Walsh University and The Ohio State University

The Effect of Load Carriage on Quadriceps Angle During Walking and Running

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Abstract

Females tend to have a larger quadriceps angle (Q angle) than males, and it has been inferred that larger Q angles can change movement patterns and increase the potential for musculoskeletal injury. Movement biomechanics can also be affected by load carriage, that is external weight added to one's body, whether in exercise or in tactical settings (e.g., soldiers, police officers, firefighters wearing equipment). However, the interaction of how Q angle and load carriage impact movement and biomechanics together is unknown, therefore the purpose was to investigate if load carriage changes the Q angle during walking and running. For the study we had 10 healthy recreational or competitive female athletes aged 18-35 participate. Demographics and previous injuries data were collected using a set of questionnaires, static Q angle was measured using a goniometer, and dynamic Q angle was measured using a motion capture system and treadmill with embedded force plates. Participants walked and ran on the treadmill at a self-selected pace for five minutes each with and without the weighted vest. Q angle changes were measured as frontal plane knee angle movement during weight acceptance throughout the gait cycle. Overall, the dynamic Q angle in females was different at a walking pace versus running pace, while unloaded and loaded variables did not significantly affect dynamic Q angle either in valgus or varus positions. Therefore, future research can provide society with more knowledge and understanding of how load carriage may affect Q angle, and thus alter movement patterns with the potential to cause patellofemoral pain syndrome as well as increase injury risk.

Keywords: Quadriceps angle, Patellofemoral pain syndrome, Load carriage, Dynamic Q angle, Gait, Intercondylar notch, Biomechanics

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Introduction

Load carriage is an activity involved in various occupations, such as soldiers, police officers, and fire fighters. Loads are commonly carried on a belt, vest, or backpack not only by these occupations, but also by students, hikers, and cross training athletes. Many lower extremity muscles are involved in load carriage, in particular, the quadriceps, gastrocnemius, hamstrings, and tibialis anterior muscles. The external added load affects the biomechanics of movement, specifically gait variability. Therefore, when carrying a load, individuals maximize efficiency and minimize energy expenditure by adjusting their stride rate, stride length, and speed. For example, as the amount of load carried increases, stride length decreases in order to provide greater stability (Boffety et al., 2019).

However, the interaction between load carriage and Q angle is unknown, particularly for females. The quadriceps angle, or Q angle, is a clinical measure of the angle formed between a line along the quadriceps from the anterior superior iliac spine to the center of the patella and then a line connecting the center of the patella with the tibial tuberosity. The Q angle impacts the quadriceps muscle force vector for movements including walking and running, as shown in Figure 1. The larger the Q angle, the greater the lateral forces on the patella. Therefore, the Q angle can be a predictor of a future lower limb injury including overuse injury, patellofemoral pain syndrome (PFPS), patellar instability and dislocation, or anterior knee pain such as an anterior cruciate ligament (ACL) injury (Shampain et al., 2018). Since larger Q angles are associated with a higher potential for injury, it is important to understand the static and dynamic biomechanical factors that contribute to Q angle and the clinical implications for risk of injury, particularly when carrying a load.

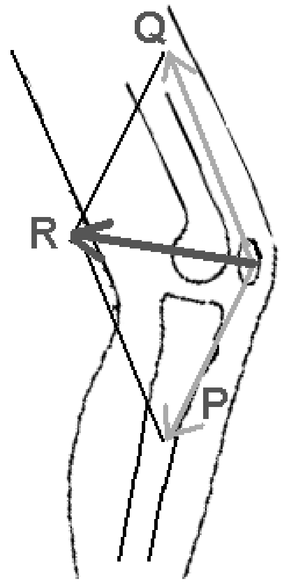

Figure 1. Forces applied to the patella by the quadriceps muscles and the patellar tendon. Vector Q = quadriceps force on the patella. Vector P = force restrained by the patellar tendon. Vector R = resultant vector composition force (Q and P equal in magnitude and length) that pulls the patella laterally.

(Thompson, 2000)

Research has demonstrated that having a larger Q angle increases the potential for injury during dynamic activities. Movement is affected by load carriage, however the interaction of how Q angle and load carriage impact movement together is unknown. Our purpose is to investigate if adding load changes the Q angle during gait and if differences exist between participants when walking and running on a treadmill with a weighted vest.

Literature Review

Quadriceps Angle

The quadriceps angle, or Q angle, was first described by Brattstrom in 1964 (Khasawneh et al., 2019) and is a clinical measure of the angle formed between a line along the quadriceps from the anterior superior iliac spine to the center of the patella and then a line connecting the center of the patella with the tibial tuberosity, as shown in Figure 2. For females, average Q angles are between 15-17°, whereas typical Q angles for males are between 10-13° (Loudon, 2016; Kuhn et al., 2002; Tillman et al., 2005).

