the end of study procedures done.

SOURCE OF FUNDING FOR THE STUDY

This research study is sponsored by the University of Hawaiʻi at Mānoa.

QUESTIONS

Contact Cris Stickley Ph.D., ATC at 808-956-3798 or Dr. Cass Nakasone at 808-522-4232 for any of the following reasons:

- if you have any questions about this study or your part in it
- if you feel you have had a research-related injury or
- if you have questions, concerns or complaints about the research

If you have questions about your rights as a research subject or if you have questions, concerns, input, or complaints about the research, you may contact:

Western Institutional Review Board® (WIRB®)
1019 39th Avenue SE Suite 120
Puyallup, Washington 98374-2115
Telephone: 1-800-562-4789 or 360-252-2500
E-mail: Help@wirb.com.

WIRB is a group of people who perform independent review of research.
WIRB will not be able to answer some study-specific questions, such as questions about appointment times. However, you may contact WIRB if the research staff cannot be reached or if you wish to talk to someone other than the research staff.
Do not sign this consent form unless you have had a chance to ask questions and have gotten satisfactory answers.
If you agree to be in this study, you will receive a signed and dated copy of this consent form for

your records.

CONSENT

I have read this consent form. All my questions about the study and my part in it have been answered. I freely consent to be in this research study.

I authorize the use and disclosure of my health information to the parties listed in the authorization section of this consent for the purposes described above.

By signing this consent form, I have not given up any of my legal rights.

____
Subject Name (printed)

<u>CONSENT SIGNATURE:</u>

Signature of Subject	Date
Signature of Person Conducting Informed Consent Discussion	Date

Activity Assessment Survey

Subject ID#: _______________ Data Collection Period 0 1 2 3 4

Please circle the number that best describes current activity level.

1. Wholly inactive, dependent on others, and cannot leave residence
2. Mostly inactive or restricted to minimum activities of daily living
3. Sometimes participates in mild activities, such as walking, limited housework and limited shopping
4. Regularly participates in mild activities
5. Sometimes participates in moderate activities such as swimming or could do unlimited housework or shopping
6. Regularly participates in moderate activities
7. Regularly participates in active events such as bicycling
8. Regularly participates in active events, such as golf or bowling
9. Sometimes participates in impact sports such as jogging, tennis, skiing, acrobatics, ballet, heavy labor or backpacking
10. Regularly participates in impact sports

Please circle the number that best answers the following question. "How does your knee affect your ability to rise from a chair?":

1. "Because of my knee I cannot rise from a chair."
2. "Because of my knee, I can only rise from a chair if I use my hands and arms to assist."
3. "I have pain when rising from the seated position, but it does not affect my ability to rise from the seated position."
4. "My knee does not affect my ability to rise from a chair."

Are you satisfied with your partial knee replacement? YES or NO

Anthropometric Data

Subject ID#: _______________ Date_________

Age________________ Gender: F / M

Data Collection Period 0 1 2 3 4

Patient's Operated leg: L / R / B Dominant Leg: L / R

Date of Surgery_________________

Weeks after Surgery________________

Vicon/Nexus Measurements

Weight (kg)	
Height (mm)	
Age (yrs)	
Left leg length (mm)	
Left knee width (mm)	
Left ankle width (mm)	
Right leg length (mm)	
Right knee width (mm)	
Right ankle width (mm)	

Data Collection Form

Subject ID#: _______________Data Collection Period 0 1 2 3 4

Patient's Operated leg: L / R / B Dominant leg: L / R

Walking Trials		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	

Stair Ascent		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	
4	R / L	
5	R / L	

Stair Descent		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	
4	R / L	
5	R / L	

Manual Muscle Testing Data Collection

Subject ID#: _______________ Data Collection Period 0 1 2 3 4

Patient's Operated leg: L / R / B Dominant Leg: L / R

Tester: ______________________

	Left Leg						**Right Leg**					
	Trial 1 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trail 2 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 3 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 1 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 2 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 3 Score (ft-lb_f)	Pain Score (HHD/Jt)
Hip abduction		/		/		/		/		/		/
Knee extension		/		/		/		/		/		/

