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Influence of Q-angle on lower extremity coordination during curve running.

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Influence of Q-angle on Lower Extremity Coordination During Curve Running

by

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ABSTRACT

The purpose of this study was to examine the effects of Q-angle on running curve kinematics and kinetics. Each dependent variable was evaluated for statistically significant differences using a two factor mixed analysis of variance with repeated measures on the track curve variable. A significance level of $P < 0.05$ was employed for the evaluations. Independent variables were curve radius and Q-angle. Dependent variables included peak vertical force, time of peak vertical force, average vertical force, vertical impulse, continuous relative phase, and variability of continuous relative phase. Three levels of curve radius were used: straightaway, 36.5 m, and 18.5 m. Q-angle had two levels: high Q-angle (HQ) and low Q-angle (LQ). Subjects exhibiting static Q-angles of 14° or less were included in the LQ group while subjects with static Q-angles of 16° or greater were included in the HQ group.

Results of the study did not identify any significant differences between groups or curve conditions for any kinetic variables. As well, there were no significant interaction effects between groups and curve conditions for any kinetic variables. However, there was a trend leading towards significant differences between Q-angle groups for time of peak force.

Continuous relative phase (CRP) was calculated to demonstrate phase relations between the knee and ankle joints while variability of CRP (vCRP) characterized the level of stability of the coordination of the two joints. There were no significant differences between groups for either CRP or vCRP. There were no significant differences in the vCRP of curve conditions, however there were significant differences

between curve conditions for CRP. This suggests that there may be stability in the coordination patterns despite being out of phase with each other.

The conclusions of this study are that runners with high Q-angles are not at greater risk for injury than runners with low Q-angles. Furthermore, all runners, regardless of individual Q-angle measurement, may be at increased risk of injury while repetitively running curves.

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CHAPTER I

INTRODUCTION

An estimated 40 million people run in North America (Reber, Perry, & Pink, 1993). Cavanagh (1987) predicted that a runner running at 3.83 m/s will contact the ground approximately 5,100 times an hour. If a runner were to log 160 km of running in a week, this would total approximately 3,000,000 foot strikes in a year (Subotnick, 1975). Munro, Miller, and Fuglevand (1987) reported ground reaction forces of 1.6 times body weight while running at 3.0 m/s which increased to 2.3 times body weight at 5.0 m/s. This is a significant amount of force that a runners lower extremities must attenuate properly to avoid injury. The number of runners incurring an injury is within the range 24-85% (Hoeberigs, 1992; Macera, 1992; van Mechelen, 1992). In a survey of 688 runners in Canada, 57% reported injuries with 58% of the injuries being new and directly attributed to running (Walter, Hart, Sutton, McIntosh, and Gauld, 1988). Overuse injury is the most common form of injury, accounting for 50 to 75% of injuries (Brill, & Macera, 1995). An overuse injury occurs when the adaptive capacity of the body is unable to withstand or adapt to the repetitive forces that are being applied to it (Taunton, McKenzie, and Clement, 1988).

Injuries lead to a decrease in training or termination of training in approximately 30 to 90% of incidences (van Mechelen, 1992). However, the exact mechanisms causing injury are not yet understood. Structural misalignment is receiving the greatest amount of attention in research currently because of the apparent associations discovered between it and injury incidence. Of particular interest is misalignment that causes excessive pronation of the subtalar joint (Heiderscheit, Hamill, and van Emmerik, 1999). Excessive pronation has been shown to be most notably related to shin splints, stress fractures, and non-specific knee pain (Reber et al., 1993; Stergiou, Bates, and James, 1999; Viitasalo, and Kvist, 1983).

Despite the information that has been collected regarding running injuries and running mechanics it is not possible to directly state which runners will become injured. This may be due in part to the experimental set ups used in research. Running conditions are simulated on treadmills and straightaways but are still inadequate for replicating true running conditions. Treadmill running has been shown to demand slightly different mechanics than running on the road, grass, or track (Nigg, DeBoer, & Fisher, 1995). In particular, the change in running mechanics to run the curve of a track is significantly different than running a straightaway (Greene, 1985) but has received little attention in research. The mechanics of running a curve places abnormal stress on the lower extremity that may lead to injury.

Heiderscheit et al. (1999) hypothesized that the approximate angle that the quadriceps muscle pulls on the patella (Q-angle) would be significantly related to differences in timing of the lower extremity segments, particularly in how they relate to each other. The authors reasoned that a large Q-angle induces pronation of the subtalar joint, which becomes excessive during running, thereby altering the timing of segment movements. Although their reasoning was logical on why there should be a relationship between relative phase of segments and injury risk, they did not find any significant differences between a group with normal Q-angles and a group with large Q-angles while running a straightaway. Considering information regarding running track curves, although limited, it would be logical to expect relative phase differences between the segments of the lower extremity while running a track curve due to the alterations in running mechanics that are imposed on the runner in order to successfully navigate the turn.

Statement of the problem.

Previous research and modelling has shown that a runner must exert centripetal force to navigate the curve successfully and that the magnitude of this force is directly related to the radius of the curve and velocity of the runner. In order for the runner to apply this force a body lean toward the centre of the curve is required which causes excessive pronation of the left foot and excessive supination of the right foot, while running counter clockwise around the curve. The mechanics of running a curve are, therefore, significantly different than running a straightaway. The normal alignment and structural design of the body is adapted to attenuate the impact forces associated with running but does not appear to be adapted for curve running. Alterations in relative timing of segment movements of the lower extremity may occur due to the altered alignment of the body and affect how impact forces are attenuated. Further changes in alignment may be a result of increased Q-angle, which has been demonstrated to be related to various injuries in running populations. Large Q-angles are indicative of altered lower extremity alignment and may cause alterations in shock absorption such that inappropriate tissues are attenuating the forces rather than the structures and tissues of the midfoot and forefoot that are adapted to it. Large Q-angles may also be related to alterations in relative timing of segment movements due to the misalignment present in the lower extremity.

The purpose of this study was to investigate the continuous relative phase of the left lower extremity while running two different radii of track curve and compare the results with the continuous relative phase of running a straightaway. Furthermore, this study involved two subject groups of differing static Q-angle in order to investigate the relationship with continuous relative phase while running a track curve.

Hypotheses.

There were three general hypotheses in this study.

H1: Significant differences in measured dependent variables will be found between Q-angle groups.

H2: Significant differences in measured dependent variables will be found between track radii.

H3: Significant interactions will be found between Q-angle and track radii as they affect the measured dependent variables.

Limitations.

The results of this study are limited to an experienced running population since only experienced runners were used. Specifically, the results are only applicable to endurance runners. Furthermore, since velocity has a large effect on centripetal force and force may require a change in running mechanics through body lean, extrapolation of the results are also limited to velocities similar to the velocities that were present in this study. As well, not all tracks are of equal radii even if they are of equal length. Therefore, the results of the curve running are limited to running curves of equal radii. Finally, since this study only tested male subjects, the results are not applicable to female runners. All results in this study are limited by the small sample size of five subjects per grouping condition. Twenty-five subjects were initially recruited, however due to individual subject schedules, subject acceptance protocol, and equipment errors, the subjects were ten in number. This may be a significant factor in reducing the power of the statistical techniques employed.

Delimitations.

Since sprint running mechanics are different than endurance running mechanics (Greene, 1987), only endurance runners were tested. Individuals wore their own training shoes to prevent changes in running mechanics that occur when running in new shoes (Vagenas, and Hoshizaki, 1992). Furthermore, although the running surface was a rubberized mondo track, the surface of the force platform was not rubberized mondo.

Computations regarding compressive strength of the mondo surface were not conducted and therefore did not enable use of mondo on the force platform surface.

Subjects were instructed to run at their own relative training pace. This decreased the change in mechanics that imposing an unaccustomed velocity may effect. Furthermore, subjects were requested to abstain from running on the day of testing prior to test period to prevent fatigue from being an applicable factor.

Track radii were chosen to simulate standard track construction as outlined by the International Amateur Athletic Federation (IAAF), the world governing body for athletics, in their regulations handbook (2000). Although the IAAF only outline the range of parameters allowed for track dimensions, W. T. Armbrust (personal communication, May 5, 2000) relayed to the author standard track radii dimensions used to build tracks and it is from this communication that the radii for testing were chosen. However, the radii used in this study were not very different than those used in previous studies (Greene, 1985; Hamill, Murphy, & Sussman, 1987; Harvie, 1998) and should still allow for comparison.

Markers attached to the lower extremities of subjects were assumed to give good representation of segment and joint movements. Reinschmidt, van den Bogert, Nigg, Lundberg, and Murphy (1997) reported that markers attached to the lower extremity had good agreement with markers attached to the bones for providing representation of skeletal movement in knee flexion/extension. The authors did not analyze ankle flexion/extension, however it was assumed to be in good agreement with skeletal movement as well.

Operational definitions.

The following definitions were used in this study.

Abnormal Q-angle. Any measurement of Q-angle that is outside the range of 10° to 15° for males.

Angular Velocity. The velocity with which the distal segment rotates relative to the proximal segment about a shared joint.

Asymmetry. The variability in the relative phase representing the asynchronous movement of the coupled of segments .

Average Ground Reaction Force. The summation of ground reaction force for the entire stance phase divided by the total time of stance phase. Normally measured in Newtons (N). It was normalized by body weight and reported as percent (%) body weight.

Centrifugal Force. The horizontal force directed away from the centre of the turn. Equal to centripetal force if there is no departure from the curved path of the runner.

Centripetal Force The horizontal force directed towards the centre of the turn. Equal to centrifugal force if there is no departure from the curved path of the runner.

Contact Phase. From the moment the heel comes in contact with the ground to maximal foot pronation. Usually constitutes the first 25% of stance phase.

Continuous Relative Phase (CRP). The assessment of segmental coupling during the entire stance phase. Demonstrates the stability and coordination between segments.

Dorsiflexion. Flexion of the ankle bringing the top of the foot towards the lower leg.

Endurance Running/Runner. A runner that trains to compete at distances at least 3000 m in length.

Eversion. Rotation of the foot or joints of the foot about the anteroposterior axis in a medial direction.

Excessive Pronation. Any pronation that occurs for a greater length of time than normal or to a greater magnitude than normal.

Excessive Supination. Any supination that occurs for a greater length of time than normal or to a greater magnitude than normal.

