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Effects of Workrate and Seat Position on Frontal Plane Knee Biomechanics in Recumbent Cycling

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Effects of Workrate and Seat Position on Frontal and Sagittal Plane Knee Biomechanics in Recumbent Cycling

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ABSTRACT

In cycling study, there were limited research on recumbent bike kinetics, especially the frontal plane. Increased internal knee abduction moment (KAbM), on the frontal plane, has been shown to be an effective predictor of knee osteoarthritis. The purpose of this study was to examine the effects of different workrates and seat positions on knee biomechanics during stationary recumbent cycling. Fifteen participants cycled on a recumbent ergometer in 6 test conditions of pedaling in far, medium and close seat positions in each of two workrates of 60 and 100 W, at the cadence of 80 RPM. A three-D motion analysis system and a pair of custom-made instrumented pedals were used to collect kinematic and kinetic data. A 3 × 2 (seat position × workrate) repeated measures analysis of variance (ANOVA) was used to examine the effect of seat positions and workrates on selected variables of interest. Increased workrates significantly increased peak KAbM and knee extension moment. Different seat positions did not change either peak KAbM or knee extension moment. Due to the larger Q-factor for the recumbent bike used in the study, future study should examine the knee biomechanics with smaller Q-factors, as well as the lower limb muscle activities in recumbent cycling.

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CHAPTER I

INTRODUCTION

Cycling is a popular mode of transportation, recreation, and sport. From 2016 to 2017, 25% of US citizens owned a road bicycle and half of them cycled on a regular basis (1, 2). Studies have shown that cycling can improve cardiorespiratory fitness (3), and strengthen knee flexor and extensor muscles (4, 5), as well as reduce cancer mortality (3), obesity morbidity (3), and depression (6). Moreover, cycling is a preferred exercise over walking or running for individuals with knee osteoarthritis (OA) and anterior cruciate ligament (ACL) injuries since it is advantageous at lessening knee joint loads (7). Despite the numerous benefits, there is a risk of suffering traumatic and non-traumatic injuries during cycling (8, 9), most commonly at the knee (10).

The recumbent bicycle has become popular in recent years due to its multiple advantages over traditional upright bikes. Several studies have reported a decreased knee load reflected in the reduced peak knee extension moments on a recumbent bike when compared to an upright bike (11-15). On a recumbent bike, the rider is allowed to pedal at a reclined position to decrease the intervertebral disc compression on the back (16, 17). With large and anatomically fitted padded areas, recumbent bike can provide a more significant weight distribution across the back and buttocks and relax arms in a neutral position (16), which would benefit cyclists with symptoms such as perineal numbness, erectile dysfunction, handlebar palsy and carpal tunnel syndrome caused by riding on a upright bicycle (16, 18-24). Additionally, a stationary recumbent cycling has been used as a rehabilitation and injury treatment method for people with disabilities such as cerebral palsy (25, 26), cerebral vascular accident (27, 28), diabetes (13), spinal cord injuries (29-31) and ankle immobilization (12).

OA is the most common joint disease in the US and over 80% of the cases affect the knee joint (32, 33). The most important variables associated with knee OA is the external knee adduction moment (KAM) [also known as the internal knee abduction moment (KAbM)]. Together with knee extension moment, they represent the medial compartment loading of the knee. During walking, the knee OA patients showed a greater than normal peak KAbM (34). Also, KAbM is an effective predictor of knee OA progression (35). Due to the reduced knee joint load, cycling is considered as a well suited exercise for OA patients (7). However, only a limited number of studies have investigated frontal plane knee biomechanics (including KAbMs) during recumbent and upright cycling (36-40). In fact, frontal plane knee kinetics has never been examined in recumbent cycling. The only data about frontal plane kinematics of recumbent cycling was reported in the study by Johnson et al. (25). The authors compared lower extremity biomechanics between teenagers with and without cerebral palsy. Subjects were asked to cycle at a cadence of 30 and 60 RPM for at least 30 seconds. The workrate and seat position were highly individualized to each subject's bodyweight and anthropometric measurements. As for results, the knee ROM was about 3 degrees in the frontal plane, ranging from 0 to 3 degrees of knee adduction in healthy subjects. During upright cycling, the knee frontal-plane ROMs were found between 6 degrees of adduction and 4 degrees of abduction (36, 37). The peak KAbM were reported to range from 7.8 Nm to 24.5 Nm while the peak knee adduction moment ranged from 2.9 to 8.1 Nm (36-40). The large variability in the KAbM may be mostly due to the large variation of workrates (80 to 225 W) used in the studies. Fang et al. (36) indicated that larger workrate increased peak KAbM and knee abduction ROM in upright cycling. Hummer et al. (41) examined KAbM in upright cycling at two workrates (80 and 120 W) and 3 saddle heights

(20, 30, and 40 degrees of maximum knee extension angle). No significant difference was found across different saddle heights at either of the workrate.

Sagittal plane knee kinetics has been widely studied in both recumbent and upright cycling. The results of these studies revealed that the recumbent cycling may create smaller knee extension moment than upright cycling, although no tendency of decreased flexion moments was shown. The sagittal plane knee kinetics reported in recumbent cycling by different studies have larger variations. The participants showed a mean peak knee extension moment of about 30 Nm with a pedaling resistance of 15 N, cadence of 60 RPM and backrest-ground angle of 40 degrees in a study by Brown et al. (11). Szecsi et al. (12) reported general muscle moments (GMMs, calculated via inverse dynamics) in recumbent cycling with participants' ankles immobilized. The peak knee extension moments were shown to be 8.6 Nm and 24.7 Nm at the workrate of 30 and 80 W, respectively, while the peak knee flexion moments were around 7.5 Nm at both workrates. Perell et al. (13) showed a mean peak knee extension moment of 1.8 Nm and flexion moment of 17.8 Nm with a cadence of 60-65 RPM and workrate of 60-65 W. With regards to upright cycling, Ericson et al. (15) reported a peak knee extension moment of 28.8 Nm and peak knee flexion moment of 11.9 Nm when the participants pedaled at 120 W and 60 RPM. Gregor et al. (42) showed that the peak knee extension reached 53 Nm with the cycling condition of 160 W and 60 RPM. In the paper of Fang et al. (36), at the cadence of 60 RPM, the peak knee extension and flexion moment ranged from 11.6 to 37.2 Nm and from 17.4 to 19.7 Nm, respectively, when the workload increased from 0.5kg to 2.5kg. In the sagittal plane, larger workrates are found to lead to increased knee extension and flexion moments in both recumbent and upright cycling (12, 14, 15, 36, 43).

Knee biomechanical variables can be influenced by the seat position as well. In a recumbent bike, the seat position is usually controlled by different notches. Therefore, the options are rather limited. The existing literatures only examined knee biomechanics when participants pedaled at different backrest angles in recumbent bike. Reiser et al. (44) did not find significant difference in knee ROM with varied angles of torso reclining. Brown et al. (11) addressed a significant increase of mean knee moment during one entire pedaling cycle when the back rest angle increased from 0 degree to 80 degree. Seat position on upright bike is usually reflected by saddle height, which is defined as the largest distance from the top of the saddle to the center of the upper pedal surface when the crank arm is in line with the seat tube (15). In the sagittal plane, saddle height affects knee kinetics (41). According to Hummer et al. (41), the peak knee flexion moment was increased and the peak knee extension moment was decreased as the saddle height increased.

STATEMENT OF PROBLEM

To our knowledge, no study has investigated how different workloads and seat positions would affect the frontal plane knee kinematics and kinetics in recumbent cycling. In fact, knee biomechanics data on the frontal plane related to recumbent cycling is nearly nonexistent. Therefore, the purpose of the study was to examine the effects of different workrates and seat positions on knee frontal and sagittal plane biomechanics during stationary recumbent bicycling among middle-aged and old cyclists.

SIGNIFICANCE

KAbM has been shown to be associated with the progression of knee OA (34, 45-47). Comprehensive understanding of knee biomechanics, especially frontal-plane joint moment, is necessary to provide guidelines for prescribing recumbent cycling as the therapeutic intervention and rehabilitation tool.

HYPOTHESIS

1. An increased workrate would result in a larger peak knee abduction moment and increased peak knee extension moment.
2. A closer seat position would not result in a different peak knee abduction moment but would result in an increased peak knee extension moment.

DELIMITATIONS

1. All participants were 50 to 70 years old.
2. All participants were free of lower extremity injuries for the past six months.
3. All participants were able to ride a stationary bike for at least 20 minutes without aid.
4. All participants cycled at least 6 hours per week on the regular basis.

LIMITATIONS

1. All tests were conducted in a laboratory setting.
2. The anatomical marker placement of the bony landmarks might not be completely accurate.
3. The tracking markers of the feet were placed on the shoes, which may not completely reflect the actual motion of the feet.
4. The accuracy of the instruments in the study might affect the accuracy of the results.
5. The cycling experience of each cyclist may vary.

CHPATER II

LITERATURE REVIEW

The purpose of the study was to investigate the effects of different workrates and seat positions of recumbent bicycle on knee frontal plane biomechanics among middle-aged and old cyclists. This literature review includes the background of cycling, injury and biomechanics of upright cycling, and advantages and biomechanics of recumbent cycling.

BACKGROUND OF CYCLING

Benefits of cycling

As an efficient and environment-friendly mode of transportation, recreation and sport, cycling is intimately connected to people's lives worldwide. In the United States, one-fourth of citizens owned a road bicycle in their household and half of them cycled regularly from 2016 to 2017 (1, 2). The popularity of cycling is not a coincidence. According to related studies, cycling has been shown to improve cardiorespiratory fitness (3), and reduce cancer mortality (3), obesity morbidity (3), depression (6), and aid brain tissue health by increasing cerebral blood flow (48). Therefore, cycling is recommended for both physically and psychologically disabled and diseased populations.

Besides the benefits listed above, cycling is also commonly used as a lower extremity strength builder and injury rehabilitation tool by health professionals. Several studies have shown that cycling, as a method of resistance training, can effectively increase the strength and power of knee extensor and flexor muscles (4, 5). Also, since cycling is advantageous at lessening knee joint loads (7), it is a preferred exercise compared to walking or running for people who suffer from knee osteoarthritis (OA) and anterior cruciate ligament (ACL) injuries.

Injury risks of cycling

Although it is recognized that cycling can result in benefits for disease prevention and mental health improvement, there is a risk of suffering common traumatic and non-traumatic injuries (8, 9). According to Kulund and Brubaker (10), the most prevalent lower limb non-traumatic injuries experienced by bicycle riders involve the knee joint. In general, knee injuries in cyclists can be classified into 3 categories: patellofemoral inflammation, patella tendinitis and iliotibial band friction syndrome, which are believed to correlate with bike-fit problems, including saddle height, pedal width, and cleat orientation, as well as other factors, such as workload and cadence (49-51). Hence, it is essential to have a thorough understanding of how these variables impact knee biomechanics in cycling.

UPRIGHT CYCLING BIOMECHANICS

Upright cycling has been widely studied by researchers in recent years, with a certain amount of opinions and knowledge being universally accepted. In the following sections, we aim for reviewing the kinematics and kinetics of upright bike.

Terminology

Throughout one upright pedaling cycle, the highest and the lowest point of the crank are called the top dead center and the bottom dead center, respectively. The top dead center is defined as the 0 degree or the 360 degrees, while the bottom dead center is defined as the 180 degrees, of the crank cycle. A full cycle of the pedal contains power phase (0 to 180 degrees) and recovery phase (180 to 0 or 360 degrees). During the power phase, the lower limb extends to produce sufficient force to overcome the pedal resistance and to assist opposite leg in elevating during its recovery phase (52, 53).

Kinematics

Researchers have investigated the kinematics in upright cycling dating back to 1980s. Ericson et al. (54) showed that during the standard ergometer cycling (120 Watt workrate, 60 RPM pedal cadence, a saddle height of 113% of the distance between the ischial tuberosity and the medial malleolus measured on each subject), the average knee range of motion (ROM) in the sagittal plane was 66 degrees, with 112 degrees of peak knee flexion and 46 degrees of peak knee extension. Bailey et al. (55) reported the average knee ROM was 67.5 degrees ranging from 41.5 to 109 degrees for healthy subjects, and 66.7 degrees ranging from 40.7 to 107.4 degrees for subjects with anterior knee pain and/or patella tendinitis. Too and Landwer (56) studied the effect of crank arm length of upright bicycle on hip, knee, ankle angles and power production, noticing a mean knee ROM of 65.8 degrees at the crank length of 145 mm, which is almost identical as the number reported by Bini et al. (43). The knee kinematics results in the sagittal plane found in studies are generally in agreement with each other. The slight differences may be caused by the different settings such as workrate, pedaling cadence and saddle height.

Both Ericson et al. (15, 54) and Bailey et al. (55) pointed out that the peak knee flexion occurred right before the bottom dead center, which is the lowest position of the crank and pedal. Ericson et al. (15) specified that during cycling, the knee extension occurred between the crank angle of 300 and 140 degrees, while knee flexion happened during the rest of the crank cycle.

As for the frontal plane, Gardner et al. (37) compared the effects of limb alignment alternations on knee biomechanics between individuals with and without knee OA, authors found that the first peak knee adduction angle was reached around 60 degrees in the crank cycle when the riders' feet were in a neutral position with a toe cage.

Several studies showed that saddle height (the distance from the top of the saddle to the pedal axle center when the crank arm is pointing down and in line with the seat tube) has a substantial impact on sagittal plane knee kinematics (54, 57). Rugg et al. (57) calculated the lower limb muscle lengths at different saddle height. They showed that compared to ankle and hip, knee joint ROM was more affected as the saddle height increased from 100% to 115% crotch height (the vertical distance from the crotch of the standing subject to the ground). Ericson et al. (54) further showed that when the saddle height was increased from 102% to 120% of the distance between the ischial tuberosity and medial malleolus, the knee extension in the power phase increased 41 degrees while the knee flexion in the recovery phase decreased 22 degrees, resulting a significant increase in the knee ROM of roughly 19 degrees.

However, with respect to the effect of workrate on knee kinematics in cycling, the studies in the literature show inconsistent results. An earlier study conducted by Ericson et al. in 1988 (54), described that when the workrate increased from 0 to 240 W, the maximum knee extension angle during power phase significantly lessened from 49 to 42 degrees, while the maximum knee flexion angle during recovery phase and knee ROM did not significantly change. The results are partially supported by the findings of several later studies. Bini et al. (43) asked their participants to perform the test at 3 saddle heights [100% trochanteric length (the length from the greater trochanter of the femur to the floor) as reference; low (-3cm) and high (+3cm)], and at 2 cadences (40 and 70 RPM) 3 workloads (0, 5 and 10 N of breaking force). They found that neither the knee extension ROM nor peak knee flexion/extension angles in the power phase were influenced when workloads were increased by 5 N of breaking force. Edeline et al. (58) also observed a non-significantly changed knee extension ROM when cyclists pedaled till fatigue, with a starting workrate of 100 W and an increase of workrate by 50 W for every 180 seconds.

Fang and her colleagues (36) focused on the effects of workload and cadence on frontal plane knee biomechanics. They used a motion analysis system and a customized pedal instrumented with two 3D force sensors to collect three-dimensional kinematics and pedal reaction force data at five workloads (0.5, 1, 1.5, 2 and 2.5 kg) at 60 RPM and three cadence conditions (70, 80 and 90 RPM) with 1 kg workload. As the workload increased, no difference at peak knee adduction angle was found, although significant but small changes in knee extension ROMs did exist (ranged from 76.9 to 80.3 degrees). The authors believed the increased knee extension ROMs might attribute to participant's trunk sway and rotation to keep up with the higher pedaling workloads.

Kinetics

Ericson et al. (15) conducted a series of experiments related to knee joint kinetics. When subjects cycled at a power output of 120 W, a cadence of 60 RPM, and a saddle height of 113% of the distance between the ischial tuberosity and the medial malleolus, the average peak knee extension moment was 28.8 Nm and peak knee flexion moment was 11.9 Nm (15). Gregor et al. (42) utilized two instrumented dynamometric pedals on both sides of the bicycle to measure the pedal reaction forces between the feet and pedals in the sagittal plane. Five participants pedaled at 60 RPM with a power output of 160 W for four minutes, revealing a mean peak knee extension moment of 53 Nm at 36 degrees of cranks cycle and a peak knee flexion moment right before the bottom dead center (same time when the peak knee flexion angle occurred discussed in the previous section). Neptune and Hull (59) created a forward dynamic model and an optimization framework to simulate steady-state ergometer cycling with submaximal effort. They identified the intersegmental joint moments when six subjects pedaled at 90 RPM and 225 W. It was shown that the peak knee extension and flexion moments were both about 30 Nm.

As for frontal plane, Ericson et al. (38) examined the knee adduction and abduction load when subjects cycled at 60 RPM and 120W. It showed that the average peak knee abduction moment was 24.5 Nm in the power phase and peak knee adduction moment was 2.9 Nm in the recovery phase. In a study by Gregersen et al. (39), participants cycled at 225 W and 90 RPM. The peak knee abduction moment was 7.8 Nm and the peak knee adduction moment was 8.1 Nm. Recently, Gardner et al. (37) showed that the average peak knee abduction moment was 9.0 Nm when healthy subjects pedaled at 60 RPM and 80 W with neutral foot position. Shen et al. (40) showed that when subjects pedaled at 60 RPM with neutral knee alignment and toe clips on, the average peak knee abduction moments were 4.8, 6.6 and 8.9 Nm with workloads of 0.5kg (40 W), 1.0kg (78W) and 1.5kg (W), respectively.