APPENDIX D

INFORMED CONSENT FORM AND DATA COLLECTION FORMS FOR UKA STAIR NEGOTIATION PROTOCOL

RESEARCH SUBJECT INFORMATION AND CONSENT FORM

TITLE:	Biomechanical Analysis of the Oxford® Unicompartmental Knee Implant Design During Level Walking and Stair Negotiation
PROTOCOL NO.:	2016-007
SPONSOR:	Cris Stickley, PhD, ATC Honolulu, Hawaii United States
INVESTIGATOR:	Cass Nakasone, M.D. 888 South King Street Honolulu, Hawaii 96813 United States
STUDY-RELATED PHONE NUMBER(S):	Cass Nakasone, M.D. 808-522-4000 Cris Stickley PhD, ATC 808-956-3798

This consent form may contain words that you do not understand. Please ask the study doctor or the study staff to explain any words or information that you do not clearly understand. You may take home an unsigned copy of this consent form to think about or discuss with family or friends before making your decision.

SUMMARY

You are being asked to be a participant in a research study. The purpose of this consent form is to help you decide if you want to be in the research study. Please read this consent form carefully. To be in a research study you must give your informed consent. "Informed consent" includes:

- Reading this consent form
- Having the study doctor or study staff explain the research study to you
- Asking questions about anything that is not clear, and
- Taking home an unsigned copy of this consent form. This gives you time to think about it and to talk to family or friends before you make your decision.

You should not join this research study until all of your questions are answered.

Things to know before deciding to take part in a research study:

- The main goal of a research study is to learn things to help patients in the future.
- The main goal of regular medical care is to help each patient.
- No one can promise that a research study will help you.
- Taking part in a research study is entirely voluntary. No one can make you take part.
- If you decide to take part, you can change your mind later on and withdraw from the research study.

- The decision to join or not join the research study will not cause you to lose any medical benefits. If you decide not to take part in this study, your doctor will continue to treat you.
- Parts of this study may involve standard medical care. Standard care is the treatment normally given for a certain condition or illness.
- After reading the consent form and having a discussion with the research staff, you should know which parts of the study are experimental (investigational) and which are standard medical care.
- Your medical records may become part of the research record. If that happens, your medical records may be looked at and/or copied by the sponsor of this study and government agencies or other groups associated with the study.

After reading and discussing the information in this consent form you should know:

- Why this research study is being done;
- What will happen during the research;
- Any possible benefits to you;
- The possible risks to you;
- How problems will be treated during the study and after the study is over.

If you take part in this research study, you will be given a copy of this signed and dated consent form.

PURPOSE OF THE STUDY

The purpose of this study is to compare the function of patients with the Oxford partial knee implant design during level walking and stair negotiation tasks.

Approximately 20 people will participate in this study.

PROCEDURES

If you decide to participate in this study you will be receiving per the physician's protocol the Oxford partial knee implant which is approved by the FDA for the type of surgery you are having and will be used according to their approved indication.

You will be asked to report to the University of Hawaiʻi at Mānoa, Kinesiology and Rehabilitation Science Laboratory (Gait Lab) (Sherriff 100) for all testing before and after your knee surgery.

Upon arrival to the Gait Lab, you will be asked to fill out one survey in reference to your current pain and activity level.

When you arrive at the Gait Lab measurements about your body will be taken and you will be asked to perform the following tasks:
(1) walk for 6 meters at a comfortable speed 6-10 times (Gait Analysis),
(2) walking up and down stairs at a comfortable speed 3-4 times, and

(3) push into stationary objects (fixed dynamometer) with your leg for three seconds for two different leg movements (Isometric Strength).
You will also be asked some questions about your daily activities. The entire visit will take approximately 60 minutes.

You will be asked to go to the Gait Lab for your first study visit before your surgery. Each visit to the Gait lab will take approximately 60 minutes. You will be asked to return to the Gait Lab four more times over the next one year to repeat the procedures listed above (please see Table 1 below for visit schedule).

Table 1. Visit Time Line

	Before Surgery	6 Weeks After Surgery	3 Months After Surgery	6 Months After Surgery	1 Year After Surgery
Gait Analysis (test)	X	X	X	X	X
Isometric Strength	X	X	X	X	X
Survey	X	X	X	X	X

RISKS AND DISCOMFORTS

There are risks associated with your knee replacement surgery. These include:
- Blood clots that can, in rare cases, be life threatening
- Complications after a blood transfusion
- Allergic reaction to the medications or materials used
- Injury to arteries in your leg
- Surgery may not reduce your pain and stiffness, possibly requiring more treatment
- Surgery may cause more pain

Due to the level of physical activity involved, there is a risk of injury. You may have pain in your affected joint during testing. You may also have some discomfort, muscle cramping or soreness during or after test sessions. Although we have people to assist you and handrails in place, there is a chance of falling during the test. There is a very remote chance of cardiac arrest and/or death. These risks are comparable to your routine rehabilitation and activities of daily living, and will not affect your recovery from the surgery.