Experienced Runner. Any runner that has been running and training regularly and consistently for at least two years.

External Rotation. Rotation of the tibia or femur about the longitudinal axis in a lateral direction.

Flight Phase. The period of time during running in which both feet are not in contact with the ground. Occurs between toe off with one foot and heel contact with the other foot.

Ground Reaction Force (GRF). The force that the ground exerts against the foot in reaction to the action force applied by the foot during stance phase.

Impact Phase. Synonymous with contact phase. Occurring for the initial 25% of stance phase.

Impact Shock. The force applied to the ground by the impact of the foot. Primarily regarded as vertical force since it has the greatest magnitude and most significance on the remainder of the body.

Injury. Any physical wound or damage resulting in cessation of or decrease in intensity of running.

Internal Rotation. Rotation of the tibia or femur about the longitudinal axis in a medial direction.

Inversion. Rotation of the foot or joints within the foot about the anteroposterior axis in a lateral direction.

Leg Length Discrepancy(LLD). The difference between left and right leg lengths, measured from the greater trochanter to the respective lateral malleolus. Deemed excessive if greater than 1.3 cm and resulted in the removal of subject from the study.

Loading Rate. The initial part of the vertical ground reaction force curve which represents the rate by which force is applied to the ground through the foot. Usually measured in units of Newtons per second (N/s) it was normalized by body weight to units of Body Weight per second (BW/s).

Maximal Vertical Ground Reaction Force. The greatest magnitude of force that occurs in the vertical direction. Measured in Newtons (N) and normalized by body weight to units of percent Body Weight (%BW).

Midstance phase. The middle 50% of stance phase during which the entire foot is in contact with the ground. Occurs from maximal pronation to maximal supination.

Misalignment. Any alignment of the lower body which decreases the ability of the lower extremity to effectively attenuate impact shock and generate propulsive force.

Muscle Fatigue. The inability of a muscle to maintain a given level of force production.

Neutral Position. The position the foot is in without experiencing either pronation or supination. Usually occurs when the subtalar joint is at 45° angle from horizontal.

Normal Q-angle. Normal Q-angle for males will be defined as the Q-angle being within the range 10° to 15°.

Overuse Injury. Occurs when a body structure or system is unable to withstand or adapt to the repetitive forces it is being subjected to.

Phase Angle. Calculated as the angle formed from the right horizontal to any given point on a phase plot.

Phase Plot. Graphical representation of segment angular velocity as a function of segment angular position.

Plantarflexion. Extension of the ankle moving the top of the foot away from the lower leg.

Pronation. The rotation of the foot around the anteroposterior axis involving the triplane motions of forefoot abduction, ankle dorsiflexion, and calcaneal eversion.

Propulsive Phase. The last 25% of stance phase wherein the foot becomes a rigid lever and force is applied to the ground to generate a positive anteroposterior impulse. From the time of maximal supination to the time of toe off, during which the heel is not in contact with the ground.

Quadriceps Femoris Angle (Q-angle). The approximation of the relative angle of pull that the quadriceps femoris has on the patellar tendon. Measured as the acute angle formed by the intersection of the line connecting the anterior superior iliac spine to the centre of the patella with the line connecting the centre of the patella to the tibial tuberosity.

Relative Phase. The influence one segment has on another characterized by synchronous movement times.

Running Velocity. A vector property reflecting displacement over the time taken for that displacement to occur. Measured in metres per second (m/s).

Segment Angle. The angle formed between two segments joined about a single joint. Calculated as the angular position of the distal segment relative to the proximal segment.

Stance Phase. The time during which any portion of the foot is in contact with the ground. Occurs from heel contact to toe off of the same foot.

Successful Trial. When the subject appears to have not altered running mechanics in order to target the force plate, and has contacted the force plate with the entire left foot.

Supination. The rotation of the foot around the anteroposterior axis involving the triplane motions of forefoot adduction, ankle plantar flexion, and calcaneal inversion.

Symmetry. The invariance of the relative phase symbolizing the synchronous movement timing of the coupled segments.

Time To Maximal Vertical Ground Reaction Force. The time at which maximal vertical ground reaction force occurs following touch down. Will be normalized to percent of total stance time.

Toe Off. Time at which toe discontinues contact with the ground, normally characterizing the end of stance phase and the beginning of flight phase.

Vertical Ground Reaction Force. The vertical component of the ground reaction force, perpendicular to the ground.

CHAPTER II

LITERATURE REVIEW

The purpose of this study was to examine the effects of Q-angle on running curve kinematics and kinetics. Risk of injury was determined by variability in the continuous relative phase of segment couplings.

Normal mechanics of the lower extremity in running.

There are two phases in the running cycle, those being stance phase and flight phase. The first part of the stance phase is the contact phase and may account for 20-25% of the running cycle. The second phase of stance is the midstance phase during which the entire foot is in contact with the ground and includes from 25-75% of the stance time. The final phase of stance is the propulsive phase when the heel is lifting from the ground and it occurs for the final 25% of stance (Subotnick, 1985). Flight phase takes place between toe off of one foot and heel contact of the other foot.

When the heel contacts the ground, it may be in a neutral (Subotnick, 1985), pronated (Tiberio, 1987), or supinated (Donatelli, 1985) position. Immediately following contact, the foot pronates. Pronation is the rotation of the foot around the anteroposterior axis involving the triplane motions of forefoot abduction, ankle dorsiflexion, and calcaneal eversion (De Wit, & De Clerq, 2000). Supination is the rotation of the foot around the anteroposterior axis involving the triplane motions of adduction, plantar flexion, and inversion (Donatelli, 1985).

Pronation of the foot occurs by pronation of the subtalar joint and takes place until the end of midstance when the foot is in full contact with the ground. Subtalar joint pronation causes eversion of the calcaneus with plantar flexion and medial movement of the talus head. Movement of the talus head medially induces a rotation of the talus body which thereby internally rotates the lower extremity (Tiberio, 1987).

Subtalar joint pronation is associated with concurrent knee joint flexion and tibial internal rotation (De Wit, & De Clerq, 2000). The position of pronation allows for shock absorption of impact forces as energy is transferred through the unlocking of the transverse tarsal joints of the calcaneal-cuboid and talonavicular joint (Adelaar, 1986). The increased flexibility of the foot at this point also allows it to adapt to the ground surface, maintain balance, and improve muscle contraction efficiency. Peak pronation occurs between 38-50% of stance phase (Neely, 1998). Knee joint flexion which is associated with pronation of the foot is also involved in impact shock attenuation (Stergiou et al., 1999) along with dorsiflexion of the ankle joint and flexion of the hip joint (Hintermann, & Nigg, 1998). Specifically, 70-80% of shock attenuation occurs before the knee (Stergiou et al.) and a further 15% reduction occurs within the knee (Kim, Voloshin, & Johnson, 1994).

Following midstance the subtalar joint begins to supinate and causes tibial external rotation accompanied by knee joint extension (De Wit, & De Clerq, 2000). Subtalar joint supination is initiated by inversion of the calcaneus (Donatelli, 1985). The talus moves laterally, in reverse motion observed during pronation, and directly causes external rotation of the tibia due to the tight articulation of the talus with the tibia (Tiberio, 1987). Supination stabilizes the midtarsal, subtalar, and talocrural joints creating a rigid lever for propulsion at the end of stance phase (Neely, 1998).

Normally a foot needs to pronate and supinate 6-8° from neutral. Calcaneal eversion is normally 10° and inversion is 20°. The neutral position is defined as occurring when the longitudinal axis of the lower limb and the vertical axis of the calcaneus are parallel (Donatelli, 1985). The movements of the subtalar joint and knee joint during stance phase are independent, with the rotation of the tibia being a required motion crucial for normal kinematic functioning of both joints (Tiberio, 1987). Timing of movements of the lower leg have been shown to become more synchronous with increased velocity (De Wit, & De Clerq, 2000).

Abnormal mechanics of the lower extremity in running.

There are many contributing factors which may lead to abnormal mechanics in running. These include muscle imbalance, lower extremity misalignment, and structural anomalies. Abnormal mechanics may not necessarily precipitate injury but may be important contributing factors when running intensity or time increases (Vagenas, & Hoshizaki, 1992). Lower extremity misalignment has been consistently implicated as the main proponent involving running related injuries (Viitasalo, & Kvist, 1983).

Excessive pronation usually occurs at the subtalar joint, generally occurring for greater than 25% of the stance phase. This occurs when the foot either begins to supinate too late or does not supinate at all. This results in an inability to effectively distribute impact shock forces (Donatelli, 1987) and does not allow the foot to become a rigid lever for propulsion (Fredericson, 1996). The inability to distribute impact forces causes excessive force to be absorbed by soft tissue and bone that may otherwise not be conditioned for it (Subotnick, 1985). Decreasing the amount of pronation will cause a transfer of impact energy to the heel, leg, knee, and hip rather than the forefoot and midfoot regions (Adelaar, 1986; Neely, 1998).

Associated with excessive pronation is tibial internal rotation. Since the knee joint begins to extend at approximately the same time as supination of the foot with normal running mechanics, transverse torsion will be experienced at the knee joint due to the prolonged tibial internal rotation (Fredericson, 1996). As well, if the tibia does not externally rotate at the correct time, due to lack of supination of the subtalar joint, then the knee joint is unable to extend without compensatory internal rotation of the femur. The compensatory internal rotation of the femur may cause an alteration in the tracking of the patella within the femoral condyles (Tiberio, 1987). McClay, and Manal (1997), reported that subjects who pronated displayed greater knee joint flexion than normal subjects. Increased knee flexion is associated with increased patellofemoral joint contact forces (McClay, & Manal).

Donatelli (1987) defines abnormal supination functionally as the inability of the foot to pronate. Excessive supination occurs during the entire stance phase or the foot pronates just before propulsion. Early supination is the equivalent of decreased pronation. Lack of supination before propulsion will not allow the joints within the foot to lock and thereby decrease the effective leverage for propulsion (Fredericson, 1996). Furthermore, deficits in supination at propulsion will cause an internal rotation of the tibia during a time that it should be externally rotating. Opposing torsional forces will result at the tibia with the distal end experiencing internal torque and the proximal end experiencing external torque. This may require the femur to internally rotate to allow continued extension of the knee joint (Fredericson).

Mechanics of running curves.