Most studies agreed that the saddle height has some impacts on the knee kinetics. (15, 43, 60) Ericson et al. (15) compared knee kinetics in the sagittal plane when subjects were cycling at saddle heights of 102, 113, and 120% of the distance between the ischial tuberosity and the medial malleolus. Although the exact magnitudes were not provided, a bar graph in the paper showed that the peak knee flexion moment was decreased and the peak knee extension moment was increased as the saddle height enlarged. In the Bini et al.'s study (43), the reference saddle height was defined as 100% of the greater trochanteric height, while the low and high saddle heights were described as 3 cm lower and higher, respectively. They did find that the knee work contribution (42% vs 38%) to the total mechanical work of the lower limb joint was inversely related to saddle height when the seat was changed from low to high, although no differences of the knee work contribution to the total mechanical work were seen when comparing the reference saddle height to the "low" and "high" heights.

Different workrates usually change knee kinetics substantially. Studies done by Ericson (15) and Bini (43) groups discussed in the previous paragraph both examined knee kinetics at several different workrates. Ericson et al. (15) used power output of 0, 120 and 240 W. When the workrate increased, both peak knee extension moment and knee flexion moment increased significantly. In particular, the external knee flexion moment had a large significant increase of 41 Nm (from 9 to 50 Nm) as the workrate being modified from 0 to 240 W. When Bini et al. (43) compared the results of cycling under workloads of 0 N, 5 N, and 10 N, they noticed that even a small increase in workload caused a significantly increase of knee joint mechanical work (11 J at 5 N, 15 J at 10 N). Significant increases of peak knee extension moment were also noticed between all pairs of workloads from 0.5 to 2.5 kg, only with 2 to 2 kg as an exception in Fang's (36) paper.

For the frontal plane moment, Fang et al. (36) manipulated different workloads to explore the biomechanical changes in the knee frontal plane. They found that the peak knee abduction moments significantly increased 3.68 Nm (from 5.82 to 9.50 Nm) and 4.18 Nm (from 10.18 to 14.36 Nm) when the workload changed from 0.5 to 1 kg and from 1.5 to 2.5 kg, respectively.

RECUMBENT CYCLING BIOMECHANICS

Studies about recumbent cycling are generally lacking in biomechanics literature, especially the frontal plane kinetics. When sitting on a recumbent bicycle, gravity influences body parts dissimilarly than that in an upright bicycle because of the different body positions, which may cause differences in joint kinematics and kinetics (61). The next sections will review the advantages, components, body positions, kinematics, kinetics of recumbent bicycle, and the biomechanics comparison of upright and recumbent bike.

Advantages of Recumbent Cycling

Recumbent bicycle has become the newest craze among today's exercise bikes. Several advantages over traditional upright bikes made this kind of bicycles popular throughout the world. According to various publications, 30%-70% of riders reported cervical, dorsal or lumbar back pain, which usually causes recreational cyclists to drop out of the sport (17, 18, 62). The pain is typically a consequence of intervertebral disc compression with the back in a prolonged flexed position (16-19, 63). On a recumbent bike, the rider is allowed to pedal at a natural and relaxed reclined position to eliminate most of the stress on the back (16). Due to the small areas of the saddle and handlebars of upright bikes, the concentration of the rider's bodyweight on the pubic area and ulnar nerves can reduce blood flows to the particular body parts, causing genital and upper extremity disorders such as perineal numbness, erectile dysfunction, handlebar palsy and carpal tunnel syndrome (16, 18-24). Recumbent bikes, with much larger and anatomically fitted padded areas, can benefit cyclists who have such symptoms by providing a more significant weight distribution across the back and buttocks and relaxing arms in a neutral position without the need to support the weight of the arm and trunk (16). Additionally, a stationary recumbent bike has been recommended as a reliable substitute for upright stationary cycle as a rehabilitation and injury treatment tool for people with physiological disabilities like cerebral palsy (25, 26), cerebral vascular accident (27, 28), diabetes (13), spinal cord injuries (29-31) and ankle immobilization (12). In terms of safety, the recumbent bike is a preferred type of transportation than the conventional upright bike as well. With a more erect, head-up riding position on a recumbent bike, the rider would be more conscious of the surrounding environment. In addition, since a recumbent bicycle is lower to the ground than the upright bicycle, the rider is more unlikely to get seriously injured when accident happens (16).

Equipment & Body Positions of Recumbent Cycling

Similar to upright cycle, a recumbent cycle is usually made up of the frame, saddle, cranks, pedals and handlebars (52). One unique component of a recumbent cycle is the seat-backrest (64, 65). There are many variations in the segments between different brands of recumbent cycles. In the studies of Szecsi et al. (12) and Telli et al. (66), the handlebar was placed in front of the rider at approximately chest level, while in many other studies (14, 67, 68), the handle bars were positioned at sides of recumbent cycles. Few brands of recumbent bicycles even have two sets of handlebars mounted at both of the positions listed above. There are some variations in how the seats can be adjusted in specific bike models. Some allow for both the seat back inclination and the seat to pedal distance (SPD) to be adjusted (12, 67), while others are restricted to only the SPD adjustments (13, 14, 53, 66). Johnston et al. (65) used a recumbent bike with adjustable-length crank arms, pedals and seat back to investigate the differences in pedal forces of adolescents with and without cerebral palsy in 2008 .

During one recumbent pedaling cycle, most literatures defined the top dead center and bottom dead center same as that of upright pedaling cycle illustrated previously (13, 14, 53). However, there are few exceptions. Johnson et al. (25) defined the zero degrees as the point at which the crank arm is at 3 o'clock and farthest away from the subject in recumbent bike.

The four critical geometrical variables to describe the body position of the rider on a recumbent bike are body configuration angle, torso angle, hip orientation angle and seat to pedal distance (Figure 1) (64, 66). Body configuration angle was defined as the angle formed by the trunk and the line connecting the hip joint and crank center with the origin at the hip joint. Torso angle, as known as the body orientation or backrest angle, is the hip-shoulder segment angle relative to the ground. Hip orientation angle is the angle between the horizontal line and the line

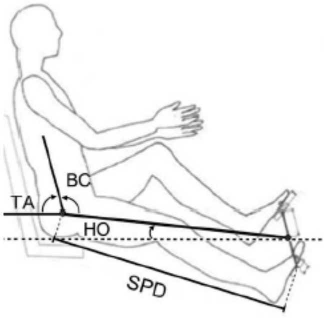


Figure 1. Four critical geometrical variables that describe the riders' positions (69): body configuration angle (BC), torso angle (TA), hip orientation angle (HO) and seat to pedal distance (SPD).

connecting hip joint and bottom bracket (64, 66). The seat to pedal distance is usually modifiable for each subject's lower limb length and reflected by the knee angle at bottom dead center (12, 13, 66). Telli et al. (66) made the seat to pedal distance 100% of trochanteric length of each subject, meaning that the knee extension at the bottom dead center was 180 degrees. However, both of the Szecsi et al. (12) and Perell et al. (13) regulated the knee to be around 20 degrees of flexion at the bottom dead center for their subjects.

Kinematics

Only a limited number of articles presented the lower limb kinematics in recumbent cycling (25, 26, 44, 70-73). Although kinematics was secondary research interests of most of these studies, the results summarized here may give us a clear picture of knee kinematics in the recumbent cycling. It is worth mentioning that when discussing about the knee ROM, none of the articles specified it as knee extension ROM or knee flexion ROM.

Reiser et al. (44) investigated the power output and kinematics in standard pedaling (upright position), as well as recumbent cycling with multiple backrest angles (at 60 RPM and 0 kg for 5 seconds, and as fast as possible for 30 seconds and 8.5% body mass). The backrest angles were defined by a fixed hip orientation angle (-15 degrees) and 5 different body configuration angles (100, 110, 120, 130 and 140 degrees). For the standard cycling position, the participants were allowed to adjust the handlebar height and rotation by their own preferences in order to cycle with comfortable angles of torso lean. The authors controlled the hip-to-pedal distance of all conditions at 105% of the standing leg length (the height from greater trochanter to floor). Interestingly, the body configuration angles of the optimal recumbent peak-power output position (ORP) for the cyclists were not different from that of the self-selected standard cycling position (SCP) (135 vs 134 degrees). As demonstrated in the literature, the lower extremity angles were not affected by how much the subjects lean backwards in recumbent positions. Yet the knee kinematics in ORP and SCP did show some significant differences although the body orientation angles did not differ. Specifically, the maximum and minimum knee angles for all five backrest angles were about 115 and 50 degrees, respectively, creating a knee ROM of around 65 degrees. In the SCP, the mean peak knee flexion and extension angles were 108 and 38 degrees, respectively, resulting in a knee ROM of 70 degrees.

A study by Kerr et al. (72) compared muscle activities and joint kinematics in recumbent cycling versus sit-to-stand and step-up movements. The extension phase of each movement was selected for comparison. In terms of recumbent cycling, the extension phase started at the time when the knee began to extend and finished at the time that the hip began to flex. Subjects were instructed to keep the cycling rate at 60 RPM and enable to have their own preferred seat position choices (workload was not specified). The average knee ROM on the recumbent bike was 51.2 degrees, with the peak knee flexion and extension angles of 80 and 28.8 degrees. These results were not significantly different from that of sit-to-stand (71.0 degrees, ranging from 9.1 to 80.1 degrees) and step-up activities (59.7 degrees, ranging from 10.5 to 70.2 degrees).

When examining recumbent and supine cycling, it is necessary to use a position-controllable cycle ergometer, as seen in a study performed by Kato et al. (71), who used this to examine the maximum muscle strength and oxygen uptake in these conditions. The backrest of the recumbent position was adjusted to a body configuration angle of 105 degrees. For both recumbent and supine conditions, the seat positions were individualized to each subject in order to let their knees slightly bent when reaching the farthest point in the crank cycle. The isokinetic leg muscle strength was tested under three angular velocities: 300, 480 and 660 degrees per second (50, 80, 110 RPM respectively). The researchers found that the knee joint angles at the peak torque (around 110 degrees) were very similar between recumbent and supine pedaling, so as among all three cycling cadences. Additionally, no significant difference of the knee range of motion (78.8 degrees for recumbent vs. 83.1 degrees for supine), peak knee extension angle (139.1 degrees for recumbent vs. 143.1 degrees for supine) and peak knee flexion angle (60.3 degrees for recumbent vs. 60.0 degrees for supine) was seen within two cycling positions.

Johnston et al. (25, 26) executed a series of studies with regard to adolescents with cerebral palsy (CP) on the recumbent bicycle, and two of the studies included and discussed kinematics. In the earlier paper (25), the authors analyzed the muscle electromyographic (EMG) activities, kinematics and power output of lower extremities of CP and typical development teenagers. Subjects were requested to cycle at a cadence of 30 and 60 RPM for at least 30 seconds. The seat position and workload were highly individualized to individual subjects. In particular, the seat-to-pedal distance was set as 85% of the distance measured from the greater trochanter to the base of the calcaneus; the seat back angle was set when the seat-to-greater trochanter distance (the distance from seat to greater trochanter) reached 15% of the distance measured from the greater trochanter to the base of the calcaneus; the crank arm length was adjusted to 30% of the tibial length of the participant; the work load was calculated by the method of Dore et al. (74). In terms of the procedure, this study is different from the other papers mentioned in this section in two main ways. First, they examined not only sagittal plane kinematics, but three dimensional (3D) kinematics data by using a 7-camera motion analysis system. Secondly, unlike the most of the recumbent cycling studies that labeled the top dead center as zero degrees of the crank cycle, the zero degree here is defined as the point at which the crank arm is parallel to the ground and farthest away from the subject. For adolescents with typical development, the knee ROM was about 35 degrees (from 95 to 130 degrees of knee extension) in the sagittal plane and 3 degrees in the frontal plane (0 to 3 degrees of knee adduction) at the cadence of 30 RPM. The peak knee extension and flexion occurred at around 15 and 180 degrees, respectively, while the peak knee adduction reached at around 120 degrees of the crank cycle.

To summarize, sagittal plane knee angles in recumbent cycling have been shown to be similar to that of upright cycling in majority of the literature. The knee ROM usually fluctuates around 60 degrees, with a peak knee flexion angle of around 100 degrees and a peak knee extension angle of around 40 degrees. Although only one article reported frontal plane knee kinematics, knee joint exhibits small adduction movement during a pedaling cycle. It is worth mentioning that the knee kinematics on the bicycle is highly related to the seat position. Since the there is a certain level of variance on the brand of the bike, the seat-to-pedal distance, the backrest angle etc. in the reviewed articles, it is understandable that their results were not exactly consistent.

Kinetics

There are only a handful of studies examining the kinetics on the knee joint in recumbent cycling (11, 12, 14, 27, 28, 65, 70, 75, 76). Despite the fact that many of the experiments were performed with diseased populations, none of them examined populations with knee diseases such as knee OA.

Brown et al. (11) investigated muscle activities, along with joint moments and angles when individuals pedaled at different orientations. Eleven healthy participants cycled at constant workrate of 80 J and cadence of 60 RPM, with the same hip and knee kinematics, and backrest angles of 0, 40 and 80 degrees relative to the ground. Pedal forces were obtained by using a pair of instrumented pedal (77) with footplates attached. Lower extremity joint moments of the sagittal plane were calculated through the pedal forces and kinematics by using the standard Newton-Euler inverse dynamics equations (78). The results showed enlarged knee extensor moments when the body was more perpendicular to the ground. In particular, the average peak knee extensor moment was around 35 Nm (estimated) when the backrest-ground angle was 80 degrees, while the backrest-ground angles of 0 and 40 degrees both showed the peak knee extensor moments around 30 Nm (estimated). The mean knee moment during the entire pedaling cycle was significantly increased when the backrest-ground angles increased from 0 degree to 80 degrees (10.5 Nm vs 15.4 Nm). Since the body orientations (backrest angles relative to the ground in this article) of 40 degrees and 80 degrees mimicked the recumbent and upright cycling, respectively, the study might suggest that the knee joint would have a less moment at a more reclined position. However, we should also keep in mind that in this study, subjects were fixed by lap harnesses on a backboard throughout the whole testing process, which was somewhat different from the traditional upright cycling position.

Gregor et al. (14) investigated effects of workrate and age in recumbent cycling, along with the comparison of the general muscle moment (GMM) between the recumbent and upright bicycle. The kinematics data was recorded by a six-camera motion capture system (Motion Analysis, Santa Rosa, CA, USA) while pedal forces were collected by a pair of customized pedals (Konigsberg, Pasadena, CA, USA) that can monitor the normal and tangential parts of the applied loads. According to Szecsi et al. (12), the GMMs, also known as the net GMMs, are the results of subtracting the passive cycling GMMs (nonzero moments caused by ligaments or joint moment forces during passive cycling) from the active cycling GMMs. The passive moments were calculated from the crank moments recorded during the passive cycling period (motor driven leg turning) by using inverse dynamics. The younger (under 35 years old) and older subjects (over 50 years old) were asked to ride a recumbent bike at a steady cadence (60-65 RPM) and two workrates (30-32.5 W and 60-65 W). The age difference did not influence lower limber GMM patterns, while the workrate had the most obvious effect on the knee moments, as the average peak knee moment were positive (extensor) at the higher workrate (1.10 Nm for the younger group and 2.72 Nm for the older group) and negative (flexor) at the lower workrate (-2.37 Nm for the younger group and -2.17 Nm for the older group). The authors additionally compared the kinetics patterns with the upright cycling study by Gregor et al. (42) in 1985. The upright and recumbent cycling had the similar timing when the knee extension and flexion moment achieved their peak values. However, the recumbent cycling had significantly smaller peak knee extensor moment values (1.91 Nm vs 100 Nm) during the first 90 degrees of the crank cycle, which was mainly due to the lower workrate applied to the subjects in the recumbent cycling study (60-65 W) than in the upright cycling study (160 W). Nevertheless, it cannot completely exclude the contribution of the different pedaling positions to the difference of knee extension moments.

In 2014, Szecsi et al. (12) provided net GMM and power patterns of healthy subjects with ankle immobilization while riding on a recumbent bike at two different workrates (30 and 80 W). The fixation of the ankle joint at the sagittal plane was to mimic the cycling procedure for patients with partial or complete paralysis in the rehabilitation process. At the workrate of 80 W, the knee produced extensor moments from the crank angle of 350 degrees to 180 degrees, with the peak value of 24.7 Nm. For the lower workrate, the corresponding knee extension phase showed a reduced range (350 to 150 degrees) with a significant lower peak knee extension moment of 8.6 Nm. Interestingly, Subjects revealed very similar peak knee flexion moments at the two workrates, which were both around 7.5 Nm. The knee GMM patterns were somewhat different from the ones provided by Gregor et al. (14). Gregor et al. (14) reported a constant knee flexor moment all through the entire crank cycle without knee extension moment in recumbent cycling with a power output of 30-32.5 W. Szecsi et al. (12) believe the previous authors (14) should have subtracted passive moments from the GMM data in order to obtain the knee extensor moments in the power phase. Power and work were also estimated in the paper, showing that knee joint extensors generated significantly more work as the workrate increased (4.5 J at 30 W vs 14.5 J at 80 W).

Hakansson and Hull (75) used forward dynamic simulations to quantify the power contribution of the lower extremity muscles during low power (50 W) recumbent cycling at different pedaling cadences (40, 50 and 60 RPM). The six-segment model previously developed by Neptune and Hull (59) via SIMM (MusculoGraphics, Inc., Santa Rosa, CA) was used to compute muscle excitation patterns of the right and left legs. At 50 RPM, three-component vastus (all three vasti muscles) (VAS) and gluteus maximus & adductor magnus (GMAX) muscle groups reached the peak power of 46.1 and 40.7 W, respectively, generating the major net mechanical work of the right leg. The knee extensor muscle groups (VAS) was shown negatively correlated to the pedaling rates, with the net mechanical work contributions of 38.4%, 33.6% and 22.3% at 40, 50 and 60 RPM, respectively.