You cannot participate in this study if you are pregnant because the information collected during the walking test may not accurately represent your normal walking characteristics. If you are unaware that you are pregnant, participation in this study will result in no more danger to the mother or fetus than normal activities of daily living. However, if you become pregnant or think you might be pregnant during the course of this study, you must inform the researchers, and you will be excluded from study participation.

NEW INFORMATION

You will be told about anything new that might change your decision to be in this study. You may be asked to sign a revised consent form if this occurs.

BENEFITS

You may not receive direct/immediate benefits. However, you will obtain information regarding your walking gait, functional activity capacity, hip muscular strength, and behavioral characteristics. Results of this study may assist physicians, physical therapists, and athletic trainers to ensure the optimal clinical outcomes to maintain the beneficial effects of knee replacement.

PAYMENT FOR PARTICIPATION

You will be given $5 that can be applied towards parking and/or transportation to the University of Hawaiʻi Gait Laboratory each time you come for a visit. The money will be given to you after you arrive to the facility so it is a reimbursement. If you do not finish the study, you will be paid only for the visits you have completed.

COSTS

There are no additional costs related to the procedures and visits that may result from your participation in this research study.

Any costs associated with parking/transportation over and above the $5 provided will be your responsibility. The fee for parking at the University of Hawaiʻi parking structure is $5 during the week and $6 on the weekends.

ALTERNATIVE TREATMENT

Your alternative is not to participate in this study.

USE AND DISCLOSURE OF YOUR HEALTH INFORMATION:

By signing this form, you are authorizing the use and disclosure of individually identifiable information. Your information will only be used/disclosed as described in this consent form and as permitted by state and federal laws. If you refuse to give permission, you will not be able to be in this research.

This consent covers all information about you that is used or collected for this study. It includes

- Past and present medical records
- Research records
- Records about your study visits.
- Information gathered for this research about:
    - Physical exams

Laboratory, x-ray, and other test results
Questionnaires

- Records about the implanted medical device.

Your authorization to use your identifiable health information will not expire even if you terminate your participation in this study or you are removed from this study by the study doctor. However, you may revoke your authorization to use your identifiable information at any time by submitting a written notification to the principal investigator, Cass Nakasone, MD at 888 S. King Street, Honolulu, HI 96813. If you decide to revoke (withdraw or "take back") your authorization, your identifiable health information collected or created for this study shall not be used or disclosed by the study doctor after the date of receipt of the written revocation except to the extent that the law allows us to continue using your information. The investigators in this study are not required to destroy or retrieve any of your health information that was created, used or disclosed for this study prior to receiving your written revocation.

By signing this consent form you authorize the following parties to use and or disclose your identifiable health information collected or created for this study:

- Cass Nakasone, MD and his research staff for the purposes of conducting this research study.
- Straub Medical Center and Hawaiʻi Pacific Health
- The University of Hawaiʻi

Your medical records may contain information about AIDS or HIV infection, venereal disease, treatment for alcohol and/or drug abuse, or mental health or psychiatric services. By signing this consent form, you authorize access to this information if it is in the records used by members of the research team.

The individuals named above may disclose your medical records, this consent form and the information about you created by this study to:

- The sponsor of this study and their designees (if applicable)
- Federal, state and local agencies having oversight over this research, such as the Office for Human Research Protections in the U.S. Department of Health and Human Services, Food and Drug Administration, the National Institutes of Health, etc.
- The University of Hawaiʻi
- Hawaii Pacific Health (HPH) Officials, the Western Institutional Review Board, and the HPH Office of Compliance for purposes of overseeing the research study and making sure that your ethical rights are being protected.

Some of the persons or groups that receive your study information may not be required to comply with federal privacy regulations, and your information may lose its federal privacy protection and your information may be disclosed without your permission.

