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Effects of Cycling Workload and Cadence on Frontal Plane Knee Load

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Effects of Cycling Workload and Cadence on Frontal Plane Knee Load

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ABSTRACT

The effect of workload and cadence on sagittal plane knee biomechanics in cycling has been widely studied, and few studies have focused on the frontal plane. Frontal plane knee biomechanics, especially knee abduction moment, is closely related to the severity and progression of knee osteoarthritis. The purpose of this study was to investigate effects of different workload and cadence on knee frontal plane kinematics and kinetics. Eighteen subjects participated in this study. A motion analysis system was used to collect 5 cycles of kinematics during 2 minutes of cycling in 8 testing conditions, which included five workload conditions of 0.5, 1, 1.5, 2, and 2.5 kg at 60 revolution per minute (RPM), and four cadence conditions of 60, 70, 80, and 90 RPM with 1 kg workload. A custom instrumented pedal was used to collect pedal reaction force (PRF). Increased workloads significantly increased knee abduction moment and knee abduction range of motion (ROM), without any change of peak knee adduction angle. Increased workloads also significantly increased medial, posterior, and vertical pedal PRF, and knee extension moment. Increased cadences had no effects on knee abduction moment. In addition, increased cadences increased anterior and vertical PRF, and knee flexion moment. We found two patterns of frontal knee moments among our subjects which deserves further investigation. Further study may be needed to demonstrate the efficacy of appropriate level of workload in the knee osteoarthritis and other deceased populations.

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CHAPTER I

INTRODUCTION

Cycling is a popular recreational activity which can provide health benefits, improve cardiovascular fitness, and decrease cancer morbidity (36). It is also a recommended exercise for individuals who have lower extremity injuries, because it allows people to work on muscle strength and lower extremity mobility while exert a smaller load on lower extremity joints compared to walking or jogging (28-30, 32, 40). Despite this low impact, the repetitive nature has made cycling a risk to overuse injuries, and the knee joint is the most affected site (2, 5).

In cycling, the majority of power and driving force come from knee extension and flexion (19). Knee movement in the sagittal plane has been widely studied, and knee moments and forces can vary due to different seat heights, cadences, and workloads used in different studies. A knee range of motion (ROM) in the sagittal plane of 66 to 67.5 degrees has consistently been found among studies, with only small variations due to the differences in seat height (3, 14, 44). Nepture and Hull (35) found intersegment knee torques to be about 30 Nm during extension and about 30 Nm during flexion when cycling at 90 revolution per minute (RPM) and 225 Watts (W). Gregor (26) found the peak knee extension moment to be 53 Nm and the peak knee flexion moment to be 34 Nm when the subjects cycled at 60 RPM at about 160 W. Tamborindeguy et al. (42) found the peak anterior tibiofemoral shear force was around 80 N and the peak tibiofemoral compressive force was around 700 N when subjects cycled at 70 RPM and 70 W.

Frontal plane knee movements exist in cycling. During the power phase, the knee adducts as it extends. This motion leads to medial translation of the knee while the knee extends (3). There are a limited number of studies examining the frontal plane knee moments in cycling and their results varies (16, 24, 25, 39). Among studies, the knee external varus (abduction) moment

ranged from 8.1 Nm to 15.3 Nm while the knee external valgus moment ranged from 2.2 Nm to 11.2 Nm. These differences can be attributed to the facts that the equations derived to calculate knee moment were not the same among studies (25, 39). In addition, different studies used subjects with different levels of cycling experiences, e.g., recreational cyclists (16), competitive cyclists (25), both recreational and competitive cyclists (39), and knee osteoarthritis (OA) patients and healthy subjects (24).

The knee moment in the mediolateral direction is important in studying some injuries (3). For example, patellofemoral syndrome, which is known as the “cyclist’s knee”, is thought to be caused by an internal abduction moment during the downward pedal stroke (8, 48). Iliotibial band syndrome, another common cycling injury, occurs at the lateral side of the knee and is often caused by the repetition of knee flexion (9, 12, 46). In addition, some studies pointed out the non-driven knee moments, which are the varus/valgus and internal/external axial moments, are substantial and they are important in understanding cycling overuse injuries (25, 39, 45).

Despite the injury potentials especially at competitive levels, cycling is recommended as an exercise for individuals with OA (30). A training study reported that after 10 weeks of stationary cycling, knee OA patients showed significant improvement in timed chair rise, 6-minute walk distance, walking speeds, amount of overall pain relief, and aerobic capacity (31). Kutzner et al. (30) showed that peak knee resultant contact force was ranging from 0.5 to 1.63 times body weight (BW) in cycling compared to 2.52 times BW in walking. The increased internal knee abduction moment (KAM), which reflects the loading to the medial compartment of the knee, has been shown to be associated with medial knee OA. Subjects with medial knee OA have been found to walk with greater than normal knee adduction moment (4), and KAM can also predict OA progression (33). However, there are limited studies relating KAM to

cycling. In a recent study (24), healthy subjects and knee OA patients cycled at 60 RPM and 80 W, and KAM was calculated when the subjects cycled in toe-in foot positions (5 and 10 degrees) and in everted positions with lateral wedges placed on the pedal (5 and 10 degrees), in a neutral foot position. The results showed that the 10-degree wedge caused significantly smaller KAM compared to the neutral condition.

Cycling workload and cadence are two variables that can influence the pedal reaction force, and further affect knee load. A number of research studies have examined the effect of workload and cadence on knee angle, moment, force, and work (6, 7, 11, 13-15, 17, 18, 30, 34, 38). However, all of them have focused only on the sagittal plane. It has been shown by most studies that neither workload nor cadence changes knee ROM or peak knee angles (7, 13, 14). For the knee kinetics, increasing workload has been found to increase knee moment, force, and work (7, 11, 15, 18, 30). However, increasing cadence does not seem to affect peak knee contact forces, which has been supported by results from inverse dynamics using an instrumented pedal (6, 18) and contact force measured using an instrumented implant (11). The effects of cadence on peak knee moments are varied (18, 34, 38). Some studies concluded that changes in cadence did not affect the magnitude of knee moment (17, 38) while one study reported that the knee net moment was decreased with increased cadence (34). The discrepancy in results may be partially related to different bicycle types of ergometers used in different studies. For example, if the bike uses an electromagnetically braked system, the resistance force decreases as the cadence increases in order to maintain constant workload which may explain the decreasing knee moment.

STATEMENT OF PROBLEM

No cycling studies have examined influences of workload and cadence on frontal plane knee kinematics and kinetics, and only a limited number of studies have reported frontal plane

knee kinematic and kinetic data (16, 24, 25, 39). It was important to study effects of the workload and cadence on frontal plane knee variables, especially KAM, to provide research evidence for prescribing cycling as a therapy for knee OA patients. In addition, most of the existing studies have used young healthy male subject or patients (3, 6, 7, 11, 13-18, 34), while cycling data in middle-aged and old populations are necessary. Furthermore, most knee OA patients are middle-aged and old adults.

Therefore, the purpose of this study was to examine the effect of changing cycling workload and cadence on knee frontal plane biomechanics in middle-aged and old adults.

HYPOTHESIS

1. Increasing cycling workload will increase peak knee abduction moment and peak knee adduction angle.
2. Increasing cadence will not change peak knee abduction moment or peak knee adduction angle.

DELIMITATIONS

1. Subjects should be men and women between the age of 40 and 79.
2. Subjects should be free from lower extremity injuries from the past six months.
3. Subjects should be able to ride a stationary bike without any assistance for sixteen minutes.
4. Kinematics was collected using a motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) and pedal force will be collected using a customized bike pedal instrumented with two 3D force sensors (1200 Hz, Type 9027C, Kistler, Switzerland).

LIMITATIONS

1. All tests were conducted in a laboratory setting.
2. Pedal reaction forces were collected on the left pedal only.

3. The accuracy of the results was limited by the accuracy of the instruments used in the study; and the accuracy of estimating joint centers was limited by the accuracy of placements of the anatomical markers.

CHAPTER II

LITERATURE REVIEW

The purpose of this study was to examine effects of different cycling workloads and cadences on knee frontal plane biomechanics in middle-aged and old adults. This literature review includes background about cycling, cycling biomechanics, and the influence of cycling workload and cadence on knee kinetics and kinematics.

BACKGROUND ABOUT CYCLING

Cycling as a recreational activity and as a rehabilitation intervention has been the focus of a great deal of research. According to a recent review, cycling has been found to improve cardiovascular fitness, gain health benefits, and decrease cancer morbidity (36). Cycling also allows people to work on lower extremity range of motion and strength while minimizing stress on joints (28, 40). Thus, cycling is a recommended exercise for individuals with physical disabilities, like people who suffer from OA, anterior cruciate ligament (ACL) injury, stroke, etc. (29, 32, 40).

Despite these benefits, cycling is associated with a high incidence of overuse injuries with the knee joint being the most affected site (10, 12). These injuries are closely related to the load being generated at the cyclists' knees. The magnitude of the load to the knee joint during cycling can be affected by many factors, such as the seat position, foot position, workload, and pedal cadence (5, 9, 24, 46). The biomechanics analysis in the sagittal plane has been widely studied in cycling, however, discrepancies among studies still exist. In addition, data on the frontal plane are lacking, although the frontal plane variables are valuable in studying certain diseases like knee OA. In the next section, cycling studies related to knee kinematics and kinetics, and cycling-related injuries and rehabilitation studies will be reviewed.

CYCLING BIOMECHANICS

Equipment

The basic components of a bicycle include the frame, seat (saddle), handlebars, cranks, and pedals (2, 46). During pedaling, the top most position of the crank and pedal is called the top dead center, while the bottom most position is the bottom dead center. To describe the position of the pedal and crank, the top dead center is defined as 0 degree or 360 degrees, and the bottom dead center is 180 degrees. A complete cycle of the pedal can be divided into a power phase and a recovery phase. The power phase begins at 0 degree position and ends at 180 degrees position. During this phase, the cyclist pushes down on the pedal and transfers the energy to move the bicycle forward. The recovery phase progresses from the 180 degrees position back to the 0 degree or 360 degrees position (2).

Knee biomechanics in cycling

As a modified hinge joint, the knee rotates mostly about the mediolateral axis in the sagittal plane. In cycling, the movements of knee extension and flexion generate majority of driving force and moment (19). Thus, early cycling studies were mainly focused on the sagittal plane. However, frontal plane knee movements also exist during cycling. During the power phase, the knee adducts as it extends. This motion leads to medial translation of the knee while the knee extends (3). Meanwhile, the ankle everts during this phase, causing an internal rotation of the tibia that increases stress on the medial knee (2). Several studies regarding the frontal plane knee movements have been conducted, and the authors of them pointed out the importance of studying about the non-driving knee moments (25, 39, 45).

Sagittal plane

Ericson et al. (14, 15, 17, 18) investigated the knee kinematics and kinetics during standard ergometer cycling (120 W, 60 RPM, and saddle height of 113% of the distance between

the ischial tuberosity and the medial malleolus). Mean knee ROM was 66 degrees ranging from 46-112 degrees (14). The mean peak knee extension moment was 28.8 Nm and peak flexion moment was 11.9 Nm. The knee extended between about 300 and 140 degrees crank angle, and flexed during the rest of the crank cycle (17). The mean peak tibiofemoral compressive force induced during knee extension was 812 N, and peak anterior tibiofemoral shear force was 37 N (18). Peak concentric power output was 110.1 W for knee extensors, and 30.0 W for knee flexors. Knee extensors contributed 39% to the total concentric work, and knee flexors contributed 10% to the total concentric work (15).

The knee ROM in the sagittal plane reported in studies shows consistent patterns, although the specific ranges vary (3, 44). The difference can be attributed to the difference in seat height. Bailey et al. (3) reported a mean knee ROM of 67.5 degrees ranging from 41.5 - 109 degrees for healthy subjects, and a mean knee ROM of 66.7 degrees ranging from 40.7-107.4 degrees for previous injured subjects. Too et al. (44) found that the mean knee ROM was 67 degrees at 110 mm crank length and 65 degrees at 145 mm crank length.

Knee moments and forces are more sensitive to manipulations of variables, such as seat height, workload and cadence (7). The difference in these variables may lead to discrepancies among studies. Nepture and Hull (35) used a forward dynamics model and found intersegment knee torque to be about 30 Nm during extension and about 30 Nm during flexion. Gregor (26) studied the knee moments when the subjects cycled at 60 RPM at about 160 W. They found the peak knee extension moment to be 53 Nm and the peak knee flexion moment to be 34 Nm. In a study by Tamborindeguy et al. (42), subjects cycled at 70 RPM and 70 W. The peak anterior tibiofemoral shear force was around 80 N and the peak tibiofemoral compressive force was

around 700 N. Despite the differences, knee extension moment in cycling is smaller than that in walking. In one study, knee extension moment was 49 Nm (47).

Frontal plane

There are a limited number of studies examining the frontal plane knee moments in cycling, with all using an instrumented pedal and inverse dynamics approach (16, 24, 25, 39). In most studies, only one sensor was used except for one performed by Gardner (24). The reported knee adduction moment was around 10 Nm except for the study by Ericson et al. (16). The knee abduction moment varied from 2.9 to 15.3 Nm among studies.

In a study by Ruby et al. (39), subjects cycled at 90 RPM and 225 W with the right pedal instrumented with a six-load-component sensor. The authors developed a five-bar linkage model and calculated the three dimensional (3D) knee joint loads using inverse dynamics. The mean peak knee varus (abduction) moment was 15.3 Nm and peak knee valgus moment was 11.2 Nm. Gregersen and Hull (25) used 3D inverse dynamics to calculate the knee load of the right leg in the frontal plane. The model inputs included the pedal force measured by a one-sensor instrumented pedal and 3D kinematic data collected by a motion capture system. When pedaling at 225 W and 90 RPM, the peak knee varus moment was 7.8 Nm during the power stroke and peak knee valgus moment was 8.1 Nm during the recovery stroke. Both moments were highly variable between subjects. The power stroke began at a crank angle of 306 degrees and ended at a crank angle of 119 degrees. Gardner (24) studied the effect of shoe wedges on knee kinetics and kinematics. The author used an instrumented pedal with two 3D force sensors to measure pedal force, calculate both anteroposterior and mediolateral pedal center pressure (COP) and knee moment used inverse dynamics. When the pedal position was neutral, 1st peak knee adduction angle was 2.2 ± 5.3 degrees, and the mean peak knee adduction moment was 9 Nm.

The use of shoe wedges didn't cause any significant changes to peak knee adduction angle or peak internal knee adduction moment.

Ericson, Nisell, and Ekholm (16) studied the varus and valgus knee loads during ergometer cycling. Subjects cycled at 60 RPM and 120 W and the left pedal was instrumented with a force-measuring transducer. The frontal plane knee load was calculated using inverse dynamics combining position data and measured pedal force data. The peak knee varus moment was 24.5 Nm and occurred at 70 degrees of knee angle, and peak knee valgus moment was 2.9 Nm. In addition, as the cyclists rode in position with the knee joints moving close to the midline of the bike, the varus moment decreased to 11.2 Nm.