Hamill et al. (1987) stated that runners must change their body position and adjust their lower extremity function to successfully navigate a track turn. They found that at the midpoint of the track turn there were asymmetries between left and right lower extremities. While running counter clockwise the right foot supinated a great deal more than the left foot on the curve and more than the right foot when running a straightaway. Furthermore, the left foot was pronated significantly more than the right foot on either the curve or the straightaway.

Essentially, the path that the runner follows in navigating a track turn is decomposed into straight segments of length equal to stride length (Behncke, 1994). To accomplish this, the runner applies a shear force away from the curve centre producing a centripetal force on the runner due to ground reaction forces. This affects a torque on the runner's body away from the vertical and away from the track centre which requires the runner to lean towards the track centre to counteract the torque. This lean causes changes in the position and mechanics of the lower extremity (Hamill et al., 1987). Thus, the lean causes the right foot to supinate and the left foot to pronate more, relative to the body, than normal. Not only does the lean change the mechanics of the lower extremity, but changes the position of the body centre of mass relative to the lower extremity, thereby applying abnormal stress on the soft tissue and bones of the lower extremity (Behncke).

The centripetal force which allows the runner to navigate the turn is a function of the mass of the runner multiplied by the velocity squared, all divided by the radius of the curve (Hamill et al., 1987). Therefore, the amount of change in lower extremity mechanics is affected by the runner's velocity. Greene (1987), found that by having the turn banked, there can be a 10% effect on the velocity of the runner. He attributed this to increased foot force being available to maintain roll stability about the anteroposterior axis of the foot. As a runner traverses the turn, extra tendon force is required to prevent anteroposterior rotation of the foot which decreases the available foot force and decreases velocity by decreasing air time and increasing contact time. However, this study consisted of sprinters and may not be applicable to endurance running.

Harvie (1998) found that males ran track turns faster than the straightaway. As well, she reported that the right foot, for both males and females, created significantly more anteroposterior propulsive force than the left foot. Furthermore, males ran the turns faster with the right foot than the left. Since running velocity affects lower extremity mechanics, it would be expected that the right and left foot experience different mechanics due to their different velocities and both feet will experience altered mechanics when running a curve compared to running a straightaway. Harvie also studied females and observed that, unlike the males, females ran the turns with equal velocity for both feet. Therefore the velocity effects on female lower extremity mechanics during curve running may be less accentuated than the males. Additionally, she found the left foot, for both males and females, to be more involved in producing centripetal force than the right. Assuming the subjects were running counter clockwise, this would be applying mediolateral force on the left foot while hyperpronated and the right foot would be applying predominantly anteroposterior force for propulsion while hypersupinated.

Running related injuries.

Incidence of running-related injuries in runners ranges from 48-57% with the lower extremity injured most frequently in 67-86% of diagnoses (Macintyre, Taunton, Clement, Lloyd-Smith, McKenzie, & Morrell, 1991; Walter et al., 1988; Walter et al., 1989). Of this number 23-48% of injuries occur in the knee, 20% in the lower leg, and 17% in the foot (Macintyre et al.; Walter et al., 1988). Magee (1987) described the knee as being most susceptible to injury since it is located at the end of two long bones, the femur and tibia, forming two long lever arms and because it can not depend on bony configuration to provide strength and stability, rather it must rely on surrounding muscles and ligaments. Furthermore, it is the largest joint in the body and only has two degrees of freedom.

The rate for runner injuries relates to military infantry injury incidents where 82% of injuries are in the lower extremity (Almeida, Williams, Shaffer, & Brodine, 1999). However, the largest occurrence is in the ankle/foot, 34%; second is the knee, 28%; and 14% in the lower leg. The contrast of infantry injuries being greater in the ankle/foot may be due to the grouping together of the ankle and foot. Macintyre et al. (1991) do not state incidence of ankle injury, nor do they state whether it is included with the information for the foot. Military infantry presented injuries most often following increase in training volume, particularly running. However only 40% of the recruits became injured during the study, slightly lower than the occurrence in runners, 48-57%.

Hintermann, and Nigg (1998) state that the factors most associated with running related injuries in the foot and ankle include improper biomechanics, anatomical abnormalities, poor flexibility, muscle imbalance, shoe type, and running surface. Walter et al. (1989) did not find any anthropometric variables that they examined to be significantly related to injury risk. Variables included femoral neck anteversion, knee and patella alignment, rearfoot valgus and pes cavus/planus. Furthermore, Nigg, Cole, and Nachbauer (1993) were unable to demonstrate a functional relationship between arch height and calcaneal eversion for running injury risk. However, the authors did find a significant relationship between arch height and transfer coefficient such that an increase in arch height related to an increase in transfer of foot eversion to internal leg rotation. Yet only 27% of the variance in transfer coefficient was explained by arch height, and therefore other factors may be involved in the transfer of calcaneal eversion to leg internal rotation (Nigg et al., 1993). Nawoczenski, Saltzman, and Cook (1998) reported similar results in a study of injured runners divided into high arch and low arch groups. Runners in both groups demonstrated similar magnitudes for calcaneal eversion but the high arch group had greater tibial medial rotation which the authors related to be the possible underlying cause of injury.

Subtalar joint eversion has been shown to be significantly higher in athletes with shin splints when compared to a control group (Viitasalo, & Kvist, 1983). The authors related that the excessive pronation as a result of subtalar joint eversion was most likely due to an increased stretch and eccentric contraction of the anterior muscles of the lower leg. This was in addition to overuse problems of the posterior muscle group of the lower leg. Although the authors did not measure anthropometric variables they felt that the longer pronation time evident in the athletes with shin splints was compensatory to misalignment of the heel-foot or leg-foot alignment. Reber et al. (1993) reported that the tibialis anterior muscle demonstrated the highest rate of continuous contraction of all muscles in the lower leg. The tibialis anterior muscles were contracted for more than 85% of the gait cycle at greater than 20% of maximum. This has been shown to cause muscle fatigue overload by Monod (1985). When muscle fatigue occurs the muscles experience a decreased ability to absorb energy, thereby causing increased energy absorption by the surrounding bones and joints (Subotnick, 1985). Muscle fatigue of the tibialis anterior muscles is the origin of tibial stress fractures and anterior tibial stress syndrome (Reber et al.). Hintermann, and Nigg (1998) state that excessive pronation is related to medial tibial stress syndrome, tibialis posterior tendonitis, Achilles bursitis or tendonitis, patellofemoral disorders, iliotibial band friction syndrome, and lower extremity stress fractures. High repetition of stress even when the stress is moderate may cause stress fractures (Taunton et al., 1988).

Increased knee flexion is associated with increased patellofemoral joint contact forces. Subjects who pronate excessively have demonstrated increased knee flexion concomitant to tibial internal rotation while running on a treadmill (McClay, & Manal, 1997). Stergiou et al. (1999) found that knee angle remains temporally unaffected by increases in impact force even though there was a significant increase in rearfoot angle. Increased velocity was the cause of increased impact force. Therefore, if a runner were to experience excessive pronation before increasing running velocity, it would be expected that there would be an additional increase in pronation which would cause increased knee flexion once velocity was increased. This would also be associated with increased contact forces due to the increased running velocity which would be experienced even greater at the knee because of the increased knee flexion associated with pronation. However, the increase in knee flexion and the increase in pronation are not of equal timing since the knee angle remains temporally unaffected with increased velocity, and therefore there would be a lack of coordination between the subtalar joint pronation and knee flexion that may be an injury mechanism (Stergiou et al.). This may not concern all runners since some are able to run at equivalent speed as others yet apply ground reaction forces that are 30% lower in magnitude (Cavanagh, & Lafortune, 1980). A change in the timing between the subtalar and knee joints also occurs with softer shoe soles. The softer the shoe the greater the amount of calcaneal pronation but this does not affect the amount of flexion or extension of the knee joint (Hamill, Bates, & Holt, 1992). Therefore shoe selection may be a concern for runners especially those who already demonstrate excessive pronation.

Hamill et al. (1987) demonstrated that both the right and left foot exhibit asymmetry and different mechanics while running track curves. The left foot lands in excessive pronation before having to pronate further to foot flat position while the right foot lands in excessive supination compared to running a straightaway. Although their study did not deal with the epidemiology of injuries, they speculated that mechanics of the right foot would result in compressive injuries of the knee and hip from impact shock absorption and the left foot would involve injuries such as plantar fascitis, metatarsalgia, and tibialis tendonitis. Furthermore they demonstrated that the stress involved in running curves increases as curve radius decreases or running velocity increases.

Quadriceps angle.

Quadriceps angle (Q-angle) is defined as the acute angle formed by the intersection of the line connecting the anterior superior iliac spine to the centre of the patella with the line connecting the centre of the patella to the tibial tuberosity (Heiderscheit et al., 1999). It represents the approximate relative angle of pull of the quadriceps femoris muscle on the patella in the frontal plane. In a study on cadavers Schulthies, Francis, Fisher, and Van De Graaff (1995) demonstrated the Q-angle to be a relevant clinical measurement. By modelling the distal vastus muscles according to the direction of muscle fibre pennation and the proximal vastus muscles along with the rectus femoris according to the muscle's respective longitudinal axes, the authors calculated that the actual angle of pull of the quadriceps femoris muscle to be 3.9° lateral to the measured Q-angle. A strong linear relationship existed between the calculated and measured values and the authors reported that 84% of the variation of the quadriceps femoris muscle force on the patella in the frontal plane could be explained by variations in the measured Q-angle. Neely (1998) stated that there is a lack of support determining whether the muscle should be in a contracted or relaxed state, what angle of flexion the knee should be in, and what position the foot should be in while Q-angle is measured. Since the subjects were cadavers in the Schulthies et al. study, the measurements were performed with the subjects in a supine, relaxed musculature position with the lateral malleoli equal distance apart as the greater trochanters which may provide opportunity for measurement differences if compared with studies of subjects in weight bearing position. Measured Q-angle in the Schulthies et al. study was 16.3° with a standard deviation of 3.5°, where all subjects were male. In studies cited by Neely normal Q-angle ranged from 11.2° to 15° for males. The difference appears to be negligible. No significant difference was found when subjects were measured in a standing extended knee position when compared to 24.3° of knee flexion (Caylor, Fites, & Worrell, 1993). Q-angle for the extended position was 12.43 ± 6.05° and 10.80 ± 5.93° for the flexed position.