Reiser et al. (70, 76) investigated the effects of the recumbent cycling position (RCP) and standard, upright cycling position (SCP) on power outputs. The authors recruited 19 recreational cyclists and asked them to pedal at 250 W and 90 RPM at RCP and SCP. For both positions, knee muscles did the majority of the work (55%), followed by hip (25%) and ankle (11%) muscle groups. Despite the similarity, for SCP, 67% of the knee positive work was done during the power phase and rest of the knee positive work (33%) was done in the recovery phase, while for RCP, only 55% of the knee positive work was done during the power phase, although the two positions did not have significantly different total amount of positive work produced by the knee extensors. A larger peak power generated by knee flexor activities in the recovery phase was also observed in the RCP, which was coupled with a smaller knee extensor moment at the power phase.

Johnston et al. (65) compared pedal forces between young adults with and without CP at the cadences of 30 and 60 RPM in recumbent cycling. The pedal force data were measured by tri-axial piezoelectric force transducers (PCB Piezotronics, Depew, NY, USA) that instrumented into cycle pedals and a seven-camera, 3D motion analysis system (Vicon Motion Analysis, Inc., UK). The vertical forces to the pedal surface were measured. The results suggested that CP subjects spent less percentage of time during a complete crank cycle to push into the pedal to create positive force than healthy subjects at 30 RPM (41.4% vs 50.4%, respectively) and 60 RPM (43.9% vs 51.9%, respectively). The reason why this pattern occurred might be because the CP subjects had weaker hip extensors and ankle plantarflexors, therefore increased hip flexion and ankle dorsiflexion, and discontinued the knee extension phase early. The larger hip flexion motion of the CP subjects can be clearly obtained from the kinematic results of the earlier study done by the same group (25). However, no joint moments were reported.

Two studies looked into the recumbent cycling mechanics of people who had experienced cerebrovascular accidents (CVAs). Hemiplegia, the impairment resulted from CVAs and one of the most commonly seen neurological symptoms, usually causes asymmetries between left and right limbs. In 1998, Perell et al. (27) examined both the affected (aka involved) and the unaffected (aka contralateral) lower limbs within CVAs population. Subjects pedaled at self-selected cadences ranging from 20 to 60 RPM and moderate resistances (28-70 W). The mean peak knee flexor moment of the involved side was larger (21.71 vs 18.29 Nm) and occurred later (189 vs 200 degrees in the crank cycle) than that of the contralateral side. In addition, the authors showed that the contralateral lower limbs of CVAs individuals shared similar patterns of knee joint moment as healthy cyclists who pedaled on an upright bike. The same research team in 2000 (28) also noticed significantly posteriorly directed tangential pedal forces when subjects with CVAs received force symmetry feedback trainings. Perell et al. (13) made the comparison of the joint kinetics in diabetic and nondiabetic men during recumbent pedaling with consistent cycling cadence of 60-65 RPM and workrate of 60-65 W. Although the groups showed the similar muscle moment patterns, they did have disparities on the magnitudes of peak joint moments. For the knee, the peak extensor moment was 1.82 Nm for healthy subjects while the diabetic subjects did not show positive peak knee extensor moment. Moreover, the diabetic group revealed a significantly increased peak knee flexion moment than healthy group (27.24 vs 17.81 Nm respectively).

In summary, the results of several studies have shown that recumbent cycling may create smaller knee extensor moment, which is an advantage over standard upright cycling, although this might be related to the fairly low workloads used in the recumbent cycling studies. However, the recumbent cycling did not seem to show decreased knee flexion moments even though lower workloads were involved. Unfortunately, no previous research reported frontal plane knee kinetics in recumbent cycling, which is a parameter that is strongly correlated to the knee OA progression (45, 46).

CHAPTER III

METHODS

PARTICIPANTS

Fifteen experienced, 50 to 70 year-old cyclists who were healthy (age: 55.5±3.7 years, height: 1.75±0.09 m, mass: 84.3±15.7 kg) participated in the study. Experienced cyclist was defined as an individual who spends at least six hours per week in cycling (41). A healthy participant was free of injury in the lower extremities for the past six months, and able to ride a stationary bike for at least 20 minutes. The participants were recruited from local cycling shops, groups and clubs by emails, flyers and social media. Before the data collection, a written informed consent that was approved by the University of Tennessee Institutional Review Board was read and signed by each participant.

A power analysis was done based on the peak knee abduction moments in the research by Hummer et al. (41). A sample size of 18 was approximated with an effect size of 0.59 with Cohen's F, alpha level of 0.05 and beta level of 0.8 in a 3 x 2 ANOVA design using G*Power (3.1).

INSTRUMENTATION

3D Motion Analysis System

A 12-camera three-dimensional (3D) motion capture system (240 Hz, Vicon, Oxford, UK) was used to collect kinematics data during the test. Reflective anatomical markers were attached to the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, greater trochanter, iliac crest, and acromion process of both sides of the body. Four non-collinear reflective tracking markers grouped as a cluster on a semi-rigid thermoplastic shell were placed to the pelvis, both thighs, and both legs. For the feet, four individual reflective tracking markers were placed at the posterior and lateral heel counter of each shoe. One pedal anatomical marker was secured in the middle of the front side of each pedal. Four pedal tracking markers were put on the pedal bilaterally, with three of them facing the lateral side and one pointing to the inferior direction of the bike (Figure 2b). One reflective marker was attached on each side of the crank axis as well as the front of the recumbent bike.

Recumbent Ergometer

A Kettler Recumbent Ergometer (Model RE7, Kettler, Ense-Parsit, Germany) with electromagnetic brake system was used in the data collection. There are 12 notches along the sloping support frame allowing seat position adjustments. The angle of recline of the backrest can be altered as well. Both the workload and cadence were shown on the bicycle computer display in front of the participants. A jig was used to secure the recumbent ergometer so that the axes of the pedal coordinate system and the lab coordinate system were aligned parallel to each other.

Customized Pedals

Two customized instrumented pedals were utilized to measure 3D pedal reaction forces and moments. To achieve that, two 3D force sensors (1200Hz, Type 9027C, Kistler, Switzerland) paired with two amplifiers (Type 5073A, Kistler, Switzerland) were mounted on each pedal in order to measure the pedal reaction force (PRF) data bilaterally (36, 37). The charge amplifiers converted the output from the force sensors to voltages and sampled simultaneously with the 3D kinematic data by the Vicon system using Nexus (Version 2.7, Vicon, Oxford, UK).

PROCEDURES

All participants wore spandex shorts, t-shirt and a pair of standard lab running shoes (Air Zoom Pegasus 34, Nike). The height and bodyweight of each participant were then recorded. Reflective anatomical and tracking markers were then placed on the participant as described previously. Before the actual data collection, a static calibration trial was taken, during which the participant stood with their arms crossed in front of the body and feet separated at shoulder width with both feet pointing forward. After each static trial, anatomical markers were removed from the participant and pedals.

For dynamic trials, a total of six conditions with 3 seat positions (close, medium and far) and 2 workrates (60 and 100 Watts) were tested in the study. The "far", "medium" and "close" seat positions had knee extension angles of 20-30 degrees, 30-40 degrees and 40-50 degrees, respectively. The seat positions were randomized first. Within a certain seat position, the randomization of the two workrates was followed. Participants were asked to grab the handlebars on the sides of the ergometer and maintain a cadence of 80 RPM (±2 RPM) during all test conditions. Before the actual testing, they were allowed to pedal at least two minute at the middle seat position with a cadence of 80 RPM and workrate of 80 W to allow participants to acclimate to the testing protocol. After the practice trials, participants then cycled one minute for each condition. The actual recording of the kinematics and kinetics data started at the 48th second until the end of each minute. The final10 seconds of the cycling movement was chosen to ensure at least five continuous pedaling cycles collected for individual trials. Participants took a minimum of 2 minutes of rest between conditions and drank water whenever they needed to minimize fatigue and dehydration. After each condition, they were asked to provide the rating of perceived exertion (RPE) (79) to evaluate the perceived intensity of the test condition.

DATA AND STATISTICAL ANALYSES

The 3D marker trajectories were first examined and processed in Nexus. The mislabeled markers were relabeled and the marker gaps were filled by the means of either rigid body fill or pattern fill, and the ghost markers were deleted. For each condition of each participant, the ten seconds of trajectory was truncated into five individual trials with each cycle starts and ends at a crank angle of 270°. This starting crank angle was chosen by examining knee, ankle and hip extension moments to ensure the peaks of these moments occurring during the power phase (first 180° of the crank cycle), which is different from the traditional starting crank angle of 0° for upright bike due to the nature of the recumbent bike.

The marker trajectory data then were exported from Nexus and imported into Visual 3D (Version 2.6, C-Motion, Inc., Germantown, MD, USA) to calculate the 3D kinematic and kinetic variables. The computation of the joint angles followed an X-Y-Z Cardan rotation sequence. A right-hand rule was applied to determine the polarity of the joint angles and moments. Positive values represented knee extension, adduction, internal rotation; ankle dorsiflexion, inversion, internal rotation and hip flexion, adduction, internal rotation angles and moments. A 4th order low-pass Butterworth filter with zero lag at a cutoff frequency of 6 Hz was used to filter both raw kinematics and PRF data (37). In order to determine the critical peak values of the important variables and organize them for statistical analyses, customized programs (VB_V3D and VB_Tables, MS VisualBASIC 6.0) were used.

A 3 × 2 (seat position × workrates) repeated measures analysis of variance (ANOVA) was used to examine the effect of seat positions and workrates on selected variables of interest (Version 25, IBM SPSS Statistics, Chicago, IL). An alpha level of 0.05 was set a priori for the ANOVAs. When a significant interaction or a seat position main effect was present, a post-hoc analysis using a pairwise t-test was followed with Bonferroni adjustments to test specific differences between seat positions at different workrates and seat positions. The adjusted p values were 0.008 for post hoc analysis for interaction, and 0.016 for post hoc analysis for seat position.

CHAPTER IV

EFFECTS OF WORKRATE AND SEAT POSITION ON FRONTAL AND SAGITTAL PLANE KNEE BIOMECHANICS IN RECUMBENT CYCLING

ABSTRACT

In cycling study, there is limited research on recumbent bike kinetics, especially in the frontal plane. Increased internal knee abduction moment (KAbM) has been shown to be an effective predictor of knee osteoarthritis. The purpose of this study was to examine the effects of different workrates and seat positions on knee biomechanics during stationary recumbent cycling. Fifteen participants cycled on a recumbent ergometer in 6 test conditions of pedaling in far, medium and close seat positions in each of two workrates of 60 and 100 W, at the cadence of 80 RPM. A three-dimensional motion analysis system and a pair of custom-made instrumented pedals were used to collect kinematic and kinetic data. A 3 × 2 (seat position × workrate) repeated measures analysis of variance was used to examine the effect of seat positions and workrates on selected variables of interest. Increased workrates significantly increased peak KAbM and knee extension moment. Different seat positions did not change either peak KAbM or knee extension moment. Due to the larger Q-factor for the recumbent bike used in the study, future study should examine the knee biomechanics with smaller Q-factors, as well as the lower limb muscle activities in recumbent cycling.

Keywords: recumbent cycling, knee OA, knee abduction moment, knee extension moment

INTRODUCTION

Cycling is a popular mode of transportation, recreation, sport and rehabilitation. Research has shown that cycling can improve cardiorespiratory fitness (3), strengthen knee flexor and extensor muscles (4, 5), reduce cancer mortality (3), obesity morbidity (3), and depression (6). According to Kutzner et al. (7), cycling is also a preferred exercise over walking or running for individuals with knee osteoarthritis (OA) and anterior cruciate ligament injuries since it is advantageous at lessening knee joint loads.

On a recumbent bike, the rider is allowed to pedal in a reclined position with large and padded backrest, in order to decrease the intervertebral disc compression and help with symptoms such as perineal numbness, erectile dysfunction, handlebar palsy and carpal tunnel syndrome (17-24). Due to its multiple advantages over traditional upright bikes, recumbent bicycle has become preferred exercise and rehabilitation tool in recent years. Several studies have reported decreased knee loads in recumbent bike compared to upright bike, reflected by the reduced peak knee extension moments (11-15). Additionally, stationary recumbent cycling has been used as a rehabilitation and injury treatment method for people with cerebral palsy (26, 65), cerebral vascular accident (27, 28), diabetes (13), spinal cord injuries (29-31) and ankle immobilization (12).

Despite the numerous benefits, there is a risk of suffering overuse injuries and diseases during stationary cycling (8, 9), most commonly at the knee (10). OA is the most common joint disease in the US and over 80% of the cases affect the knee joint (32, 33). The most important variable that is associated with knee OA is the external knee adduction moment, also known as the internal knee abduction moment (KAbM). Together with knee extension moment, they represent the medial compartment loading of the knee. During walking, knee OA patients showed a greater than normal peak KAbM (34), which makes KAbM an effective predictor of knee OA progression (35). Due to the reduced knee joint load, cycling is considered as a well-suited exercise for OA patients. However, only a very limited number of studies has investigated frontal plane knee biomechanics (including KAbMs) during upright cycling. Knee frontal plane kinetics has never been examined in recumbent cycling. Johnson et al. (25) only reported data about frontal plane kinematics in recumbent cycling in teenagers with and without cerebral palsy. During upright cycling, the peak KAbM were reported to range from 7.8 Nm to 24.5 Nm while the peak knee adduction moment ranged from 2.9 to 8.1 Nm (36-40). The large variability in the KAbM may be mostly due to the large variation of workrates (80 to 225 W) used in the studies. Fang et al. (36) indicated that an increased workrate increased peak KAbM in upright cycling. Besides the effect of workrate, knee biomechanical variables can be influenced by the seat position as well. In a recumbent ergometer, the seat position is usually controlled by different notches. Therefore, the options of seat adjustments are limited. The existing literatures only examined knee biomechanics when participants pedaled at different backrest angles in recumbent bike. Reiser et al. (44) did not find significant difference in knee ROM with varied angles of torso reclining. Brown et al. (11) showed a significant increase of mean knee moment during one entire pedaling cycle when the back rest angle increased from 0 degree to 80 degree and a mean peak knee extension moment of about 30 Nm with a pedaling resistance of 15 N and cadence of 60 RPM. In an upright bike study, Hummer et al. (41) examined KAbM in upright cycling at two workrates (80 and 120 W) and 3 saddle heights (20, 30 and 40 degrees of maximum knee extension angle). No significant differences were found across different saddle heights at either of the workrate. For the sagittal plane, the peak knee flexion moment was increased and the peak knee extension moment was decreased as the saddle height increased (41).

To our knowledge, no study has investigated how different workloads and seat positions affect frontal plane knee kinetics and kinematics in recumbent cycling. Comprehensive understanding of knee biomechanics, especially frontal plane joint moments, is necessary to provide evidence for prescribing recumbent cycling as the therapeutic intervention and rehabilitation tool. Therefore, the purpose of the study was to examine effects of different workrates and seat positions on knee frontal and sagittal plane biomechanics during stationary recumbent cycling. It was first hypothesized that an increased workrate would result in a larger peak knee abduction moment and extension moment. It was also hypothesized that a closer seat position would result in no changes in peak knee abduction moment but an increased knee extension moment.

METHODS

Participants

Fifteen experienced and healthy cyclists (age: 55.5±3.7 years, height: 1.75±0.09 m, mass: 84.3±15.7 kg) participated in the study. All participants were free of injury in the lower extremities for the past six months. Each participant spent at least six hours in cycling on a weekly basis. A sample size of 18 was approximated with an effect size of 0.59, alpha level of 0.05 and beta level of 0.8 in a 3 x 2 ANOVA design using G*Power (3.1) based on the knee abduction moment data of Hummer et al. (41). A written informed consent approved by the University of Tennessee Institutional Review Board was read and signed by each participant before the data collection.

Instrumentation

A 12-camera three-dimensional (3D) motion capture system (240 Hz, Vicon, Oxford, UK) was used to collect kinematics data during the test. Reflective anatomical markers were attached to the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, greater trochanter, and iliac crest of both sides of the body. Four non-collinear reflective tracking markers grouped as a cluster on a semi-rigid thermoplastic shells were placed on the pelvis, both thighs, and both legs. The two-marker clusters were placed on the pelvis anteriorly due to the need of proper tracking. For the feet, four individual reflective tracking markers were placed at the posterior and lateral heel counter of each shoe. One pedal anatomical marker was secured in the middle of the front side of each pedal. Three pedal tracking markers were put on the lateral side of each pedal, one additional tracking marker was placed on the anterior-interior side of pedal. One reflective marker was attached on each side of the crank axis as well as the front of the recumbent bike (Figure 2b).

A Recumbent Ergometer (RE7, Kettler, Ense-Parsit, Germany) with electromagnetic brake system was used in the data collection (Figure 2a). There are 12 notches along the sloping support frame allowing seat position adjustments. The angle of recline of the backrest can be altered as well but was kept at the default angle. Both the workload and cadence were shown on the bicycle monitor in front of the participants. A customized jig was used to secure the recumbent ergometer to the floor so that the axes of the pedal coordinate system and the lab coordinate system were aligned parallel to each other. Two customized instrumented pedals were utilized to measure 3D pedal reaction forces and moments (Figure 1b). Two 3D force sensors (Type 9027C, Kistler, Switzerland) paired with two amplifiers (Type 5073A, Kistler, Switzerland) were mounted on the each pedal in order to measure the pedal reaction force (PRF) data bilaterally (36) (37). The outputs from the force sensors were converted to voltages by the charge amplifiers and sampled at 1200 Hz simultaneously with the 3D kinematic data using Nexus (2.7, Vicon, Oxford, UK).