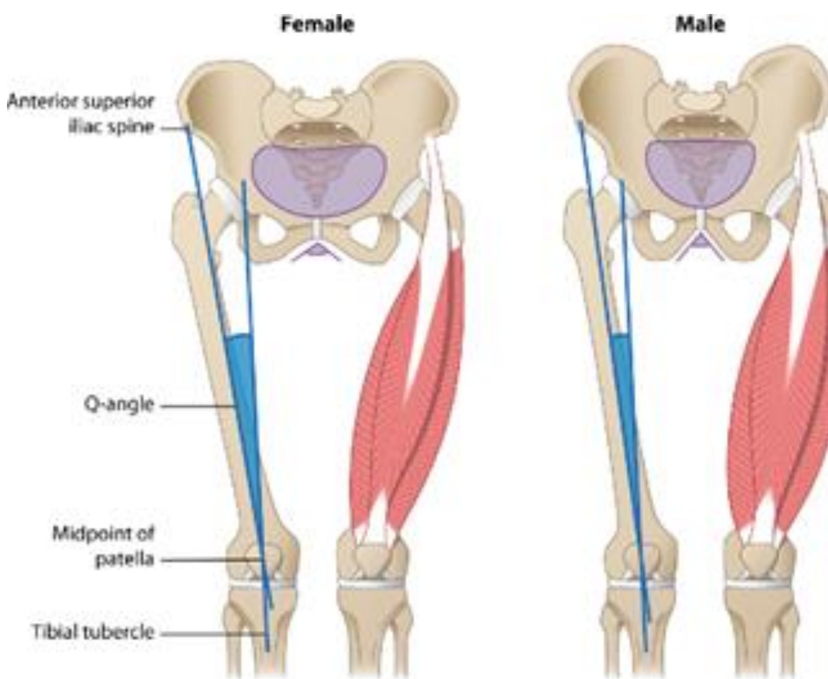

Figure 2. Q angle, muscle mass, and pelvic differences in males and females.

(Shampain et al., 2018)

The larger the Q angle, the greater the lateral forces on the patella. Therefore, the Q angle can be a predictor of a future lower limb injury including overuse injury, PFPS, patellar instability and dislocation, or anterior knee pain such as an ACL injury (Livingston & Spaulding, 2002; Nguyen et al., 2009). Since the larger the Q angle the greater the potential of injury, it is important to understand the static and dynamic anatomical factors that contribute to Q angle and the clinical implications for risk of injury.

The Q angle impacts the quadriceps muscle force vector on the frontal plane, specifically the patella and tibial tuberosity, for movements including knee extension, jump landing, walking, and running, as shown in Figure 3. Along with the effect on the patella and tibia, the magnitude of the Q angle varies when the foot position moves internally or externally, increasing or decreasing as the foot rotates (Livingston & Spaulding, 2002; Heiderscheit et al., 2000). In addition to the pelvis, patella, and foot position, it has been proposed that hip and tibial rotation may also contribute to magnitude of the Q angle (Nguyen et al., 2009). During motion, the Q angle can change dynamically due to the contraction of the quadriceps, extensibility of the connective tissue around the patella, and the geometry of the patella and trochlear groove. Therefore, as the knee goes through flexion, the relative angle of the articulating surface of the patella changes throughout its range of motion, resulting in a dynamically changing Q angle (Loudon, 2016).

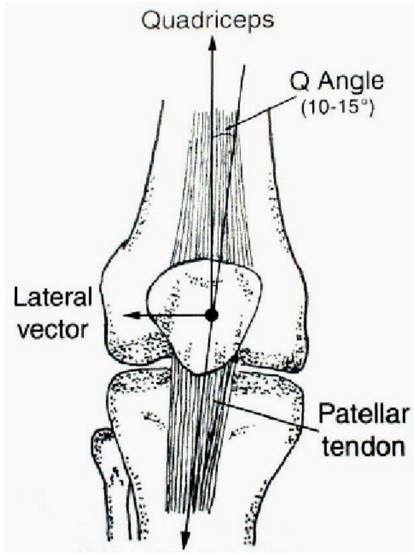

Figure 3. Patellofemoral joint reaction force due to the lateral forces on the patella. The lateral vector represents the lateral forces during a landing/movement.

(Wu & Wu, 2005)

Q angle can be measured statically or dynamically using various methods. The goniometer is the most common tool used to measure the angle of a joint and is used throughout the medical field in diverse clinical settings, especially in orthopedics and physical therapy. Measuring range of motion helps clinicians evaluate and assess a patient's care plan and progress. Instead of measuring a single Q angle in a constant position, the use of a goniometer can measure a Q angle in various postures (supine, standing, sitting) and different knee flexion angles (Türkmen et al., 2015). Dynamically, 2D and 3D motion capture systems accurately measure marker positions to determine Q angle. However, there is not a standardized Q angle protocol for measuring static or dynamic values, as indicated in Table 1, summarizing various Q angle measurement methods referenced in the research literature.

Table 1. Q angle measurement methods.

Author	Measurement Tool	Method/Protocol
Almeida et al., 2016	Goniometer	Measured in dorsal decubitus with the knee and hip extended, and the hip and foot in neutral position.
de Oliveira Silva et at., 2015	3D Motion Analysis System	Determined reliability and capability of three Q angle measurements: static clinical test, peak dynamic knee valgus during stair ascent, and a static measurement.
Grelsamer et al., 2005	Customized Translucent Protractor	Measured Q angle on right leg of each subject.
Heiderscheit et al., 2000	Goniometer & 3D Motion Capture System	Measured Q angle of the right lower extremity in standing position with goniometer. Motion capture lower extremity kinematics while running.
Imhoff et al., 2021	Gait Videography & Radiographic Imaging (MRI)	Clinical and radiographic measurements taken; internal and external rotation, Q angle, tubercle sulcus angle, femoral anteversion, tibial tubercle-trochlear groove distance, and frontal leg axis.
Khasawneh et al., 2019	Full Circle Manual Goniometer	Measured Q angle with the subject standing in the erect weight-bearing position while supine.

Livingston & Spaulding, 2002	OPTOTRAK 2D LED Motion Measurement System	Compared Q angles under static weight bearing conditions with the feet positioned in self-selected versus standardized stance positions.
Nguyen et al., 2009	Goniometer, Inclinometer, Ruler, & Anthropometric Caliper	Measured static alignment of the left lower extremity to determine the impact of lower extremity alignment on the magnitude of Q angle while standing.
Rauh et al., 2007	Goniometer	Measured static, standing position with quadriceps relaxed.
Sanchez et al., 2014	Computerized Biophotogrammetry	Measured supine positions with parallel feet, supine with abduction (external rotation of lower limbs) and standing position with parallel feet and with external rotation.
Türkmen et al., 2015	Goniometer	Measured right leg of each subject in supine, sitting, and standing positions.