COMPENSATION FOR INJURY

In the event of any physical injury from the research, only immediate and essential medical treatment is available. First Aid/CPR and a referral to a medical emergency room will be provided. In the event of any emergency incidence outside the lab as a result of this research, contact your medical doctor and inform the study coordinator: Cris Stickley Ph.D., ATC, at 808-956-3798. You should understand that if you are injured in the course of this research process that you or your medical insurance will be billed for the costs of treating your injuries.

VOLUNTARY PARTICIPATION AND WITHDRAWAL

Your participation in this study is voluntary. You may decide not to participate or you may leave the study at any time. Your decision will not result in any penalty or loss of benefits to which you are entitled.

Your participation in this study may be stopped at any time by the study doctor or the sponsor without your consent for any of the following reasons:

- it is in your best interest;
- you do not consent to continue in the study after being told of changes in the research that may affect you;
- you become pregnant;
- or for any other reason.

If you leave the study before the planned final visit, you may be asked by the study doctor to have some of the end of study procedures done.

SOURCE OF FUNDING FOR THE STUDY

This research study is sponsored by the University of Hawaiʻi at Mānoa.

QUESTIONS

Contact Cris Stickley Ph.D., ATC at 808-956-3798 or Dr. Cass Nakasone at 808-522-4232 for any of the following reasons:

- if you have any questions about this study or your part in it
- if you feel you have had a research-related injury or
- if you have questions, concerns or complaints about the research

If you have questions about your rights as a research subject or if you have questions, concerns or complaints about the research, you may contact:

Western Institutional Review Board® (WIRB®)
1019 39th Avenue SE Suite 120
Puyallup, WA 98374-2115
Telephone: 1-800-562-4789 or 360-252-2500
E-mail: Help@wirb.com.

WIRB is a group of people who perform independent review of research.
WIRB will not be able to answer some study-specific questions, such as questions about appointment times. However, you may contact WIRB if the research staff cannot be reached or if you wish to talk to someone other than the research staff.

Do not sign this consent form unless you have had a chance to ask questions and have gotten satisfactory answers.
If you agree to be in this study, you will receive a signed and dated copy of this consent form for your records.

CONSENT

I have read this consent form. All my questions about the study and my part in it have been answered. I freely consent to be in this research study.

I authorize the use and disclosure of my health information to the parties listed in the authorization section of this consent for the purposes described above.

By signing this consent form, I have not given up any of my legal rights.

____
Subject Name (printed)

CONSENT SIGNATURE:

____ ________________
Signature of Subject Date

____ ________________
Signature of Person Conducting Informed Consent Discussion Date

Activity Assessment Survey

Subject ID#: _______________ Data Collection Period 0 1 2 3 4

Please circle the number that best describes current activity level.

11. Wholly inactive, dependent on others, and cannot leave residence
12. Mostly inactive or restricted to minimum activities of daily living
13. Sometimes participates in mild activities, such as walking, limited housework and limited shopping
14. Regularly participates in mild activities
15. Sometimes participates in moderate activities such as swimming or could do unlimited housework or shopping
16. Regularly participates in moderate activities
17. Regularly participates in active events such as bicycling
18. Regularly participates in active events, such as golf or bowling
19. Sometimes participates in impact sports such as jogging, tennis, skiing, acrobatics, ballet, heavy labor or backpacking
20. Regularly participates in impact sports

Please circle the number that best answers the following question. "How does your knee affect your ability to rise from a chair?":

5. "Because of my knee I cannot rise from a chair."
6. "Because of my knee, I can only rise from a chair if I use my hands and arms to assist."
7. "I have pain when rising from the seated position, but it does not affect my ability to rise from the seated position."
8. "My knee does not affect my ability to rise from a chair."

Are you satisfied with your partial knee replacement? YES or NO

Anthropometric Data

Subject ID#: _______________ Date_________

Age________________ Gender: F / M

Data Collection Period 0 1 2 3 4

Patient's Operated leg: L / R / B Dominant Leg: L / R

Date of Surgery_________________

Weeks after Surgery________________

Vicon/Nexus Measurements

Weight (kg)	
Height (mm)	
Age (yrs)	
Left leg length (mm)	
Left knee width (mm)	
Left ankle width (mm)	
Right leg length (mm)	
Right knee width (mm)	
Right ankle width (mm)	

Data Collection Form

Subject ID#: _______________Data Collection Period 0 1 2 3 4

Patient's Operated leg: L / R / B Dominant leg: L / R

Walking Trials		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	

Stair Ascent		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	
4	R / L	
5	R / L	