Procedures

A static calibration trial was taken before the actual data collection. A total of six test conditions with 3 seat positions (far, medium and close) and 2 workrates (60 and 100 Watts) were tested in the study. The far, medium and close seat positions were determined to target the peak knee extension angle to fall between 20-30 degrees, 30-40 degrees and 40-50 degrees, respectively. The order of the testing condition was determined such that the seat positions were randomized first, followed by the randomization of the two workrates for each seat position. Participants were asked to grab the handlebars on the sides of the ergometer and maintain a cadence of 80 RPM (±2 RPM) during all test conditions. Before the actual testing, participants were allowed to pedal at least one minute at the preferred seat position with a cadence of 80 RPM and workrate of 60 W to acclimate to the testing protocol. After the practice, participants then cycled one minute for each condition. The actual recording of the kinematics and kinetics data started at the 48th second until the end of each minute to obtain at least five continuous pedaling cycles. Participants took at least 1 minute of rest between conditions and drank water whenever they needed to minimize fatigue and dehydration. After each condition, they were asked to provide the rating of perceived exertion (RPE) (79) to evaluate the perceived intensity of the test condition.

Data and Statistical Analysis

The 3D marker trajectories were first examined and processed in the Nexus of the Vicon system. For each condition of each participant, the 12 seconds of trajectory data were truncated into five individual cycles/trials for analysis. Each cycle starts and ends at the 270° of the crank angle. This starting crank angle was chosen by examining knee, ankle and hip extension moments to ensure these peaks of these moments occurring during the power phase (first 180° crank cycle), which is different from the traditional starting crank angle of 0° for upright bike due to the nature of the recumbent bike.

The marker trajectory data then were exported from the Nexus to Visual 3D (Version 2.8, C-Motion, Inc., Germantown, MD, USA) to calculate the 3D kinematic and kinetic variables. The computation of the joint angles was computed following an X-Y-Z Cardan rotation sequence. A right-hand rule was applied to determine the polarity of the joint angles and moments. Positive values represented knee extension, adduction, internal rotation; ankle dorsiflexion, inversion, internal rotation and hip flexion, adduction, internal rotation angles and moments. A 4^{th} order low-pass Butterworth filter with zero lag was used to filter both raw kinematics and PRF data at a cutoff frequency of 6 Hz (37). In order to determine the critical peak values of the important variables and organize them for statistical analyses, customized programs (VB_V3D and VB_Tables, MS VisualBASIC 6.0) were used.

A 3 × 2 (seat position × workrate) repeated measures analysis of variance (ANOVA) was used to examine the effect of seat positions and workrates on selected variables of interest (Version 25, IBM SPSS Statistics, Chicago, IL). An alpha level of 0.05 was set a priori for ANOVA. When a significant interaction or a seat position main effect was present, a post-hoc analysis using a pairwise t-test was followed with Bonferroni adjustments to test specific differences between seat positions at different workrates and seat positions. The adjusted p values were 0.008 for post hoc analysis for interaction, and 0.016 for post hoc analysis for seat position.

RESULTS

Significant main effects of workrate and seat position were found for RPE (Table 2). The RPE was larger at 100 W than 60 W ($p< 0.001$). However, post hoc comparisons did not show significant difference of RPE between specific seat positions. A significant main effect of workrate was found for peak vertical, anterior and medial PRF (Table 2). These peak forces were higher at 100 W compared to 60 W.

A significant main effect of workrate was found for peak knee extension moment, peak knee abduction moment, peak ankle plantarflexion moment, peak ankle abduction moment, and peak hip abduction moment (all $P \leq 0.008$, Table 3). The magnitudes of all these variables were all higher at 100 W compared to 60 W. There was a significant main effect of seat position only for peak knee flexion moment (Table 3). The post hoc comparison showed that peak knee flexion moment was higher in the far seat position compared to medium and close seat position (both $p < 0.001$). In addition, the peak flexion moment was higher in the medium seat position than close position ($p < 0.001$).

There were significant main effects of workrate and seat position on peak knee extension angle (Table 4). The peak knee extension angle was greater at 60 W than 100 W. Post hoc comparison showed that the peak knee extension angle was higher in the close position compared to medium and far positions (both $p < 0.001$). Moreover, the peak knee extension angle was higher in the medium position than far position ($p < 0.001$). Significant main effects of workrate and seat position were also found on peak knee extension ROM (Table 4). Knee extension ROM was greater at 100 W than 60 W. Post hoc results indicated that the peak knee extension ROM was greater in the far position than the medium and the close position (both $p < 0.001$), and was higher in the medium position than the close position ($p < 0.001$). Lastly, there was a significant main effect of seat position for the knee abduction ROM (Table 4). Post hoc results showed that the knee abduction ROM was significantly higher in the far position than close position (p = 0.004). In addition, knee abduction ROM was significantly higher in the medium position than close position (p = 0.002).

DISCUSSION

The purpose of the study was to examine the effects of different workrates and seat positions on knee biomechanics during stationary recumbent cycling amongst middle aged and old cyclists. We first hypothesized that an increased workrate would result in an increased peak KAbM and peak knee extension moment. The hypothesis was supported by our results.

Our results showed that peak KAbM increased as the workrate increased. Peak KAbM moment, along with peak knee extension moment, represent the knee medial compartment loading. The increased KAbM accompanied with increased peak knee extension moment caused the possible increased medial compartment loading with the increased workrate. The increased peak knee extension moments are also reflected in the increased peak vertical (the major component with larger magnitude) and medial PRF. Studies on frontal plane knee joint kinetics in recumbent back is lacking in the literature. Several studies have reported the frontal plane knee kinetics in upright stationary cycling. Fang et al. (36) examined the effects of workrate and cadence on frontal plane knee biomechanics. At the workload of 1kg, the mean KAbMs were 7.0 Nm when the participants cycled at 80 RPM. The peak KAbM increased 63%, 7%, 14%, and 24% when the workload increased from 0.5 to 1 kg, 1 to 1.5 kg, 1.5 to 2 kg, and 2 to 2.5 kg, at cadence of 60 RPM, respectively. Hummer et al. (41) also reported a significant increase of peak KAbM when the cycling workrate increased from 80 W to 120 W at cadence of 80 RPM. The peak KAbM ranged from 10.2 to 13.7 Nm when participants pedaled at different workrates (80-

120 W) and saddle heights (peak knee extension angle of 20-40 degrees). These results about how peak KAbM tends to change with increased workrate are in agreement with our results.

Several other studies (37, 39, 40) showed that the peak KAbM ranged from 4.8 to 9.0 Nm on the upright bike (with workrate of 40-225 W and cadence of 60-90 RPM). In our study, the peak KAbM ranged from 10.8 to 15.6 Nm across all conditions (workrate of 60-100 W and seat position of 24.3-46.4 degrees of peak knee extension angle), which did not show advantages over upright bikes. The slightly larger KAbM in recumbent bike might be caused by the several factors. Thorsen et al. (80) showed that an increased Q-factor (the intra-pedal distance between the outside surface of one crank arm to the outside surface of the crank arm on the opposite side) caused increases in KAbM in upright stationary cycling. The Q-factor of the recumbent ergometer in our study is 20.3 cm while it is only 14.5 cm for the upright cycle ergometer (Excalibur Lode Ergometer) used in the study by Hummer et al. (41). Another potential contributor to the larger KAbM is body mass of the participants. In our study, the average mass of the participants of 84.3 kg was larger than that in most of the upright cycling literature (ranged from 73.1 kg to 80.1 kg) (36, 37, 39, 40). Since our cycling moment values were not normalized to body mass, it is possible that these moment values may be more affected by the body mass. The KAbM is a predictor of knee OA progression (35) and therefore the recommendation of recumbent bike usage for knee OA patients should consider Q-factor. Further study on the frontal plane knee loads in recumbent bike with different Q-factors is recommended.

Peak knee extension moment also increased significantly as the workrate increased. During the power phase of the cycling, the knee extends to produce sufficient torque to overcome the pedal resistance and to assist the opposite leg during its recovery phase. Our results also showed that both vertical and posterior PRF significantly increased with the increased workrate, explaining the increased peak knee extension moment. During cycling, the knee extension moment is the most important and driving moment that powers the cycling motion. When the workrate increased, the participants needed to exert greater knee extension moment to overcome increased resistance.

A previous recumbent cycling study showed a mean peak knee extension moment of about 30 Nm with a pedaling resistance of 15 N and cadence of 60 RPM (11). Another study reported a peak knee extension moment of 24.7 Nm at the workrate of 80 W (12). These results are similar to the peak knee extension moments found in our study, which ranged from 20.7 Nm (far seat position at 60 W) to 34.6 Nm (close seat position at 100 W). In upright cycling, the peak knee extension moment could be as high as 53 Nm (42). The main contributor of the knee extension moment in recumbent and upright cycling are somewhat different. In the studies by Fang et al. (36) and Hummer et al. (41), the magnitudes of the vertical PRF were about 3 times as large as that of the posterior PRF. However, our results show that in recumbent cycling, the magnitudes of the vertical PRF and posterior PRF were very similar and in most of the conditions, the posterior PRFs were even slightly larger. The primary pedaling direction of power phase in recumbent cycling is mostly horizontal whereas the primary pedaling direction during the same power phase in upright cycling is vertical, which are reflected by the different magnitudes in the respective vertical and posterior PRFs. The increase peak extension moments are also supported by the increased RPEs reported by the participants. Our participants reported their RPEs ranging from 7 to 10, representing very light to light exertion. It is also worth mentioning that all the participants are experienced cyclists, and therefore their perceived ratings could be lower than what the regular population would report. For patients with knee pathologies, RPE in riding a recumbent ergometer at similar workrate, cadence and seat conditions could be higher than the light exertion reported by experienced older cyclists. However, it is unknown if perceived exertions would be lower in riding a recumbent ergometer compared to riding an upright ergomter.

Our second hypothesis stating that a closer seat position (with more knee flexion) would not result in a different peak KAbM, but would result in an increased peak knee extension moment. This hypothesis was partially supported by the results in that a decreased seat position did not result in significant changes in peak KAbM and peak knee extension moment. In upright cycling, Hummber et al. (41) reported that when healthy participants cycled at 80 RPM, no significant differences were found in peak KAbMs at three different saddle heights (20, 30 and 40 degrees of knee angles), which is in line with our results. Main contributors of the KAbM are vertical and medial PRFs, and neither of these variables was significantly changed by seat positions. However, the medial PRF at the close position did show a 17.2% increase than that at the far position. This result suggests that patients with knee OA may have some flexibility when they pick seat positions in exercise on a recumbent bike without worrying about increased medial knee loading.

Peak knee extension moment did not change with seat positions. However, Hummer et al. (41) reported that the knee extension moment significantly decreased when participants pedaled at a more knee extended position. A factor for this result may be related to the different peak knee extension angles reached in the current study: at the “far”, “medium” and “close” positions, the peak knee extension angles are 25.4, 34.2 and 45.1 degrees. Even though we had the similar increment about 10 degrees as the upright cycling saddle positions of 20, 30 and 40 degrees (desired positions), it is difficult to make direct comparisons as Hummer et al. (41) did not report the actual peak knee angles achieved in the three saddle height positions. The seat position in our recumbent ergometer is controlled by equal distance notches (2.7 cm between each notch). This design feature made adjustments of seat positions limited and more difficult to achieve desired peak knee angle for each of the three seat positions, as the desired knee angle is not only influenced by the seat position but also by different body height, and relative thigh and leg lengths of our participants.

Another interesting finding is that there was a main effect of seat position on the peak knee flexion moment in recumbent bicycle. At the workrate of 60 and 100 W and cadence of 80 RPM, the peak knee flexion moment was only 7.3 Nm at the flexed position, but was up to 18.9 Nm at the far position, which is almost 257% of the magnitude. In recumbent cycling, when the seat gets farther away from the pedal, the knee extensors would have difficulty in completing the transition from the power phase to the recovery phase. At the same time, the contralateral limb needs to rely on knee flexors to exert flexion moment and drive the pedal forward to transition from the recovery phase to power phase. During recumbent cycling, the peak knee flexion moment occurred around 50% of the crank cycle (Figure 3d), while the peak ankle planterflexion moment occurred almost at the same time to assist the transition from the power into the recovery phase (Figure 3e). In addition, a larger knee flexion moment is usually coupled with increased muscle activation of knee flexors. In the study by Hummer et al. (41), a more extended (farther) seat position showed a significantly larger the knee flexion moment, along with increased semitendinosus muscle activity. Future studies may be needed to investigate electromyographic activities of knee extensors and flexors in recumbent cycling at different workrates and seat positions.

There are a few limitations of this study. As mentioned before, the seat position of the recumbent ergometer is controlled by fixed notches, which made it difficult for us to control each participant’s peak knee extension angle same at respective positions (close, medium and far). The number of cyclists participated did not fully meet the desired sample size, reducing the statistical power of the key variables (e.g. peak knee extension moment). In addition, even though all the participants were experienced cyclist, some of them had more experience than the others, which might have led to different pedaling habits and techniques.

CONCLUSION

The findings of this study indicate that increased workrate significantly increased KAbM and peak knee extension moment. However, as seat position was adjusted, neither KAbM nor peak knee extension moment was changed. This study is the first study to examine the effects of workrate and seat position on frontal plane knee biomechanics in recumbent cycling. For patients with knee OA, a low workrate should be selected in recumbent cycling exercises, and the seat position should be chosen based on personal preference. In addition, using a recumbent ergometer with smaller Q-factor could be more beneficial. Future study should investigate the knee biomechanics with different Q-factors as well as the lower limb muscle activities in recumbent cycling.

LIST OF REFERENCES

1. [Internet]. [cited 2018. Available from: https://www.statista.com/topics/1686/cycling/.
2. What kind of bicycle do you have in your household? 2017.
3. Oja P, Titze S, Bauman A et al. Health benefits of cycling: a systematic review. *Scand J Med Sci Sports*. 2011;21(4):496-509.
4. Chen CL, Hong WH, Cheng HY, Liaw MY, Chung CY, Chen CY. Muscle strength enhancement following home-based virtual cycling training in ambulatory children with cerebral palsy. *Res Dev Disabil*. 2012;33(4):1087-94.
5. Macaluso A, Young A, Gibb KS, Rowe DA, De Vito G. Cycling as a novel approach to resistance training increases muscle strength, power, and selected functional abilities in healthy older women. *J Appl Physiol (1985)*. 2003;95(6):2544-53.
6. Salmon P. Effects of physical exercise on anxiety, depression, and sensitivity to stress: a unifying theory. *Clin Psychol Rev*. 2001;21(1):33-61.
7. Kutzner I, Heinlein B, Graichen F et al. Loading of the knee joint during ergometer cycling: telemetric in vivo data. *J Orthop Sports Phys Ther*. 2012;42(12):1032-8.
8. Clarsen B, Krosshaug T, Bahr R. Overuse injuries in professional road cyclists. *Am J Sports Med*. 2010;38(12):2494-501.
9. Dettori NJ, Norvell DC. Non-traumatic bicycle injuries : a review of the literature. *Sports Med*. 2006;36(1):7-18.
10. Kulund D, Brubaker C. Injuries in the Bikecentennial tour. *Phys Sportsmed*. 1978;6(6):74-8.
11. Brown DA, Kautz SA, Dairaghi CA. Muscle activity patterns altered during pedaling at different body orientations. *J Biomech*. 1996;29(10):1349-56.
12. Szecsi J, Straube A, Fornusek C. Leg general muscle moment and power patterns in able-bodied subjects during recumbent cycle ergometry with ankle immobilization. *Med Eng Phys*. 2014;36(11):1421-7.
13. Perell KL, Scott G, Kim G, Rushatakankovit S. Comparison of cycling kinetics during recumbent bicycling in subjects with and without diabetes. *Journal of rehabilitation research and development*. 2002;39(1):13.
14. Gregor SM, Perell KL, Rushatakankovit S, Miyamoto E, Muffoletto R, Gregor RJ. Lower extremity general muscle moment patterns in healthy individuals during recumbent cycling. *Clin Biomech (Bristol, Avon)*. 2002;17(2):123-9.
15. Ericson MO, Bratt A, Nisell R, Nemeth G, Ekholm J. Load moments about the hip and knee joints during ergometer cycling. *Scand J Rehabil Med*. 1986;18(4):165-72.
16. Nolan PK [Internet]. Available from: http://www.bikeroute.com/Recumbents/BentMedBenefits.php.
17. Salai M, Brosh T, Blankstein A, Oran A, Chechik A. Effect of changing the saddle angle on the incidence of low back pain in recreational bicyclists. *Br J Sports Med*. 1999;33(6):398-400.
18. Mellion MB. Common cycling injuries. Management and prevention. *Sports Med*. 1991;11(1):52-70.
19. Schwellnus MP, Derman E. Common injuries in cycling: Prevention, diagnosis and management. *South African Family Practice*. 2005;47(7):14-9.
20. Huang V, Munarriz R, Goldstein I. Bicycle riding and erectile dysfunction: an increase in interest (and concern). *J Sex Med*. 2005;2(5):596-604.
21. Jeong SJ, Park K, Moon JD, Ryu SB. Bicycle saddle shape affects penile blood flow. *Int J Impot Res*. 2002;14(6):513-7.