As outlined in Table 1, there are many studies that have measured Q angles using various protocols. For example, Rauh et al. (2007) used a goniometer to measure Q angle and found that runners with large or asymmetric Q angles may be more susceptible to injury while running. On the other hand, de Oliveira Silva et at. (2015) used 3-dimensional analysis to measure Q angle and discovered that in the patellofemoral pain group, peak dynamic knee valgus was found to be greater. Because of the higher muscular and mechanical demands required to complete movement activities, PFPS mechanisms can be better examined in dynamic rather than static conditions (de Oliveira Silva et al., 2015).

There are many static and dynamic anatomical differences between males and females. Static refers to the differences present at rest whereas dynamic refers to the differences that occur during movement (Shampain et al., 2018). For recreational and competitive athletes, these differences can become a greater concern as to whether an athlete can continue to participate. These reasons can include or lead to overuse injuries, PFPS, or anterior cruciate ligament (ACL) injuries causing the athlete to be sidelined. When comparing female and male collegiate athletes, overuse injuries such as stress fractures occur twice as frequently in females (Shampain et al., 2018). This is due to the anatomical fact that males generally have a greater muscle mass than females which protects against stress fractures (Shampain et al., 2018). With regular heavy resistance training, females can increase bone density and muscle mass in order to decrease the risk of overuse injuries. When it comes to PFPS in females, researchers have determined that weakness in the hip abductor, external rotator, and extensor is associated with increased hip adduction and internal rotation. This promotes a dynamic Q angle which changes as the knee is flexed, creating a lateral force on the patella (Louden, 2016) which can contribute to patellofemoral pain. ACL injuries occur two to eight times more often in females than in males (Shampain et al., 2018; Tillman et al., 2005). Many factors can contribute to ACL injuries, but one conjecture is that a greater Q angle increases the quadriceps lateral force resulting in increased valgus tension across the knee, thus making females more susceptible to ACL injury (Shampain et al., 2018; Tillman et al., 2005). This can occur as a result of patellar instability, subluxation, and dislocation.

Statically, females tend to have a lower center of gravity, shorter and smaller limbs, as well as greater laxity and flexibility of their joints. Whereas males have more muscle mass and tend to be taller (Shampain et al., 2018; Grelsamer et al., 2005; Khasawneh et al., 2019). In addition, as they mature, female's intercondylar notch size does not widen as much as males, (Shampain et al., 2018), as seen in Figure 4. Due to a smaller intercondylar notch volume, smaller ACL volumes and steeper posterior slopes in the lateral tibial plateau, females ACL injury risk is increased 2.4-9.5 times (Simon et al., 2010). It is hypothesized that a narrower notch results in a greater force on the ACL (Geng et al., 2016).

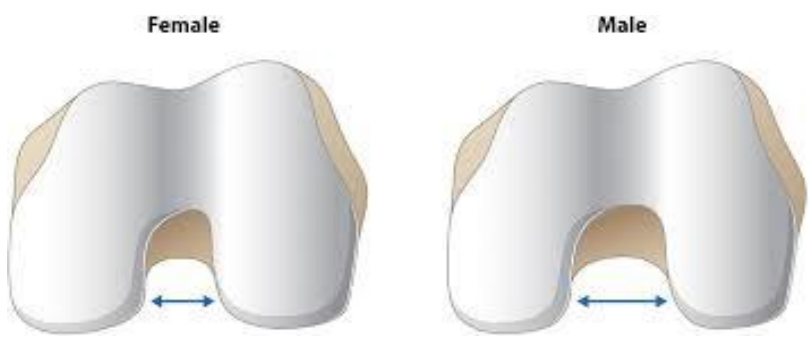

Figure 4. Male and female differences in the intercondylar notch of the femur. Average femoral intercondylar notch width for males is 13.9 +/- 2.2mm while females is 15.9 +/- 2.5mm. (Shampain et al., 2018; Shelbourne et al., 1998)

Post-puberty, there are non-anatomical differences that can affect posture and movement. Females have higher estrogen levels, resulting in increased laxity, diminished neuromuscular control of the knee, and delayed hamstring and quadriceps contraction when planting their foot (Shampain et al., 2018). Dynamically, for example when landing from a jump, increased activity of the rectus femoris, decreased activity of the gluteus maximus, and more upright posture consequently make females place weight on their forefoot increasing their risk of injury such as overuse, patellofemoral pain syndrome, or ACL injuries compared to males (Shampain et al., 2018; Zazulak et al., 2005).

One of the many important differences between male and female athletes is that female athletes have wider pelvises, increasing their hip varus angle (angle between head and shaft of the femur), knee valgus angle (angle between the femur and the ankle, seen when knee collapses inward), and femoral anteversion (knees and feet turn inward, commonly referred to as “pigeon-toed”). These static and dynamic anatomical differences contribute to females having a greater Q angle (Shampain et al., 2018).

Interestingly, race may play a slight factor in Q angle differences between females. According to Hill et al., (2021) higher Q angles have been noted in Nigerian females compared to Caucasian females due to the fact that Nigerian females have a larger ratio between tibia length and femur length. This anthropometric difference is also seen in African American females and white American females (Hill et al., 2021).