Knee overuse injuries and rehabilitation

The most prevalent injuries among cyclist are the knee overuse injuries. Overuse injury often occurs when submaximal loading repeatedly exerts on a tissue (2, 5). In cycling, the most common injury is the patellofemoral syndrome, or “cyclist’s knee”, which can cause anterior knee pain. Riding in high gears can develop this injury, because high workload might generate excessive pressure across the patellofemoral joint (46). Iliotibial band syndrome is a common overuse injury in the mediolateral direction of the knee, and most pain occurs on the lateral side. Unlike the patellofemoral syndrome, the repetition of knee flexion instead of pedal force is more of a concern, and cycling with high cadence may cause this injury (9, 12, 46).

Despite the injuries, cycling is recommended as part of a rehabilitation program following ACL surgery (21) and exercise for individuals with OA (30) due to the reason that it exerts smaller load on the knee compared to walking. Studies showed that proper cycling can decrease the applied strain on an ACL graft while decreasing patellofemoral joint stress (21), thus enhancing the healing and recovery process (32). It has been shown by in vivo data that

stationary bicycling is a rehabilitation exercise that can increase muscle activity by increasing the power level without increasing ligament or ligament graft strains (22). One study reported that after 10 weeks of stationary cycling, knee OA patients showed significant improvement in timed chair rise, 6-minute walk distance, walking speeds, amount of overall pain relief, and aerobic capacity (31). Kutzner et al. (30) showed that peak knee resultant contact force was ranged from 0.5 to 1.63 times body weight (BW) in cycling, compared to 2.52 times BW in walking. This study also concluded that the lowest forces can be accomplished by cycling at a low workload, a high cadence, and a high seat height.

Based on the above findings, the magnitude of the load at the knee and the repetition of the load are crucial in determining whether one gets injury or gains health. In cycling, the magnitude and frequency of the load are mainly determined by two factors: workload and cadence. We will examine the literature related to these topics in the next section.

INFLUENCE OF CYCLING WORKLOAD AND CADENCE ON LOWER BODY BIOMECHANICS

Effect of cycling workload on knee biomechanics

Kinematics

Most cycling studies (7, 13) have shown that cycling workload has very little effect on knee ROM or peak knee angles. Bini et al. (7) studied the influence of changing workload on knee kinematics in the sagittal plane in cycling. The participants rode at two cadences (40 and 70 RPM) and three saddle heights (reference height at 100% of trochanteric height; high, +3 cm; low, -3 cm), and the workload was set at 0 N, 5 N, and 10 N of braking force under all conditions. It showed that neither the mean knee angle nor knee ROM was affected by different workloads. Ediline et al. (13) studied the ankle, knee, and hip kinematics under different workloads while collecting 3D kinematics data. However, only the sagittal plane knee joint data was reported. The

cyclists performed the test at 90 RPM and workload was changed from 100 W to exhaustion, with an increase of workload by 50 W every three minutes. The results indicated that there was no difference in knee ROM when cycling at different workloads, with a mean peak knee flexion of 71 degrees, peak knee extension of 138 degrees and a mean knee ROM of 67degrees under all conditions.

Ericson et al. (14) conducted the only study that reported a significant change of peak knee angle under different workloads. They added weights of 0, 2, and 4 kg to the brake generating the workloads of 0, 120, and 240 W, respectively. The results showed that the maximum knee extension angle was significantly decreased with increased workload. The maximum knee flexion angle and mean knee ROM were, however, not affected, which supported findings of with the other studies.

Kinetics

A direct relationship between cycling workload and knee moment has been reported in previous studies. Ericson et al. (17) discovered that during cycling at 60 RPM with workloads of 0, 120 and 240 W, both peak knee extension moment and peak flexion moment significantly increased as the workload increased. Mornieux and Guenette (34) studied the effect of changing workload on relative net moment of each lower extremity joint. Net moment indicates the average of the summed absolute moment over the pedaling cycle. The test was conducted at 80 RPM with workloads of 150, 250, and 350 W. As the workload increased, the total net moment generated at the ankle, knee, and hip increased from 86 Nm to 152 Nm, and the contribution of knee net moment significantly decreased from 30% to 25%. Thus the knee net moment actually increased from 25.8 Nm to 38 Nm with increased workload.

The change of knee compressive contact force with respect to workload showed the same trend in all studies, with either significant or small increases in peak contact force being associated with increased workload. In Ericson et al.'s study (18), subjects pedaled at 60 RPM with workloads of 0, 120 and 240 W, both the peak tibiofemoral compressive force and the peak anterior tibiofemoral shear force increased significantly with increased workload. In a study by Kutzner et al. (30), the authors used an instrumented implant to measure tibiofemoral contact force. Subjects were pedaling at 40 RPM with the seat height set at 2 cm below the pedal from the shoe sole when the subject stretched the leg. As the power levels were set at 50, 75, 95, and 120 W, the measured peak knee resultant contact forces were 0.65, 0.96, 1.18, and 1.31 BW, respectively. When cycling at 40 RPM, the peak knee resultant force significantly increased from 0.5 to 1.63 BW as the power increased from 25 to 95 W. The authors found a highly significant correlation between peak knee force magnitude and power. D'Lima et al. (11) used an instrumented stem with strain gauges to measure knee contact force and found that the peak knee compressive force were both around 1.03 BW when the workload set at level two and three. However the measured knee contact forces were slightly larger at level three although the difference was not significant. This might be attributed to the low workloads used in this study.

In the study by Bini et al. (7) described earlier in the kinematics section, the authors also calculated the joint work under different workloads. As the workload increased, both the total mechanical work of lower extremity joints and the knee work increased significantly. In addition, the contribution of the knee to total mechanical work of lower extremity joints (knee work ratio) was also significantly increased with increased workload.

Effects of cycling cadence on knee biomechanics

Kinematics

Most studies have found no effect of cycling cadence on knee kinematics, except for one study reporting significant changes of knee ROM under different cadences. In that study, subjects cycled with a free chosen cadence (FCC), a cadence 20% higher than FCC (FCC + 20%), and a cadence 20% lower than FCC (FCC – 20%). The knee ROM decreased with increased cadence and the knee ROMs at FCC – 20%, FCC, and FCC +20% were 64.3, 62.5, and 58.6 degrees, respectively. The difference between FCC – 20% and FCC, between FCC – 20% and FCC + 20%, and between FCC and FCC + 20% were significant (6).

Bini et al. (7) performed a study that examined the relationship between cycling cadence and knee kinematics. In the study, two pedaling cadences, 40 and 70 RPM, were selected and subjects cycled under three workloads of 0, 5, and 10 N braking force. They found that the cadence did not affect mean knee angle or knee ROM in any condition. In a study by Ericson et al. (14), subjects cycled at 40, 60, 80, and 100 RPM with 2 kg workload. When cycling at 120 W and 60 RPM, the mean knee ROM was 66 degrees (46 - 112 degrees). When the pedal cadence increased, the maximum knee flexion angle, extension angle, and mean knee ROM were not influenced.

Kinetics

Changes in cadence do not affect peak knee contact force has been shown in previous studies (5, 11, 18). Ericson and Nisell et al. (18) used pedal cadence of 60, 80, 100 and 120 RPM with 2 kg workload. They showed that neither the peak tibiofemoral compressive force nor the peak anterior tibiofemoral shear force was affected by changing cadence. D’Lima et al. (11) used a total knee replacement instrumented with strain gauges to measure the three orthogonal forces

at the knee joint. They asked subjects to ride a bike at 60, 70, 80, and 90 RPM. The results showed that the peak knee compressive force was about 1.03 BW and the anterior tibiofemoral shear force was about 0.21 BW for all conditions, and no difference was found between any conditions. In the study by Bini et al. (6), the knee resultant forces did not differ significantly between conditions. The measured knee resultant forces at FCC - 20%, FCC, and FCC + 20% were 106.6 N (0.149 BW), 107.8 N (0.151 BW), and 90.3 N (0.127 BW), respectively.

It is hard to summarize the relationship between peak knee moment and cycling cadence as there is much discrepancy among limited studies. Redfield and Hull (38) used a five-bar linkage model to calculate one subject's knee moment. In the experiment, the cadence was increased from 63 to 80 and 100 RPM with a constant power of 98 W. They concluded that changes in cadence did not affect the magnitude of knee moment. However, based on the estimation from the curves in the study, the peak knee flexion moment and extension moment decreased with increased cadence as the peak knee extension moments were 52, 42, and 39 Nm and peak knee flexion moments were 39, 30, and 24 Nm at 63, 80, and 100 RPM, respectively. Ericson and Nisell et al. (17) used cadences of 60, 80, 100 and 120 RPM with 2 kg workload and found that peak knee flexion moment increased with increased cadence, while the peak extension moment was not affected. Mornieux and Guenette (34) examined the influence of cadence on relative joint net moment. Subjects in the study cycled at 60, 80 and 100 RPM with a workload of 250 W. This study found that the cadence increased, the total net moment generated by the ankle, knee, and hip decreased from 142 Nm to 94 Nm and the contribution of knee net moment significantly increased from 26% to 30%. It was estimated that the knee net moment was decreased from 36.9 Nm to 28.2 Nm.

Three studies also reported the effect of cadence on knee work and their results were conflicting. In one study (7), when the pedaling rates were set at 40 and 70 RPM, neither the total mechanical work of the lower extremity nor knee mechanical work was influenced by cadence. In another study (6), changing cadence did not affect knee contribution to the total mechanical work but the knee joint work decreased with increased cadence. The knee joint work at FCC - 20%, FCC, and FCC + 20% were 71.7, 65.8, and 55.3 J, respectively and the differences between FCC and FCC + 20% and between FCC - 20% and FCC + 20% were significant. Hoshikawa et al. (27) showed the relationship between cadence and relative joint power. The average relative knee power was decreased with increased cadence when pedaling at 40, 60, 90 and 120 RPM and 200 W.

The effect of changing cadence can be attributed to the differences in bikes used among the studies. In general, there are two types of bicycle ergometer. One type uses a weighted brake system and the workload does not change with changes of pedal cadence (6, 7, 11, 18). The other type uses an electromagnetically braked system and the workload changes automatically with cadence to maintain constant power (27, 34, 38). In the second scenario, the workload decreases as the cadence increases, this can explain the decreasing knee work and knee moment in some studies.

SUMMARY

Most studies (7, 13) showed that cycling workload has very little effect on knee ROM and knee angles. For joint kinetics, most studies (7, 11, 17, 18, 30) have shown that the increasing workload leads to greater peak knee extension moment, peak flexion moment, knee work, and peak knee contact force. Changes in cycling cadence do not affect peak knee extension or flexion angle (14). As to the knee ROM during cycling, a consensus cannot be fully reached. Some studies (7, 14) found that knee ROM was not affected by cadence while one study

(6) reported that knee ROM decreased with increased cadence. On the other hand, changes in cadence do not affect knee contact force (6, 11, 18). The results on the relationship between knee moment and cadence are not consistent. With increased cadence, one study showed the net knee moment decreased (34) while the other showed the total knee moment unchanged (7). Finally, another study showed the peak knee extension moment increased and the peak flexion moment unchanged with increased cadence (17).

The interests of previous cycling research have not been focused on the frontal plane and no study has reported the effect of changing workload or cadence on frontal plane knee biomechanics. A limited number of studies that focused on the frontal plane used one force sensor, which may not be capable of measuring movement of center of pressure in frontal plane. No previous studies has reported pedal forces, which made it harder to compare values of knee variables among studies.

CHAPTER III

METHODS

PARTICIPANTS

Eighteen healthy male and female subjects of ages between 40 to 79 (age: 55.78 ± 11.02 yrs, height: 1.80 ± 0.10 m, weight: 78.80 ±16.31 kg) with recreational cycling experience participated in this study. Middle age was defined as 40 to 64 years of age, and old was defined as 65 to 79 years of age based on the age classification by American College of Sports Medicine (ACSM) (1). The subjects were free from lower extremity injuries within the past six months, and were able to ride a stationary bike without any aid. Prior to testing, each subject was asked to read and sign an informed consent that was approved by the University of Tennessee Institutional Review Board.

A sample size of 16 was estimated in a power analysis with an effect size of 0.25, a β level of 0.8 and α level of 0.05 (G*Power 3.1) (20).

INSTRUMENTATION

3D Motion analysis system: A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to collect three dimensional (3D) kinematic and kinetic data. Reflective anatomical markers were placed bilaterally on the subject at the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, great trochanters, iliac crests, and acromion processes. A pedal anatomical marker was placed on the midpoint of the front edge of both pedals. Semi-rigid thermoplastic shells with four non-collinear reflective tracking markers were attached to the trunk, pelvis, thighs, and shanks. Reflective tracking markers were placed on the outer surface of the shoe at the superior, inferior, and lateral heel. Three pedal tracking markers were placed on the lateral side of both pedals, and a crank tracking marker was placed on the crank axis of both cranks. The Vicon Nexus software suite was used to

collect the kinematic and pedal force data simultaneously. The participants wore standard lab shoes (Noveto, Adidas).

Bicycle ergometer: A Monark Ergometer (Model 828E, Monark, Varberg, Sweden) with adjustable seat was used for the cycle testing. The ergometer is equipped with a weighted brake so that the workload can be fixed despite any changes in cadence. The pedals of the ergometer can be removed, and the location of handlebars and seat height can be modified to fit each rider. In addition, the seat position can adjusted anteroposterially.

Instrumented pedal: A customized bike pedal instrumented with two 3D force sensors (1200 Hz, Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A and 5072A, Kistler, Switzerland) was placed on the ergometer to measure 3D forces and moments (24). The charge amplifiers can convert the charges measured by the force sensors to voltage values used by the Vicon Nexus. The kinetic data from the instrumented pedal was recorded by the Vicon Nexus software suite simultaneously with the 3D kinematic data.

Visual 3D: Visual 3D (C-Motion Inc.) was used to process signal and compute 3D kinematic and kinetic data.

PROCEDURES

The subjects were asked to wear spandex shorts and lab running shoes, with height and weight recorded after they changed the clothes. The seat height was set so that the subject's knee angle was at 30 degrees flexion when the crank is at bottom dead center. This seat height was chosen due to the reason that the knee flexion angle method can standardize the kinematics of the knee (5), and knee flexion between 25 degrees and 30 degrees has been reported to reduce the risk of knee injuries (48). The horizontal seat depth was set so that the knee was in line with the pedal when the crank was at the 90 degrees position (8). The position of the handlebars was modified so that the angle between the subject's trunk and thigh was 90 degrees. After the seat

and handlebar positions was set, the subject cycled for three minutes on the ergometer to get used to the bike and position.