22. Lowe BD, Schrader SM, Breitenstein MJ. Effect of bicycle saddle designs on the pressure to the perineum of the bicyclist. *Med Sci Sports Exerc*. 2004;36(6):1055-62.
23. Schwarzer U, Sommer F, Klotz T, Cremer C, Engelmann U. Cycling and penile oxygen pressure: the type of saddle matters. *Eur Urol*. 2002;41(2):139-43.
24. Capitani D, Beer S. Handlebar palsy--a compression syndrome of the deep terminal (motor) branch of the ulnar nerve in biking. *J Neurol*. 2002;249(10):1441-5.
25. Johnston TE, Barr AE, Lee SC. Biomechanics of submaximal recumbent cycling in adolescents with and without cerebral palsy. *Phys Ther*. 2007;87(5):572-85.
26. Johnston TE, Barr AE, Lee SC. Biomechanics of recumbent cycling in adolescents with cerebral palsy with and without the use of a fixed shank guide. *Gait Posture*. 2008;27(4):539-46.
27. Perell KL, Gregor RJ, Scremin AE. Lower limb cycling mechanics in subjects with unilateral cerebrovascular accidents. *Journal of Applied Biomechanics*. 1998;14(2):158-79.
28. Perell KL, Gregor RJ, Scremin AE. Bicycle pedal kinetics following force symmetry feedback training in subjects with unilateral cerebrovascular accident. *Journal of Applied Biomechanics*. 2000;16(2):124-41.
29. Hooker SP, Figoni SF, Rodgers MM et al. Physiologic effects of electrical stimulation leg cycle exercise training in spinal cord injured persons. *Arch Phys Med Rehabil*. 1992;73(5):470-6.
30. Nash MS, Bilsker S, Marcillo AE et al. Reversal of adaptive left ventricular atrophy following electrically-stimulated exercise training in human tetraplegics. *Paraplegia*. 1991;29(9):590-9.
31. Faghri PD, Glaser RM, Figoni SF. Functional electrical stimulation leg cycle ergometer exercise: training effects on cardiorespiratory responses of spinal cord injured subjects at rest and during submaximal exercise. *Arch Phys Med Rehabil*. 1992;73(11):1085-93.
32. Murray CJ, Atkinson C, Bhalla K et al. The state of US health, 1990-2010: burden of diseases, injuries, and risk factors. *JAMA*. 2013;310(6):591-608.
33. Vos T, Flaxman AD, Naghavi M et al. Years lived with disability (YLDs) for 1160 sequelae of 289 diseases and injuries 1990-2010: a systematic analysis for the Global Burden of Disease Study 2010. *Lancet*. 2012;380(9859):2163-96.
34. Baliunas AJ, Hurwitz DE, Ryals AB et al. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis Cartilage*. 2002;10(7):573-9.
35. Miyazaki T, Wada M, Kawahara H, Sato M, Baba H, Shimada S. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. *Ann Rheum Dis*. 2002;61(7):617-22.
36. Fang Y, Fitzhugh EC, Crouter SE, Gardner JK, Zhang S. Effects of Workloads and Cadences on Frontal Plane Knee Biomechanics in Cycling. *Med Sci Sports Exerc*. 2016;48(2):260-6.
37. Gardner JK, Zhang S, Liu H et al. Effects of toe-in angles on knee biomechanics in cycling of patients with medial knee osteoarthritis. *Clin Biomech (Bristol, Avon)*. 2015;30(3):276-82.
38. Ericson M, Nisell R, Ekholm J. Varus and valgus loads on the knee joint during ergometer cycling. *Scand J Sports Sci*. 1984;6(2):39-45.

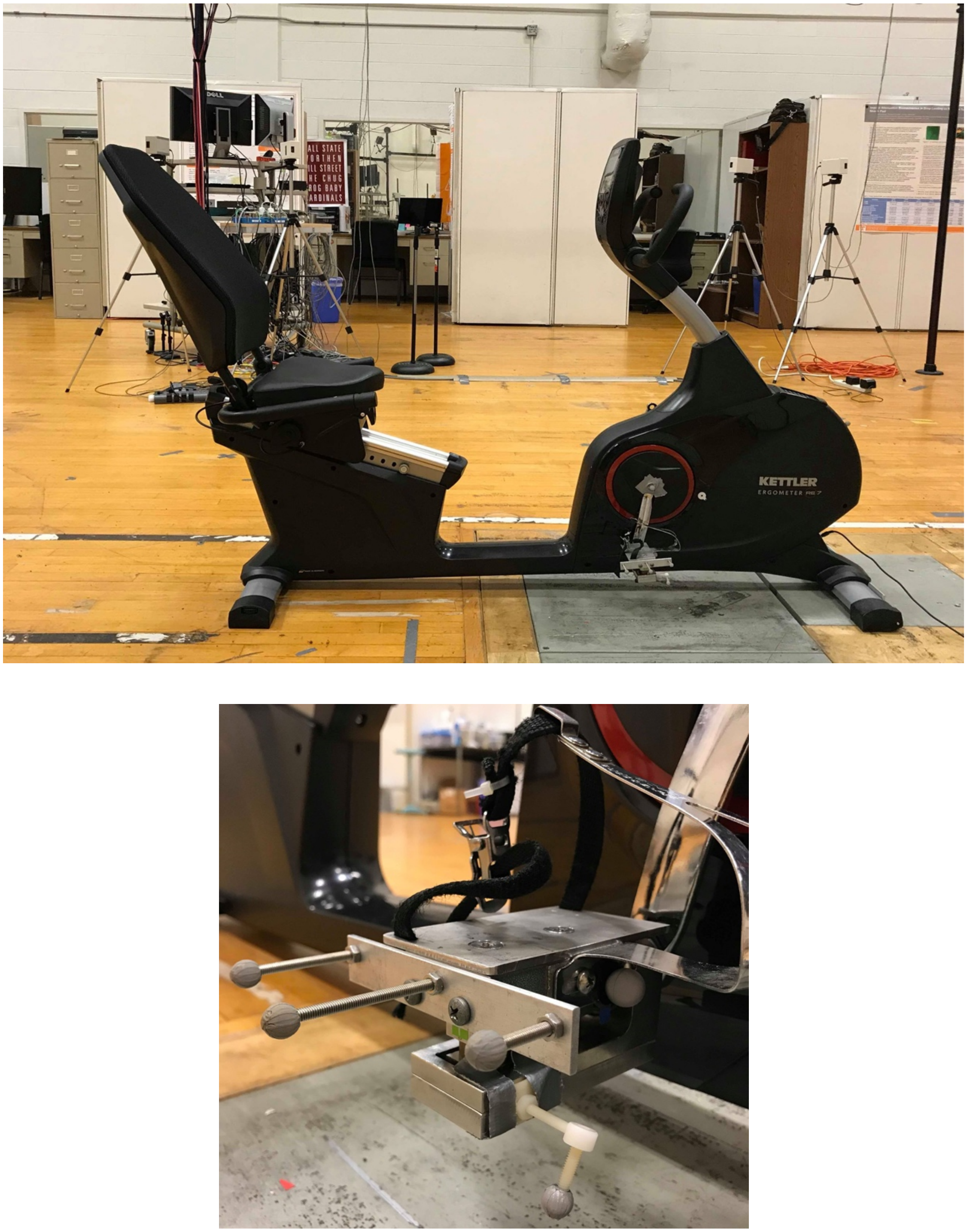
39. Gregersen CS, Hull ML. Non-driving intersegmental knee moments in cycling computed using a model that includes three-dimensional kinematics of the shank/foot and the effect of simplifying assumptions. *J Biomech*. 2003;36(6):803-13.
40. Shen G, Zhang S, Bennett HJ, Martin JC, Crouter SE, Fitzhugh EC. Effects of Knee Alignments and Toe Clip on Frontal Plane Knee Biomechanics in Cycling. *J Sports Sci Med*. 2018;17(2):312-21.
41. Hummer ET, Thorsen TA, Zhang S. Effects of saddle height and workrate on frontal-plane knee joint biomechanics. *2019 Annual Conference of American College of Sports Medicine, Orlando, Florida*. 2019.
42. Gregor RJ, Cavanagh PR, LaFortune M. Knee flexor moments during propulsion in cycling--a creative solution to Lombard's Paradox. *J Biomech*. 1985;18(5):307-16.
43. Bini RR, Tamborindeguy AC, Mota CB. Effects of saddle height, pedaling cadence, and workload on joint kinetics and kinematics during cycling. *J Sport Rehabil*. 2010;19(3):301-14.
44. Reiser RF, Peterson ML, Broker JP. Anaerobic cycling power output with variations in recumbent body configuration. *Journal of Applied Biomechanics*. 2001;17(3):204-16.
45. Foroughi N, Smith R, Vanwanseele B. The association of external knee adduction moment with biomechanical variables in osteoarthritis: a systematic review. *Knee*. 2009;16(5):303-9.
46. Hunt MA, Birmingham TB, Giffin JR, Jenkyn TR. Associations among knee adduction moment, frontal plane ground reaction force, and lever arm during walking in patients with knee osteoarthritis. *J Biomech*. 2006;39(12):2213-20.
47. Sharma L, Hurwitz DE, Thonar EJ et al. Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. *Arthritis Rheum*. 1998;41(7):1233-40.
48. Hiura M, Nariai T, Ishii K et al. Changes in cerebral blood flow during steady-state cycling exercise: a study using oxygen-15-labeled water with PET. *J Cereb Blood Flow Metab*. 2014;34(3):389-96.
49. Bohlmann JT. Injuries in Competitive Cycling. *Phys Sportsmed*. 1981;9(5):117-24.
50. Gregor RJ, Wheeler JB. Biomechanical factors associated with shoe/pedal interfaces. Implications for injury. *Sports Med*. 1994;17(2):117-31.
51. Holmes JC, Pruitt AL, Whalen NJ. Lower extremity overuse in bicycling. *Clin Sports Med*. 1994;13(1):187-205.
52. Asplund C, St Pierre P. Knee pain and bicycling: fitting concepts for clinicians. *Phys Sportsmed*. 2004;32(4):23-30.
53. Hakansson NA, Hull ML. Functional roles of the leg muscles when pedaling in the recumbent versus the upright position. *J Biomech Eng*. 2005;127(2):301-10.
54. Ercison MO, Nisell R, Nemeth G. Joint Motions of the Lower Limb during Ergometer Cycling. *J Orthop Sports Phys Ther*. 1988;9(8):273-8.
55. Bailey MP, Maillardet FJ, Messenger N. Kinematics of cycling in relation to anterior knee pain and patellar tendinitis. *J Sports Sci*. 2003;21(8):649-57.
56. Too D, Landwer GE. The effect of pedal crank arm length on joint angle and power production in upright cycle ergometry. *J Sports Sci*. 2000;18(3):153-61.
57. Rugg S, Gregor R. The effect of seat height on muscle lengths, velocities and moment arm lengths during cycling. *Journal of Biomechanics*. 1987;20(9):899.

58. Edeline O, Polin D, Tourny-Chollet C, Weber J. Effect of workload on bilateral pedaling kinematics in non-trained cyclists. *Journal of Human Movement Studies*. 2004;46(6):493-517.
59. Neptune RR, Hull ML. Evaluation of performance criteria for simulation of submaximal steady-state cycling using a forward dynamic model. *J Biomech Eng*. 1998;120(3):334-41.
60. Tamborindeguy AC, Bini RR. Does saddle height affect patellofemoral and tibiofemoral forces during bicycling for rehabilitation? *Journal of bodywork and movement therapies*. 2011;15(2):186-91.
61. Davis BL, Cavanagh PR. Simulating reduced gravity: a review of biomechanical issues pertaining to human locomotion. *Aviat Space Environ Med*. 1993;64(6):557-66.
62. Weiss BD. Nontraumatic injuries in amateur long distance bicyclists. *Am J Sports Med*. 1985;13(3):187-92.
63. Kolehmainen I, Harms-Ringdahl K, Lanshammart H. Cervical spine positions and load moments during bicycling with different handlebar positions. *Clin Biomech (Bristol, Avon)*. 1989;4(2):105-10.
64. Too D. The effect of seat-to-pedal distance on anaerobic power and capacity in recumbent cycling. 1993.
65. Johnston TE, Prosser LA, Lee SC. Differences in pedal forces during recumbent cycling in adolescents with and without cerebral palsy. *Clinical Biomechanics*. 2008;23(2):248-51.
66. Telli R, Seminati E, Pavei G, Minetti AE. Recumbent vs. upright bicycles: 3D trajectory of body centre of mass, limb mechanical work, and operative range of propulsive muscles. *J Sports Sci*. 2017;35(5):491-9.
67. Too D, Williams CD. Determination of the Crank-arm Length to Maximize Power Production in Recumbent Cycle Ergometry. *Human Power*. 2000.
68. Leong CH, McDermott WJ, Elmer SJ, Martin JC. Chronic eccentric cycling improves quadriceps muscle structure and maximum cycling power. *Int J Sports Med*. 2014;35(7):559-65.
69. Reiser RF, 2nd, Peterson ML. Lower-extremity power output in recumbent cycling: A literature review. *Human Power*. 1998;13(3):6-13.
70. Reiser RF, 2nd, Broker JP, Peterson ML. Knee loads in the standard and recumbent cycling positions. *Biomed Sci Instrum*. 2004;40:36-42.
71. Kato M, Tsutsumi T, Yamaguchi T, Kurakane S, Chang H. Characteristics of maximum performance of pedaling exercise in recumbent and supine positions. *J Sports Sci Med*. 2011;10(3):491-7.
72. Kerr A, Rafferty D, Moffat F, Morlan G. Specificity of recumbent cycling as a training modality for the functional movements; sit-to-stand and step-up. *Clin Biomech (Bristol, Avon)*. 2007;22(10):1104-11.
73. Adrian MJ, Hamilton NP. Biomechanical investigation of recumbent bicycling. *Journal of Biomechanics*. 1988;21(10):880.
74. DorÉ E, Bedu M, FranÇa NM, Diallo O, DuchÉ P, Van Praagh E. Testing peak cycling performance: effects of braking force during growth. *Medicine & Science in Sports & Exercise*. 2000;32(2):493.

75. Hakansson NA, Hull ML. Influence of pedaling rate on muscle mechanical energy in low power recumbent pedaling using forward dynamic simulations. *IEEE Trans Neural Syst Rehabil Eng*. 2007;15(4):509-16.
76. Reiser RF, 2nd, Peterson ML, Broker JP. Understanding recumbent cycling: instrumentation design and biomechanical analysis. *Biomed Sci Instrum*. 2002;38:209-14.
77. Newmiller J, Hull ML, Zajac FE. A mechanically decoupled two force component bicycle pedal dynamometer. *J Biomech*. 1988;21(5):375-86.
78. Hull ML, Jorge M. A method for biomechanical analysis of bicycle pedalling. *J Biomech*. 1985;18(9):631-44.
79. Borg GA. Psychophysical bases of perceived exertion. *Med Sci Sports Exerc*. 1982;14(5):377-81.
80. Thorsen TA, Zhang S. The Effects of increased q-factor on knee biomechanics during cycling. *UNIVERSITY OF TENNESSEE, KNOXVILLE*. 2018.

APPENDICES

APPENDIX A: FIGURES AND TABLES FOR CHAPTER FOUR



a)

b)

Figure 2. The recumbent ergometer (a) and the instrumented pedal, and anatomical and tracking markers (b) used in the study.

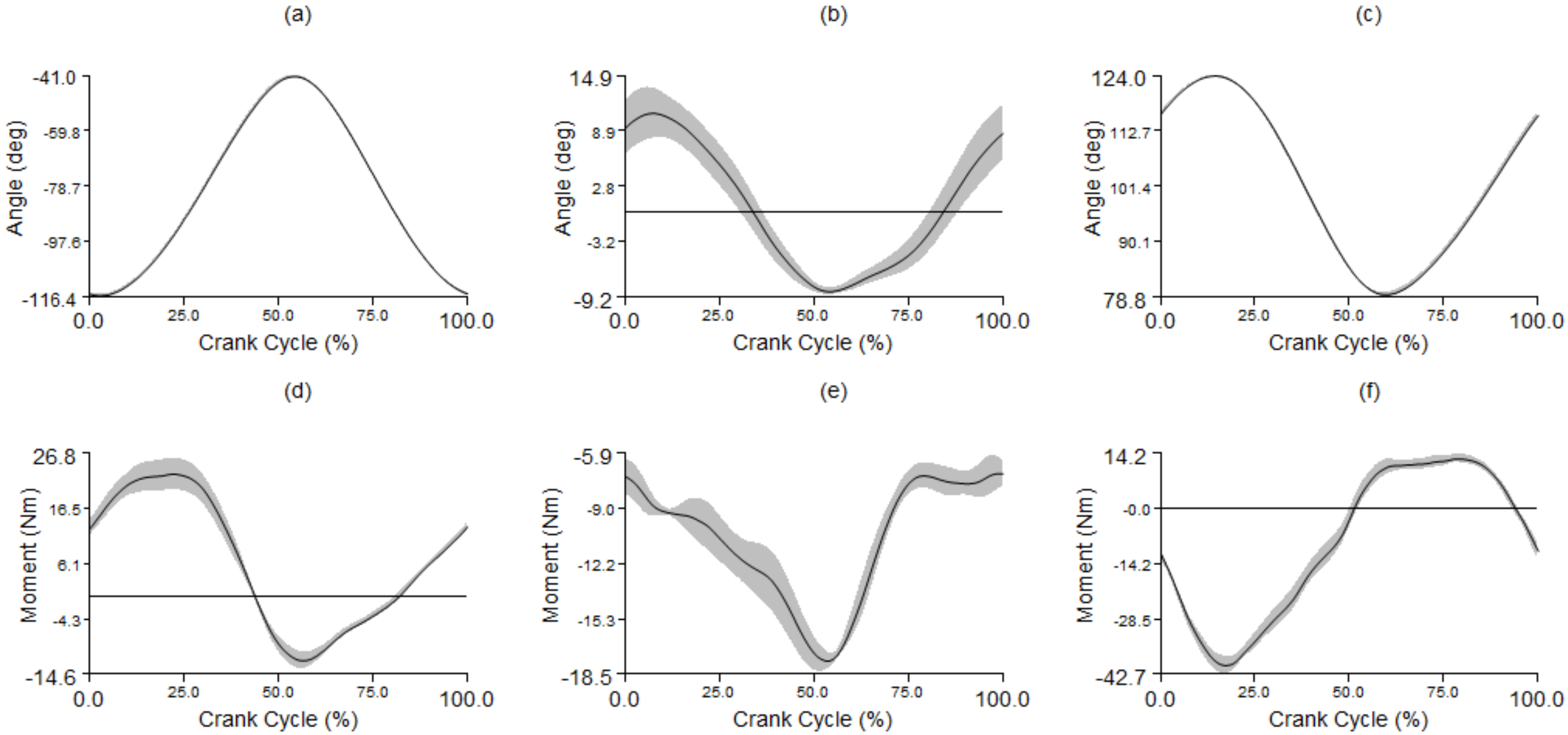


Figure 3. Representative ensemble curves of knee, ankle, hip angle and moment in sagittal plane at workrate of 100 W and medium seat position of a representative subject: (a) knee angle, (b) ankle angle, (c) hip angle, (d) knee moment, (e) ankle moment and (f) hip moment.