Patellofemoral Pain Syndrome

Patellofemoral Pain Syndrome (PFPS), also referred to as runner’s knee (Shampain, 2018), is one of the most common lower extremity injuries that can become chronic for one third of runners (Davis, 2020). It is described as anterior or retro-patellar knee pain around the patella with excessive patellofemoral joint stress (Louden, 2016). A major risk factor for PFPS is poor or improper running mechanics including excessive hip adduction, hip internal rotation, and greater knee internal rotation (Noehren, 2012).

PFPS is hypothesized to be caused by fatigue, which causes the gluteus medius and maximus to lose their ability to control hip adduction and internal rotation (Noehren, 2012). In addition to strengthening lower extremity muscles, reducing hip adduction, increasing forward lean, transitioning to a forefoot strike pattern, and increasing cadence are used to address PFPS (Davis, 2020).

In a study of high school cross-country runners, Rauh et al. (2007) concluded that runners with larger or more asymmetric Q angles had greater risk for developing a running injury, as well as having greater time lost from participation. Girls (right: 15.8° $\pm$ 4.1°; left: 15.0° $\pm$ 3.8°) portrayed significantly higher mean Q angles compared to boys (right: 12.7° $\pm$ 3.7°; left: 12.1° $\pm$ 3.5°). A Q angle of $\geq$20° puts runners at a 1.7 times greater risk of injury compared to a Q angle in the 10-15° range (Rauh et al., 2007). Rauh et al. (2007) also found asymmetry ($\geq$ 4° difference between right and left Q angles) contributed to a 1.8 times greater risk of injury. It is important to note that increasing the Q angle by 10% increased the load on the patellofemoral joint by 45% (Almeida et al., 2016). Because of the increased load, there may be a rise in PFPS as well as long term degeneration of the joint cartilage of the patella.

Females are twice as likely to have more patellofemoral pain compared to males (Loudon, 2016; Noehren et al., 2012). Studies have indicated that large Q angles may be a risk factor for lower extremity running injuries due to the greater quadriceps forces acting at the knee and patellofemoral joints (Nguyen et al., 2009; Ramskov et al., 2013). In segmenting out the contributing components of the Q angle, studies have reported large Q angles are associated with calcaneal eversion (foot pronation) and internal tibial rotation (Kuhn et al., 2002; Daneshmandi et al., 2011; Tillman et al., 2005). It is hypothesized that the increased calcaneal eversion is produced by increased valgus at the knee caused by the greater Q angle (Tillman et al., 2005).

Tillman et al. (2005) showed a greater association between tibial rotation and calcaneal movement than seen in previous studies. Livingston and Spaulding (2002) demonstrated that Q angle magnitude varies with changes in foot position, increasing as the foot rotates internally (pronation) and decreasing as the foot rotates externally (supination). It is therefore suggested that there is a relationship between foot position and dynamic Q angle during gait.

Some research suggests that hyper-pronation is a factor in lower extremity pain (Kuhn et al., 2002) as lower extremity motion results from the ground foot strike position (Dugan & Bhat, 2005; Loudon, 2016). Abnormal foot rotations cause corresponding lateral displacement of the patella, changing the force vector of the quadriceps and thus increasing the Q angle (Kuhn et al., 2002). However, there are studies that have not found a correlation between static Q angle and PFPS due to pronation, but it has been suggested that a better approach is to evaluate Q angle dynamically using video analysis (Loudon, 2016). Movement retraining has been shown to be an effective intervention used in physical therapy to reprogram motor function with the purpose of reducing pain or injury risk and has been used to address patellofemoral pain (Davis, 2020).

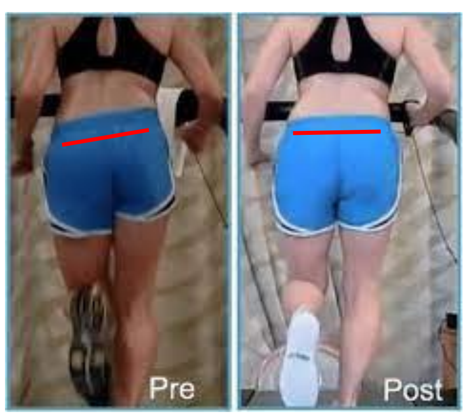

Figure 5. Pre and post hip mechanics after retraining and strengthening.

(Figure adapted from Davis, 2020)

With evidence that suggests the magnitude of the Q angle can be altered with a change in foot position (Livingston & Spaulding, 2002), the authors proposed that gait retraining to correct pronation will lower an individual's dynamic Q angle and therefore be less prevalent to injuries.

Load Carriage

Load carriage relates to Q angle through its effect on gait characteristics. Loads are commonly carried on a belt, vest, or backpack by students, hikers, cross training athletes, and warfighters/soldiers; this is referred to as load carriage. According to Boffety et al. (2019), individuals change their stride rate, stride length, speed when carrying a weight, attempting to enhance efficiency while minimizing energy expenditure as opposed to when carrying no load. Load should be carried or distributed as close as possible to the center of mass in order to minimize load movement which can shift by 40% due to its rotational inertia while walking, resulting in more stress and load (Golriz et al., 2018). As a result of the body's physical adjustment, the head tends to lean forward therefore flexing the trunk, stressing the ligaments and intervertebral discs of the lumbar region as well as hyperextending the cervical vertebrae, removing the shock-absorbing curvature that exists naturally (Dahl et al., 2016). This could impact the low back, shoulders, and neck causing chronic pain. Besides postural compensations like musculoskeletal imbalances and pain there are also internal demands that arise, for example, an increase in oxygen demand, energy cost, VO_2 (volume of oxygen), minute ventilation, heart rate, movement economy, and overall exertion (Golriz et al., 2018).