Reflective markers were then being placed on subject's segments as described above. All anatomical markers were removed after recording the static trial. Then the dynamic cycling trials were recorded. The participants cycled in 8 testing conditions which included five workload conditions with workloads of 0.5, 1, 1.5, 2, and 2.5 kg at 60 RPM, and four cadence conditions of 60, 70, 80, and 90 RPM with 1 kg workload. The condition of 60 RPM with 1 kg workload was an overlap by 5 workload conditions and 4 cadence conditions, it was performed only once in real data collection. The cycling conditions were randomized. Each cycling condition was performed for 2 minutes. Data were collected on at least 5 consecutive pedaling cycles from top dead center (0˚) to top dead center (360˚), which began during the last 30 seconds of each trial. After each condition, participants were asked to identify the intensity of the cycling condition using a rated perceived exertion (RPE) scale. Participants were given sufficient time of rest between conditions (Figure 2).

The range of workload in this study was set in such a way so that they correspond to work rates of 30, 60, 70, 80, 90, 100, 120, and 150 W, which covered the light, moderate, and high intensity for middle-aged and old adults. The cadence of 60 RPM was selected due to the reason that a low cadence was recommended for recreational cyclists during endurance training (1) , and was frequently used in biomechanics studies of cycling (11, 17, 24, 38). According to ACSM, a workout at 2.0-3.9 metabolic equivalents (METs) is considered to be light in intensity, 4.0-5.9 METs is considered to be moderate in intensity, and 6.0-8.4 METs is considered to be high intensity activity (1). For example, a 75 kg middle-aged adult cycles at 60 W equals to

workout at 4.5 METs, and cycles at 160 W equals to work at 8.6 METs. The following equation was used to calculate METs with respect to workload (43).

$$METs = \left(10.8 \times Workload\ (Watts) \div Body\ Mass\ (Kg)\right) + 7) \div 3.5 \quad (1)$$

DATA AND STATISTICAL ANALYSES

The signals from the two pedal sensors were calculated to get the forces, moments of force, and center of pressure (COP) of the right pedal using following equations:

$$F_x = F_{x1} + F_{x2} \quad (2)$$

$$F_y = F_{y1} + F_{y2} \quad (3)$$

$$F_z = F_{z1} + F_{z2} \quad (4)$$

$$M_x' = a_{z0} \times F_y \quad (5)$$

$$M_y' = (a \times F_{z1} - a \times F_{z2}) - a_{z0} \times F_x \quad (6)$$

$$M_z' = -a \times F_{y1} + a \times F_{y2} \quad (7)$$

$$a_x = \frac{-M_{y\prime}}{F_z} \quad (8)$$

$$a_y = \frac{M_{x\prime}}{F_z} \quad (9)$$

Where F_{x1}, F_{y1} and F_{z1} are the forces measured by Sensor 1 in the x, y, and z direction, respectively; F_{x2}, F_{y2} and F_{z2} are the forces measured by Sensor 2 in the x, y, and z direction, respectively; a is half the distance between two sensors, and a_{z0} is the distance from the sensors to the top of the pedal; F_x is the mediolateral pedal reaction force, F_y is the anteroposterior pedal reaction force, and F_z vertical pedal reaction force; M_x', M_y', M_z' are the moment at the top of the pedal about x-axis, y-axis, and z-axis, respectively; a_x and a_y are COP in the x and y direction, respectively (Figure 1).

The consecutive pedal cycles were separated to obtain 5 individual trials from the top dead center (0 degrees) to top dead center (360 degrees) in Vicon Nexus. Original kinematic and kinetic data was filtered using a low-pass 4th order Butterworth filter with zero lag at a cutoff frequency of 6 Hz (25, 49). Visual 3D (C-Motion Inc.) was used to compute pedal reaction forces, lower extremity joint kinematics and kinetics. Peak angles, velocities, moments and powers were determined using a customized program (VB_V3D) and selected variables were further organized for statistical analysis and reports using another customized program (VB_Table). It should be noted that the pedal force and joint moment variables were not normalized as the majority of the subject's weight was carried by the seat and handlebars.

Two separate one-way repeated measures analyses of variance (ANOVA) were employed to detect influences of cadences and workloads on selected variables, respectively. If a main effect was significant, a pairwise t-test was performed in the post hoc analysis with Bonferroni adjustments to determine differences. An alpha level of 0.05 was set a priori.

CHAPTER IV

EFFECTS OF CYCLING WORKLOAD AND CADENCE ON FRONTAL PLANE KNEE LOAD

ABSTRACT

The effect of workload and cadence on sagittal plane knee biomechanics in cycling has been widely studied, and few studies have focused on the frontal plane. Frontal plane knee biomechanics, especially knee abduction moment, is closely related to the severity and progression of knee osteoarthritis. The purpose of this study was to investigate effects of different workload and cadence on knee frontal plane kinematics and kinetics. Eighteen subjects participated in this study. A motion analysis system was used to collect 5 cycles of kinematics during 2 minutes of cycling in 8 testing conditions, which included five workload conditions of 0.5, 1, 1.5, 2, and 2.5 kg at 60 revolution per minute (RPM), and four cadence conditions of 60, 70, 80, and 90 RPM with 1 kg workload. A custom instrumented pedal was used to collect pedal reaction force (PRF). Increased workloads significantly increased knee abduction moment and knee abduction range of motion (ROM), without any change of peak knee adduction angle. Increased workloads also significantly increased medial, posterior, and vertical pedal PRF, and knee extension moment. Increased cadences had no effects on knee abduction moment. In addition, increased cadences increased anterior and vertical PRF, and knee flexion moment. We found two patterns of frontal knee moments among our subjects which deserves further investigation. Further study may be needed to demonstrate the efficacy of appropriate level of workload in the knee osteoarthritis and other deceased populations.

Keywords: knee abduction moment, knee flexion moment, knee OA, cycling pattern

INTRODUCTION

Cycling is a popular recreational activity which can provide health benefits, improve cardiovascular fitness, and reduce cancer morbidity (36). It is also a low impact exercise which allows people to work on muscle strength and lower extremity mobility while exert a smaller load on lower extremity joints compared to walking or jogging (28, 30, 40). According to one study, peak knee contact force was ranging from 0.5 to 1.63 body weight (BW) in cycling compared to 2.52 BW in walking (30). Cycling is also a recommended exercise for individuals with knee osteoarthritis (30, 31, 33). A training study reported that after 10 weeks of stationary cycling, knee osteoarthritis patients showed significant improvement in pain relief, physical functions, and aerobic capacity (31).

In cycling, the majority of power and driving force comes from knee extension during power phase and flexion during recovery phase (19). Knee movement in the sagittal plane has been widely studied. A knee sagittal plane ROM of 66 to 67.5 degrees has consistently been found among studies, with only small variations due to the differences in seat height (3, 14, 44). Knee extension moments have been shown to have a positive correlation with workload (15, 17, 30). Nepture and Hull (35) found intersegment knee torques to be about 30 Nm during extension and about 30 Nm during flexion when cycling at 90 revolution per minute (RPM) and 225 Watts (W). Gardner (24) found peak knee extension moment to be 26.27 Nm when cycling at 60 RPM and 80 W. One study reported knee flexion and extension moments to be 34 Nm and 53 Nm, respectively when cycling at 60 RPM and 160 W.

Cycling workload and cadence are two variables that can influence the PRF and further affect knee load. A number of research studies has examined the effect of workload and cadence on sagittal plane knee angle, moment, force, and work (6, 7, 11, 13-15, 17, 18, 30, 34, 38). It has been shown by most studies that neither workload nor cadence changes knee ROM or peak knee

angles (7, 13, 14). For the knee kinetics, increasing workload has been found to increase knee moment (15, 17, 34), force (11, 18, 30), and work (7, 15). Ericson et al. (15, 17) found that knee extension moment and knee flexion moment significantly increased as the workload increased from 0 to 2, and to 4 kg. Another study reported knee net moment increased significantly when workload changed from 1.9 to 3.1 and to 4.4 kg (34). However, the effects of cadence on peak knee moments are varied in literature (17, 38). Redfield and Hull (38) reported that the knee extension moment or flexion moment was not significantly different when subjects cycled at 63, 80, and 100 RPM at 98 W. Ericson et al. (17) used cadences of 40, 60, 80, and 100 RPM at 2 kg workload, and found increased knee flexion moments and unchanged knee extension moment across the different cadences. The discrepancy in results may be partially related to different bicycle types of ergometers used in different studies. In general, there are two types of bicycle ergometer. One type uses a weighted brake system and the brake force does not change with changes of pedal cadence (17). The other type uses an electromagnetically braked system and the brake force changes automatically with cadence to maintain constant power (38). In the second scenario, the brake force decreases as the cadence increases, which can explain the decreased knee moment (34).

The increased internal knee abduction moment in level walking, which reflects the loading to the medial compartment of the knee, has been shown to be associated with severity and progression of medial knee OA (33). Subjects with medial knee OA have been found to walk with greater than normal knee abduction moment (4). In a recent study, healthy and knee OA subjects cycling at 60 RPM and 80 W showed that the 10-degree wedge caused reduced knee abduction moment compared to a neutral condition (24).

During the power phase, the knee moves medially as it extends (3). There is a limited number of studies examining the frontal plane knee moments in cycling and their results varied (16, 24, 25, 39). The knee abduction moment and adduction moment have been reported to reach 24.5 and 2.9 Nm at 60 RPM and 120 W (16), 7.8 and 8.1 Nm at 90 RPM and 225 W (25), and 15.3 and 11.2 Nm at 90 RPM and 225 W (39), respectively. One study (24) showed the knee abduction moment to be 9 Nm when cycling at 60 RPM and 80 W. The difference in magnitude of knee frontal plane moment may be caused by different cadence and workload. In addition, different studies used subjects with different levels of cycling experiences, e.g., recreational cyclists (16), competitive cyclists (25), both recreational and competitive cyclists (39), and knee osteoarthritis (OA) patients and healthy subjects (24), which may contribute to the discrepancy found in frontal plane moments.

No cycling studies have examined influences of different workloads and cadences on frontal plane knee kinematics and kinetics. Only a limited number of studies have reported frontal plane knee kinematic and kinetic data (16, 24, 25, 39). It was important to study effects of the workload and cadence on frontal plane knee variables, especially internal knee abduction moment, to provide research evidence for prescribing cycling as a therapy for knee OA and other knee orthopedic patients. In addition, most of the existing studies have used young healthy male subject or patients (3, 6, 7, 11, 13-18, 34), while cycling data in middle-aged and old populations are lacking. Furthermore, most knee OA patients are middle-aged and old adults. Therefore, the purpose of this study was to examine effects of different cycling workloads and cadences on knee frontal plane biomechanics in middle-aged and old adults. It was hypothesized that increasing cycling workload would increase peak knee abduction moment and peak knee

adduction angle; and increasing cadence would not change peak knee abduction moment or peak knee adduction angle.

METHODS

Participants

Eighteen healthy male and female subjects of ages between 40 to 79 (age: 55.78 ± 11.02 yrs, height: 1.80 ± 0.10 m, weight: 78.80 ±16.31 kg) with recreational cycling experience participated in this study. The subjects were free from lower extremity injuries within the past six months and were able to ride a stationary bike without any aid. A sample size of 16 was estimated in a power analysis with an effect size of 0.25, a β level of 0.8 and α level of 0.05 (G*Power 3.1) (20). Each subject was asked to read and sign an informed consent that was approved by the Institutional Review Board.

Instrumentation

A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to collect three dimensional (3D) kinematic and kinetic data. Reflective anatomical markers were placed bilaterally on the subject at the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, great trochanters, iliac crests, and acromion processes. A pedal anatomical marker was placed on the midpoint of the front edge of both pedals. Semi-rigid thermoplastic shells with four non-collinear reflective tracking markers were attached to the trunk, pelvis, thighs, and shanks. Reflective tracking markers were placed on the outer surface of the shoe at the superior, inferior, and lateral heel. Three pedal tracking markers were placed on the lateral side of both pedals, and a crank tracking marker was placed on the crank axis of both cranks. The Vicon Nexus software suite was used to collect the kinematic and pedal force data simultaneously. The participants wore standard lab shoes (Noveto, Adidas).

A Monark Ergometer (Model 818E, Monark, Varberg, Sweden) was used for the cycle testing. The ergometer was equipped with a weighted brake so that the resistance force can be fixed despite any changes in cadence. The pedals of the ergometer can be removed, and the location of handlebars and seat height can be modified to fit each rider.

A customized bike pedal instrumented with two 3D force sensors (1200 Hz, Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A and 5072A, Kistler, Switzerland) was placed on the left side of the ergometer to measure 3D forces and moments (24). The charge amplifiers can convert the charges measured by the force sensors to voltage values used by the Vicon Nexus. The kinetic data from the instrumented pedal was recorded by the Vicon Nexus software suite simultaneously with the 3D kinematic data. A dummy pedal with the same mass and design was used on the right side.

Experimental Protocol

Upon arrival to the laboratory, the subject cycled for three minutes on the ergometer to get used to the bike and position. The seat height was set so that the subject's knee angle was at 150 degrees of flexion when the crank is at the bottom dead center (5, 48). The position of the handlebars was modified so that the angle between the subject's trunk and thigh was 90 degrees.

The participants cycled in 8 testing conditions which included five workload conditions with workloads of 0.5, 1, 1.5, 2, and 2.5 kg at 60 RPM, and four cadence conditions of 60, 70, 80, and 90 RPM with 1 kg workload. The condition of 60 RPM with 1 kg workload was an overlap by 5 workload conditions and 4 cadence conditions, it was performed only once in real data collection. The cycling conditions were randomized. Each cycling condition was performed for 2 minutes. Data were collected on at least 5 consecutive pedaling cycles from top dead center (0˚) to top dead center (360˚), which began during the last 30 seconds of each trial. After each condition, participants were asked to identify the intensity of the cycling condition using a rated

perceived exertion (RPE) scale. Participants were given sufficient time of rest between conditions (Figure 2).

Data and statistical analysis

The consecutive pedal cycles were separated to obtain 5 individual trials from the top dead center (0 degrees) to top dead center (360 degrees) in Vicon Nexus. Original kinematic and kinetic data was filtered using a low-pass 4th order Butterworth filter with zero lag at a cutoff frequency of 6 Hz (25, 49). Visual 3D (C-Motion Inc.) was used to compute pedal reaction forces, lower extremity joint kinematics and kinetics. Peak angles, velocities, moments and powers were determined using a customized program (VB_V3D) and selected variables were further organized for statistical analysis and reports using another customized program (VB_Table). It should be noted that the pedal force and joint moment variables were not normalized as the majority of the subject's weight was carried by the seat and handlebars.

Two separate one-way repeated measures analyses of variance (ANOVA) were employed to detect influences of cadences and workloads on selected variables, respectively. If a main effect was significant, a pairwise t-test was performed in the post hoc analysis with Bonferroni adjustments to determine differences. An alpha level of 0.05 was set a priori.