Messier, Davis, Curl, Lowery, and Pack (1991) reported Q-angle to be approximately 11° for a control group compared to 17° for an injured group, exhibiting a significant difference demonstrating that Q-angle is a strong discriminator between injured and non injured runners. No other anthropometric variables, including leg length discrepancy, arch height, ankle joint flexibility, and knee joint flexibility, were significantly different between the groups. Subject groups were comprised of both male and female runners, and subjects diagnosed with patellofemoral pain were chosen for the injured group. Unfortunately, the authors did not describe subject position for the measurement of Q-angle. Only one study cited by Neely (1998) reported a value lower than 11° to be normal for combined groups of male and female subjects while values from the remainder of the studies ranged from 14° to 20°. This is supported by Caylor et al. (1993) who cited most studies as having 14° to 15° as normal for male Q-angle. Therefore the Q-angle value measured by Messier et al. may be slightly lower than expected and may be due to differences in subject orientation during the measuring process.

Q-angle was demonstrated to be related to risk of overuse injury in male infantry trainees (Cowan, Jones, Frykman, Polly, Harman, Rosenstein, & Rosenstein, 1996). Angles greater than 15° were associated with increased number of stress fractures but not significantly related to risk of muscle strain. Similar results were found in injured basketball players who exhibited right lower extremity Q-angles greater than 10° (Shambaugh, Klein, & Hebert, 1991). Average Q-angle for the injured players was 14.1°. However, basketball player injuries involved the ankle joint, knee joint, shin splints, and muscle strains but did not include any stress fractures.

Large Q-angles may cause excessive pronation (Subotnick, 1975). Increased knee flexion may also occur but instead there may be a disruption in the temporal aspects of lower extremity function. This would be expected to cause a change in the variability of the lower extremity coordination and thereby be a factor involved in injury development (Heiderscheit et al., 1999). Lower extremity coordination is achieved through maintaining relative phase between segments by maintaining a consistent relationship of angular phase between two segments. However, a group of runners with Q-angles greater than 15° did not demonstrate significant differences in lower extremity variability when compared to a group with angles less than 15° while running along a straightaway (Heiderscheit et al.). Static Q-angles were significantly different between the groups yet the kinematic relationship between lower extremity segments remained consistent between the groups during running.

The majority of Q-angle measurements are static measurements that may not relate to the dynamics of movement. Subjects may display an excessive Q-angle in a static position but during the stance phase internal rotation of the tibia will decrease the angle (Tiberio, 1987). During the last 30° of knee extension, the tibia is externally rotating and causes the Q-angle to increase (Caylor et al., 1993). Tubercle-sulcus angle (TSA) has been used instead of Q-angle in assessing patellar position since it is engaged in the trochlear groove during measurement. Measurement involves the subject seated with knee flexed at 90°, then a horizontal line is drawn between the lateral and medial epicondyles along the frontal plane, followed by a vertical line passing through the patella centre perpendicular to the horizontal line, and finally a third line is drawn from the patella centre to the tibial tubercle. The TSA is formed by the intersection of the vertical line with the third line. The TSA was shown to be significantly related to ankle joint injuries but not with knee joint injuries (Caylor et al.). However, the authors state that the TSA is difficult to measure and therefore unreliable. Furthermore, no intra-observer reliability check was performed, nor was there an inter-rater agreement verification. Therefore, the merit of TSA versus Q-angle has not been shown reliable yet.

Continuous relative phase.

Relative phase refers to the organization between two segments and the resultant angular phase relations. It is the observable influence that one segment has on another segment during movement (Kelso, 1995; Wagenaar, & van Emmerik, 1994). Increased stability in the coordination pattern of the two segments is demonstrated by increased symmetry between the segments. Symmetry is defined as the invariance of the phase vector representing the phases of the segments involved during movement (van Emmerik, & Wagenaar, 1996). Stability is measured as the standard deviation of the relative phase (Wagenaar, & van Emmerik). An order parameter represents the cooperation between two segments and may be expressed as phase relations between segments. Order parameters characterize how a pattern is formed. Control parameters, such as velocity or frequency, affect order parameters if coordination stability is insufficient to maintain the movement pattern (Wagenaar and van Emmerik). Stability of coordination is determined by the direction of joint rotation rather than muscle group combinations (Kelso, 1995). There are two types of transitions that a dynamic system may undergo when affected by a control parameter. Abrupt transitions between stable patterns may occur due to small control parameter changes and are called discontinuous transitions. Continuous transitions occur due to large control parameter changes and are smooth transitions from one stable pattern to another (van Emmerik, & Wagenaar).

Kelso (1984) demonstrated the existence of relative phase variability in bimanual coordination. By increasing frequency as subjects flexed one wrist while extending the other a shift in the relative phase from antiphase to in-phase occurred at a critical frequency of movement. The original mode of antiphase was stable at first but as frequency of movement increased the pattern became unstable before being replaced by another stable mode of in-phase. Kelso (1984) speculated that the new mode was more energetically favourable at the given frequency than the previous mode since there was a slight, but non-significant, decrease in cycle energy. This is supported by the author's citation of studies that have shown systematic relationships between energy utilization and mode pattern for horses and gnus wherein they will choose a gait velocity that requires minimum oxygen expenditure.

van Emmerik, and Wagenaar (1996) demonstrated a significant increase in continuous relative phase (CRP) between the rotations of the human pelvis and thorax while increasing and decreasing walking velocity. The pattern between thoracic and pelvic rotations changed from in-phase to more antiphase as walking velocity increased. As well, thoracic trunk rotations increased but total range of rotation decreased. However it was found that stability was least within the intermediate range of walking velocity. Similarly, there was a coordination pattern of arms swinging in synchronization with step frequency that changed to synchronization with stride frequency in humans when walking velocity exceeded 1.0 m/s (Wagenaar, & van Emmerik, 1994). Conversely, patients with Parkinson's disease exhibited difficulties in transition of coordination patterns, shown by decreased values for relative phase variability (Wagenaar, & van Emmerik), which suggests that adaptation during locomotion is related to coordination variability (Heiderscheit et al., 1999).

Examination of coupling parameters in runners with normal and excessive pronation was studied by McClay, and Manal (1997). The results showed that the eversion to tibial internal rotation excursion ratio was decreased for the subjects presenting excessive pronation. Eversion to tibial internal rotation excursion ratio was determined by measuring the amount of calcaneal eversion and the amount of tibial internal rotation. Since the subtalar joint axis is inclined at 45° from the sagittal plane, the amount of calcaneal eversion and tibial internal rotation should be equal, providing an excursion ratio of 1.0. The finding that this ratio was lower in the excessive pronation subjects resulted from eversion being relatively consistent between the two subject groups but tibial internal rotation being significantly greater for the excessive pronation group. Furthermore, times to peak eversion, knee joint flexion, and knee internal rotation were more closely matched in the control group demonstrating greater pattern stability. As well, the relative time difference between peak events was greater in the excessive pronation group, demonstrating further the instability of the movement pattern utilized compared to subjects who exhibit normal pronation. However, utilizing peak times and values offers only discrete measures which do not represent relative segment kinematic coordination (Heiderscheit et al., 1999).

Heiderscheit et al. (1999) investigated the influence of Q-angle on the variability of running kinematics on a straightaway. The authors measured the magnitude of thigh abduction/adduction, foot eversion/inversion, and thigh flexion/extension, coupling each with the magnitude of leg rotation. Subjects were separated into high and low Q-angle groups with 15° being the discerning value. No significant difference between groups was found for CRP for any segment couplings. The authors concluded that Q-angle does not affect the variability of coordination of segments of the lower extremity despite exhibiting a significant difference between the two groups for static Q-angle measurement. Therefore it would seem that there would not appear to be a link between static Q-angle and risk of running related injuries. However, the high Q-angle group consistently demonstrated greater CRP variability for all segment couplings for all stance intervals. Possible lack of significance may be due to a lack of manipulation of a control parameter. Different scaling processes of the control parameter can lead to different coordination dynamics. This may involve the rate of change of velocity or the duration over which the control parameter is maintained (Haken, 1977, cited by van Emmerik, & Wagenaar, 1996), or perhaps even the curvature of the path followed by the runners. Since McClay, and Manal (1997) reported that the calcaneal eversion was equal between injured and non-injured groups but there was increased tibial internal rotation for the injured group, one would expect that Heiderscheit et al. would have found a difference in the CRP variability of leg internal/external rotation coupled with calcaneal eversion/inversion between groups if Q-angle were a descriptor for injury risk in runners. However, the effects of Q-angle on running mechanics may only manifest during running that exerts changes in the demands on the lower extremity mechanics. It is conceivable that had Heiderscheit et al. manipulated the curvature of the path followed by the runners, the velocity of the runners, or the stride rate of the runners, they would have found CRP variability differences between their groups.

Summary.

Misalignment of the lower extremity appears to have important consequences for running related injuries. This misalignment may be responsible for changes in the relative timing of segment movements. Curve running has been shown to require changes in the lower extremity alignment including excessive pronation of the left foot and excessive supination of the right foot. Runners with increased Q-angle may be at greater risk of injury due to the apparent misalignment that a high Q-angle represents. Therefore, increased Q-angle and curve running may have a synergistic effect with implications for injury. Risk of injury can be characterized by continuous relative phase.

CRP has previously been used to demonstrate relative timing of segments during movement.

CHAPTER III

METHODOLOGY

The purpose of this study was to examine the effects of Q-angle on running curve kinematics and kinetics. Risk of injury was determined by variability in the continuous relative phase of segment couplings.

Subjects.

Ten male subjects were selected, five for each grouping. Subjects were volunteers from the local running community and were comprised of competitive endurance runners. Informed consent forms, approved by the University of Windsor ethics committee, were signed prior to data collection by each subject in accordance with university policy. Subjects were permitted to withdraw from the experiment at any time. Since running in new shoes may accentuate subtle pronation asymmetries each subject wore their own training shoes (Vagenas, & Hoshizaki, 1992). It has previously been observed by Messier et al. (1991) that subjects demonstrated little observable difference between training groups for shoe preference. Subjects who experienced an injury within the 6 months prior to data collection were excluded from the study. The time frame of six months allowed inclusion of runners who may have experienced earlier injuries but were healthy at the time of the study. This allowed individuals with potential lower extremity pathologies to be included to identify possible alterations in segment coupling coordination (Heiderscheit et al., 1999). As well, subjects presenting a leg length discrepancy (LLD) of 1.3 cm or greater were excluded since a greater difference has been hypothesized to affect maximum eversion angle (Hamill et al., 1992). Subjects were separated into two groups based on Q-angle measurement. Based upon the results of Messier et al.(1991), and Moss, DeVita, and Dawson (1992) Q-angles of greater than 16° were included in the high Q-angle group (HQ), and angles of 14° or less were included in the control group of low Q-angle (LQ).