The combination of load carriage and fatigue compromise gait resulting in the significant decrease of gait width variability, hip range of motion, trunk range of motion, flexion/extension of knee, and pelvic rotation (Qu & Yeo, 2011). Qu & Yeo (2011) found a backpack load of 15kg resulted in increased step width variability and suggested that the fatigued muscles controlling the gait characteristics may be a factor. Overall, this can cause instability and compensation of one's stance, therefore negatively affecting gait pattern which could have an impact on injuries.

Hypothesis

We hypothesize that the dynamic Q angle will be greater during running than walking and the dynamic Q angle will be greater during loaded conditions than unloaded conditions.

Methods

Participants

This study included 10 healthy female participants between the ages of 18 and 35. Participants were included if they were recreational or competitive female athletes, had no injury within the last 3 months or lower extremity surgery within the last year, had no leg length discrepancies of $\geq$ 1.27cm (Heiderscheit et al., 2000), and did not use an orthotic. These inclusion criteria were selected to minimize the risk of injury as well as control for factors that can affect gait.

Participants were selected to only be female due to research showing that musculoskeletal issues related to Q angle measurements impact the female population more than the male population (Almeida et al., 2016; Shampain et al., 2018).

Additionally, participants could not be pregnant because the added load could cause extra stress on the female's body affecting the fetus. Also, gait biomechanics are altered when a woman is pregnant. Women over the age of 35 were excluded so that the outcomes of this research are comparable to previous research. This also limits age related variation in gait patterns so that our findings are related to load carriage and not aging in women.

Participants' rights and welfare were always protected within this study. All participation was voluntary, and all participants were free to withdraw from participation at any point in the study. This study was conducted under the approval of the Institutional Review Board (IRB) at The Ohio State University.

Recruitment and Selection

Subjects were recruited from The Ohio State University, Walsh University, and surrounding areas near Columbus, Ohio. When reaching out to students/employees, they were contacted via email as well with the use of flyers posted around campus and the study was explained to them.

Questionnaires

Participant demographics and previous injuries were recorded using custom questionnaires. The international Physical Activity Questionnaire long form (IPAQ) measured physical activity levels in recreation and professional environments. Low back pain was assessed using the Oswestry Low Back Disability Questionnaire (ODI) and the Fear-Avoidance Beliefs Questionnaire (FABQ). Questionnaire data for this research was collected through a paper survey.

After reviewing all the questionnaire variables, the relevant data extracted were the participants' responses regarding age, height, weight, competitive running experience, recreational running experience, load experience, and lower back pain within the last 12 months.

Testing Procedure

Institutional review board approved informed consent were signed and obtained from participating subjects. For this study, participants completed a set of questionnaires in order to determine participant characteristics.

Measurements were taken for each participant in one session by the same researcher. Participants performed all tasks in their personal running shoes and spandex/compression shorts. Q angle was measured both statically and dynamically. Static measurements were taken standing in an anatomical position using a goniometer with and without the weighted vest, as shown in Figure 6. The load worn was 10% of the participant's body weight. The goniometer is a common non-invasive tool used to measure the angle of a joint and is used throughout the medical field in diverse clinical settings, especially in orthopedics and physical therapy. However, when assessing the accuracy and reliability of a standard, short goniometer it is considered inaccurate and even compared to the accuracy and reliability of visual estimation (Hancock et al., 2018) due to user error. Therefore, to minimize the variability that measuring with a goniometer can have, the researcher was instructed by trained personnel, practiced many measurements before beginning data collection, and used stickers placed on the right and left anterior superior iliac spine, midpoint of the patella, and tibial tuberosity in order to measure the static Q angle. The goniometer axis was placed at the center of the patella, with the stationary arm of the goniometer aligned to the anterior iliac spine and the moving arm of the goniometer aligned to the tibial tuberosity. Two measurements were taken and averaged for each variable.

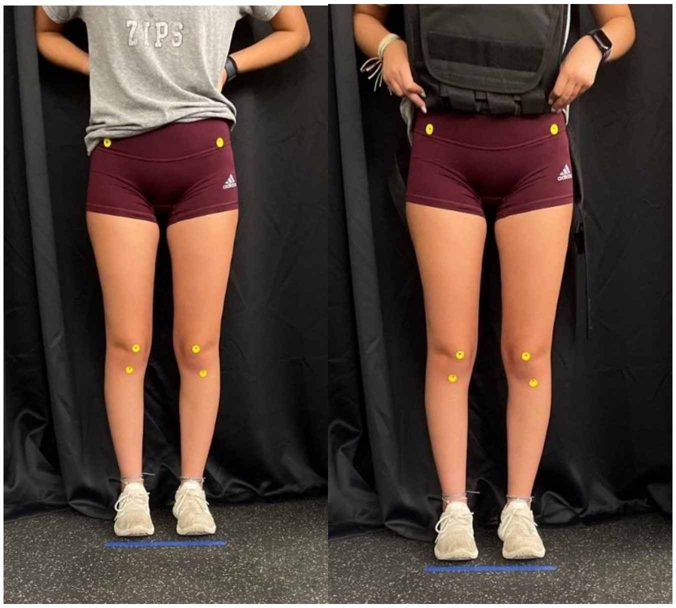

Figure 6. Static Q angle measurements taken using a goniometer with and without the weighted vest.