RESULTS

Pedal Reaction Forces

The repeated measures of variance (ANOVA) revealed a significant effect of cadence on peak anterior PRF ($F(3,15) = 26.52$, $p<0.001$). Post hoc comparisons showed significant difference between 60 and 70 RPM ($p=0.034$), 80 RPM ($p<0.001$), 90 RPM ($p<0.001$); between 70 and 80 RPM ($p=0.023$), 90 RPM ($p<0.001$), and 90 RPM ($p=0.001$, Table 1, Figure 3). The cadence had a significant effect on peak posterior PRF ($F(3,15) = 7.66$, $p=0.002$) and significant

differences existed between 60 and 70 RPM ($p=0.003$), 80 RPM ($p=0.007$), and 90 RPM ($p=0.005$, Table 1).

There was a significant effect of workload on peak medial PRF ($F(4,13) = 11.8$, $P<0.001$), and significant differences existed between 0.5 and 1 kg ($p=0.047$), 1.5 kg ($p=0.002$), 2 kg ($p=0.002$), and 2.5 kg ($p<0.001$); between 1 and 1.5 kg ($p=0.01$) and 2 kg ($p=0.001$). There was a significant effect of workload on peak posterior PRF ($F(4,12) = 34.80$, $p<0.001$), and significant differences existed between 0.5 and 1 kg ($p<0.001$), 1.5 kg ($p<0.001$), 2 kg ($p<0.001$), and 2.5 kg ($p<0.001$); between 1 and 1.5 kg ($p=0.01$), 2 kg ($p<0.001$), and 2.5 kg ($p<0.001$); between 1.5 and 2 kg ($p<0.001$), and 2.5 kg ($p<0.001$). Workload also had a significant effect on peak vertical PRF ($F(4,13) = 47.90$, $p<0.001$) and significant differences existed between each pair of the workloads ($p<0.001$, Table 2).

Knee kinematics and kinetics

Cadence revealed a significant effect on knee abduction ROM ($F(3,15) = 3.88$, $p=0.031$). However, the post hoc analysis showed no significant results (Table 1). Peak knee flexion moment significantly increased with increased cadence ($F(3,15) = 12.52$, $p<0.001$). There were significant differences between 60 and 80 RPM ($p<0.001$), and 90 RPM ($p=0.003$); and 70 and 90 RPM ($p=0.001$).

Workload revealed a significant effect on knee extension ROM ($F(4,14) = 7.78$, $p=0.002$), and significant differences were observed between 0.5 and 1.5 kg ($p=0.036$), 2 ($p<0.001$) and 2.5 kg ($p=0.01$); between1 and 2 kg ($p<0.001$), and 1.5 and 2 kg ($p=0.044$, Table 2). Workload had a significant effect on knee abduction ROM ($F(4,14) = 9.48$, $p = 0.001$), and there were significant differences between 0.5 and 2.5 kg ($p=0.007$), and 1 and 2.5 kg ($p=0.028$). Peak knee extension moment significantly increased with increased workload ($F(4,14)=33.043$,

p<0.001). Significant differences were revealed between all pairs of workload ($p \leqq 0.003$, Table 2), except for difference between 2 and 2.5 kg (Table 2). Workload had a significant effect on peak knee abduction moment ($F(4,3) = 10.944$, $p=0.039$). Significant differences were observed between 0.5 and 1.5 kg ($p=0.027$), and 2.5 kg ($p=0.007$), 1 and 2.5 kg ($p=0.028$), and 1.5 and 2.5 kg ($p=0.031$) (Figure 4, Table 2).

DISCUSSION

The purpose of this study was to examine the effects of cycling cadence and workload on knee kinematics and kinetics in the frontal plane. The hypothesis was that increasing workload would change knee adduction angle and abduction moment in the frontal plane; and increasing cadence would not change knee adduction angle or abduction moment in the frontal plane. Our hypothesis about workload was supported by our results which showed that increasing workload significantly increased peak knee abduction moment and changed knee adduction ROM; and changing cadence did not change knee adduction ROM or peak knee abduction moment.

Peak knee extension moment increased with increasing workload. The peak extension moment increased 74%, 29%, 31% and 9% with workload increased from 0.5 to 1.0 kg, 1.0 to 1.5 kg, 1.5 to 2.0 kg, and 2.0 to 2.5 kg, respectively. These results were similar to findings of previous studies (17, 34). Peak knee extension moment was shown to increase 314% from 0 to 2.0 kg, while it increased 195% from 0.5 to 2.0 kg in our study. A previous study from our lab showed knee extension moment of 26.27 Nm when cycling at 60 RPM and 80 W, which is very close to 26.04 Nm of the current study when cycling at 60 RPM with 1.5 kg workload (90 W) (24). Ericson et al. (17) reported a knee extension moment of 30 Nm when riding at 60 RPM with 2 kg workload, which is also similar to 34.23 Nm when cycling at 60 RPM with 2 kg in our study. The knee extension moment increased to overcome the increased workload, which can be

reflected in PRF. Our vertical and posterior PRFs increased significantly with the increased workload.

Workload significantly changed knee extension ROM, which contradict with previous study (7, 13). Bini et al. (7) reported that knee ROM was not changed when the workload was at 0, 0.5, and 1 kg. Edline et al. (13) reported that knee ROM was not changed when subjects cycled from 100 W to exhaustion at 90 RPM. The changing angle in our study might have been caused by the slightly increased trunk movement during data collection. As the workload increased, some subjects might increase their trunk sway to keep up with the higher workload, which may increase hip joint movement, and therefore increase the knee extension ROM slightly.

Peak knee abduction moment changed significantly with increased workload. Peak knee abduction moment increased 63%, 7%, 14% and 24% for workload increase from 0.5 to 1 kg, 1 to 1.5 kg, 1.5 to 2 kg and 2 to 2.5 kg, respectively. The increased abduction moment indicated higher loading to the medial knee when the workload increased. This result may have clinical implications for certain patient population, e.g. people with medial compartment knee OA may want to minimize knee abduction moment.

On the other hand, the values of peak knee abduction moment are relatively low compared to knee loading in sagittal plane (i.e., knee extensor moment). In walking, knee abduction moment has been shown to be associated with medial knee OA (33). A review has reported peak external adduction moment of 2.23 - 5.1 % BW × Ht among knee OA patients and of 2.6 - 3.16 % BW × Ht among healthy subjects in walking (23). Using mean height (1.80 m) and mean weight (773.03 N) in this study, the knee abduction moment would be 0.41 to 1.03 % BW × Ht for workloads between 0.5 to 2.5 kg at 60 RPM, which is much smaller than the lower

bound of KAM of healthy subjects in walking. A recent stair ascent study reported a higher knee abduction moment of 2.1 % BW × Ht when healthy subjects using their preferred speed compared to our results (37). However, the actual knee joint loading may be higher due to greater muscle contractions due to high mechanical demands in the power phase in cycling compared to walking (12). Further study is needed to use musculoskeletal modeling to estimate actual knee contact force in cycling.

Another interesting finding is that the subjects in this study demonstrated two different frontal plane moment patterns. Seven of them showed abduction moment (Figure 5a) while 11 demonstrated an adduction moment (Figure 5b). The contributing factor to this discrepancy is the direction of the pedal reaction force in relation to the knee in frontal plane. If the PRF vector is directed to the medial side of the knee, it generates an external knee adduction moment and hence internal knee abduction moment (Figure 6a). In contrast, if the vector is directed to the lateral side of knee joint, it generates an external knee abduction moment and internal knee adduction moment (Figure 6b). As for the PRF vector, two variables may influence its direction, ankle and knee positions in the frontal plane. Furthermore, the two factors that can lead to four combinations, everted ankle and abducted knee, everted ankle and adducted knee, inverted ankle and abducted knee, and inverted ankle and adducted knee. Ericson et al. (16) has reported that cycling with knee joints moving close to the midline of the bicycle can decrease knee adduction moment. Five of our subjects who cycled with everted ankle and abducted knee did generate peak knee adduction moment. However, other subjects with peak abduction moment cycled at inverted ankle and abducted knee, or inverted ankle and adducted knee, which indicates that some other variables, such as foot position and pelvis width might also have an influence on frontal plane knee moment.

Gardner (24) used lateral wedges of two different degrees to keep the ankle in more everted position in cycling, and the results showed that the peak knee abduction moment was decreased among both healthy subjects and knee OA patients. However, the vertical PRF increased when the wedge was used, this may somewhat negate the benefit of decreased peak knee abduction moment. In our study, we compared the vertical PRF of subjects who generated knee abduction moment and knee adduction moment and both group generate similar vertical PRF. Thus compared to using wedges, modification of cycling patterns using other methods might be other alternatives to decrease knee abduction moment. Further study may be needed to demonstrate the efficacy of higher level of workload in the knee OA and other deceased populations.

The peak extension moment did not increase with increased cadence during the power phase. However, the peak knee flexion moment increased with increasing cadence during the recovery phase, which was partly supported by findings of previous studies (15). Ericson et al. (15) used cadences of 40, 60, 80, and 100 RPM at 2 kg workload, and found that the peak knee flexion moment increased across the different cadences. The peak flexion moment of both studies occurred at the beginning of recovery phase. However, Ericson et al. reported knee flexion moment at 60 and 80 RPM (at 2 kg) to be 11.9 and 15 Nm, which were smaller than the 16 and 20.67 Nm at 60 and 80 RPM of 1 kg in our study. Redfield and Hull (38) reported that the knee moment was not significantly different when cycling at 63, 80, and 100 RPM at 98 W. The difference might be caused by the different modes of workload/power being used. In our study, the workload was fixed at 1 kg regardless of cadence settings. Redfield and Hull (38) used an electrically braked cycle with a constant power, which caused workload to decrease as cadence increased. At constant power of 98 W, the workloads were 1.56, 1.23, and 0.98 kg at 63, 80, and

100 RPM, respectively. When the cadence increased, the workload decreased, which would decrease PRF, and further negate the increase of knee flexion.

An increasing knee flexion moment often accompanies with higher activation of knee flexor muscles. Takaishi et al. (41) reported an abrupt increase of normalized integrated electromyography values (iEMG) for biceps femoris among cyclists when cadence increased from 75 to 90 and 105 RPM. The values were significantly increased from 75 to 90 RPM at 200 W. However, no increase was seen in non-cyclists with the increased cadence. The iEMG results may reflect that cyclists generated larger knee flexion moment as the cadence increased while non-cyclists did not. The authors concluded that cyclists have utilized a certain skill by positively using knee flexor muscles to deal with higher cadences. Both current and Ericson et al.'s (17) studies saw increased flexion moments with increased cadences as both used recreational cyclists. Takaishi et al. (41) also suggested that the increase of knee flexors muscle activities might assist knee extensors of the contralateral side. When the knee flexes is at the beginning of the recovery phase, it is also when the power phase starts on the contralateral side. Thus the increased knee flexion moment on the recovery side can decrease the extensor requirements on the other side. This can explain the difference of knee moment magnitude between ours and Ericson's study. The knee net moments of both sides (assuming the contralateral side's extensors would generate the same amount of extension moment as the ipsilateral side) were actually very similar between two studies, with 40.7 and 45.0 Nm for Ericson et al.'s study and 36.9 and 43.0 Nm for our study.

On the frontal plane, cadence did not have any effect on peak joint moments or ROMs. Gardner (24) reported that when healthy subjects cycled at 80 W and 60 RPM, knee abduction moment was 9 Nm, which was greater than 7.03 Nm when our subjects cycled at 1 kg workload

and 80 RPM (80 W). The small moment in our study might be related to the small knee adduction movement at the beginning of power phase. In our study, subject had 0.37° of knee adduction ROM, compared to 2.2° in Gardner's study. If the peak adduction angle occurred at the same time, a smaller adduction ROM may cause a smaller frontal plane moment arm for the frontal plane GRF which may cause a smaller abduction moment. Another reason is the muscle strength difference as the knee moments on three planes were greater in Gardner's study. It needs mention that the workload was different and it was larger (1.3 vs. 1 kg) in Gardner's study. This may also be a cause for the larger abduction moment in their study.

There are a few limitations of this study. As two distinctive patterns were observed in some variables among our subjects (e.g., knee abduction and adduction moment), the statistical power was reduced for these variables. Readers are encouraged to interpret the results with caution. Also, though all subjects were recreational cyclists, it is unclear if they all have similar experience in stationary bike riding.

CONCLUSION

The findings of this study indicate that workload significantly increased peak knee abduction moment and knee abduction ROM, and cadence did not have any effect on peak knee abduction moment or knee adduction ROM. We found that cycling pattern is an interesting topic worth of further investigation, as there are distinctive differences, and the differences are closely related to knee loading and knee injury prevention. Further study may be needed to demonstrate the efficacy of appropriate level of workload in the knee OA and other deceased populations.

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APPENDICES

APPENDIX A: TABLES AND FIGURES

Table 1. Peak pedal reaction force, knee kinematics, and peak knee kinetics variables at 1 kg workload (mean ± SD).

Variable	60 RPM	70 RPM	80 RPM	90 RPM	p Value
Pedal Reaction Force					
Medial PRF (N)	21.67±15.94	23.32±17.18	20.81±15.95	20.03±12.16	0.398
Anterior PRF (N)	34.25±12.40[2,3,4]	38.85±12.57[3,4]	44.18±13.84[4]	52.04±17.70	**<0.001**
Posterior PRF (N)	-66.98±21.98[2,3,4]	-76.73±26.49	-77.44±23.69	-82.60±26.89	**0.002**
Vertical PRF (N)	196.50±51.18	202.50±55.08	202.16±51.45	203.60±50.93	0.316
Knee Kinematics					
Extension ROM (°)	77.74±5.68	78.38±5.33	77.54±5.60	76.04±7.99	0.15
Peak Adduction Angle (°)	5.35±5.95	5.42±5.89	6.21±6.27	5.96±5.97	0.094
Abduction ROM (°)	-10.06±3.91	-10.10±3.78	-10.68±3.73	-9.86±3.60	**0.031**
Knee Kinetics					
Extension Moment (Nm)	20.23±7.82	22.26±8.79	22.33±7.98	24.50±10.63	0.126
Flexion Moment (Nm)	-16.69±8.15[3,4]	-18.07±7.82[4]	-20.67±8.62	-21.51±8.42	**<0.001**
Abduction Moment (Nm)[*]	-9.49±5.92	-7.07±3.53	-7.03±4.30	-7.54±4.57	0.754
Adduction Moment (Nm)[^]	6.12±7.09	5.92±9.41	7.64±8.98	8.34±7.05	0.396

[1]: Significantly different from 60 RPM; [2]: Significantly different from 70 RPM; [3]: Significantly different from 80 RPM; [4]: Significantly different from 90 RPM. [*]: 7 subjects out of 18 showed this pattern; [^]: 10 subjects out of 18 showed this pattern. Positive values refer to medial, anterior or vertical PRF, and knee extension and adduction angle, and knee extension and adduction moment.