Subject anthropometric data are shown in Table 1. There was a significant difference between the groups for left lower extremity Q-angle ($F = 7.40$, $p < 0.05$) where the HQ and LQ groups demonstrated means of 18.2° and 10.6°, respectively. As well, there was a significant difference between groups for stature ($F = 3.440$, $p < 0.05$) with the HQ group having a greater mean stature of 185.4 cm than the LQ group with 174.8 cm. No other significant differences were present.

Table 1
Subject anthropometric data group means

Group	Age (yrs)	Mass (kg)	Stature (cm)	LLD (cm)	Left Q-angle (°)	Race pace (m/s)	University training experience (yrs)
HQ	24.8 (5.8)	71.7 (6.7)	185.4 (1.8)**	0.88 (0.38)	18.2 (2.9)*	5.0 (0.4)	3.4 (0.9)
LQ	21.6 (21.6)	64.6 (6.5)	174.8 (7.7)**	0.74 (0.46)	10.6 (1.5)*	4.8 (0.3)	3.0 (1.0)

SD in parentheses.
* ($F = 7.40$, $p < 0.05$)
** ($F = 3.440$, $p < 0.05$)

Independent variables.

There were two independent variables. The first independent variable was the radius of the curve which had three conditions: 36.5 m, 18.5 m, and straightaway. The straightaway was the control and also allowed for validity comparison with similar studies. The second independent variable was Q-angle which had two groups: high Q-angle and low Q-angle.

Dependent variables.

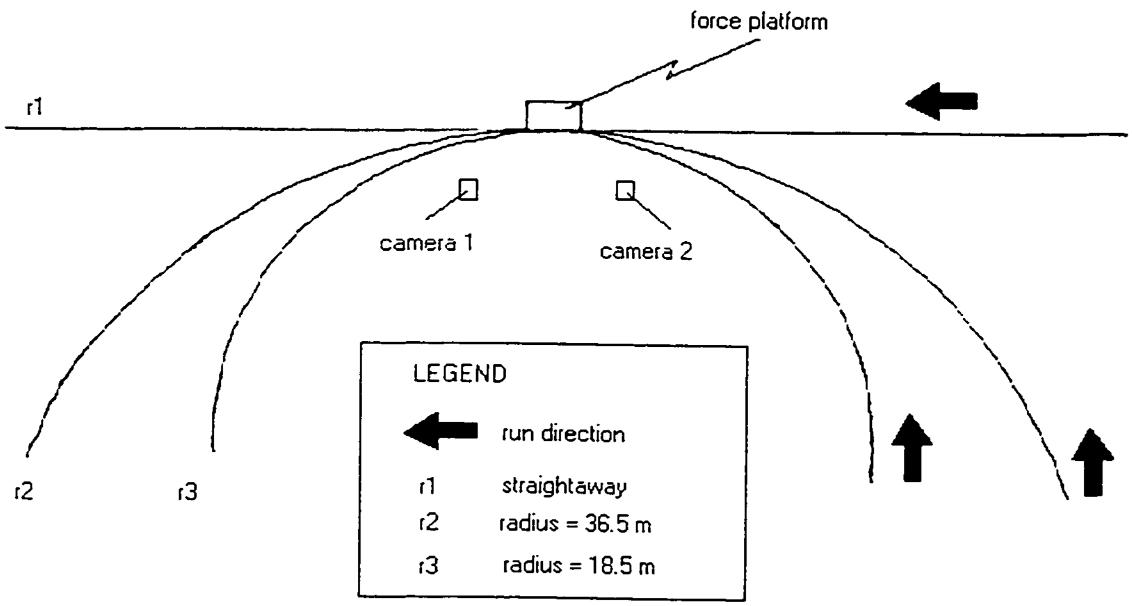
There were five dependent variables. From the kinetic data of vertical ground reaction force, dependent measures included peak force, the relative time occurrence of peak force, average force, and total impulse. From the kinematic data there was the dependent measure of the variability of continuous relative phase for the coupling: knee flexion/extension with ankle dorsi flexion/plantar flexion.

Experimental set up.

The experimental set up is shown in Figure 1. Kinematic data were collected at a rate of 60 Hz using two video camera recorders. The room coordinate system was defined using a cube of known dimensions containing 12 markers of known coordinates.

Kinetic data was collected using a force platform (Advanced Mechanical Technologies, Inc., Newton, MA) embedded in and flush with the running surface, centred at the apex of each running curve. The force platform was connected to a computer for the data collection. Stance time was defined as the period of time during which the vertical GRF exceeded 16 N (Munro et al., 1987). Running curves were measured and clearly marked on the running surface with radii 36.5 m and 18.5 m to simulate outdoor and indoor track radii, respectively (W. T. Armbrust, May 5, 2000). The force plate was tested statically and dynamically prior to each testing session for reliability.

Figure 1. Experimental set up of testing area.

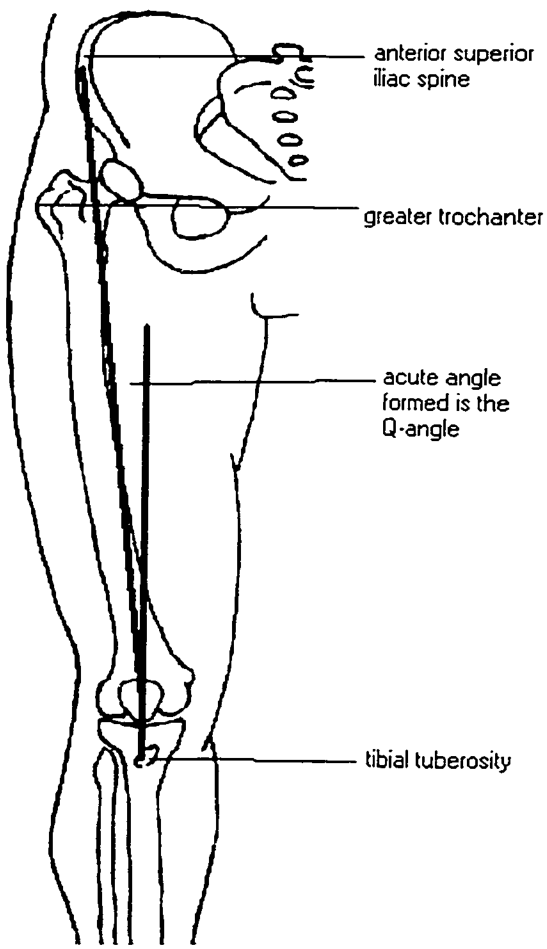


Procedure.

Since excessive pronation has received the greatest amount of attention in studies on running mechanics, the left lower extremity was studied because it has been shown to experience excessive pronation while running a track curve (Greene, 1987). Figure 2 shows the Q-angle measurement. The researcher measured Q-angle since Caylor et al. (1993) demonstrated that inexperienced testers are able to accurately reproduce Q-angle measurements following training. Measurement was recorded for the left lower extremity in a relaxed weight-bearing position with knee joint extended using a goniometer. A line was drawn from the palpated left anterior superior iliac spine (ASIS) to the patella centre and another line from the patella centre to the palpated tibial tuberosity. These lines were used to line up the arms of the goniometer using the patella centre for the point of origin and the acute angle formed served as the Q-angle measurement. Leg length discrepancy was measured using a standard tape measure with the subject supine on a table. Measurement was between the greater trochanter and respective lateral malleolus of each leg.

Subjects were provided sufficient time to warm up and stretch prior to data collection and self reported readiness for data collection. Subjects ran three trials for each of the three conditions at their own respective training velocity. Stergiou et al. (1999) theorized that the self selection of a velocity by the subject may be based on both the energy requirements and a functional requirement for maximizing coordination between joints to avoid injury. Therefore, an unaccustomed pace was not imposed on the subjects and their own training pace was used to simulate training conditions that are experienced by the athlete regularly. Subjects were also requested to abstain from running on the day of testing prior to the test period since fatigued muscles have been shown to cause a change in running mechanics (Reber et al., 1993). Due to equipment difficulties the day of testing, velocity was not monitored online at the time. However, since the athletes were all competitive at the university level, with some of them being National champions, it was decided that they would be able to run their respective training paces without great variability. Subject velocities were calculated from the videotape using the kinematic analysis software. All subjects ran each trial of each condition within ± 5% for each respective condition and each reported that the pace was their own respective training pace effort on the day of testing.

Figure 2. Measurement of Q-angle. Note: diagram shows right leg whereas left leg was measured and studied.



Successful run trials were defined as being when the entire left foot contacted the force platform without any visible alteration in running mechanics. Trials were repeated if necessary until successful to ensure reliability. Instructions were provided to the subjects to not target the force platform with the left foot since this may alter mechanics of the lower extremity. If it appeared that a runner had altered running stride to target the force platform the trial was repeated. Since curve running has been found to alter running pace (Greene, 1985) it was not expected that the runners would run all conditions at the same velocity because this may impose an unaccustomed pace on the runner and thereby cause an alteration in lower extremity running mechanics. Subjects ran counter clockwise on the curves to simulate training and racing conditions.

Data reduction.

Video images were loaded into a computer using Matrox Marvel G200 software (Matrox Graphics Inc., Dorval, PQ). Ariel Performance Analysis System (Ariel Dynamics, Inc., Trabuco Canyon, CA) was used to digitize, filter, and transform the film data. Digitization began four frames prior to stance phase and ended four frames following stance phase to minimize filtering effects (Heiderscheit et al., 1999). Filtering of data involved use of a cubic spline curve fit. Direct Linear Transformation was utilized to combine respective images from both video camera recorders into three-dimensional images calibrated using the calibration cube.