Dynamic measurements were taken using an 8-camera motion capture system (Vicon Motion Systems, Centennial, CO) while walking and running on an instrumented treadmill with embedded force plates (Bertec Corporation, Columbus, OH) at a self-selected pace for five minutes each with and without the weighted vest. To capture the dynamic measurements, retro-reflective marker clusters were attached to the participants on the head, upper and lower back, bilateral upper arm, bilateral forearm, bilateral hand, bilateral thigh, bilateral shank, and bilateral foot, as shown in Figure 8. Once the clusters were secured, the participant stood on the treadmill to determine body weight. Then the same researcher collected the participant's height and identified key anatomical landmarks: nasal bridge, bilateral medial and lateral segments of the elbow, bilateral ulna, bilateral radius, bilateral tip of third phalange, bilateral anterior superior iliac spine, bilateral posterior superior iliac spine, bilateral medial and lateral femoral epicondyle, bilateral medial and lateral malleolus, bilateral calcaneus, and bilateral second metatarsal head, as shown in Figure 9. After the participant's skeleton was built, they were fit into the harness for safety to complete their five-minute warm-up. Each participant's activity order (unloaded walk, unloaded run, loaded walk, and loaded run) was block randomized because the loaded and unloaded conditions were always paired together. The participant received a three-minute rest period to recover after each walking and running activity. After either the unloaded or loaded activity, the participant added or took off the weighted vest and then the system was re-calibrated to begin their final activities. The study took about 2 hours to collect data for each participant.

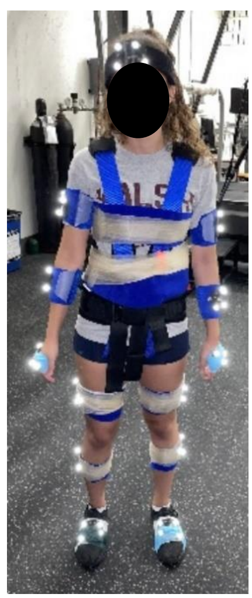

Figure 8. Dynamic Q angle measurements captured using reflective marker clusters.

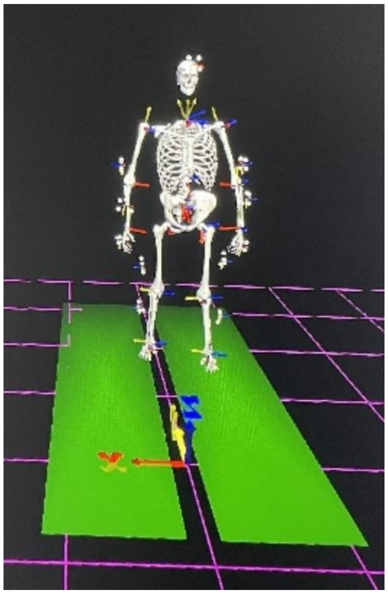

Figure 9. Key anatomical landmarks of the participant's skeleton.

Data Analysis

As previously defined, dynamic refers to the differences that occur during movement (Shampain et al., 2018). Therefore, dynamic movement was measured during the stance phase at heel strike, midstance, and toe off for walking and at heel strike and midstance for running, as shown in Figure 10. Specifically for running, we subtracted the knee deviation data points for toe off to midstance and midstance to heel strike whereas for walking, we subtracted the knee deviation data points for heel strike to toe off, toe off to midstance, and midstance to heel strike. After each trial, an average was taken for the right and left of each variable.

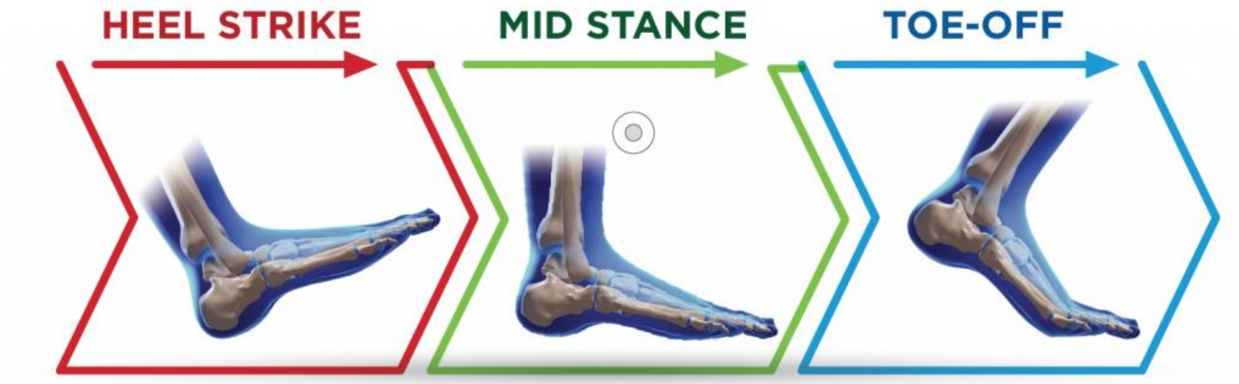

Figure 10. Gait cycle during stance phase.

(Wong, 2019)

The Motion Monitor (The MotionMonitor, Chicago, IL) was used to provide an interface for collecting synchronous data from the force plates as well as biomechanics data cleaning and data filtering/processing. The combination of this information provided precise measurement of static postures and dynamic movements.

Statistical Analysis

Statistical comparisons were performed with repeated measures ANOVAs in SPSS (Statistical Package for the Social Sciences) to compare static Q angle measurements (Table 3) for standing and dynamic Q angle measurements (Table 4) for both walking and running with and without load as well as reporting means and standard deviations for participant demographics and questionnaires (Table 2).