Table 2. Peak pedal reaction force, knee kinematics, and peak knee kinetics variables at 60 RPM (mean ± SD).

Variable	0.5 kg	1 kg	1.5 kg	2 kg	2.5 kg	p Value
Pedal Reaction Force						
Medial PRF (N)	7.97±19.83[b,c,d,e]	20.44±15.53[c,d]	27.58±17.58	34.82±21.37	36.01±27.84	**<0.001**
Anterior PRF (N)	35.21±12.30	34.25±12.40	32.80±11.39	32.58±11.40	34.56±11.86	0.128
Posterior PRF (N)	-48.16±18.79[b,c,d,e]	-64.64±21.39[c,d,e]	-81.22±25.99[d,e]	-98.21±29.20	-105.30±30.86	**<0.001**
Vertical PRF (N)	153.02±42.34[b,c,d,e]	194.16±51.76[c,d,e]	229.90±59.61[d,e]	272.46±65.50[e]	304.00±74.76	**<0.001**
Knee Kinematics						
Extension ROM (°)	76.87±5.33[c,d,e]	77.74±5.68[d]	78.93±6.23[d]	80.31±5.89	79.93±5.95	**0.002**
Peak Adduction Angle (°)	5.89±6.44	5.83±5.80	6.56±5.88	6.64±5.69	7.16±5.96	0.076
Abduction ROM (°)	-9.62±3.37[e]	-10.06±3.91[e]	-10.78±4.23	-11.00±4.39	-11.65±4.03	**0.001**
Knee Kinetics						
Extension Moment (Nm)	11.61±6.84[b,c,d,e]	20.23±7.82[c,d,e]	26.04±8.68[d,e]	34.23±10.90	37.16±13.11	**<0.001**
Flexion Moment (Nm)	-17.41±9.60	-16.69±8.15	-17.02±9.17	-16.57±9.30	-19.70±8.96	0.189
Abduction Moment (Nm)[*]	-5.82±3.26[c,e]	-9.50±5.92[e]	-10.18±5.21[e]	-11.60±6.74	-14.36±6.30	**0.039**
Adduction Moment (Nm)[^]	9.52±10.32	6.12±7.09	9.09±8.37	12.68±12.04	16.00±13.50	0.266

[a]: Significantly different from 0.5 kg; [b]: Significantly different from 1 kg; [c]: Significantly different from 1.5 kg; [d]: Significantly different from 2 kg; [e]: Significantly different from 2.5 kg.

[*]: 7 subjects out of 18 showed this pattern; [^]: 10 subjects out of 18 showed this pattern. Positive values refer to medial, anterior or vertical PRF, and knee extension and adduction angle, and knee extension and adduction moment.

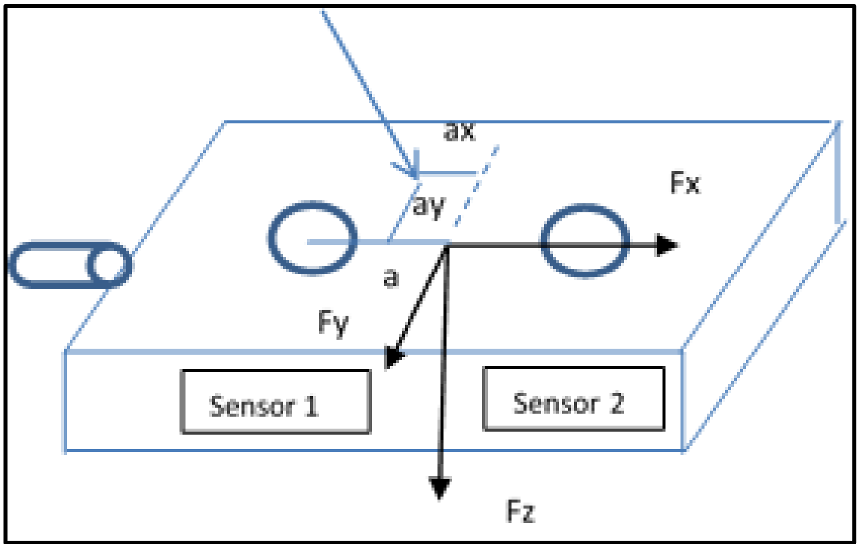


Figure 1. The local coordinate system and arrangement of the two force sensors on the right instrumented pedal.

Figure 2. Testing equipment setup.

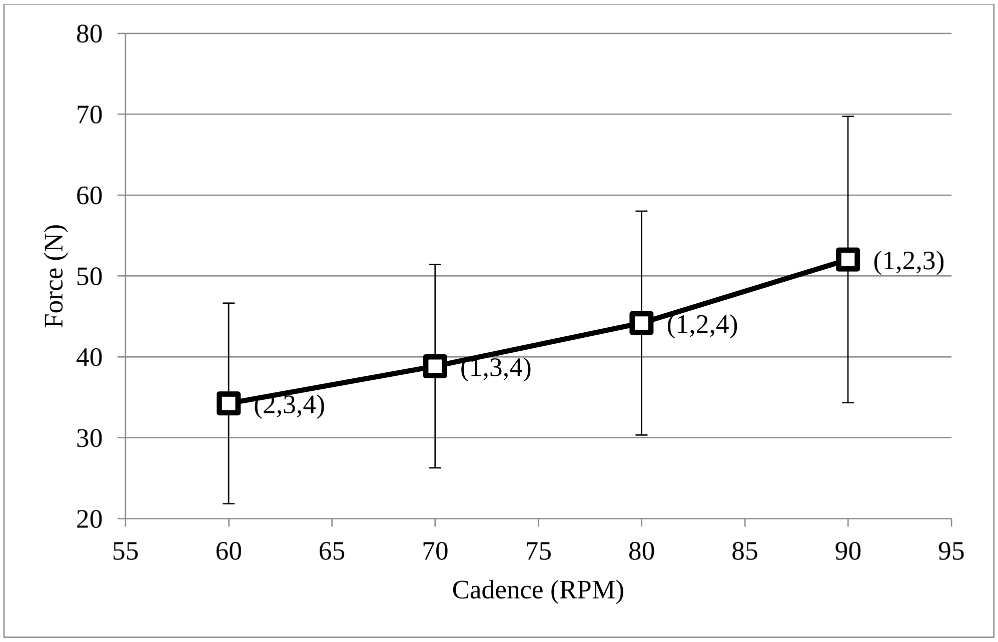


Figure 3. Anterior pedal reaction forces. 1: Significantly different from 60 RPM; 2: Significantly different from 70 RPM; 3: Significantly different from 80 RPM; 4: Significantly different from 90 RPM.

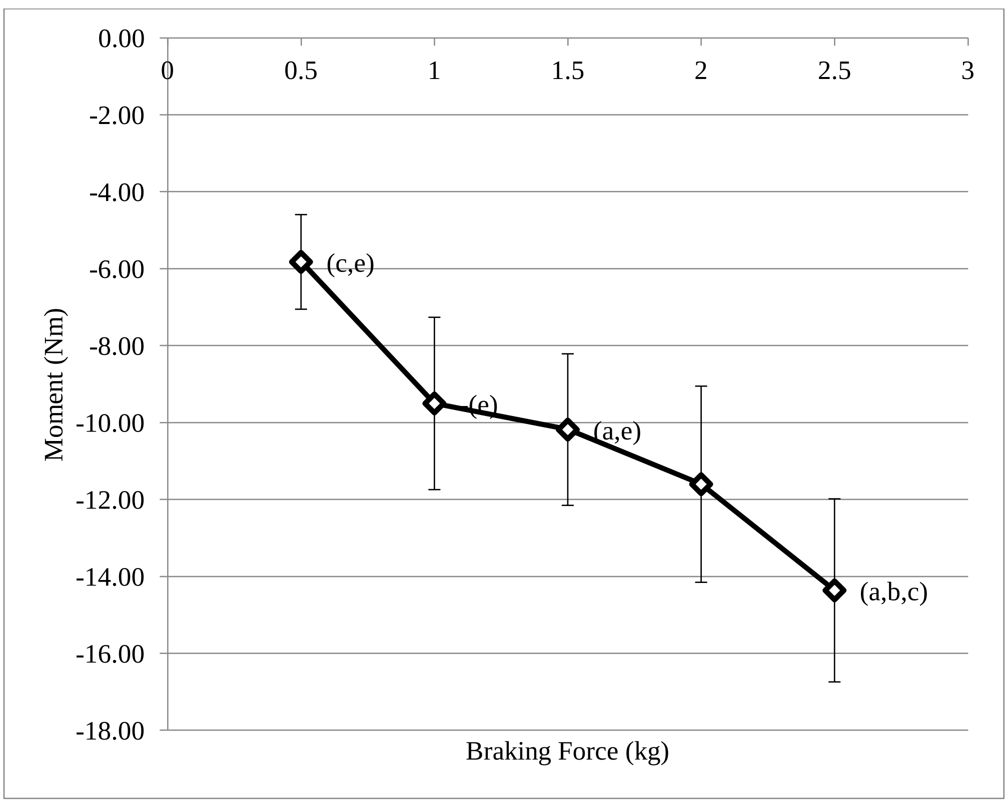
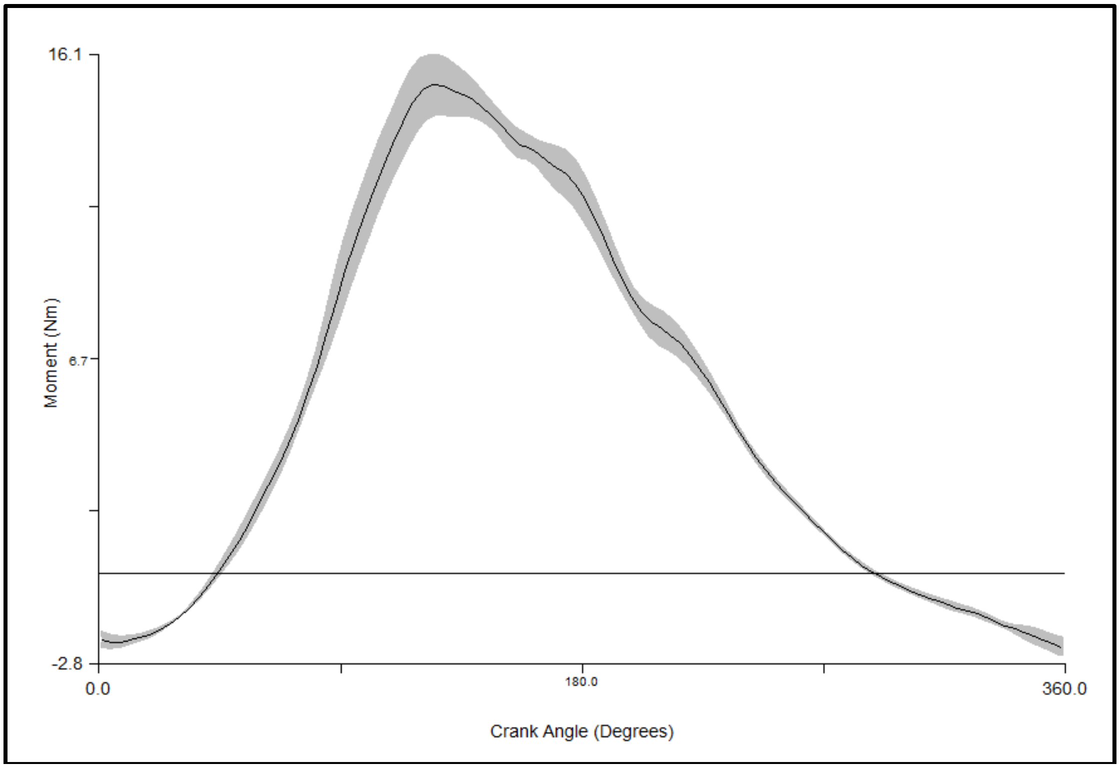
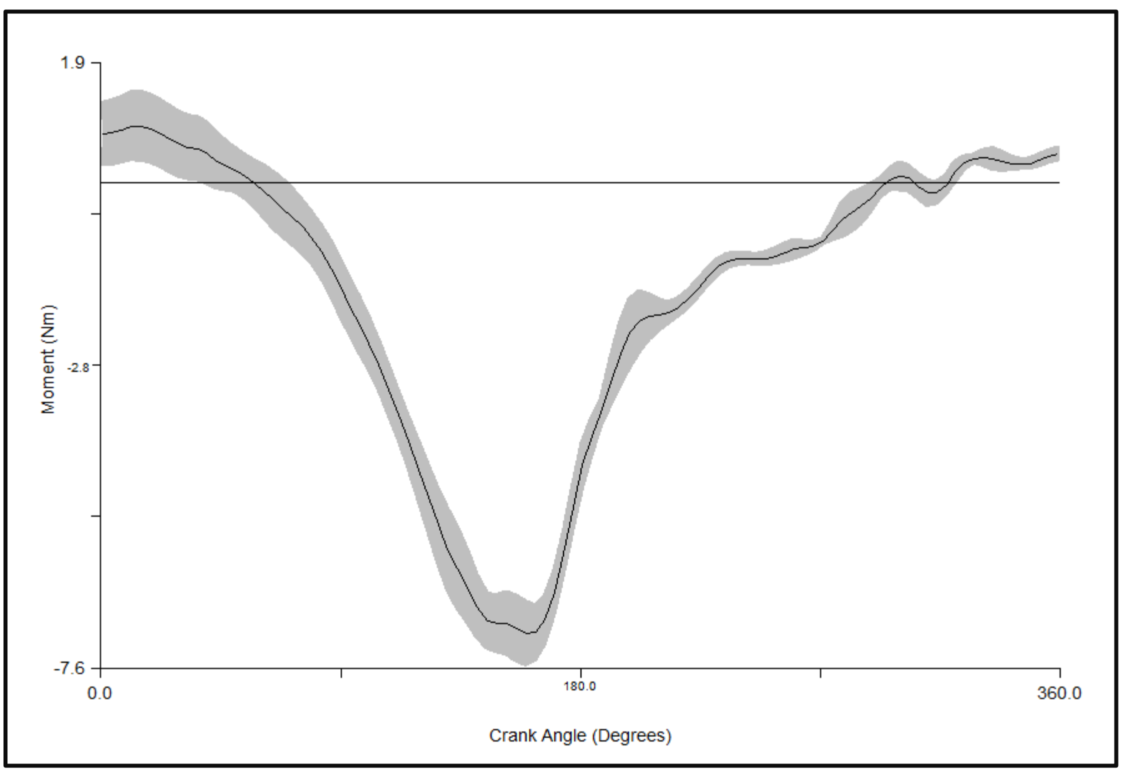


Figure 4. Peak knee abduction moments. a: Significantly different from 0.5 kg; b: Significantly different from 1 kg; c: Significantly different from 1.5 kg; d: Significantly different from 2 kg; e: Significantly different from 2.5 kg.