Stair Descent		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	
4	R / L	
5	R / L	

Manual Muscle Testing Data Collection

Subject ID#: _______________ Data Collection Period 0 1 2 3 4

Patient's Operated leg: L / R / B Dominant Leg: L / R

Tester: ______________________

	Left Leg						Right Leg					
	Trial 1 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trail 2 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 3 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 1 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 2 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 3 Score (ft-lb_f)	Pain Score (HHD/Jt)
Hip abduction		/		/		/		/		/		/
Knee extension		/		/		/		/		/		/

APPENDIX E

INFORMED CONSENT FORM AND DATA COLLECTION FORMS FOR ACROSS LIFESPAN PROTOCOL

RESEARCH SUBJECT INFORMATION AND CONSENT FORM

TITLE: Biomechanical Analysis During Level Walking and Stair Climbing Tasks in a Healthy Control Population

INVESTIGATOR: Cris Stickley, PhD, ATC
Honolulu, Hawaii
United States

SITE(S): University of Hawaii at Mānoa
PE/A Complex Room 231
1337 Lower Campus Road
Honolulu, Hawaii 96822
United States

STUDY-RELATED PHONE NUMBER(S): Cris Stickley PhD, ATC
808-956-3798

This consent form may contain words that you do not understand. Please ask the study doctor or the study staff to explain any words or information that you do not clearly understand. You may take home an unsigned copy of this consent form to think about or discuss with family or friends before making your decision.

SUMMARY

You are being asked to be a participant in a research study. The purpose of this consent form is to help you decide if you want to be in the research study. Please read this consent form carefully. To be in a research study you must give your informed consent. "Informed consent" includes:

- Reading this consent form
- Having the study staff explain the research study to you
- Asking questions about anything that is not clear, and
- Taking home an unsigned copy of this consent form. This gives you time to think about it and to talk to family or friends before you make your decision.

You should not join this research study until all of your questions are answered.

Things to know before deciding to take part in a research study:

- The main goal of a research study is to learn things about walking gait and stair climbing in a healthy population.
- No one can promise that a research study will help you.
- Taking part in a research study is entirely voluntary. No one can make you take part.
- If you decide to take part, you can change your mind later on and withdraw from the research study.

- Parts of this study may involve standard medical care. Standard care is the treatment normally given for a certain condition or illness.
- After reading the consent form and having a discussion with the research staff, you should know which parts of the study are experimental (investigational) and which are standard medical care.

After reading and discussing the information in this consent form you should know:

- Why this research study is being done;
- What will happen during the research;
- Any possible benefits to you;
- The possible risks to you;
- How problems will be treated during the study and after the study is over.

If you take part in this research study, you will be given a copy of this signed and dated consent form.

PURPOSE OF THE STUDY

The purpose of this study is to compare biomechanical variables during level walking and stair climbing tasks in a healthy control population.

PROCEDURES

You will be asked to report to the University of Hawai‘i at Mānoa, Kinesiology and Rehabilitation Science Laboratory (Gait Lab) (Sherriff 100) for a one-time data collection.

Upon arrival to the Gait Lab, you will be asked to fill out a health questionnaire.

When you arrive at the Gait Lab measurements about your body will be taken and you will be asked to perform the following tasks:
(1) walk for 6 meters at a comfortable speed 6-10 times (Gait Analysis),
(2) walking up and down stairs at a comfortable speed 3-5 times, and
(3) push into stationary objects (fixed dynamometer) with your leg for three seconds for two different leg movements (Isometric Strength).

The entire visit will take approximately 60 minutes.

RISKS AND DISCOMFORTS

Due to the level of physical activity involved, there is a risk of injury. You may also have some discomfort, muscle cramping or soreness during or after test sessions. Although we have people to assist you and handrails in place, there is a chance of falling during the test. There is a very remote chance of cardiac arrest and/or death. These risks are comparable to your routine rehabilitation and activities of daily living, and will not affect your recovery from the surgery.

You cannot participate in this study if you are pregnant because the information collected during the walking test may not accurately represent your normal walking characteristics. If you are unaware that you are pregnant, participation in this study will result in no more danger to the mother or fetus than normal activities of daily living. However, if you think you might be pregnant during the course of this data collection, you must inform the researchers, and you will be excluded from study participation.