The measured static Q angles portrayed a slightly lower mean compared to the literature findings. For example, Rauh et al. (2007) found that females (right: 15.8° $\pm$ 4.1°; left: 15.0° $\pm$ 3.8°) had significantly higher mean Q angles compared to males (right: 12.7° $\pm$ 3.7°; left: 12.1° $\pm$

3.5°). However, our findings found that females (right: 8.45° $\pm$ 1.7°; left: 9.6° $\pm$ 2.4°) portrayed a slightly lower mean Q angles compared to the male findings. This could be due to goniometer user error which may lead to skewed results in the subject's data. Static Q angle measurements on the right side were smaller than on the left side for both unloaded and loaded variables. This could be due to the fact that all 10 participants were right leg dominant leading to the right side of their pelvis, and specifically with the iliopsoas muscle being tighter and not as wide as the left. According to Lifshitz et al. (2020), the iliopsoas muscle is the primary hip flexor which assists the pelvis in order to tilt anteriorly and initiates the swing phase during running. On the left side, the static Q angle was smaller without the vest and slightly larger with the vest as hypothesized, however on the right side, the static Q angle was roughly the same with and without the weighted vest.

During dynamic Q angle measurements positive values are considered valgus and negative values are considered varus. Values were largest in right knee deviation while unloaded running. Knee deviation is the difference between frontal plane knee angle at heel strike, midstance, and toe off for walking and at heel strike and midstance for running. During walking on the right and left, the participants were in valgus, whereas during running on the right and left the participants were in varus. Therefore, dynamic Q angle is different at a walking pace versus running pace but the 10% body weight load was not found to affect whether the participant was in either valgus or varus. The dynamic Q angle difference at a walking versus running pace could be due to weak hip muscles causing the knee to collapse inward during running (Powers, 2010). However, since we did not test muscle strength, it is hard for us to tell if this finding is accurate within the data. To be considered proper running form, the runner's knees are supposed to stay in line with their hips, but if the hip muscles are weak, they will not be able to support the runner's body weight, causing the knees to collapse inward, potentially leading to the varus position which may contribute to patellofemoral pain syndrome in the future.

Interestingly, unloaded to loaded variables did not significantly affect dynamic Q angle. Although there was no significant difference overall, the main effect for speed was nearing significant ($p = 0.067$) for the left knee, as shown in Figure 11. This means that there was a trend towards dynamic Q angle decreasing when participants went from walking to running. In the graph, the lines look like they change, but the values are so small that they are essentially parallel lines with no difference. However for the right knee, there was a significant difference between walking and running ($p = 0.002$), as shown in Figure 12, where the dynamic Q angle decreased during running compared to walking. During walking, the knee was going into more valgus through the motion (deviation is a positive value) whereas in running the knee was moving into a more varus position throughout the movement (deviation is a negative value). In both cases, the degrees of change are relatively small (~1°). Looking at Figure 11 and Figure 12, we can interpret that during left knee deviation speed and load did not reach significance whereas right knee deviation almost reached significance (if only the lines crossed leading to an interaction). Therefore, with more subjects we could continue to see a trend of load and speed. Also, on the left, angle had a low effect size whereas on the right, speed had a high effect size (right = 0.754). So potentially the left side was trending in the same direction as the right side, even though it did not reach significance. Speed was not by chance since we manipulated it by having specific walking and running variables, however load was not manipulated since it was specific for each participant's body weight.

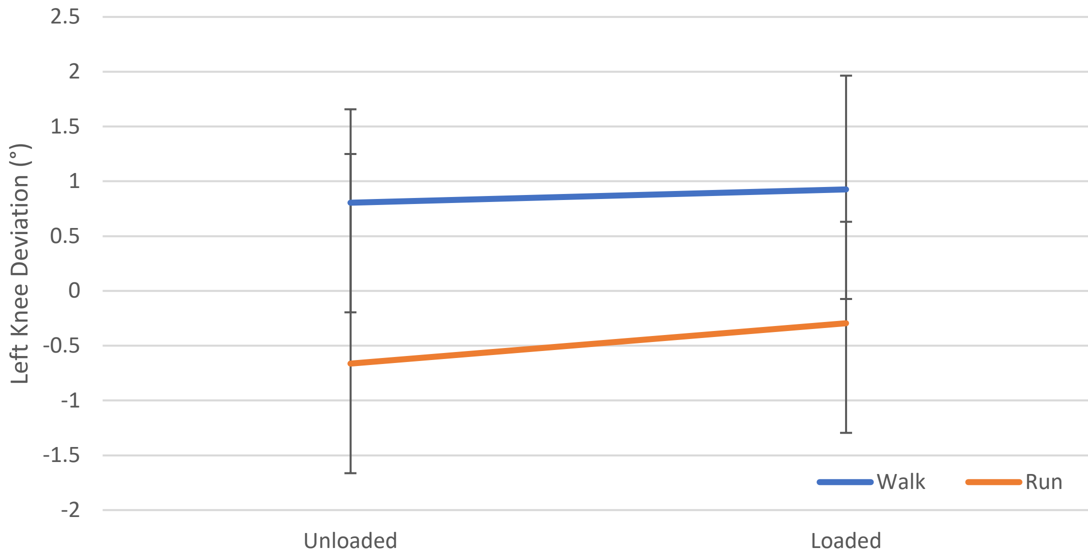

Figure 11. Left knee deviation between load and speed. The main effect for speed was nearing significant (p = 0.067) meaning that there was a trend towards dynamic Q angle decreasing when particiapnts went from walking to running.