Phase plots were used to determine lower extremity segment coordination variability. Analysis involved comparison of knee flexion/extension with ankle dorsi flexion/plantar flexion. Phase plots included normalized segment angles plotted against normalized angular velocity. Segment angles and angular velocity were extracted from the Ariel Performance Analysis System. Normalization of segment angles was achieved by setting the maximum angle to be 1, the minimum angle to be -1, and the midpoint angle to be 0. As well, angular velocity was normalized to the respective absolute maximal value. This allowed calculation of phase angles without differing amplitudes confounding the results (Heiderscheit et al., 1999). Phase angles were calculated using phase plots for each subject for each trial before calculation of means. The phase angle was defined as the angle formed by the right horizontal and a line drawn from the origin to the specific data point within the range 0° to 180°. CRP was defined as the difference between the phase angles of ankle and knee joints during the entire stance phase. The difference was calculated by subtracting the phase angle of the distal joint from the proximal joint in the coupling during the entire stance phase. Using linear interpolation, CRP was rescaled to a percent of stance phase for each trial. The standard deviation of the trials for each condition for each subject was then calculated. The variability of CRP (vCRP) was calculated as the mean standard deviation for each condition for each subject. The mean group vCRP was then calculated for each condition for each group.

Furthermore, vertical GRF was measured and compared between the groups and across conditions to examine whether runners in HQ and LQ groups were experiencing different relative magnitudes of impact force. This allowed the analysis of GRF as a cause of differing lower extremity mechanics should any be detected. Vertical GRF was normalized by each individual body weight and by percent stance time prior to group means being calculated. Other aspects of GRF while running track curves have been studied and reported elsewhere (Harvie, 1998) and are therefore beyond the scope of this study.

Statistical procedures.

The data for each dependent variable was evaluated for statistically significant differences using a two factor mixed analysis of variance (ANOVA) with repeated measures on the track curve variable. This model tested for both significant difference and interactions and resulted in the following F-ratios:

I. $\mathbf{F}_{radius}$

II. $\mathbf{F}_{Q\text{-}angle}$

III. $\mathbf{F}_{radius \times Q\text{-}angle}$

Each F-ratio was evaluated at a significance level of $p < 0.05$. To determine specific differences evidenced between curve conditions, data were analyzed using paired t-tests. In the event of statistical analyses requiring paired testing, the Bonferroni correction was employed to prevent increased Type I errors.

Null hypotheses.

For each dependent variable the null hypothesis comparing Q-angle groups was

$$H_0 : X_1 = X_2$$

For each dependent variable the null hypothesis comparing track curve was

$$H_0 : Y_1 = Y_2 = Y_3$$

Reliability.

One subject was randomly chosen to repeat the experimental procedures on two separate days for test-retest reliability. Paired t-test analyses were performed on the data. There was no significant difference between trial days for velocity of the reliability subject for the straightaway or the 18.5 m curve condition. However, there was a significant difference for velocity between the different days for the 36.5 m curve condition. There were no significant differences evident for the reliability subject between test days for CRP or vCRP. Therefore, it appears that the velocity differences did not affect the CRP or vCRP values. The experimental protocol is considered reliable based on the findings for both CRP and vCRP.

Validity.

Due to the limited number of studies on running flat track curves, comparison of the measurements of kinetic and kinematic data for the 18.5 m and 36.5 m curves was limited to Greene (1985), Hamill et al. (1987), and Harvie (1998). Similarly, the data for CRP while running the straightaway was limited to Heiderscheit et al. (1999) since there does not appear to be any other studies involving CRP and the lower extremity during running. However, the CRP data for the 18.5 m and 36.5 m curves was not validated by comparison to other studies since there does not appear to be any other studies involving CRP and the lower extremity during curve running. Considering the extensive research that has involved GRF during running, the kinetic data for running the straightaway was validated by comparison with other studies, particularly Cavanagh, and Lafortune (1980), Harvie (1998), and Munro et al. (1987). Furthermore, validity of the measured Q-angles was conducted through comparison with Caylor et al. (1993), Heiderscheit et al. (1999), and Horton, and Hall (1989).

CHAPTER IV

RESULTS AND DISCUSSION

The purpose of this study was to examine the effects of Q-angle on running curve kinematics and kinetics. Risk of injury was determined by variability in the continuous relative phase of segment couplings.

Velocity.

Subject velocity values are shown in Tables 2-4 and are represented in Figures 3 and 4. There was no significant difference between group velocities for any curve condition ($F = 0.019$, $p = 0.894$). As well, there were no significant interaction effects for group and curve condition ($F = 0.218$, $p = 0.810$). However there was a significant difference between the 36.5 m curve and 18.5 m curve conditions overall with the 36.5 m curve being run at a faster velocity ($F = 7.041$, $p < 0.05$). Furthermore, there was a trend towards significance between the straightaway and 18.5 m curve conditions overall with the straightaway being run at a faster velocity ($F = 7.041$, $p = 0.032$).

Subjects were instructed to run trials at their own training pace. This was chosen to optimize coordination between joints rather than have a change in coordination due to unaccustomed velocity (Stergiou et al., 1999). As well, this allowed for closer simulation to actual conditions since training pace is most often the velocity used by the athletes while running. It was expected that both groups run all curve conditions at approximately the same velocity since there was not a significant difference ($F = 0.101$, $p = 0.492$) present between group race paces (Table 1). Therefore, it was not expected that there would be a significant difference between curves for running velocity. Harvie (1998) did not find any significant differences between track curve velocities for the left foot of 9 male runners. She did discern a trend approaching significance in that the males ran the 30 m and 18 m curves faster than the straightaway. However the results reported here show the 18.5 m curve to be significantly slower, with the wider curve approximately the same as the straightaway for running velocity. Possibly, the running velocity chosen by the subjects may be a function of mechanical and/or physiological economy such that the body chooses a running pace that is least likely to interrupt the coordination pattern (Kelso, 1984). This may also happen to decrease the probability of an injury occurring.

Stergiou et al. (1999) reported an increase in vertical impact force with increased running velocity. However, that was not supported by this study since there were no significant differences between curve conditions for peak or average vertical force (tables 5 and 7). It is possible that the results are limited by the small subject numbers and that there would have been significant differences in the average and peak vertical forces between conditions had there been larger subject groups and that the differences would have followed those of the velocity data, however it is not possible to ascertain that with the current results.

Table 2
Mean velocity of the centre of gravity for subject groups.

Group	Velocity (m/s)
HQ	3.3
LQ	3.3

Table 3
Mean velocity of the centre of gravity for curve conditions.

Curve	Velocity (m/s)	SD
Straightaway	3.4	0.469
36.5 m	3.3*	0.444
18.5 m	3.2*	0.380

* (F = 7.041, $p < 0.05$)

Table 4
Mean velocity of the centre of gravity for each group on each curve condition.

Group – Curve condition	Velocity (m/s)	SD
HQ		
Straightaway	3.4	0.610
36.5 m	3.4	0.625
18.5 m	3.3	0.515
LQ		
Straightaway	3.3	0.343
36.5 m	3.3	0.230
18.5 m	3.2	0.243

Figure 3. Mean velocity of the centre of gravity for curve conditions.

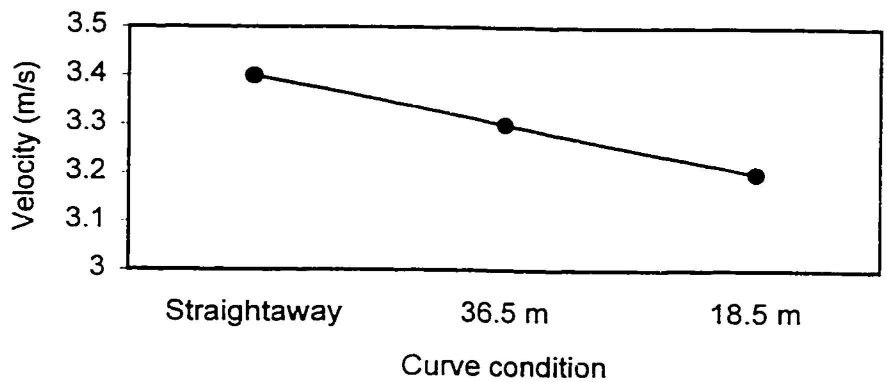
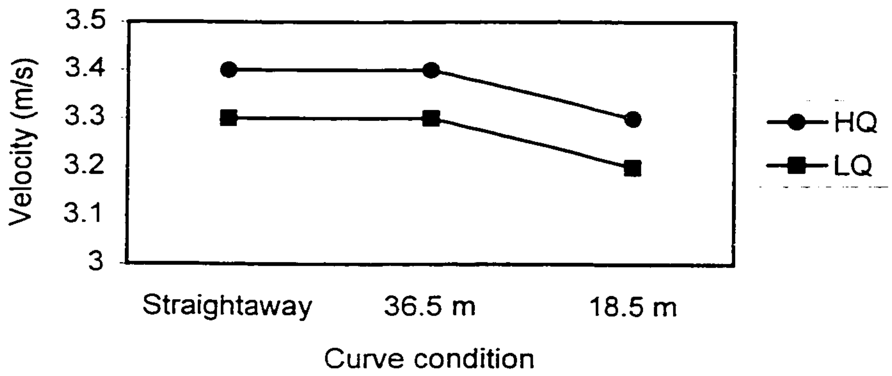


Figure 4. Mean velocity of the centre of gravity for each group on each curve condition.



Peak vertical force (Fz).

Tables 5-7 show the mean peak Fz for subject groups and curve conditions. The relationships for peak Fz are shown in Figures 5 and 6. There were no significant differences between groups for peak Fz ($F = 0.438$, $p = 0.527$). As well, there were no significant differences between curve conditions ($F = 1.020$, $p = 0.408$) or any significant effects for the interaction between group and curve condition ($F = 1.099$, $p = 0.385$). Lack of interaction effects may reflect large variability in the raw data.

Lack of differences between groups is not unexpected. Hamill et al. (1987) did not find any significant differences between running a straightaway and running a curve with radius 31.5 m. Peak vertical force values reported were 30.4 N/kg of body mass for the straightaway, and 29.8 N/kg of body mass for the left foot on the curve while running counter clockwise. These values compare closely with the values in the present study that would convert to 30.5, 28.9, and 29.8 N/kg of body weight for each of the straightaway, 36.5 m, and 18.5 m curves, respectively. Values reported by Cavanagh and Lafortune (1980) ranged from 270 to 280 % of body weight while running a straightaway. Although the values in the present study are slightly higher than those reported by Cavanagh and Lafortune, they do appear to be similar enough to be valid. Furthermore, Harvie (1998) did not find any significant differences between three different curve conditions for peak Fz. However, Harvie did not report the values recorded in the study for peak Fz.