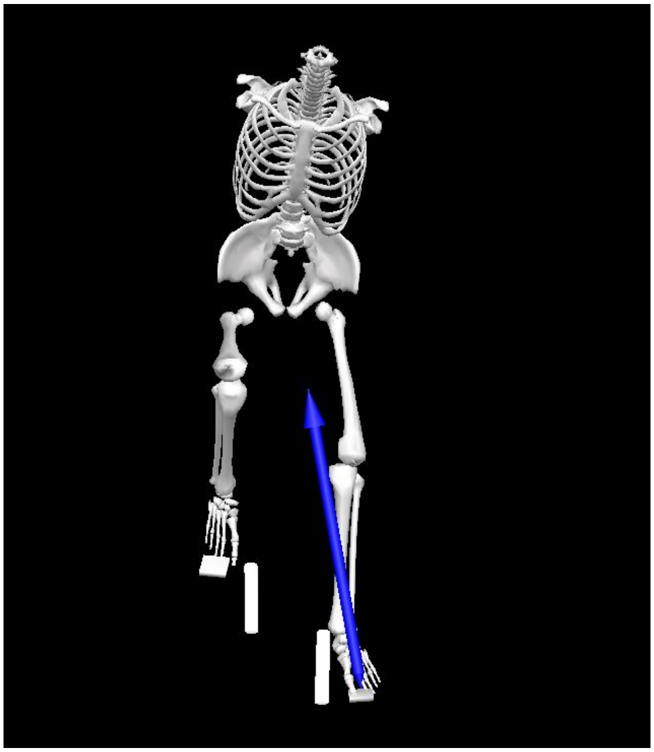


a)

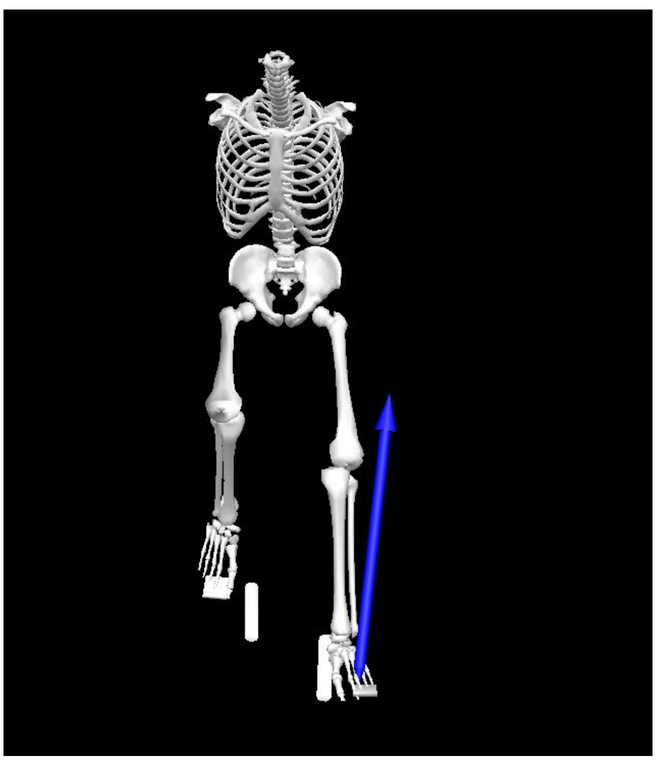


b)

Figure 5. Representative curves of a) knee abduction moment and b) knee adduction moment.



a)



b)

Figure 6. a) Pedal reaction force vector is directed to the lateral side of the knee, generating external knee abduction moment. b) Pedal reaction force vector is directed to the medial side of the knee, generating external knee adduction moment.

APPENDIX B: INDIVIDUAL SUBJECT CHARACTERISTICS

Table 3. Individual subject characteristics

Subject	Gender	Age (years)	Height (m)	Weight (kg)	BMI (kg/m^2)
1	Female	44	1.68	64.6	23.03
2	Male	50	1.85	105.5	30.93
3	Male	78	1.71	68.5	23.56
4	Female	62	1.67	55.8	20.13
5	Male	50	1.77	75.3	24.17
6	Male	71	1.91	80.74	22.25
7	Male	66	1.85	98.9	28.90
8	Male	40	1.88	86.09	24.49
9	Male	58	1.72	70.31	23.91
10	Male	70	1.77	80.59	25.72
11	Female	57	1.72	63.96	21.62
12	Female	57	1.70	68.6	23.88
13	Male	45	1.89	65.32	18.29
14	Female	40	1.66	54.43	19.75
15	Male	44	1.89	100.92	28.40
16	Male	55	1.85	80.6	23.68
17	Male	56	1.95	100.7	26.48
18	Male	61	1.90	97.5	27.15
Mean±SD		55.78±11.02	1.78±0.1	78.8±16.31	24.24±3.30

APPENDIX C: INFORMED CONSENT FORM

Informed Consent Form

Effects of Cycling Workload and Cadence on Frontal Plane Knee Load

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Introduction

You are invited to participate in this research study. You should be free of injury at the time of participation and should not have any lower extremity injuries within past six months. You should be able to ride a stationary cycle ergometer without assistance, have no cardiovascular disease or primary risk factor. You should have recreational experience and participate in physical activities (including cycling) at least three times a week. The purpose of this study is to examine effects of changing workload and cadence on knee frontal plane biomechanics in middle-aged adults. Please ask the study staff to explain any words or information that you do not clearly understand. Before agreeing to be in this study, it is important that you read and understand the following explanation of the procedures, risks, and benefits. The duration of the test will be approximately 90 minutes.

Testing Protocol

Upon arrival to the lab, you will read and sign the informed consent form and fill out Physical Activity Readiness Questionnaire prior to the testing. You will be asked to wear tight fitting workout clothing (i.e., spandex), if you do not own this type of clothing, lab spandex shorts will be supplied. Your height and weight will be recorded. The position of the seat height and handlebars will be adjusted, so that the angle between your trunk and thigh will be 90 degrees, and the knee angle will be 30 degrees when the crank is at the bottom dead center. The chosen knee angle has been reported to reduce the risk of knee injuries during cycling. You will be given at least 3 minutes to warm up and get used to the riding position.

When you finish warm-up, principal investigator will attach markers on your body using double sided tape and hook and loop wraps. Reflective anatomical markers will be placed on your trunk and both thighs, legs and feet in order to capture your motions during cycling. None of the instruments will impede your ability to engage in normal and effective motions during the test.

You will be asked to ride in a total of 8 cycling conditions: five different loads at 60 rpm and four different cadences at 1 kg of breaking force. You will cycle in each condition for 2 minutes and rate your perceived exercise intensity using the Rated Perceived Exertion Scale at minute one. After you finish all cycling conditions, you will be asked to perform 5 successful walking trials at 1.3 m/s (± 5%). You will be given at least 2 minutes of rest between each condition.

Potential Risks

Risks for participating in the study are minimal. Since cycling is a non weight bearing activity, the loading to knee joints will be minimal. You will be required to cycle for no more than 25 minutes

including the warm up during the testing session. You may experience delayed onset muscle soreness (DOMS) in which the muscles are sore for a day or two following the exercise session. However, these conditions are normal for any person who is not accustomed to regular physical activity, and you will be allowed to take sufficient break between conditions, and you will be able to end the test at any time if you feel uncomfortable. In the population of middle-aged adults, there is a risk for a cardiovascular event to occur due to physical activity. Individuals in the study will be excluded if they answer "yes" to any question in the Par-Q. The work rates of the test conditions used in this study are considered as being moderate to vigorous intensity aerobic exercises for most adults based on the Absolute Intensity (MET) by Age.

Benefits of Participation

Results from the proposed study will help establish appropriate cycling protocols for middle-aged healthy adults. The findings will be helpful to illustrate the role of cycling as an exercise for knee osteoarthritis population.

Voluntary Participation and Withdrawal

Your participation is entirely voluntary and your refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled. You may withdraw from the study at any time without penalty. It is your obligation to ask questions regarding any aspect of this study that you do not understand. You acknowledge that you have been offered the opportunity to have any questions answered. Your participation in this study may be stopped if you fail to follow the study procedures or if the investigator feels that it is in your best interest to stop participation.

Confidentiality

Your identity will be held in strict confidence through the use of a coded subject number during data collection, data analysis, and in all references made to the data, both during and after the study, and in the reporting of the results. The results will be disseminated in the form of presentations at conferences, and publications in journals. Only the principal investigators, faculty advisor, Biomechanics/Sports Medicine Laboratory personnel, and the individual subject will have access to the respective subject information and data. Data will be stored on hard drives of password protected computers in the Biomechanics/Sports Medicine Lab and will be backed up onto CDs/DVDs and/or data backup disks, and erased from the hard drives after the completion of the study. All subject data will be coded numerically and referred to only by the code and not by subject name.

The information sheets including the consent forms, and other forms containing subject's identity information will be destroyed three years after the completion of the study. If a subject decides to withdraw from the study, his/her information sheet, consent form and data with the identity and injury history will be destroyed. The cameras used in the study do not capture images of the subjects. If you have any questions about the study at any time or if you experience adverse effects as a result of participating in this study you can contact Ying Fang at 1914 Andy Holt Ave. HPER Bldg, the University of Tennessee (865-974-2091). Questions about your rights as a participant can be addressed to Compliance Officer in the Office of Research at the University of Tennessee at (865) 974-3466.

Consent Statement

The study has been explained fully to my satisfaction and I agree to participate as described. I have been given the opportunity to discuss all aspects of this study and to ask questions. Answers to such questions, if any, were satisfactory. I am eighteen years of age or older, in good health, am qualified for

the study and freely give my informed consent to serve as a subject in this study. I have received a copy of this form.

Subject's Name: ________________________________ Date: _________
Subject's Signature: ______________________________________
Investigator's Signature: ________________________________ Date: __________
Subject # ______

APPENDIX D: FLYER

THE UNIVERSITY of TENNESSEE UT KNOXVILLE

Kinesiology, Recreation, and Sport Studies

UT Biomechanics/Sports Medicine Lab is recruiting subjects for a cycling study.

If you:

- are between the ages of 40 and 64
- are free from leg or foot injuries
- have recreational cycling experience
- participate in physical activities (including cycling) at least 3 times a week

In our lab, you can:

- get an AVI video about your cycling animation
- experience the advanced motion capture system and other technology
- learn how to ride for knee safety

For participation or more information:

Please call **Ying Fang** at the UT Biomechanics / Sports Medicine Lab:

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APPENDIX E: PHYSICAL ACTIVITY READINESS QUESTIONNAIRE (PAR-Q)

Physical Activity Readiness
Questionnaire - PAR-Q
(revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition <u>and</u> that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of <u>any other reason</u> why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to <u>all</u> PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

<u>Informed Use of the PAR-Q</u>: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ____

SIGNATURE ____ DATE ______________________________

SIGNATURE OF PARENT ___________________________________ WITNESS ___________________________
or GUARDIAN (for participants under the age of majority)

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

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continued on other side...

APPENDIX F: INDIVIDUAL RESULTS FOR SELECTED VARIABLES

Table 4. Peak medial PRF (N).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	19.665±0.836	34.009±1.501	43.390±3.814	57.068±5.512	64.521±9.843	31.564±1.618	28.391±3.651	36.192±9.741
2	20.906±11.069	32.379±2.477	40.159±5.364	56.862±6.935	58.369±4.816	36.028±6.526	28.537±6.560	32.288±5.919
3	24.726±5.438	44.161±7.600	49.095±8.351	49.518±3.452	68.752±12.391	29.035±8.321	31.580±6.205	28.227±8.728
4	20.249±5.992	26.100±6.358	39.248±2.784	49.085±9.323	64.196±7.103	29.180±7.022	25.542±2.844	26.936±4.579
5	6.990±0.923	14.090±1.902	18.150±0.936	20.468±1.252	26.512±2.190	12.453±1.054	12.737±0.953	13.112±0.961
6	-14.574±1.325	-12.634±1.632	-12.038±0.424	-11.818±3.134	40.093±1.307	20.920±1.944	-15.747±0.605	17.039±1.679
7	-55.533±0.892	10.585±2.722	17.346±1.352	42.115±2.945	-20.898±0.660	-19.771±0.719	-5.217±3.284	-2.941±3.256
8	23.089±8.017	42.533±8.823	23.373±6.057	-0.253±5.31	22.554±1.823	36.997±4.828	25.729±8.505	25.986±4.523
9	16.175±3.589	31.541±5.536	23.774±3.551	32.017±0.995	39.756±1.423	24.751±0.805	23.875±3.994	16.856±8.129
10	3.323±0.415	26.335±0.474	36.528±2.533	52.273±1.379	56.575±1.941	31.484±1.475	31.256±3.141	28.631±6.775
11	-8.414±0.481	-7.303±0.391	-5.712±0.411	-5.651±0.522	-5.140±0.775	-7.488±0.548	-5.886±0.157	-7.013±0.731
12	18.894±1.320	35.506±1.726	40.303±3.623	57.713±10.564	57.374±6.349	36.676±4.383	29.289±2.815	24.957±5.523
13	12.569±0.536	8.538±1.396	27.629±0.773	38.677±0.870	-3.920±0.959	52.222±8.426	49.563±3.435	8.710±11.516
14	5.146±1.350	7.004±3.898	14.813±3.333	21.847±1.998	23.446±9.894	6.009±3.452	12.021±0.760	14.360±0.448
15	8.403±1.709	24.202±1.622	40.528±2.899	20.140±2.942	10.226±2.178	26.035±8.087	25.719±4.275	30.175±6.837
16	12.478±0.930	16.740±3.072	15.150±0.958	17.830±1.128	23.205±1.844	13.958±1.993	19.605±1.364	10.221±1.056
17	28.429±3.454	34.574±8.577	38.465±4.977	50.034±4.948	50.926±9.299	34.558±4.628	30.221±3.533	32.149±4.854
18	16.122±1.005	21.646±1.623	42.002±3.949	43.815±0.653	58.115±5.179	25.146±3.670	27.420±3.628	24.612±3.669
Mean±SD	8.813±19.561	21.667±15.940	27.345±17.087	32.874±22.322	35.259±27.197	23.320±17.184	20.813±15.948	20.028±12.159