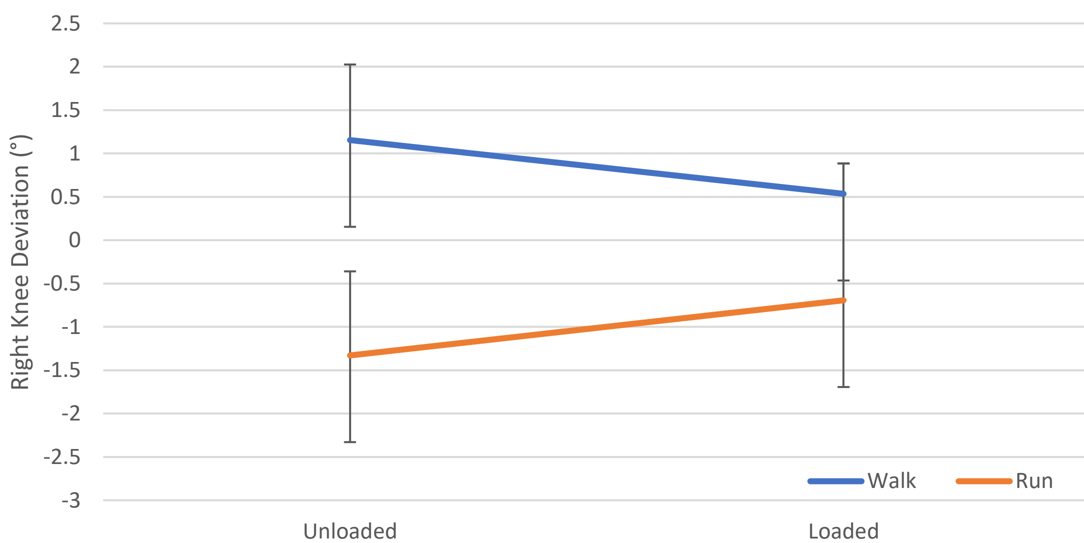

Figure 12. Right knee deviation between load and speed. There was a significant difference between walking and running (p = 0.002) meaning that the dynamic Q angle decreased during running compared to walking.

Subject Characteristics

This study included 10 healthy female participants between the ages of 18 and 35. Table 2 summarizes the group descriptive data between all 10 participants for age, height, weight, competitive running experience, recreational running experience, and load experience. It was noted that 3 out of the 10 participants reported lower back pain within the last 12 months but reported no pain at the time of the study. The 3 participants that reported lower back pain within the last 12 months scored 1.67 ± 1.25 on the ODI Questionnaire. Low back pain could be impacted by load carriage, but the changes did not impact the results on the knee. Load carriage experience included school backpacks to recreational hiking with packs.

Results

Table 2. Participant characteristics for study variables.

	Mean	**Median**
Age (yr)	21.8 ± 3.2	21
Height (in)	64.1 ± 2.3	63.8
Weight (lbs)	134.7 ± 18.5	125.5
Competitive Running Experience (yr)	5.2 ± 5.4	3
Recreational Running Experience (yr)	6.7 ± 3.9	8

Table 3. Static Q angle measurements (means and standard deviations).

	Left	**Right**
Unloaded	9.6° ± 2.4°	8.5° ± 1.7°
Loaded	10.3° ± 2.2°	8.4° ± 1.6°

Table 4. Dynamic Q angle measurements (means and standard deviations).		
	Left	**Right**
Unloaded Walk	0.80° ± 0.85°	1.15° ± 0.87°
Unloaded Run	-0.68° ± 1.80°	-1.39° ± 0.96°
Loaded Walk	0.92° ± 0.98°	0.83° ± 1.00°
Loaded Run	-0.25° ± 0.88°	-0.79° ± 1.44°

Discussion

Research has demonstrated (Livingston & Spaulding, 2002; Nguyen et al., 2009; Ramskov et al., 2013) that having a larger Q angle increases the potential for injury during dynamic activities. Movement is affected by load carriage, however the interaction of how Q angle and load carriage impact movement together is unknown. This paper discusses the importance of understanding how larger Q angles can alter movement patterns with the potential to cause patellofemoral pain syndrome as well as increase injury risk. Our purpose was to investigate if adding load changes the Q angle during gait and if differences exist between participants when walking and running on a treadmill with a weighted vest. These results supported our hypothesis that the dynamic Q angle will be greater during running than walking, but the results did not support the hypothesis that the dynamic Q angle will be greater during loaded conditions than unloaded conditions. In terms of females who walk and run with or without load, these results can be applied to future research in order to provide more knowledge and understanding of how load carriage may affect Q angle, and thus alter movement patterns with the potential to cause PFPS as well as increase injury risk.

Limitations

The most significant limitation of this study was the sample size. The researchers aimed for a sample size of 30 participants, however only 10 qualified subjects volunteered to take part in the study. The limited sample size decreases the representation of the overall population this study was aiming to study; therefore, this study should be considered a pilot study for more future research. Also, it is important to note that there was not much variability in age or height within the participants. Another limitation present is the duration of training experience each athlete had prior to conducting the study. The range of running experience in the participants of the study ranged from one to ten years of prior training and the range of load carriage experience in the participants of the study ranged from zero to eighteen years of prior training before the study which could lead to skewed results depending on their training experience. For safety of the participants, the added load was limited to 10% (average = 12.55 lbs.) of their body weight compared to military load carriage packs of 60-100 pounds. We chose 10% of the participants body weight because increasing the Q angle by 10% increased the load on the patellofemoral joint by 45% (Almeida et al., 2016) therefore we wanted to minimize the increased risk of patellofemoral pain while also trying to identify whether if adding load changes the Q angle during gait. This amount of weight may have not been enough to affect the participants' gait. When measuring the static Q angle standing, the researchers used a goniometer to measure the participants. This technique produces a possibility for user error which may lead to skewed results in the subject's data. However, the possibility for skewed results was limited because the same method of measuring was used for each participant and done by the same researcher.

Conclusion

In summary, dynamic Q angle in females is different at a walking pace versus running pace however unloaded and loaded variables do not significantly affect dynamic Q angle either in valgus or varus positions. Future research should be conducted with larger samples in order to continue to see a trend between load and speed as well as further assess the effect of load carriage on a Q angle during walking and running.

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