NEW INFORMATION

You will be told about anything new that might change your decision to be in this study. You may be asked to sign a revised consent form if this occurs.

BENEFITS

You may not receive direct/immediate benefits. However, you will obtain information regarding your walking gait, functional activity capacity, hip muscular strength, and behavioral characteristics.

PAYMENT FOR PARTICIPATION

You will be given $5 that can be applied towards parking and/or transportation to the University of Hawaiʻi Gait Laboratory each time you come for a visit. The money will be given to you after you arrive to the facility so it is a reimbursement. If you do not finish the study, you will be paid only for the visits you have completed.

COSTS

There are no additional costs related to the procedures and visits that may result from your participation in this research study.

Any additional costs associated with parking/transportation over and above the $5 provided will be your responsibility. The fee for parking at the University of Hawaiʻi parking structure is $5 during the week and $6 on the weekends.

USE AND DISCLOSURE OF YOUR HEALTH INFORMATION:

By signing this form, you are authorizing the use and disclosure of individually identifiable information. Your information will only be used/disclosed as described in this consent form and as permitted by state and federal laws. If you refuse to give permission, you will not be able to be in this research.

This consent covers all information about you that is used or collected for this study. It includes

- Research records
- Records about your study visit

Your authorization to use your identifiable health information will not expire even if you terminate your participation in this study or you are removed from this study by the study doctor. However, you may revoke your authorization to use your identifiable information at any time by submitting a written notification to the principal investigator, Dr. Cris Stickely University of Hawaiʻi at Mānoa, Honolulu, HI 96813. If you decide to revoke (withdraw or "take back") your authorization, your identifiable health information collected or created for this study shall not be used or disclosed by the study staff after the date of receipt of the written revocation except to the extent that the law allows us to continue using your information. The investigators in this study are not required to destroy or retrieve any of your health information that was created, used or disclosed for this study prior to receiving your written revocation.

By signing this consent form you authorize the following parties to use and or disclose your identifiable health information collected or created for this study:

- Cris Stickley and his research staff for the purposes of conducting this research study.
- University of Hawaiʻi at Mānoa.

The individuals named above may disclose this consent form and the information about you created by this study to:

- The sponsor of this study and their designees (if applicable)
- Federal, state and local agencies having oversight over this research, such as the Office for Human Research Protections in the U.S. Department of Health and Human Services, Food and Drug Administration, the National Institutes of Health, etc.
- The University of Hawaiʻi for purposes of overseeing the research study and making sure that your ethical rights are being protected.

Some of the persons or groups that receive your study information may not be required to comply with federal privacy regulations, and your information may lose its federal privacy protection and your information may be disclosed without your permission.

COMPENSATION FOR INJURY

In the event of any physical injury from the research, only immediate and essential medical treatment is available. First Aid/CPR and a referral to a medical emergency room will be provided. In the event of any emergency incidence outside the lab as a result of this research, contact your medical doctor and inform the study coordinator: Cris Stickley Ph.D., ATC, at 808-956-3798. You should understand that if you are injured in the course of this research process that you or your medical insurance will be billed for the costs of treating your injuries.

VOLUNTARY PARTICIPATION AND WITHDRAWAL

Your participation in this study is voluntary. You may decide not to participate or you may leave the study at any time. Your decision will not result in any penalty or loss of benefits to which you are entitled.

Your participation in this study may be stopped at any time by the study staff or the sponsor without your consent for any of the following reasons:

- it is in your best interest;

- you do not consent to continue in the study after being told of changes in the research that may affect you;
- you become pregnant;
- or for any other reason.

SOURCE OF FUNDING FOR THE STUDY

This research study is sponsored by the University of Hawaiʻi, Mānoa.

QUESTIONS

Contact Cris Stickley Ph.D., ATC at 808-956-3798 for any of the following reasons:

- if you have any questions about this study or your part in it
- if you feel you have had a research-related injury or
- if you have questions, concerns or complaints about the research

Do not sign this consent form unless you have had a chance to ask questions and have gotten satisfactory answers.

If you agree to be in this study, you will receive a signed and dated copy of this consent form for your research.

CONSENT

I have read this consent form. All my questions about the study and my part in it have been answered. I freely consent to be in this research study.

I authorize the use and disclosure of my health information to the parties listed in the authorization section of this consent for the purposes described above.

By signing this consent form, I have not given up any of my legal rights.