Table 1: Subject age (years), height (m), mass (kg), BMI (kg/m^2) and cycling time per week (hr): Mean ± STD

	Mean ± STD
Age	55.53±3.68
Height	1.75±0.09
Mass	84.33±15.68
BMI	27.44±3.73
Cycling Time/Week	7.47±2.29

Table 2: RPE and Mean Peak Pedal Reaction Force (N) at three seat positions and two workrates (W): Mean ± STD

Variables	Workrate	Far	Medium	Close	Interaction	Seat Position	Workrate
RPE	60	8.20±2.54	7.46±1.81	7.93±2.05	0.606	**0.023**	**< 0.001**
	100	10.13±2.62	9.27±2.15	10.07±2.22			
Vertical PRF	60	131.0±28.6	132.8±29.6	140.1±31.1	0.475	0.134	**<0.001**
	100	144.0±30.3	147.7±28.8	148.4±28.3			
Posterior PRF	60	-135.3±26.4	-148.4±27.6	-154.2±34.6	0.943	0.167	**<0.001**
	100	-174.2±29.4	-185.2±21.0	-192.2±31.5			
Medial PRF	60	-30.3±10.1	-35.3±12.9	-35.9±14.4	0.784	0.091	**<0.001**
	100	-41.9±12.7	-48.2±13.4	-48.7±15.3			

Table 3: Peak knee, ankle and hip moment (Nm) at three seat positions and two workrates (W): Mean ± STD

Variables	Workrate	Far	Medium	Close	Interaction	Seat Position	Workrate
Knee Extension Moment	60	20.7±5.6	22.9±6.6	25.2±8.8	0.725	0.132	**<0.001**
	100	28.2±7.3	31.1±6.8	34.6±9.2			
Knee Flexion Moment$^{\#,\$,\%}$	60	-17.8±7.1	-12.3±6.1	-7.20±6.78	0.216	**<0.001**	0.160
	100	-20.0±8.6	-13.9±7.7	-7.35±7.87			
Knee Abduction Moment	60	-10.8±4.0	-12.0±5.2	-12.2±5.9	0.769	0.592	**<0.001**
	100	-14.6±5.5	-15.4±6.4	-15.6±7.5			
Ankle Plantarflexion Moment	60	-17.5±5.1	-17.8±5.1	-18.5±5.3	0.112	0.676	**<0.001**
	100	-20.0±5.2	-20.4±5.2	-19.4±4.6			
Ankle Abduction Moment	60	-3.4±2.1	-3.6±2.4	-3.5±2.5	0.991	0.452	**0.008**
	100	-3.9±2.4	-4.1±2.7	-4.0±3.0			
Hip Flexion Moment	60	-16.7±8.1	-18.5±8.1	-20.6±11.1	0.439	0.434	0.785
	100	-18.8±9.5	-18.4±9.4	-19.3±9.7			
Hip Abduction Moment	60	-14.2±8.3	-14.1±9.2	-14.0±9.2	0.998	0.982	**<0.001**
	100	-17.9±10.3	-17.8±10.6	-17.7±10.1			

Note:

$^{\#}$: significant difference between Far and Medium, $^{\$}$: significant difference between Far and Close, $^{\%}$: significant difference between Medium and Close

Table 4: Peak Knee Angle (deg) and Knee ROM (deg) at three seat positions and two workrates (W): Mean ± STD

Variables	Workrate	Far	Medium	Close	Interaction	Seat Position	Workrate
Extension Angle$^{\#,\$,\%}$	60	26.5±7.3	35.1±10.0	46.5±10.3	0.797	**<0.001**	**<0.001**
	100	24.3±8.7	33.3±9.0	43.8±10.6			
Abduction Angle	60	2.6±3.7	3.5±4.9	5.9±6.7	0.689	0.058	0.063
	100	1.7±4.3	3.2±5.0	5.4±7.0			
Extension ROM$^{\#,\$,\%}$	60	79.7±5.2	74.3±5.0	68.5±5.0	0.892	**<0.001**	**<0.001**
	100	81.6±5.7	76.1±4.9	70.7±5.4			
Abduction ROM$^{\$,\%}$	60	9.0±4.3	8.3±5.2	5.7±4.2	0.420	**0.003**	0.067
	100	10.0±5.5	8.4±4.3	5.9±4.1			

Note:
$^{\#}$: significant difference between Far and Medium, $^{\$}$: significant difference between Far and Close, $^{\%}$: significant difference between Medium and Close

APPENDIX B: INDIVIDUAL PARTICIPANT CHARACTERISTICS

Table 5: Individual participant characteristics.

Subject	Gender	Age (years)	Height (m)	Weight (kg)	BMI (kg/m^2)	Cycling Time/Week (h)
1	M	60	1.81	95.25	29.07	8
2	M	55	1.72	73.48	24.84	6
3	M	56	1.81	102.05	31.15	6
4	M	61	1.81	81.64	24.92	6.5
5	M	50	1.75	106.59	34.80	7
6	F	52	1.65	54.40	19.98	6
7	M	59	1.83	102.05	30.47	7
8	F	53	1.57	70.31	28.52	6.5
9	F	54	1.57	63.50	25.76	7
10	M	53	1.75	89.81	29.33	9.5
11	M	56	1.83	90.72	27.09	6.5
12	M	53	1.75	78.02	25.48	8
13	M	61	1.78	73.48	23.19	15
14	M	51	1.78	81.64	25.77	7
15	M	59	1.81	102.05	31.15	6
Mean±STD		55.53±3.68	1.75±0.09	84.33±15.68	27.44±3.73	7.47±2.29

APPENDIX C: INFORMED CONSENT FORM

Consent for Research Participation

Research Study Title: Effects of Workrate and Seat Position on Frontal Plane Knee Biomechanics in Recumbent Cycling
Researcher(s): Tianyi Lu, University of Tennessee, Knoxville
Tanner Thorsen, University of Tennessee, Knoxville
Faculty Advisor: Dr. Songning Zhang, University of Tennessee, Knoxville

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Why am I being asked to be in this research study?

We are asking you to be in this research study because you have met all the inclusion and exclusion criteria and we believe you will be a good candidate for this study.

What is this research study about?

The purpose of this study is to investigate the effects of different workrates and seat positions of recumbent bicycle on knee frontal plane biomechanics among middle-aged and old adults. The exclusion and inclusion criteria of the study are:

Inclusion criteria:

- Being between the ages of 50 and 70 years old
- Spending about 6 hours per week in cycling

Exclusion criteria:

- Suffering from lower extremity injuries in the past 6 months
- Not being able to ride a stationary bike for at least 20 minutes without aid
- Answering “No” to any question on Par-Q form

How long will I be in the research study?

If you agree to participate, your participation will last approximately 1-1.5 hours.

What will happen if I say “Yes, I want to be in this research study”?

If you agree to be in this study, we will ask you to:

- Change into appropriate clothing provided by either yourself or the lab.
- Complete a brief 2-minute cycling warmup on a recumbent bicycle.
- Be fitted with retroreflective markers and have a calibration trial taken.
- Complete 1 minute of successful cycling trials per each of 6 test conditions, including 2 workrates and 3 seat positions. The 2 workrates are 80 and 120W and the 3 seat positions are close, middle and far positions.
- Take a minimum of 2-minute rest between conditions and drink water whenever you need to minimize fatigue and dehydration.

What happens if I say “No, I do not want to be in this research study”?

Being in this study is up to you. You can say no now or leave the study later at any time.

What happens if I say “Yes” but change my mind later?

Even if you decide to be in the study now, you can change your mind and stop at any time.

If you decide to stop before the study is completed, please inform the primary investigator to end your participation. Once the primary investigator is informed, your collected data, and any data identifying you directly will be destroyed immediately.

Are there any possible risks to me?

Potential risk associated with this study is minimal. Since recumbent 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 20 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. You will be able to end the test at any time if they feel uncomfortable. The attachment of the reflective markers to skin will unlikely cause skin irritation. The researchers are also certified in first aid to render care if needed. It is also possible that someone could find out you were in this study or see your study information, but we believe this risk is small because of the procedures we use to protect your information. These procedures are described later in this form.

Are there any benefits to being in this research study?

There is a possibility that you may benefit from being in the study, but there is no guarantee that will happen. Possible benefits include the identification of any possible abnormalities of cycling pattern as a result of their participation in the study which may serve as valuable information for correcting these abnormalities. Even if you don’t benefit from being in the study, the data collected from you will help provide a better understanding of how different seat positions and workrates would affect the knee frontal plane biomechanics in recumbent cycling. Comprehensive understanding of knee biomechanics, especially frontal-plane joint moment, is necessary to provide guidelines for prescribing recumbent cycling as a therapeutic intervention and rehabilitation tool. We hope the knowledge gained from this study will benefit others in the future.

Who can see or use the information collected for this research study?

We will protect the confidentiality of your information by de-identifying data such that only subject numbers will be collected and attributed to your data. Only the principal investigators and Biomechanics/Sports Medicine Laboratory personnel will have access to the respective subject information and data. The de-identified data will be stored on hard drives of password protected computers in the Biomechanics/Sports Medicine Lab for a minimum of three years after the completion of the study and will be backed up onto DVDs, flash drives, and/or data backup cartridges, and then deleted from all hard drives. All subject data will be coded numerically and referred to only by the code and not by subject name at the time of data collection. Identity of the subjects will be held in strict confidence through the use of the coded subject numbers during data collection, analysis, and in all references made to data, both during and after the study, and in the reporting of the results. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used. We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information or what information came from you. Although it is unlikely, there are times when others may need to see the information we collect about you. These include:

- People at the University of Tennessee, Knoxville oversee research to make sure it is conducted properly.
- Government agencies (such as the Office for Human Research Protections in the U.S. Department of Health and Human Services), and others responsible for watching over the safety, effectiveness, and conduct of the research.
- If a law or court requires us to share the information, we would have to follow that law or final court ruling.

What will happen to my information after this study is over?

We will not keep your information to use for future research purposes. Your name and other information that can directly identify you will be deleted from your research data collected as part of the study.

We may share your research data with other researchers without asking for your consent again, but it will not contain information that could directly identify you.

Who can answer my questions about this research study?

If you have questions or concerns about this study, or have experienced a research related problem or injury, contact the researchers, Tianyi Lu via email at tlu3@vols.utk.edu, or via phone at (865) 765-7511. You may also contact my faculty advisor, Dr. Songning Zhang via email at szhang@utk.edu.

For questions or concerns about your rights or to speak with someone other than the research team about the study, please contact:

Institutional Review Board
The University of Tennessee, Knoxville
1534 White Avenue
Blount Hall, Room 408
Knoxville, TN 37996-1529
Phone: 865-974-7697
Email: utkirb@utk.edu

STATEMENT OF CONSENT

I have read this form and the research study has been explained to me. I have been given the chance to ask questions and my questions have been answered. If I have more questions, I have been told who to contact. By signing this document, I am agreeing to be in this study. I will receive a copy of this document after I sign it.

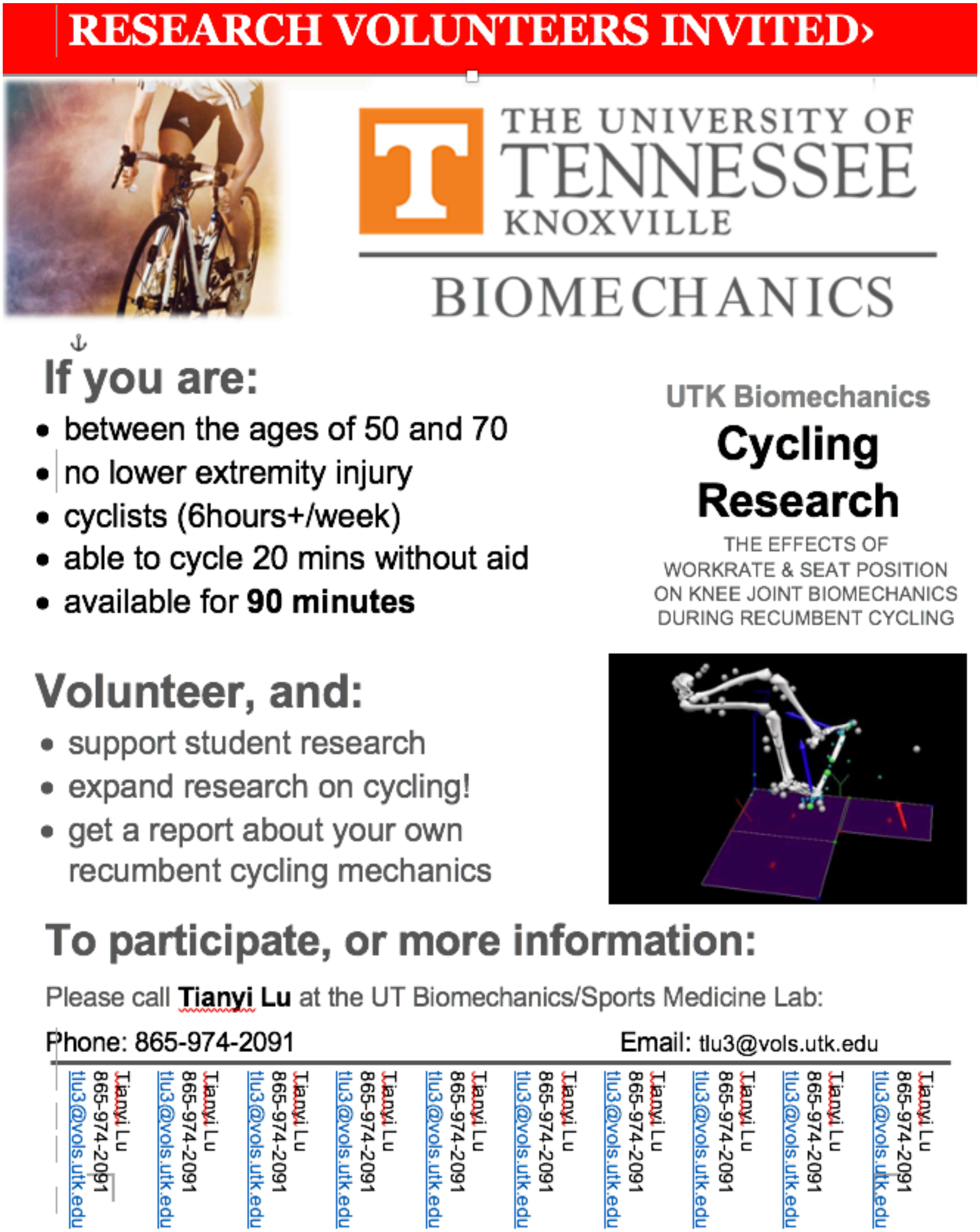
Name of Adult Participant	Signature of Adult Participant	Date

Researcher Signature (to be completed at time of informed consent)

I have explained the study to the participant and answered all of his/her questions. I believe that he/she understands the information described in this consent form and freely consents to be in the study.