Table 5. Peak anterior PRF (N).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	26.543±1.888	21.633±1.225	25.566±3.252	34.443±2.028	42.056±3.414	28.888±2.794	29.691±1.872	39.704±3.857
2	49.698±4.724	56.920±0.998	48.020±2.360	42.652±3.124	42.694±3.173	55.937±3.630	58.125±2.945	63.637±3.716
3	47.395±1.542	40.912±2.026	43.524±1.842	26.575±2.477	43.904±3.692	46.071±2.310	54.596±2.278	60.282±3.716
4	28.105±1.995	26.244±1.053	23.213±2.685	33.640±3.114	35.119±0.987	34.319±2.227	42.543±2.433	45.083±3.028
5	17.126±0.635	17.137±0.371	17.238±0.694	15.266±0.380	18.738±0.590	18.461±0.659	21.023±1.227	26.077±0.935
6	43.246±1.544	44.138±1.794	39.356±5.672	37.407±2.940	35.364±2.038	34.708±1.743	59.380±3.782	54.795±1.513
7	31.169±2.415	35.338±1.570	28.362±2.516	42.181±0.487	36.783±1.284	36.715±0.956	42.850±2.542	47.108±1.380
8	29.204±2.447	26.527±6.123	24.247±5.466	20.897±5.097	22.524±3.887	40.429±1.394	42.519±3.497	48.894±4.538
9	25.441±1.091	23.186±1.159	28.936±2.385	28.889±1.250	27.926±1.196	33.285±1.045	34.168±0.657	49.102±1.351
10	45.242±1.339	41.991±0.695	36.664±1.869	27.273±2.648	33.542±1.917	42.768±1.468	48.701±0.542	58.424±1.961
11	43.723±2.260	37.845±1.918	39.363±2.018	37.921±3.177	40.836±4.725	40.249±2.373	43.797±2.597	57.594±1.957
12	33.747±1.024	37.237±2.416	26.319±1.192	28.972±1.045	29.105±1.873	36.464±2.364	44.700±3.082	48.739±5.854
13	33.409±1.251	34.020±3.798	33.706±2.257	38.973±10.178	39.046±2.424	32.931±1.718	47.018±2.333	49.573±2.680
14	13.414±0.381	15.919±1.438	14.955±1.251	13.158±1.794	13.457±1.904	24.400±1.154	18.939±0.332	18.305±1.163
15	45.944±5.025	43.223±8.374	45.348±3.869	44.714±9.478	37.645±5.991	53.637±6.930	52.505±2.489	70.954±3.353
16	18.587±1.428	16.204±0.621	17.858±0.702	15.225±0.346	14.571±1.058	20.529±0.646	26.203±0.563	29.622±0.564
17	53.311±1.650	49.154±2.432	50.154±2.560	52.082±2.760	52.157±2.137	65.165±2.582	63.684±2.892	81.334±2.436
18	48.498±2.023	48.833±1.344	47.594±1.093	46.234±2.522	56.535±3.328	54.248±4.064	64.870±2.213	87.415±1.299
Mean±SD	35.211±12.302	34.248±12.401	32.801±11.386	32.584±11.403	34.556±11.862	38.845±12.569	44.184±13.836	52.036±17.697

Table 6. Peak posterior PRF (N).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	-44.289±1.637	-57.598±2.152	-77.923±4.657	-97.356±6.927	-112.470±20.695	-59.672±3.066	-70.876±7.994	-75.979±10.396
2	-87.496±11.978	-95.874±4.381	-123.009±12.070	-141.808±9.281	-143.298±6.204	-109.914±3.754	-104.853±11.101	-143.201±7.933
3	-35.515±5.501	-68.834±4.054	-69.459±10.518	-110.730±4.859	-88.898±23.788	-64.532±4.801	-61.759±14.913	-52.676±8.264
4	-62.981±3.595	-74.877±5.712	-98.292±7.911	-110.577±9.396	-123.260±2.124	-85.322±11.186	-84.456±1.758	-92.930±12.021
5	-28.555±3.112	-39.505±1.386	-48.015±1.268	-58.381±2.197	-69.841±3.462	-45.281±1.788	-49.224±1.790	-49.263±2.266
6	-47.682±2.467	-57.459±4.166	-83.889±1.007	-101.491±3.423	-109.015±4.062	-65.507±4.798	-74.653±1.401	-74.864±2.649
7	-84.699±3.445	-105.168±9.567	-117.980±7.983	-140.577±11.700	-169.357±9.197	-128.668±3.165	-123.853±5.018	-95.393±9.205
8	-76.828±24.287	-102.541±14.919	-100.983±3.851	-140.012±18.982	-162.388±11.537	-112.023±10.814	-91.224±4.943	-106.021±10.698
9	-51.130±9.162	-78.145±9.222	-95.873±3.955	-99.921±5.875	-114.642±11.186	-72.968±2.489	-98.145±6.379	-101.873±4.328
10	-61.974±1.296	-84.320±3.260	-112.668±5.329	-128.297±0.937	-123.131±1.868	-110.922±1.820	-112.720±4.850	-114.838±2.535
11	-41.375±3.957	-55.489±3.337	-69.474±7.113	-82.014±4.865	-84.990±3.714	-64.311±6.723	-71.118±3.908	-75.175±5.452
12	-42.373±3.437	-65.764±2.646	-82.824±4.343	-94.361±8.190	-109.466±9.753	-78.994±7.252	-73.782±5.412	-77.063±6.840
13	-52.693±6.511	-70.371±3.524	-73.652±4.427	-112.561±19.960	-97.357±14.440	-96.434±6.680	-89.096±4.504	-98.336±2.444
14	-21.665±2.616	-23.586±2.584	-28.284±2.749	-41.098±3.861	-47.954±4.125	-32.369±3.231	-34.603±1.788	-34.098±1.790
15	-30.205±7.587	-72.104±14.048	-91.114±6.973	-124.940±12.254	-130.279±16.580	-80.153±16.445	-68.127±9.551	-79.736±10.130
16	-28.703±2.639	-41.470±4.881	-46.303±3.417	-55.375±1.027	-60.057±2.655	-40.217±2.714	-38.928±2.323	-43.266±2.363
17	-45.799±2.826	-68.502±14.257	-71.659±7.323	-96.015±6.387	-92.284±11.122	-70.548±8.843	-72.341±3.690	-83.591±4.786
18	-38.906±2.437	-44.054±3.092	-78.515±6.911	-86.639±2.971	-97.432±10.297	-63.384±2.876	-74.222±1.824	-88.512±4.348
Mean±SD	-49.048±19.194	-66.981±21.983	-81.662±25.035	-101.231±29.233	-107.562±32.283	-76.734±26.494	-77.443±23.691	-82.601±26.887

Table 7. Peak vertical PRF (N).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	125.664±3.968	156.697±5.114	210.767±11.329	266.852±14.554	332.795±47.504	169.552±10.347	170.494±16.272	222.671±14.593
2	215.430±22.943	258.394±9.745	300.230±26.103	342.794±22.515	372.228±13.228	254.760±11.571	272.268±7.808	274.622±14.286
3	169.158±17.763	222.076±18.264	297.367±32.033	317.675±30.662	386.995±59.858	227.364±10.263	239.125±28.373	199.181±29.408
4	143.710±3.900	178.322±9.980	205.996±17.879	268.300±22.112	319.117±8.505	177.243±16.104	193.087±5.989	210.071±18.356
5	87.847±4.854	112.931±1.357	130.750±2.226	146.859±6.076	170.964±5.381	117.367±2.066	123.294±2.472	133.019±3.616
6	167.917±7.980	210.739±5.351	230.120±6.765	297.980±5.643	290.937±7.106	183.834±10.817	199.398±4.441	197.748±4.154
7	189.805±4.730	259.059±14.837	279.818±11.407	323.539±25.986	365.644±20.688	253.589±7.610	240.071±7.058	202.284±6.034
8	192.010±32.901	236.233±19.427	249.675±10.683	295.545±29.251	333.389±16.788	262.294±21.349	224.184±7.484	217.206±19.441
9	138.854±8.352	169.810±8.049	213.223±7.468	264.773±8.288	288.497±12.089	176.658±2.650	192.843±5.281	199.689±4.475
10	184.785±4.488	234.579±4.486	266.190±6.699	302.455±4.760	340.635±6.919	234.541±4.285	230.595±6.383	247.025±6.527
11	173.516±12.612	198.581±7.474	239.467±8.393	286.651±14.715	317.750±7.281	206.379±10.196	212.198±11.992	218.778±19.295
12	151.484±3.544	226.810±6.522	243.939±10.779	280.599±9.497	325.109±25.197	231.940±15.972	203.638±11.799	204.090±9.289
13	163.832±15.007	187.622±7.563	241.056±8.322	292.831±38.722	306.986±42.125	211.220±9.624	205.035±6.915	191.158±4.372
14	66.944±7.219	88.826±4.057	106.840±4.200	137.355±8.458	152.329±13.453	83.640±4.202	92.875±4.245	83.710±3.744
15	146.822±11.50	222.710±27.80	283.406±13.18	343.602±30.56	398.532±17.15	241.985±34.36	210.488±14.34	232.379±27.85
16	83.276±3.741	109.980±9.585	126.497±5.800	148.335±4.177	162.571±6.299	107.618±5.974	105.593±4.535	110.188±3.579
17	202.607±8.180	235.717±29.570	247.140±22.898	306.218±16.478	301.547±24.788	266.312±29.849	258.275±16.978	241.252±15.405
18	189.677±7.714	227.881±6.025	285.462±12.096	305.049±7.076	335.236±14.615	238.675±13.368	265.384±9.221	279.754±12.139
Mean±SD	155.185±42.087	196.498±51.181	230.997±58.016	273.745±63.775	305.626±72.856	202.499±55.081	202.158±51.453	203.601±50.934

Table 8. Knee extension ROM (°).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	83.270±3.099	86.689±0.784	84.505±1.871	86.506±0.607	82.202±1.583	84.261±1.160	86.085±0.734	85.155±1.533
2	69.318±1.917	69.200±1.128	74.065±0.652	74.451±0.896	72.933±1.662	69.947±1.288	68.319±0.318	70.715±0.693
3	81.954±0.714	84.619±0.639	84.348±1.106	83.811±1.215	85.087±1.649	82.108±1.511	80.554±1.203	80.318±1.325
4	81.079±0.908	80.493±0.887	82.946±1.244	81.313±1.617	79.664±0.869	81.024±1.218	80.126±0.388	79.748±1.939
5	77.888±0.890	79.610±0.576	84.607±0.328	84.679±0.543	88.553±1.137	82.106±0.753	81.059±0.502	79.498±0.739
6	71.181±0.235	70.948±0.471	71.086±0.450	72.970±0.892	72.688±0.506	71.428±0.586	69.489±0.472	70.751±0.551
7	72.118±0.603	72.092±0.524	73.361±0.412	74.688±0.833	74.725±0.542	73.982±0.735	72.529±0.703	71.364±0.465
8	71.683±1.825	71.550±1.186	67.882±3.935	73.484±2.577	74.945±1.150	72.649±1.496	71.962±0.827	52.188±20.684
9	88.665±1.143	87.712±2.137	90.457±1.795	91.463±0.405	93.694±1.691	87.873±0.514	88.828±2.420	90.885±0.578
10	81.761±0.703	82.048±1.224	87.479±0.661	88.730±0.552	84.930±0.477	83.411±0.413	80.713±0.630	79.478±0.325
11	77.625±0.638	74.868±0.608	75.548±0.310	77.416±0.942	76.278±0.787	77.041±1.039	77.492±0.602	75.735±0.892
12	78.766±0.967	80.199±1.012	80.169±1.032	81.810±1.388	81.763±1.663	80.444±0.951	78.369±0.578	78.748±1.600
13	77.808±2.051	81.369±2.256	81.630±0.708	85.966±4.267	81.203±0.641	85.243±2.169	81.784±0.698	79.036±0.717
14	77.991±1.539	78.400±1.205	79.637±1.476	80.791±1.684	80.892±0.857	77.233±3.967	80.272±1.639	79.134±0.841
15	70.618±1.108	74.317±0.787	74.504±1.812	78.144±1.752	80.286±3.816	77.008±0.758	75.514±1.320	74.799±1.931
16	77.577±0.497	80.143±0.696	81.449±0.296	82.874±0.972	82.931±1.121	79.590±0.853	78.139±0.405	78.292±0.444
17	70.550±0.981	71.292±1.205	73.021±0.800	72.214±0.678	72.316±1.042	71.085±1.196	71.327±1.030	70.023±0.182
18	73.761±0.381	73.673±0.369	73.988±1.295	74.257±0.626	73.574±1.590	74.454±0.736	73.136±0.392	72.903±0.455
Mean±SD	76.867±5.330	77.735±5.684	78.927±6.226	80.309±5.885	79.926±5.945	78.383±5.330	77.539±5.601	76.043±7.989

Table 9. Peak knee adduction angle (°).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	1.482±0.531	5.507±0.407	2.733±0.704	6.647±0.561	9.091±0.782	4.020±0.428	4.017±0.720	4.592±1.054
2	7.636±0.523	7.107±0.437	8.746±0.263	7.327±0.206	8.989±1.014	8.014±0.765	7.330±0.716	9.753±0.420
3	3.079±0.870	-0.714±0.583	2.387±0.333	0.472±0.349	3.047±0.375	1.393±0.510	1.477±0.807	4.439±1.426
4	-3.473±0.221	-0.603±0.574	0.655±0.507	1.588±0.700	-2.082±0.450	-3.455±0.416	-0.789±0.472	-2.636±0.738
5	-4.604±0.390	-2.114±1.452	-2.583±0.242	-1.696±0.448	-1.999±0.434	-4.830±0.569	-2.443±0.705	-2.877±0.331
6	5.673±0.201	6.319±0.477	6.116±0.452	5.995±0.290	6.289±0.787	5.912±0.377	6.856±0.164	7.646±0.214
7	-2.868±0.665	-2.813±0.384	-1.839±0.462	-0.652±0.731	0.041±0.574	-1.749±0.450	-3.563±0.321	-2.418±0.246
8	12.460±0.643	14.046±1.081	10.291±0.647	12.220±1.010	13.597±0.352	11.807±0.901	11.252±0.912	11.015±0.360
9	9.216±0.471	8.930±0.269	8.918±0.445	9.977±0.514	9.566±0.593	6.738±0.622	9.507±0.633	7.849±0.170
10	13.254±0.292	12.697±0.208	13.739±0.278	15.901±0.231	15.465±0.439	14.156±0.322	14.741±0.503	14.273±0.592
11	6.305±0.881	5.079±0.601	6.146±0.591	5.140±0.725	6.286±0.440	5.590±0.660	5.872±0.227	7.545±0.534
12	0.154±0.721	-2.312±0.503	-2.930±0.645	-2.726±0.783	-1.636±0.239	-0.536±0.373	1.473±0.466	-0.112±0.590
13	13.557±0.783	10.505±0.537	14.399±0.464	12.480±0.565	13.563±0.671	12.325±0.998	15.669±0.342	11.309±0.372
14	6.489±0.738	6.939±1.032	6.556±0.651	7.019±1.124	8.175±0.425	7.171±0.345	7.902±0.461	7.721±0.563
15	7.904±1.173	6.211±0.666	12.816±0.579	8.429±1.653	8.457±0.811	7.961±0.706	8.139±1.751	7.795±0.650
16	2.588±0.255	3.250±0.550	2.791±0.257	2.714±0.416	2.536±0.912	2.576±0.699	2.446±0.166	1.660±0.160
17	-5.534±0.453	-4.314±0.110	-4.388±0.173	-4.084±0.323	-3.881±0.421	-4.344±0.341	-3.679±0.221	-4.851±0.162
18	16.533±0.424	15.120±0.536	14.958±0.622	15.043±0.704	15.385±0.518	14.608±1.097	16.962±0.822	15.798±0.690
Mean±SD	5.276±6.731	5.350±5.954	5.528±6.293	5.655±6.069	6.160±6.305	4.853±6.202	5.732±6.414	5.472±6.156