Conversely, De Wit et al. (2000) reported peak Fz to be 190 % of body weight for nine male runners while running 3.5 m/s, and further reported this value to increase to 230 and 280 % at velocities of 4.5 and 5.5 m/s, respectively. Comparatively, the values in the present study are closer to those reported for 5.5 m/s although the present subjects did not average greater than 3.4 m/s. It can not be determined why such a difference between reported values is present, they may be due to methodological differences. It is possible that since the subjects in the present study freely chose to run at an average velocity of 3.4 m/s their bodies subconsciously optimized velocity and vertical force to produce the most efficient movement for them. In the study conducted by De Wit et al. subjects had the velocities imposed on them that may not have allowed them to optimize efficiency of movement, hence generating values differing from the present study.

While studying male running velocities ranging from 3.0 to 5.0 m/s, Munro et al. (1987) demonstrated peak Fz to increase from 251 to 283 %, at 3.0 and 5.0 m/s, respectively. Specifically, they reported peak Fz to be 262 % at 3.5 m/s. Although this value may be similar to values reported in the present study, differences may again be attributed to the imposed run velocity. As well, data were collected for both the right and left foot by Munro et al. and was grouped together whereas the present study only collected data for the left foot. Within individual asymmetries have been reported within individuals (Cavanagh et al., 1985) and may be a factor in the differences between reported peak Fz values and present study values.

Table 5
Mean normalized peak Fz values for subject groups

Group	Mean peak Fz (%)
HQ	285.7
LQ	297.5

Table 6
Mean normalized peak Fz values for curve conditions

Curve	Mean peak Fz (%)	SD
straightaway	299.0	36.7
36.5 m	283.8	28.1
18.5 m	291.9	39.7

Table 7
Mean normalized peak Fz values for each subject group for each curve condition

Group – Curve condition	Mean peak Fz (%)	SD
HQ		
Straightaway	291.5	42.1
36.5 m	289.0	27.3
18.5 m	276.5	37.8
LQ		
Straightaway	306.6	33.3
36.5 m	278.5	31.0
18.5 m	307.3	39.1

Figure 5. Mean normalized peak Fz for curve conditions.

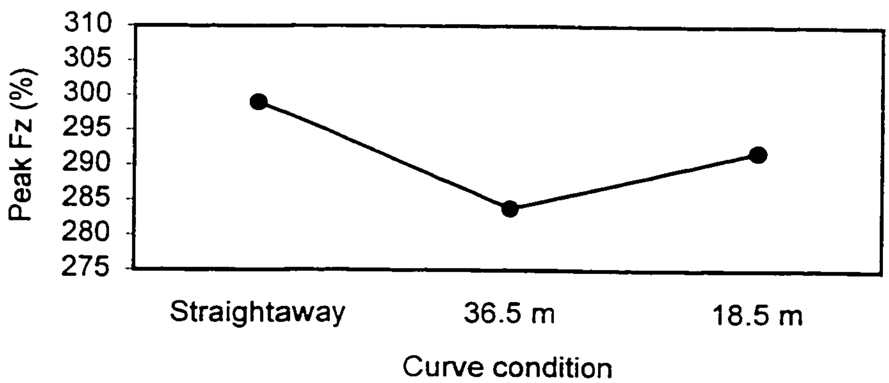
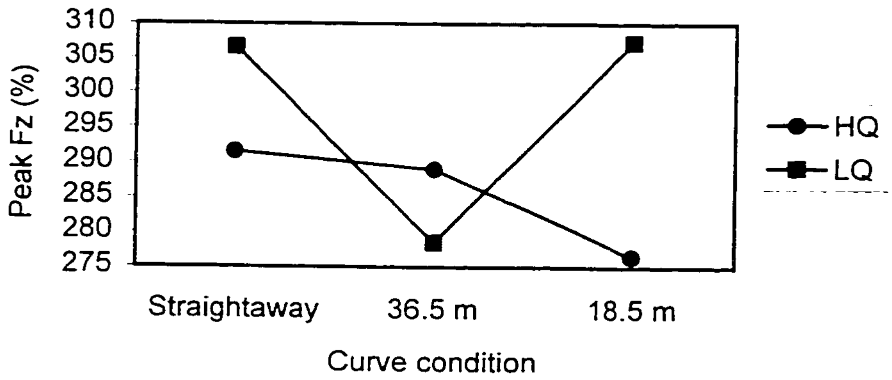


Figure 6. Mean normalized peak Fz for each subject group for each curve condition.



Time of peak vertical force (t @ peak Fz).

Tables 8-10 show the mean time of peak Fz for subject groups and curve conditions. Time of peak force is illustrated in Figures 7 and 8. Although results demonstrated a trend towards significance, no significant difference between groups for time of peak Fz was apparent (F = 4.506, p = 0.67). As well, no significant difference between curve conditions (F = 1.442, p = 0.299) or effects of the interaction of curve and group for time of peak Fz were present (F = 2.721, p = 0.134). Large variability in the raw data may be reflected by the lack of interaction effects.

De Wit et al. (2000) reported t @ peak Fz to occur significantly earlier in stance phase as velocity increased from 3.5 to 5.5 m/s. Normalizing by their reported values for mean contact time produces values of 15.1, 16.5, and 17.1 % for 3.5, 4.5, and 5.5 m/s running velocities, respectively. Differences between values calculated in the current study and those of De Wit et al. may be due to the constraints of imposing velocities on the runners, rather than the self selection of pace as experienced by runners in the current study. Indeed, although the values for the HQ group appear to be different from those reported by Hamill et al. (1987), the LQ group does not seem to differ greatly. Hamill et al. recorded t @ peak Fz values to be 41.9, and 42.6 % when running a straightaway, and 31.5 m curve, respectively. For the straightaway condition in the current study, the LQ group recorded a mean t @ peak Fz value of 40.1, despite running at a velocity of 3.3 m/s, which is strikingly different from the 6.3 m/s utilized in the Hamill et al. study. The HQ group recording a value of 30.9 for t @ peak Fz compared to the runners in the Hamill et al. study supports the apparent trend towards significance between groups that was evident in the current study.

Since De Wit et al. (2000) demonstrated that t @ peak Fz occurs later in stance phase as velocity increases, it is not surprising that t @ peak Fz values in the Harvie (1998) study increased as curve radius decreased considering the runners in that study increased velocity as curve radius decreased. Harvie reported the t @ peak Fz values to be 41.6, 41.8, and 43.1 % for the straightaway, 30 m, and 18 m curves, respectively. In the current study velocity decreased as curve radius increased and therefore it was expected that t @ peak Fz values would decrease as curve radius increased. Although this did occur for the LQ group, a reverse trend for the HQ group was evident which influenced the overall t @ peak Fz values to demonstrate a reverse trend as well. The LQ group recorded a value of 40.1 % for the straightaway condition, comparable to the 41.6 % reported by Harvie for the same condition. As well, the value of 39.1 % for the 36.5 m condition does not appear to be greatly different than the 41.8 % recorded for runners running the 30 m curve in the study by Harvie. Conversely, the values for the HQ group for both the straightaway and the 36.5 m conditions appear to be greatly different from those reported by Harvie and further supports the apparent trend towards significant difference evident between the HQ and LQ groups. It is important to note this trend towards significance since specific tissues are designed to attenuate GRF and are adapted to absorb those forces while in specific positions. If peak Fz occurs at the wrong time, those tissues may not fully absorb the force but instead transfer it to inappropriate tissues that are not designed for it or attenuate it itself while in the wrong position, thereby resulting in injury to either one of the tissues.

Table 8
Mean normalized time of peak Fz for subject groups

Group	Mean t @ peak Fz (%)
HQ	30.9
LQ	38.8

Table 9
Mean normalized time of peak Fz for each curve condition

Curve	Mean t @ peak Fz (%)	SD
straightaway	31.7	13.4
36.5 m	33.5	11.6
18.5 m	39.5	6.77

Table 10
Mean normalized time of peak Fz for each subject group for each curve condition

Group – Curve condition	Mean t @ peak Fz (%)	SD
HQ		
straightaway	23.2	14.1
36.5 m	27.9	11.4
18.5 m	41.8	3.8
LQ		
straightaway	40.1	5.2
36.5 m	39.1	9.8
18.5 m	37.2	8.7

Figure 7. Mean t @ Fz for curve conditions.

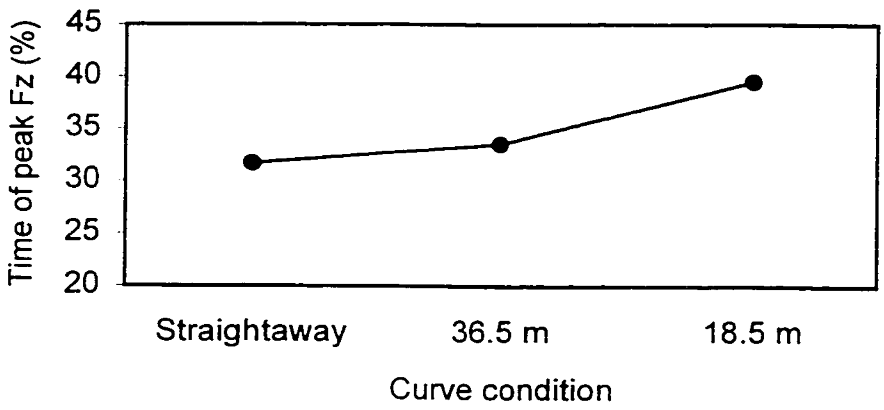
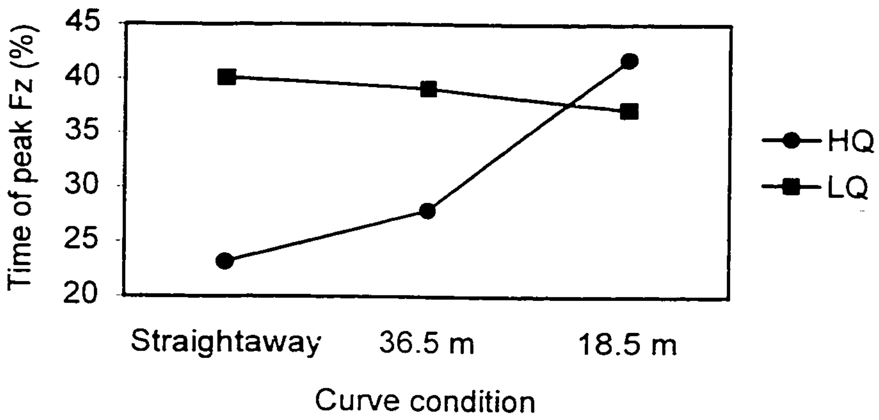


Figure 8. Mean t @ Fz for each subject group for each condition.