Name of Research Team Member	Signature of Research Team Member	Date

APPENDIX D: FLYER



RESEARCH VOLUNTEERS INVITED›

THE UNIVERSITY OF
TENNESSEE
KNOXVILLE

BIOMECHANICS

If you are:

- between the ages of 50 and 70
- no lower extremity injury
- cyclists (6hours+/week)
- able to cycle 20 mins without aid
- available for **90 minutes**

UTK Biomechanics

Cycling Research

THE EFFECTS OF
WORKRATE & SEAT POSITION
ON KNEE JOINT BIOMECHANICS
DURING RECUMBENT CYCLING

Volunteer, and:

- support student research
- expand research on cycling!
- get a report about your own recumbent cycling mechanics

To participate, or more information:

Please call **Tianyi Lu** at the UT Biomechanics/Sports Medicine Lab:

Phone: 865-974-2091

Email: tlu3@vols.utk.edu

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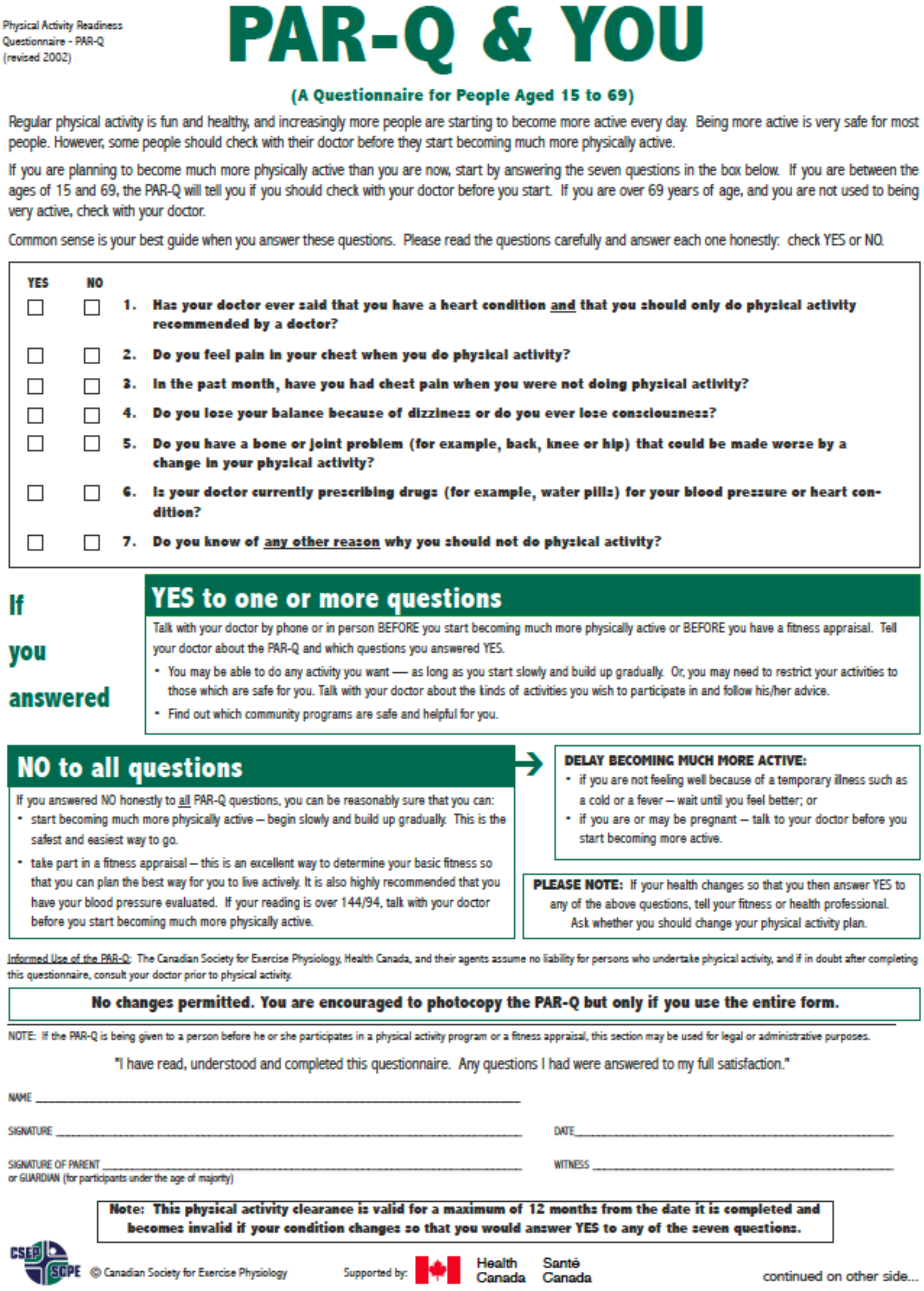
APPENDIX E: PHYSICAL READINESS QUESTIONNAIRE (PAR-Q)



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 and 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 any other reason 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 all 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.

Informed Use of the PAR-Q: 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.

Supported by: Health Canada Santé Canada

continued on other side...

APPENDIX F: INDIVIDUAL RESULTS FOR SELECTED VARIABLES

Table 6: Individual mean peak vertical PRF (N).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	159.155±6.734	170.534±13.641	155.977±4.025	166.352±9.918	164.764±25.166	167.791±32.097
2	80.395±4.127	87.750±8.506	98.729±3.369	120.221±2.178	111.533±6.307	128.009±3.234
3	137.676±21.495	177.444±24.006	149.630±14.242	176.800±14.723	203.084±7.549	183.448±8.109
4	153.691±11.801	141.438±5.354	158.217±8.372	172.670±5.065	144.967±6.502	167.639±11.109
5	188.362±9.416	187.065±10.409	181.778±6.377	204.219±15.706	191.350±14.447	207.915±14.802
6	96.399±4.705	104.132±6.606	87.804±8.739	99.212±6.778	101.297±3.078	127.168±5.805
7	138.168±19.798	153.925±12.800	160.761±4.203	171.615±26.962	157.890±5.739	157.773±8.636
8	101.818±11.009	118.600±8.806	106.941±11.087	128.142±8.694	117.922±3.878	136.085±5.974
9	102.744±7.320	149.761±22.683	84.380±7.117	120.587±17.066	88.130±3.327	91.266±2.794
10	128.231±3.333	144.038±9.298	139.946±6.085	143.720±10.775	144.264±11.401	144.611±7.889
11	159.407±5.500	192.700±12.104	154.054±4.272	178.270±5.226	148.538±5.124	167.715±5.077
12	131.681±20.658	134.255±12.877	109.860±5.402	132.889±18.998	141.896±11.581	128.854±9.761
13	111.129±4.196	114.126±4.835	123.515±3.405	126.498±1.052	126.998±1.358	135.013±1.762
14	141.439±3.669	130.627±3.328	130.983±5.358	137.904±13.288	129.443±5.271	129.199±2.277
15	134.644±9.714	153.427±18.510	148.963±29.729	137.023±13.391	129.593±10.480	153.569±7.603
Mean	130.996±28.579	143.988±30.285	132.769±29.551	147.741±28.831	140.111±31.103	148.404±28.323

Table 7: Individual mean peak posterior PRF (N).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-157.368±7.561	-192.388±10.042	-164.101±4.762	-199.757±19.739	-126.232±56.389	-192.732±25.572
2	-116.743±7.918	-165.336±18.820	-117.647±3.574	-157.372±15.536	-122.961±3.156	-178.046±14.745
3	-121.212±55.889	-196.173±9.048	-179.747±18.904	-205.681±21.983	-219.633±17.608	-238.481±21.008
4	-154.278±2.133	-157.638±9.436	-171.477±11.612	-218.994±22.401	-154.387±15.962	-203.066±29.701
5	-162.880±17.544	-198.911±9.447	-174.368±14.918	-197.439±12.788	-188.145±11.233	-199.287±19.560
6	-90.768±11.479	-111.958±6.740	-102.207±14.021	-140.745±26.711	-99.030±6.835	-132.887±20.021
7	-144.548±16.271	-160.269±12.506	-143.440±76.872	-199.634±13.582	-197.058±39.536	-257.140±54.489
8	-135.176±17.213	-183.646±12.310	-137.850±14.325	-181.562±20.260	-150.799±4.506	-193.524±14.442
9	-141.205±14.028	-189.289±27.915	-106.706±12.294	-181.117±21.021	-114.015±8.842	-148.906±16.736
10	-111.212±26.762	-191.760±65.549	-119.185±18.320	-164.265±57.434	-121.226±18.766	-156.125±7.778
11	-162.041±7.823	-188.317±6.624	-149.952±9.239	-164.432±86.854	-149.346±9.828	-184.155±8.658
12	-84.116±21.243	-108.243±10.705	-143.348±3.814	-182.740±26.069	-176.929±16.092	-191.758±5.674
13	-128.035±5.092	-175.498±9.953	-163.395±7.809	-192.084±4.033	-164.382±6.649	-208.392±5.499
14	-172.175±20.506	-191.714±12.492	-193.387±6.883	-204.027±9.149	-187.858±15.275	-197.984±15.239
15	-147.723±14.668	-201.428±22.367	-159.357±12.989	-188.134±25.428	-140.252±13.345	-201.051±23.765
Mean±STD	-135.299±26.371	-174.171±29.362	-148.411±27.568	-185.199±21.041	-154.150±34.597	-192.236±31.467

Table 8: Individual mean peak medial PRF (N).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-29.648±5.575	-36.153±4.876	-26.469±2.745	-38.589±8.162	-23.251±1.930	-41.532±8.272
2	-38.133±2.381	-56.329±5.515	-42.998±1.825	-58.374±6.018	-38.298±1.953	-63.065±3.780
3	-38.769±7.218	-48.769±3.006	-46.134±4.030	-54.269±5.441	-50.405±7.719	-67.816±5.753
4	-41.243±2.493	-35.466±0.821	-36.742±3.658	-51.712±3.396	-33.678±4.787	-40.460±19.860
5	-40.741±9.818	-46.291±7.390	-49.335±10.312	-53.054±10.625	-49.759±7.784	-56.066±13.492
6	-17.671±3.722	-26.389±2.278	-23.261±4.749	-38.080±8.533	-23.675±2.857	-33.862±3.023
7	-28.334±3.513	-47.536±3.280	-41.275±11.028	-46.848±9.151	-43.786±13.036	-62.081±22.334
8	-36.824±2.758	-50.979±4.472	-39.423±6.669	-50.477±7.644	-39.139±2.373	-52.579±4.096
9	-44.765±7.101	-63.194±11.631	-32.031±6.362	-61.177±8.166	-39.270±5.087	-50.486±7.502
10	-20.177±8.285	-43.789±22.371	-24.046±8.400	-46.951±26.993	-29.402±9.319	-37.032±5.497
11	-8.082±2.196	-10.594±2.098	-12.102±1.963	-17.120±3.439	-14.794±2.343	-21.579±3.455
12	-28.204±4.084	-44.779±7.510	-60.310±11.905	-75.478±16.822	-66.362±8.618	-79.472±4.048
13	-29.668±4.179	-47.252±4.204	-41.395±3.978	-50.343±1.469	-42.554±4.079	-51.245±1.903
14	-28.474±4.678	-34.153±4.448	-36.612±3.497	-48.221±9.532	-33.239±3.654	-37.877±3.988
15	-23.693±4.936	-37.334±6.835	-16.872±7.968	-32.923±7.597	-11.033±2.644	-35.873±6.240
Mean±STD	-30.295±10.090	-41.934±12.702	-35.267±12.903	-48.241±13.352	-35.910±14.373	-48.735±15.282

Table 9: Individual mean peak knee extension angle (deg).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-10.688±0.684	-4.923±1.447	-9.672±0.907	-12.307±1.480	-18.050±0.933	-14.014±1.208
2	-22.216±0.503	-17.712±0.729	-45.760±0.295	-41.383±0.374	-50.366±0.559	-49.627±0.702
3	-26.613±2.804	-23.784±1.736	-32.889±1.640	-28.199±2.275	-40.931±1.395	-38.160±1.048
4	-24.287±0.910	-22.013±1.209	-42.908±0.355	-38.216±1.333	-52.515±0.662	-51.035±0.849
5	-27.277±1.362	-25.270±0.977	-34.455±0.668	-32.827±0.505	-53.447±0.814	-52.219±0.612
6	-31.236±0.322	-33.607±0.962	-38.962±1.053	-39.165±10.711	-51.066±1.942	-45.242±0.672
7	-32.507±2.282	-32.422±2.653	-41.854±2.468	-43.409±2.039	-47.180±2.512	-38.997±1.987
8	-22.840±2.123	-20.017±0.651	-23.769±1.778	-27.064±1.223	-39.905±0.358	-35.920±0.706
9	-24.738±1.090	-17.584±2.222	-29.133±1.625	-24.355±1.957	-40.502±0.675	-41.995±1.108
10	-21.241±1.450	-20.157±2.229	-35.053±0.398	-36.810±1.773	-58.779±0.659	-57.427±0.473
11	-38.151±1.836	-39.989±0.957	-45.204±0.811	-41.827±0.456	-44.934±0.360	-45.529±0.639
12	-25.968±1.157	-24.171±2.174	-34.038±1.683	-28.533±1.295	-42.636±0.839	-42.209±0.772
13	-28.314±1.025	-25.942±0.423	-40.096±0.457	-36.846±0.028	-59.344±0.358	-57.133±0.208
14	-20.238±1.416	-21.307±0.909	-25.885±1.086	-23.813±1.033	-42.115±0.629	-38.567±0.735
15	-40.504±1.004	-35.884±1.282	-47.170±0.856	-44.339±1.506	-56.059±1.917	-48.308±0.733
Mean±STD	-26.455±7.325	-24.319±8.662	-35.123±10.035	-33.273±9.040	-46.522±10.297	-43.759±10.635

Table 10: Individual mean peak knee abduction angle (deg).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	2.762±0.587	-0.339±0.551	2.154±0.722	6.894±7.352	6.238±0.722	4.024±0.576
2	7.590±0.599	4.140±0.374	11.252±0.341	10.703±0.409	10.420±0.347	10.648±0.163
3	9.180±0.874	9.641±1.684	10.621±0.617	10.166±0.669	15.355±3.977	19.954±0.491
4	6.201±0.712	8.456±0.966	8.653±0.942	4.914±1.837	7.272±1.705	7.331±0.296
5	4.142±1.449	3.530±0.383	4.699±1.017	4.906±0.743	7.920±2.037	8.347±0.923
6	-2.033±0.829	-3.788±0.142	-0.287±0.857	-0.264±1.566	1.489±0.485	2.270±0.152
7	1.661±2.001	0.957±1.475	-0.455±0.716	-1.622±0.903	-1.894±1.603	-5.047±0.604
8	-4.898±0.440	-6.969±0.420	-8.111±0.628	-8.760±0.799	-7.630±0.361	-7.675±0.623
9	-0.389±0.718	-1.814±1.502	2.134±1.481	-0.737±0.553	5.265±1.181	4.712±0.622
10	3.215±0.403	1.724±0.616	1.751±0.582	0.316±0.863	-3.746±0.779	-4.032±0.358
11	-0.379±1.083	0.035±0.295	1.411±0.823	2.587±0.549	8.910±0.393	7.462±1.136
12	0.901±2.046	1.063±2.839	4.999±1.450	5.581±1.670	8.802±1.922	8.099±2.006
13	4.379±0.693	2.079±0.622	8.123±0.668	7.793±0.374	16.935±0.953	11.588±0.351
14	1.357±0.418	1.395±0.282	2.438±0.352	2.346±0.663	8.554±0.520	5.518±0.883
15	5.122±1.676	5.362±1.683	3.445±0.844	3.696±1.288	4.831±1.458	7.775±2.345
Mean±STD	2.587±3.725	1.698±4.286	3.522±4.924	3.235±5.032	5.915±6.655	5.398±7.013

Table 11: Individual mean peak knee extension ROM (deg).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	82.560±0.987	87.134±1.460	83.539±1.141	82.550±1.486	77.872±1.003	80.584±1.342
2	85.239±0.618	89.171±0.834	70.598±0.471	74.137±0.758	70.073±0.513	70.170±0.526
3	74.435±1.758	76.271±1.471	72.626±1.433	74.688±2.296	67.864±1.277	72.290±1.387
4	82.159±1.035	82.698±1.109	69.615±0.653	75.249±1.233	65.268±0.532	66.054±0.815
5	74.906±1.063	77.174±0.715	70.354±0.722	71.968±0.525	59.228±1.092	59.860±1.173
6	78.410±0.320	75.680±1.343	72.200±1.813	70.569±10.496	66.366±1.516	69.835±0.853
7	77.250±2.265	76.750±2.831	70.917±2.536	71.048±2.006	69.444±2.127	75.985±1.979
8	75.015±2.061	80.827±1.043	76.984±2.219	77.046±1.002	68.525±1.338	73.605±0.931
9	85.751±1.157	92.253±1.873	82.208±1.803	86.713±1.919	74.551±1.562	74.804±1.185
10	89.023±1.691	89.159±2.333	76.683±0.544	75.977±1.388	62.723±0.568	64.567±0.488
11	80.544±2.087	77.696±1.039	73.898±0.788	76.684±0.594	74.169±0.527	74.574±0.832
12	74.301±0.970	76.352±2.539	70.323±1.594	75.452±1.116	67.164±0.973	67.148±1.289
13	76.676±1.205	78.909±0.545	70.830±0.565	72.314±0.239	63.607±0.498	64.789±0.370
14	86.502±1.666	85.784±0.505	83.158±0.933	84.328±0.957	74.118±0.660	74.839±0.906
15	73.347±0.931	77.527±1.839	71.354±0.799	72.634±1.557	66.367±1.872	71.371±0.571
Mean±STD	79.741±5.171	81.559±5.676	74.352±4.970	76.091±4.853	68.489±5.046	70.698±5.420

Table 12: Individual mean peak knee abduction ROM (deg).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-17.194±0.768	-21.129±0.579	-18.874±1.269	-13.996±7.652	-14.813±0.965	-15.921±0.767
2	-8.061±0.453	-11.386±0.389	-1.072±0.257	-2.284±0.302	-0.747±0.447	-0.437±0.248
3	-12.475±1.694	-14.083±1.397	-14.291±0.529	-13.871±1.114	-9.265±3.736	-8.409±0.760
4	-2.824±0.579	-1.486±0.704	-0.603±1.027	-2.827±1.329	0.226±1.544	0.398±0.617
5	-10.132±0.944	-11.023±0.604	-11.263±1.563	-11.464±0.314	-9.290±2.283	-9.198±1.089
6	-2.620±1.024	-3.251±0.540	-0.972±1.100	-2.468±1.293	0.292±0.353	-0.716±0.472
7	-7.100±2.396	-8.168±1.411	-7.992±1.006	-9.880±1.851	-10.270±1.616	-9.516±0.597
8	-3.917±1.422	-3.444±1.255	-5.722±1.095	-6.082±0.682	-3.660±1.436	-4.161±0.874
9	-7.711±1.150	-10.333±1.952	-10.505±1.874	-9.877±1.414	-9.476±1.413	-7.421±0.781
10	-7.685±0.916	-7.633±0.657	-3.673±1.000	-4.487±0.693	-4.917±0.599	-4.850±0.347
11	-9.157±0.999	-6.092±0.688	-7.672±0.775	-5.993±0.555	-4.124±0.481	-4.779±1.202
12	-12.623±1.950	-14.363±2.623	-11.484±1.209	-13.625±2.460	-6.165±1.619	-7.127±2.001
13	-15.180±0.518	-17.616±0.594	-11.209±0.524	-12.427±0.440	-4.275±0.964	-6.196±0.633
14	-12.405±0.705	-12.478±0.397	-11.630±0.407	-10.632±0.681	-4.499±0.573	-6.248±0.693
15	-6.428±1.804	-7.329±1.873	-7.683±1.310	-6.657±1.450	-5.042±1.691	-4.063±2.435
Mean±STD	-9.034±4.331	-9.988±5.472	-8.310±5.245	-8.438±4.282	-5.735±4.232	-5.910±4.139