____
Subject Name (printed)

<u>CONSENT SIGNATURE:</u>

Signature of Subject	Date
Signature of Person Conducting Informed Consent Discussion	Date

**Biomechanical Analysis During Level Walking and
Tasks in a Healthy Control Population**

ID #: ______________________________ **DATE**: ____________________

GENDER: M / F **AGE:** ____________________

Participant Health Questionnaire:			
1	Has your doctor ever said that you have a heart condition and that you should only perform physical activity recommended by a doctor?	**YES**	**NO**
2	Do you feel pain in your chest when you perform physical activity?	**YES**	**NO**
3	In the past month, have you had chest pain when you were not exercising?	**YES**	**NO**
4	Do you lose your balance because of dizziness?	**YES**	**NO**
5	Have you ever been diagnosed with Parkinson's Disease?	**YES**	**NO**
6	Do you have a history of fainting?	**YES**	**NO**
7	Have you ever been diagnosed with a neurological disorder?	**YES**	**NO**
8	Do you have diabetes mellitus?	**YES**	**NO**
9	Do you have a bone or joint problem that could be made worse by physical activity?	**YES**	**NO**
10	Has a doctor ever diagnosed you with rheumatoid arthritis or osteoarthritis?	**YES**	**NO**
11	Within the last year, have you experienced an injury to your knee or experience any severe knee pain?	**YES**	**NO**
12	Have you had a previous hip, knee, ankle or foot surgery?	**YES**	**NO**

Activity Assessment Survey

Subject ID#: _______________ Age: __________

1. Wholly inactive, dependent on others, and cannot leave residence
2. Mostly inactive or restricted to minimum activities of daily living
3. Sometimes participates in mild activities, such as walking, limited housework and limited shopping
4. Regularly participates in mild activities
5. Sometimes participates in moderate activities such as swimming or could do unlimited housework or shopping
6. Regularly participates in moderate activities
7. Regularly participates in active events such as bicycling
8. Regularly participates in active events, such as golf or bowling
9. Sometimes participates in impact sports such as jogging, tennis, skiing, acrobatics, ballet, heavy labor or backpacking
10. Regularly participates in impact sports

Please use the above scale to circle the most appropriate response that describes your activity level:

	INACTIVE---VERY ACTIVE									
CURRENTLY	1	2	3	4	5	6	7	8	9	10
PRIOR TO HIGH SCHOOL	1	2	3	4	5	6	7	8	9	10
DURING HIGH SCHOOL	1	2	3	4	5	6	7	8	9	10
DURING AGES 20-29	1	2	3	4	5	6	7	8	9	10
DURING AGES 30-39	1	2	3	4	5	6	7	8	9	10
DURING AGES 40-49	1	2	3	4	5	6	7	8	9	10
DURING AGES 50-59	1	2	3	4	5	6	7	8	9	10
DURING AGES 60-69	1	2	3	4	5	6	7	8	9	10

Anthropometric Data

Subject ID#: _______________ Date_________

Age________________ Gender: F / M

Dominant Leg: L / R

Vicon/Nexus Measurements

Weight (kg)	
Height (mm)	
Age (yrs)	
Left leg length (mm)	
Left knee width (mm)	
Left ankle width (mm)	
Right leg length (mm)	
Right knee width (mm)	
Right ankle width (mm)	

Data Collection Form

Subject ID#: ________________________ Date:______________

Dominant leg: L / R

Walking Trials-Self Selected		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	

Walking Trials-Increased Velocity		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	

Walking Trials-Decreased Velocity		
Trial	Which foot hit the plate	Walking Pace (s)
1	R / L	
2	R / L	
3	R / L	

Stair Ascent	
Trial	Which foot hit the plate
1	R / L
2	R / L
3	R / L
4	R / L
5	R / L

Stair Descent	
Trial	Which foot hit the plate
1	R / L
2	R / L
3	R / L
4	R / L
5	R / L

Manual Muscle Testing Data Collection

Subject ID#: ________________________________ Date:_________________

Dominant Leg: L / R

Tester: _______________________

	Left Leg						Right Leg					
	Trial 1 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trail 2 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 3 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 1 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 2 Score (ft-lb_f)	Pain Score (HHD/Jt)	Trial 3 Score (ft-lb_f)	Pain Score (HHD/Jt)
Hip abduction		/		/		/		/		/		/
Knee extension		/		/		/		/		/		/

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