Table 10. Knee abduction ROM (°).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	-6.932±0.394	-10.682±0.550	-9.338±0.665	-10.206±0.667	-11.676±0.727	-7.400±0.198	-8.455±1.028	-8.923±1.808
2	-11.915±0.604	-12.160±0.353	-12.948±0.720	-12.633±0.515	-13.215±1.084	-12.071±1.052	-11.694±0.803	-12.363±0.778
3	-12.677±1.110	-13.610±1.433	-13.236±0.770	-12.418±1.282	-14.083±0.756	-13.488±0.556	-12.905±1.322	-12.221±1.214
4	-10.299±0.585	-11.802±0.441	-14.138±0.685	-13.831±1.170	-13.023±0.613	-11.422±0.621	-12.515±0.791	-11.413±1.096
5	-15.413±0.622	-16.112±1.075	-16.193±0.545	-16.916±0.697	-16.951±0.734	-15.090±0.546	-15.206±0.715	-12.919±0.731
6	-9.083±0.510	-10.080±0.686	-9.806±0.474	-9.891±0.558	-9.906±0.415	-9.848±0.475	-11.720±0.426	-11.864±0.274
7	-10.703±0.557	-11.833±0.675	-12.456±0.554	-13.393±0.808	-14.197±0.754	-11.603±0.448	-11.133±0.239	-10.317±0.555
8	-11.016±1.839	-11.700±1.600	-5.320±1.647	-6.182±2.208	-8.352±0.600	-9.906±1.054	-9.728±0.944	-5.429±3.006
9	-11.843±0.739	-11.585±0.202	-12.619±0.592	-14.668±0.343	-14.538±0.576	-11.273±0.922	-12.390±0.570	-11.474±0.463
10	-15.123±8.439	-16.508±0.529	-18.930±0.407	-21.423±0.177	-20.852±0.250	-18.736±0.459	-18.197±0.783	-17.768±0.540
11	-2.790±0.818	-2.673±1.207	-3.733±0.643	-4.187±0.488	-5.282±0.556	-2.822±0.843	-3.339±0.463	-4.388±1.076
12	-10.420±1.155	-8.996±1.099	-7.465±0.630	-7.766±0.840	-9.291±0.553	-9.824±0.252	-11.121±0.702	-12.024±0.683
13	-10.810±1.127	-9.185±0.294	-12.558±0.217	-11.849±0.315	-12.933±0.903	-11.471±0.382	-13.168±0.632	-9.789±1.419
14	-7.630±0.414	-9.771±1.186	-15.703±1.583	-11.928±0.901	-12.579±0.375	-10.248±0.756	-12.293±0.332	-11.116±0.903
15	-6.470±1.358	-2.350±1.673	-6.655±1.184	-5.297±2.752	-7.164±1.361	-5.274±0.731	-5.309±2.921	-5.537±1.232
16	-5.049±0.425	-5.567±0.388	-5.181±0.245	-5.987±0.523	-5.193±0.418	-5.201±0.784	-4.610±0.297	-3.914±0.318
17	-5.755±0.380	-6.028±0.338	-7.424±0.212	-7.734±0.283	-8.053±0.272	-6.493±0.277	-7.391±0.243	-5.843±0.189
18	-9.138±0.328	-10.344±0.791	-10.291±0.774	-11.629±0.277	-12.368±1.020	-9.619±0.704	-11.072±0.652	-10.122±0.764
Mean±SD	-9.615±3.371	-10.055±3.906	-10.777±4.233	-10.997±4.391	-11.647±4.027	-10.099±3.781	-10.680±3.725	-9.857±3.597

Table 11. Peak knee extension moment (Nm).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	6.970±0.647	11.612±0.820	13.480±1.367	18.865±1.913	23.710±4.049	12.106±0.926	13.820±2.671	12.639±3.374
2	23.903±3.857	27.903±2.294	37.467±3.897	45.213±5.397	51.738±4.065	30.761±0.644	29.160±3.641	38.408±3.435
3	12.570±2.068	22.693±2.762	24.538±4.523	36.241±2.438	26.242±6.453	21.822±1.276	20.865±5.893	14.136±3.534
4	22.823±2.702	28.835±0.844	36.348±2.808	39.132±4.174	50.158±1.143	29.298±5.070	27.351±1.878	30.937±5.454
5	5.031±0.542	10.949±1.562	14.674±0.958	19.587±1.963	23.616±2.178	9.675±0.487	8.382±0.620	7.806±0.474
6	15.495±0.804	19.755±1.678	34.997±0.861	42.790±3.403	49.169±3.042	32.418±3.129	29.231±0.586	37.122±1.170
7	3.482±1.295	22.116±4.857	33.448±7.563	41.709±2.457	47.533±1.363	19.677±0.733	29.702±2.707	22.351±2.628
8	25.634±9.305	37.702±6.795	30.533±3.186	50.048±5.933	64.944±5.590	39.314±3.698	28.595±2.062	38.971±2.729
9	10.101±3.079	25.026±2.648	33.393±2.308	32.655±1.989	38.159±2.927	22.794±1.031	29.468±3.137	28.493±1.793
10	9.538±0.130	16.443±0.945	27.695±1.847	36.523±1.278	32.773±1.418	25.558±0.456	26.975±2.123	26.125±0.737
11	8.611±0.931	15.190±1.631	20.852±2.638	29.432±2.630	31.122±2.826	18.848±3.143	19.577±1.956	21.373±2.153
12	8.400±1.604	16.731±0.993	27.397±1.857	30.273±3.508	34.387±1.442	20.736±2.200	21.043±2.853	21.592±2.247
13	12.393±1.637	17.744±1.707	21.035±2.616	37.725±8.922	32.274±6.791	27.029±2.017	25.509±2.505	31.486±3.030
14	5.436±0.628	7.085±0.762	10.253±1.384	12.964±1.745	14.576±1.647	7.827±1.660	8.564±0.733	9.225±0.560
15	7.988±3.594	23.973±5.151	22.602±2.619	34.100±10.641	30.467±6.439	19.731±5.339	15.946±1.895	21.081±1.468
16	6.001±1.189	11.345±1.804	14.917±1.693	18.054±0.523	23.907±1.037	10.251±1.581	9.400±1.346	9.702±1.441
17	17.980±3.444	29.379±10.051	35.599±4.620	48.518±3.070	50.150±6.707	32.995±7.336	32.005±3.276	38.259±1.870
18	6.616±1.593	19.627±4.759	29.427±3.860	42.243±2.414	43.967±3.653	19.804±2.499	26.376±1.824	31.360±3.855
Mean±SD	11.609±6.839	20.228±7.819	26.036±8.676	34.226±10.899	37.161±13.112	22.258±8.786	22.332±7.976	24.504±10.630

Table 12. Peak knee flexion moment (Nm).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	-14.858±1.126	-14.147±0.868	-17.647±1.178	-21.711±1.136	-21.922±2.509	-17.275±1.877	-19.180±1.472	-20.173±3.998
2	-11.905±1.894	-16.553±1.555	-3.733±0.839	-2.023±1.208	-16.556±4.388	-16.997±2.367	-18.395±0.818	-23.812±1.113
3	-18.714±0.928	-15.293±2.855	-19.086±2.088	-11.874±3.008	-21.047±2.268	-17.586±1.787	-24.157±2.434	-25.906±3.917
4	-7.114±0.593	-7.438±0.604	-6.060±1.225	-9.781±1.702	-12.779±0.777	-11.332±1.239	-13.047±0.573	-16.532±0.878
5	-6.808±0.483	-5.138±0.797	-4.589±0.332	-5.339±0.513	-5.627±0.579	-3.879±0.470	-5.432±0.587	-8.073±0.873
6	-32.450±1.241	-33.467±2.529	-31.657±2.084	-23.727±1.013	-15.229±1.216	-24.606±1.884	-39.456±1.233	-25.724±1.180
7	-33.291±2.045	-18.803±1.953	-17.508±2.777	-15.315±0.651	-25.145±0.358	-26.102±0.342	-25.414±1.207	-31.141±1.881
8	-8.177±2.980	-7.702±5.434	-6.397±3.217	-3.242±3.981	-8.234±1.368	-9.429±6.081	-14.746±4.132	-14.514±2.194
9	-19.669±0.820	-18.068±0.573	-23.693±1.118	-26.215±1.172	-23.970±0.718	-24.713±1.046	-23.891±0.736	-27.858±1.481
10	-26.901±0.607	-24.112±1.148	-26.446±0.972	-24.300±0.396	-29.702±1.295	-23.726±0.528	-26.857±1.012	-28.897±0.817
11	-11.705±0.765	-16.952±1.061	-16.214±1.874	-14.621±1.030	-17.920±1.374	-15.498±1.446	-17.308±2.701	-18.657±1.395
12	-10.976±0.672	-15.197±2.281	-15.498±1.399	-21.424±2.075	-19.677±3.720	-16.801±1.148	-18.012±1.317	-13.817±1.452
13	-19.062±1.178	-20.321±1.148	-23.652±1.746	-26.760±4.864	-23.714±1.118	-16.020±1.132	-24.455±0.704	-23.781±1.468
14	-4.304±0.632	-5.995±0.576	-7.335±1.500	-5.624±0.418	-9.075±1.417	-8.078±1.191	-7.880±0.738	-5.058±0.641
15	-27.395±2.189	-22.048±1.483	-25.575±2.065	-26.018±1.077	-36.457±2.068	-28.790±0.763	-32.410±1.458	-29.587±4.081
16	-6.466±0.256	-6.028±0.394	-7.154±0.276	-6.501±0.459	-7.192±0.484	-8.051±0.488	-10.322±0.694	-10.958±0.821
17	-25.322±0.958	-24.318±1.996	-27.309±1.003	-26.793±2.099	-27.646±3.581	-28.438±1.059	-23.787±2.001	-29.755±1.353
18	-28.300±0.634	-28.805±1.872	-26.765±3.047	-26.965±1.410	-32.784±2.367	-27.901±0.922	-27.310±0.914	-32.974±0.926
Mean±SD	-17.412±9.594	-16.688±8.147	-17.018±9.167	-16.569±9.298	-19.704±8.963	-18.068±7.816	-20.670±8.623	-21.512±8.420

Table 13. Peak knee adduction moment (Nm).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
2	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
3	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
4	2.555±0.696	1.781±1.026	5.215±1.856	2.149±0.720	2.586±1.369	2.699±0.175	8.659±2.377	4.046±1.263
5	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
6	18.718±0.587	20.898±0.979	25.056±1.459	32.737±2.231	14.018±1.321	2.871±0.231	27.470±0.725	4.022±0.358
7	33.712±1.231	4.318±0.953	6.138±1.101	11.215±2.805	39.898±2.815	29.159±0.965	12.482±4.469	11.460±0.595
8	4.075±1.945	0.685±1.534	6.214±3.524	22.223±4.635	13.771±2.546	0.312±0.817	2.896±1.924	1.439±1.640
9	9.004±0.670	2.024±0.973	18.662±1.152	24.233±0.838	22.256±5.201	2.893±0.460	3.589±0.317	18.573±1.002
10	5.562±0.281	0.354±0.287	0.330±0.241	0.980±0.192	1.152±0.942	-0.679±0.081	0.046±0.266	8.078±5.252
11	14.455±0.733	15.308±0.854	18.064±0.914	24.362±1.378	24.251±0.395	16.021±1.555	17.218±0.475	17.735±0.655
12	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
13	5.699±1.583	10.794±0.963	6.202±1.016	2.945±2.193	32.571±4.868	1.189±0.626	0.851±0.377	15.747±1.851
14	1.058±0.416	2.782±1.416	2.193±1.027	1.021±0.345	1.218±0.278	3.799±1.136	1.763±0.549	0.000±0.000
15	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
16	0.373±0.106	2.300±0.601	2.827±0.749	4.939±0.519	8.270±0.795	0.907±0.112	1.402±0.098	2.340±1.054
17	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
18	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
Mean±SD	9.522±10.318	6.124±7.088	9.09±8.373	12.681±12.035	16±13.501	5.917±9.416	7.638±8.979	8.345±7.055

Table 14. Peak knee abduction moment (Nm).

Subject	60 RPM					1.0 kg		
	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	70 RPM	80 RPM	90 RPM
1	-0.801±0.222	-4.176±0.377	-4.352±0.732	-6.527±1.222	-8.052±2.093	-1.883±0.317	-1.664±0.329	-4.749±2.329
2	-9.692±0.979	-13.221±0.898	-18.369±1.806	-25.262±1.239	-20.564±1.435	-13.014±0.966	-14.035±0.797	-13.879±0.633
3	-8.332±1.006	-20.936±6.748	-14.388±2.847	-13.543±3.431	-23.273±6.308	-6.802±2.171	-7.577±0.835	-12.147±4.038
4	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
5	-5.262±0.268	-7.135±0.456	-8.438±0.205	-9.382±0.604	-11.190±0.746	-6.824±0.409	-6.779±0.141	-8.365±0.548
6	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
7	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
8	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
9	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
10	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
11	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
12	-2.226±0.362	-4.219±0.467	-4.056±1.060	-4.607±1.269	-6.381±1.556	-4.128±0.642	-2.324±0.582	-1.143±0.370
13	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
14	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
15	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
16	.±.	.±.	.±.	.±.	.±.	.±.	.±.	.±.
17	-7.054±0.664	-9.226±1.263	-9.624±1.717	-11.912±0.954	-14.144±1.107	-8.855±1.002	-6.623±0.869	-3.865±0.327
18	-7.362±0.462	-7.543±0.836	-12.014±1.505	-9.979±0.369	-16.944±0.978	-8.007±0.811	-10.216±1.246	-8.630±1.141
Mean±SD	-5.817±3.255	-9.496±5.921	-10.178±5.211	-11.601±6.74	-14.3614±6.299	-7.073±3.538	-7.031±4.299	-7.541±4.574

VITA

Ying Fang was born in Wuhu, China, to the parents of Ping Fang and Ying Sun. She grew up and attended elementary through high school in her hometown. After she graduated from Wuhu No. 1 high school, she went to Shanghai University of Sports, and received her Bachelor of Science degree in Kinesiology. After her graduation, she got accepted by the University of Tennessee and started to pursue a master's degree in exercise science with a concentration in Biomechanics. She graduated with a Master of Science degree in 2014.