Table 13: Individual mean peak knee extension moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	13.047±1.247	16.065±1.635	14.985±0.977	24.852±4.797	15.556±2.304	22.915±5.772
2	15.394±1.553	20.207±2.509	17.832±0.790	23.295±2.659	17.824±0.997	32.207±2.528
3	20.579±5.974	17.890±2.789	18.477±4.278	20.576±2.668	24.015±2.171	33.106±4.589
4	35.981±2.241	40.760±1.926	36.487±2.479	48.743±9.583	31.560±3.221	45.784±11.014
5	14.139±5.276	27.533±2.556	19.863±5.450	28.536±5.632	27.934±1.989	27.377±6.655
6	19.640±4.082	24.700±1.958	21.238±2.384	24.171±4.199	17.733±1.195	23.119±3.598
7	17.471±3.399	22.821±2.843	27.182±7.515	32.325±5.651	33.726±11.362	55.540±17.487
8	22.970±5.035	32.393±3.525	19.555±3.975	32.481±6.476	25.792±1.438	40.181±3.918
9	26.306±2.839	33.120±3.590	20.322±4.904	33.613±4.090	19.396±2.717	28.892±2.208
10	21.527±7.507	37.322±16.118	16.580±4.034	31.054±17.791	13.375±5.676	23.141±2.537
11	20.071±1.596	30.800±5.170	18.105±1.820	37.006±3.849	18.444±3.967	37.076±3.957
12	18.273±2.985	26.654±2.701	32.237±6.005	32.665±5.929	44.274±6.546	40.819±2.090
13	18.930±1.663	31.224±2.212	31.898±2.422	35.241±1.150	31.293±1.773	40.717±1.106
14	22.610±4.068	24.801±2.716	28.373±1.199	30.192±6.900	36.304±2.981	35.360±5.450
15	23.197±7.607	36.239±9.470	19.956±3.953	31.363±5.853	20.728±2.802	32.188±3.968
Mean±STD	20.676±5.551	28.169±7.252	22.873±6.603	31.074±6.770	25.197±8.819	34.561±9.164

Table 14: Individual mean peak knee flexion moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-28.919±1.691	-33.007±2.636	-28.393±3.626	-29.788±3.456	-25.476±10.874	-32.109±9.573
2	-13.431±0.967	-15.763±2.055	-11.921±1.357	-12.117±1.558	-9.042±1.747	-10.002±1.776
3	-22.898±8.787	-31.303±2.554	-15.354±4.928	-24.834±1.777	-4.196±0.951	-7.976±1.678
4	-17.390±1.616	-17.143±0.959	-3.754±1.123	-4.939±1.545	-3.073±1.415	0.640±2.227
5	-20.976±5.591	-21.796±5.222	-13.305±3.744	-16.754±8.635	-6.184±4.373	-10.474±5.611
6	-21.518±0.681	-22.847±1.411	-13.501±3.200	-19.169±2.129	-11.772±1.452	-10.260±0.572
7	-19.503±4.113	-32.316±5.431	-17.539±9.565	-14.536±10.174	-12.566±11.421	-7.463±4.485
8	-11.306±2.205	-16.095±1.925	-16.513±2.100	-13.230±3.659	-5.004±2.661	-4.336±2.336
9	-4.516±2.407	-10.928±2.270	-4.385±2.576	-5.929±3.257	-0.004±1.407	-1.682±2.144
10	-14.237±1.424	-17.072±3.811	-13.890±2.370	-10.800±1.922	-14.902±5.711	-10.051±3.408
11	-9.840±1.396	-8.392±3.070	-10.081±0.691	-3.942±1.678	-6.057±1.256	-0.708±0.624
12	-27.729±3.416	-21.715±3.815	-9.611±2.388	-11.288±3.201	-0.943±2.729	-1.241±2.968
13	-20.231±1.747	-17.126±0.865	-6.630±0.598	-11.485±0.752	-3.379±1.325	-3.196±0.504
14	-25.256±1.411	-29.183±1.276	-13.032±1.066	-22.954±3.858	0.418±0.763	-8.075±2.389
15	-9.796±9.438	-5.212±2.798	-6.937±3.115	-6.074±1.464	-5.748±4.565	-3.289±2.352
Mean±STD	-17.836±7.144	-19.993±8.624	-12.323±6.141	-13.856±7.655	-7.195±6.789	-7.348±7.868

Table 15: Individual mean peak knee abduction moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-15.958±2.793	-18.921±1.936	-14.948±1.894	-20.376±3.784	-12.901±1.567	-20.401±4.049
2	-12.662±0.746	-19.488±2.589	-10.184±0.572	-15.326±1.957	-9.878±0.517	-16.270±1.196
3	-13.179±2.785	-18.945±0.795	-17.294±3.035	-20.594±2.143	-20.233±1.426	-30.890±3.916
4	-12.504±1.716	-11.398±0.983	-12.748±2.059	-13.109±3.003	-8.770±1.055	-12.715±1.854
5	-17.275±4.815	-19.874±2.342	-17.230±3.640	-21.007±3.236	-18.870±2.834	-21.605±2.583
6	-1.625±1.174	-2.152±0.193	-2.850±0.727	-4.537±1.275	-2.859±1.027	-3.716±0.659
7	-10.153±1.119	-15.620±1.289	-14.532±3.512	-13.467±2.607	-15.511±5.649	-17.043±6.621
8	-8.422±1.006	-8.182±1.383	-5.735±1.043	-6.608±1.175	-4.007±0.235	-4.021±0.859
9	-10.592±2.144	-15.844±3.124	-8.546±1.789	-15.350±2.778	-10.684±1.597	-12.472±2.325
10	-9.471±3.807	-16.528±7.198	-8.593±2.564	-12.440±5.804	-9.737±1.767	-9.134±0.474
11	-4.152±0.406	-5.140±0.345	-5.195±0.230	-7.089±0.562	-7.196±0.772	-9.636±1.087
12	-9.014±1.326	-15.315±2.194	-19.329±1.868	-27.867±7.574	-22.150±2.286	-25.389±1.248
13	-12.252±1.865	-19.468±2.141	-17.652±1.110	-21.620±1.057	-18.144±1.248	-20.621±0.770
14	-12.232±1.944	-14.869±1.534	-15.996±1.327	-17.434±3.216	-15.141±1.205	-13.963±1.810
15	-11.696±2.153	-17.610±4.181	-8.916±2.329	-13.760±3.171	-7.580±1.337	-16.684±2.899
Mean±STD	-10.746±4.008	-14.624±5.508	-11.983±5.198	-15.372±6.360	-12.244±5.914	-15.637±7.517

Table 16: Individual mean peak ankle plantarflexion moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-24.270±0.803	-25.040±1.464	-23.851±1.430	-25.000±1.606	-25.641±4.645	-22.165±7.753
2	-11.812±0.614	-17.698±2.888	-14.286±0.692	-17.815±0.546	-17.257±0.965	-19.222±0.765
3	-24.042±3.583	-31.471±3.360	-25.909±4.082	-28.795±2.263	-28.846±1.841	-27.197±1.300
4	-19.702±1.449	-18.392±0.608	-19.430±0.900	-26.150±1.109	-21.296±0.471	-23.939±1.669
5	-22.912±2.243	-21.785±2.043	-19.437±0.940	-22.562±3.104	-19.694±3.728	-18.062±2.910
6	-11.815±0.476	-12.276±0.526	-9.048±0.532	-11.708±0.828	-10.133±0.475	-12.986±0.641
7	-19.612±2.394	-23.521±1.771	-19.513±1.071	-26.128±4.563	-17.874±1.916	-21.559±3.830
8	-7.857±1.252	-10.957±0.442	-11.586±0.599	-13.045±0.993	-11.499±0.987	-14.887±1.605
9	-10.084±1.075	-13.894±2.090	-9.620±1.043	-12.626±1.908	-9.815±0.607	-10.629±0.433
10	-17.967±0.980	-21.152±1.120	-21.176±0.773	-20.720±1.497	-20.902±1.461	-20.845±1.352
11	-19.371±0.680	-22.621±1.805	-18.543±0.748	-22.058±0.990	-19.867±0.843	-19.885±1.000
12	-19.758±1.489	-19.097±2.298	-14.621±1.814	-17.100±1.763	-16.208±2.687	-16.218±1.439
13	-16.504±0.607	-18.249±1.044	-19.598±0.702	-20.347±0.091	-18.777±0.561	-20.674±0.582
14	-17.102±0.907	-20.120±0.698	-17.059±0.687	-21.294±4.282	-17.248±0.641	-16.513±1.543
15	-20.120±3.994	-23.254±4.230	-23.459±6.330	-21.007±2.660	-21.823±1.355	-25.712±4.833
Mean±STD	-17.528±5.048	-19.968±5.222	-17.809±5.092	-20.424±5.167	-18.459±5.273	-19.366±4.600

Table 17: Individual mean peak ankle abduction moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-6.226±0.761	-6.109±0.200	-5.475±0.426	-5.471±0.381	-4.133±1.863	-5.904±1.316
2	-1.226±0.540	-1.576±0.324	-0.382±0.573	-0.216±0.097	0.530±0.071	0.228±0.172
3	-3.581±1.336	-4.104±1.088	-4.717±0.925	-4.377±0.578	-4.356±0.367	-4.994±0.845
4	-4.602±0.234	-4.443±0.488	-4.293±0.318	-4.784±0.262	-4.249±0.248	-4.498±0.329
5	-6.443±0.783	-5.835±0.568	-5.388±0.333	-6.374±0.827	-5.970±1.097	-6.080±0.854
6	-0.662±0.225	-0.662±0.075	-0.455±0.234	-0.468±0.288	-0.861±0.285	-0.955±0.541
7	-6.511±0.979	-8.772±0.385	-8.371±0.911	-9.447±0.806	-8.821±0.828	-11.274±2.096
8	-0.788±0.271	-1.113±0.220	-0.803±0.255	-0.918±0.086	-1.016±0.126	-0.973±0.118
9	-2.224±0.260	-3.427±0.755	-2.195±0.310	-2.867±0.457	-2.200±0.249	-2.247±0.338
10	-2.864±0.884	-5.072±2.064	-3.836±0.432	-4.454±0.862	-3.758±0.666	-4.517±0.327
11	-0.639±0.221	-0.021±0.575	-0.712±0.346	-0.749±0.152	-0.518±0.308	-0.119±0.193
12	-4.619±0.342	-6.000±1.000	-6.001±0.730	-8.104±2.007	-6.208±0.444	-6.006±0.651
13	-2.457±0.178	-3.263±0.255	-2.645±0.182	-4.338±0.299	-2.871±0.253	-2.955±0.111
14	-3.493±0.458	-3.506±0.791	-4.520±0.198	-3.696±0.370	-3.799±0.620	-3.698±0.386
15	-5.200±1.501	-4.716±1.078	-4.089±0.581	-5.209±1.734	-4.036±0.423	-5.458±1.654
Mean±STD	-3.436±2.110	-3.908±2.370	-3.592±2.351	-4.098±2.738	-3.484±2.453	-3.964±2.978

Table 18: Individual mean peak hip flexion moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-28.878±2.729	-30.215±1.322	-25.716±2.338	-24.566±2.348	-31.575±1.180	-28.341±0.688
2	-24.847±2.045	-27.996±1.223	-34.854±1.777	-40.502±2.251	-33.568±1.724	-35.642±1.357
3	-16.323±6.874	-30.669±2.287	-21.237±3.712	-26.781±0.733	-34.863±2.528	-22.915±3.084
4	-25.515±1.680	-27.518±4.933	-16.454±0.886	-16.614±1.385	-15.271±1.216	-15.742±1.426
5	-23.934±3.422	-22.738±3.344	-28.577±3.430	-30.393±3.263	-38.156±4.082	-37.391±5.881
6	-15.483±1.579	-24.805±1.969	-11.741±2.952	-17.499±4.425	-21.727±2.365	-22.790±1.805
7	-10.174±5.123	-25.980±3.218	-20.303±5.843	-10.989±4.890	-8.671±3.799	-12.075±2.086
8	-10.625±1.803	-15.112±2.427	-14.095±0.805	-12.678±2.933	-14.684±1.706	-11.854±2.764
9	-3.144±0.944	-2.720±3.470	-7.027±1.217	-5.367±1.970	-14.344±1.858	-8.236±2.272
10	-1.544±3.462	-6.213±3.594	-24.765±1.388	-21.635±5.109	-30.100±1.558	-24.406±1.858
11	-18.574±3.615	-20.735±1.284	-24.559±4.347	-19.224±1.748	-25.259±3.042	-21.602±2.298
12	-22.762±3.766	-13.350±8.303	-8.540±3.599	-10.537±4.148	-11.638±2.039	-8.871±1.865
13	-18.565±2.101	-17.566±1.977	-16.458±1.644	-19.840±1.147	-19.961±1.337	-21.530±1.871
14	-11.987±3.793	-13.248±1.289	-10.794±0.481	-12.226±3.227	-5.569±0.921	-12.245±2.243
15	-17.904±6.084	-3.252±5.172	-12.511±4.927	-7.068±4.100	-4.272±1.759	-5.939±0.000
Mean±STD	-16.684±8.048	-18.808±9.536	-18.509±8.050	-18.395±9.425	-20.644±11.121	-19.305±9.701

Table 19: Individual mean peak hip abduction moment (Nm).

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	-19.174±3.099	-18.116±8.285	-15.628±1.719	-19.876±3.143	-16.074±2.950	-17.016±2.062
2	-6.921±1.645	-9.869±1.062	-4.794±0.562	-7.441±0.566	-6.429±0.613	-5.971±0.576
3	-20.074±3.527	-25.928±3.497	-20.786±1.947	-24.569±2.304	-22.032±1.313	-28.912±2.727
4	-19.814±0.746	-16.649±1.501	-16.332±1.733	-20.200±1.076	-13.059±2.332	-19.034±2.924
5	-28.163±4.247	-31.422±3.508	-26.975±2.779	-32.344±4.105	-25.335±2.458	-28.151±4.600
6	-3.490±1.531	-5.517±0.444	-3.408±0.934	-7.011±2.084	-0.988±0.466	-2.417±0.775
7	-16.567±2.587	-22.622±1.722	-20.633±3.842	-21.610±5.091	-28.481±6.791	-32.451±10.307
8	-16.207±1.485	-22.161±1.870	-14.559±2.745	-19.000±3.176	-11.829±0.800	-18.030±1.338
9	-17.705±2.966	-25.479±4.473	-12.766±3.828	-19.361±2.955	-12.973±2.585	-14.903±1.967
10	-12.804±3.694	-22.394±10.108	-13.795±1.133	-18.275±6.738	-15.180±2.326	-16.327±0.987
11	3.059±0.860	2.063±1.500	3.030±0.551	3.601±0.942	1.848±0.645	2.038±0.953
12	-16.181±2.991	-22.271±3.452	-24.152±2.205	-29.113±5.839	-22.756±2.475	-26.425±0.813
13	-17.628±1.310	-24.957±0.966	-19.197±1.065	-26.240±0.497	-19.442±1.459	-22.213±1.544
14	-19.365±3.695	-24.506±2.688	-22.061±1.978	-25.584±2.797	-17.082±2.010	-17.520±1.695
15	-2.218±3.519	1.513±1.209	0.955±1.486	0.564±4.044	-0.801±1.982	.±.
Mean±STD	-14.217±8.285	-17.888±10.256	-14.073±9.157	-17.764±10.572	-14.041±9.192	-17.707±10.135

Table 20: Individual RPE scores.

Subject	Far		Medium		Close	
	60 W	100 W	60 W	100 W	60 W	100 W
1	11	12	9	11	9	10
2	8	12	6	12	7	11
3	6	8	6	6	6	6
4	14	15	12	13	13	13
5	6	8	6	7	11	13
6	11	13	9	11	9	11
7	6	10	6	10	6	11
8	7	8	6	7	6	7
9	6	7	6	7	6	8
10	6	6	6	7	6	7
11	10	12	7	10	7	12
12	7	10	9	10	9	11
13	7	8	8	9	8	9
14	11	13	9	11	8	10
15	7	10	7	8	8	12
Mean±STD	8.20±2.54	10.13±2.62	7.46±1.81	9.27±2.15	7.93±2.05	10.07±2.22

VITA

Tianyi Lu was born in Nanjing, China, to the parents of Xiaoping Lu and Xiaorong Yan. He attended elementary through high school in his hometown. After his graduation from Nanjing Foreign Language School, he attended University of Rochester and received his Bachelor of Science degree in Biomedical Engineering. After his graduation, he was accepted by the Department of Kinesiology, Recreation, and Sports Science at the University of Tennessee and started to pursue a master's degree in exercise science with an emphasis in biomechanics. He graduated with a Master of Science degree in 2019.