Average vertical force (avg Fz).

Figures 9 and 10 show the relationships for average Fz. Tables 11-13 show the mean normalized average Fz for subject groups and curve conditions. There was no significant difference between groups for average Fz ($F = 1.079$, $p = 0.329$). As well, there were no significant differences between curve conditions ($F = 2.945$, $p = 0.118$) or significant effects for the interaction of curve and group ($F = 0.456$, $p = 0.652$).

The average vertical force is a measure of the force applied throughout the stance phase. Munro et al. (1987) relate it to have less intra-individual variability than the other vertical GRF variables and any differences reflect change in running pattern. While studying twenty males running at velocities ranging from 3 to 5 m/s, Munro et al. demonstrated a significant increase in avg Fz. While running at 3.25 m/s, the subjects in that study averaged 144 % of body weight, whereas the subjects in this study averaged 180 % while running the straightaway. The slight difference between studies may be due to the subjects in the present study being well-trained athletes competing at the national level and the subjects studied by Munro et al. being joggers. With increased training, it may be that the athletes have adapted to applying greater force for the similar velocity. However, it is not possible to determine specifically why there is a difference in avg Fz between the two studies.

Harvie (1998) reported avg Fz values of 162 % body weight for males running a straightaway. Furthermore, Harvie found the average Fz to be less for curve conditions than the straightaway. Average Fz for the 30, and 18 m curves were 157 and 158 %, respectively, whereas the values in the present study were 171 and 174 %, respectively. Although the present study demonstrated the same trend of decreased avg Fz for curves versus the straightaway, the values are lower than those reported by Harvie and did not present as significantly different. The values do not appear to be that dissimilar between the two studies and seem to be comparable between the studies. However, the subjects in this study ran the curves at a slower velocity than the straightaway, whereas the subjects in the study conducted by Harvie ran the curves faster than the straightaway.
Considering that velocity is a component of force, one would expect greater force

generation with increased velocity. It was therefore expected that the subjects in the present study would run with a lower normalized average vertical force. Since there was a significant change in velocity but not in avg Fz between curve conditions and the straightaway this may exemplify the change in running pattern that Munro et al. (1987) theorize average Fz to represent. Hamill et al. (1987) reported average force values to be 163 and 157 % for straightaway and 31.5 m curve running, despite running at a velocity of 6.3 m/s. Hamill et al. did not find any significant differences for average force between the two conditions, similar to the current study, whereas Harvie did find a significant difference between running the straightaway and both track curves. Considering the implications that imposed running velocities may have on running mechanics, the results for average Fz in the present study appear to be valid based on the similarity of the trends with Hamill et al., despite the difference in the magnitude of the forces.

Table 11
Mean normalized average Fz for subject groups

Group	Mean average Fz (%)
HQ	169.7
LQ	180.8

Table 12
Mean normalized average Fz for curve conditions

Curve	Mean average Fz (%)	SD
straightaway	180.6	24.0
36.5 m	171.2	18.1
18.5 m	173.9	20.8

Table 13
Mean normalized average Fz for each subject group for each curve condition

Group – Curve condition	Mean average Fz (%)	SD
HQ		
straightaway	173.7	22.7
36.5 m	169.4	18.9
18.5 m	165.9	17.0
LQ		
straightaway	187.5	25.7
36.5 m	172.9	19.4
18.5 m	182.0	23.0

Figure 9. Mean normalized average Fz for curve conditions.

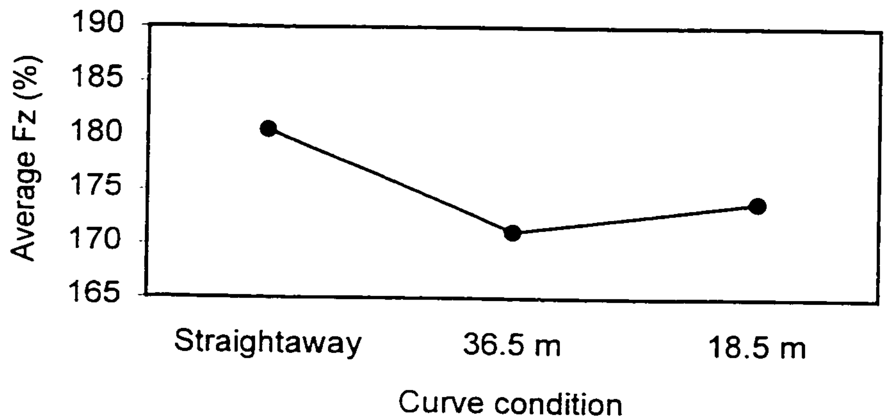
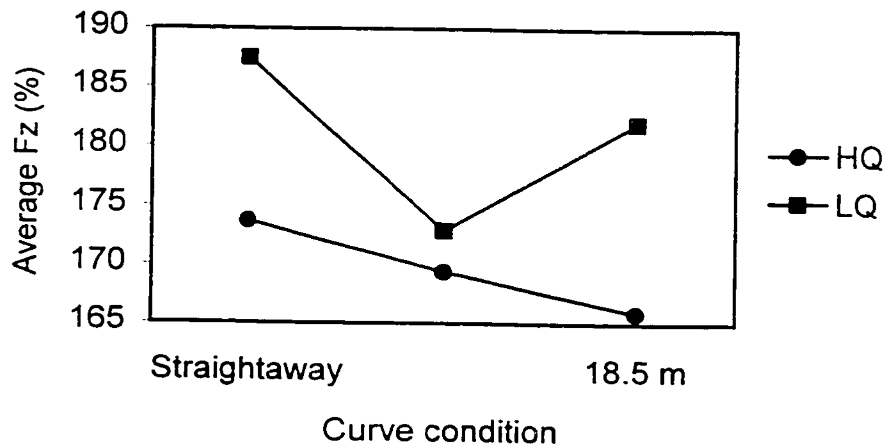


Figure 10. Mean normalized average Fz for each subject group for each curve condition.



Impulse.

Figures 11 and 12 demonstrate the impulse relationships. Tables 14-16 show the mean normalized impulse for subject groups and curve conditions. There were no significant differences between groups ($F = 1.251$, $p = 0.296$) or curve conditions for impulse ($F = 1.445$, $p = 0.298$). Furthermore, there was no significant interaction effects of group and curve condition ($F = 0.169$, $p = 0.848$).

Harvie (1998) also reported a lack significant difference when comparing the impulse of males running three different levels of curvature. Unfortunately the values reported by Harvie range between 260 and 300 Ns/N/s. Although they are reported as normalized, they appear to be raw total impulse and therefore can not be compared to the values in the current study. Hamill et al. (1987) did not find any significant differences when comparing running a straightaway with a 31.5 m curve. However, the values reported by Hamill et al. are over 50 % greater than the values in this study. For running a straightaway, and 31.5 m curve the impulses reported were 3.112 and 3.067 Ns/N/s, respectively. Since the runners in this study averaged 3.3 m/s, the discerning difference between the studies may be that the athletes in the study conducted by Hamill et al. ran at a velocity of 6.3 m/s, nearly twice that of the athletes in the current study. Since impulse is the change in momentum divided by the time it took to change, it would be expected that the athletes travelling a faster velocity would have a greater impulse because stance time, or the time it takes to change momentum, decreases with increasing velocity (Munro, Miller, & Fuglevand, 1987). Therefore, the values in the current study for impulse appear to valid when compared to those reported by Hamill et al.

Table 14
Mean normalized impulse for subject groups

Group	Mean impulse (Ns/N/s)
HQ	1.755
LQ	1.875

Table 15
Mean normalized impulse for curve conditions

Curve	Mean impulse (Ns/N/s)	SD
straightaway	1.846	0.266
36.5 m	1.808	0.228
18.5 m	1.792	0.207

Table 16
Mean normalized impulse for each subject group for each curve condition

Group – Curve condition	Mean impulse (Ns/N/s)	SD
HQ		
straightaway	1.792	0.225
36.5 m	1.750	0.187
18.5 m	1.722	0.173
LQ		
straightaway	1.899	0.318
36.5 m	1.865	0.271
18.5 m	1.862	0.232

Figure 11. Mean normalized impulse for curve conditions.

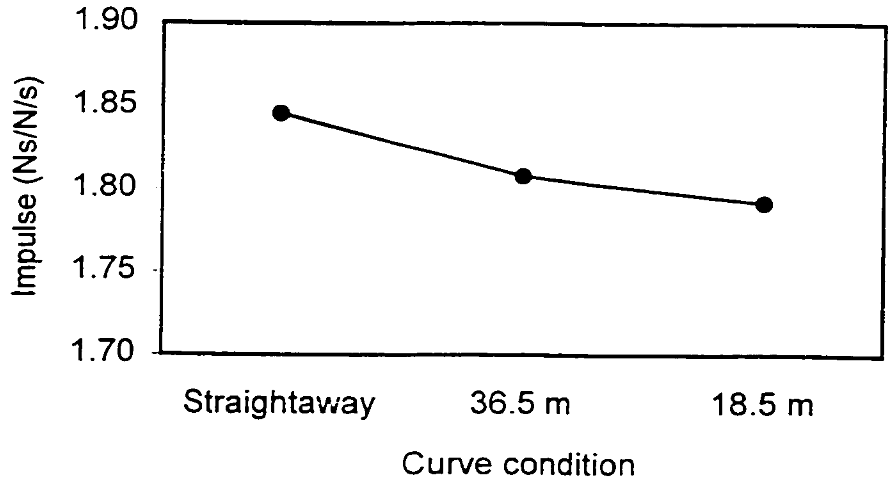
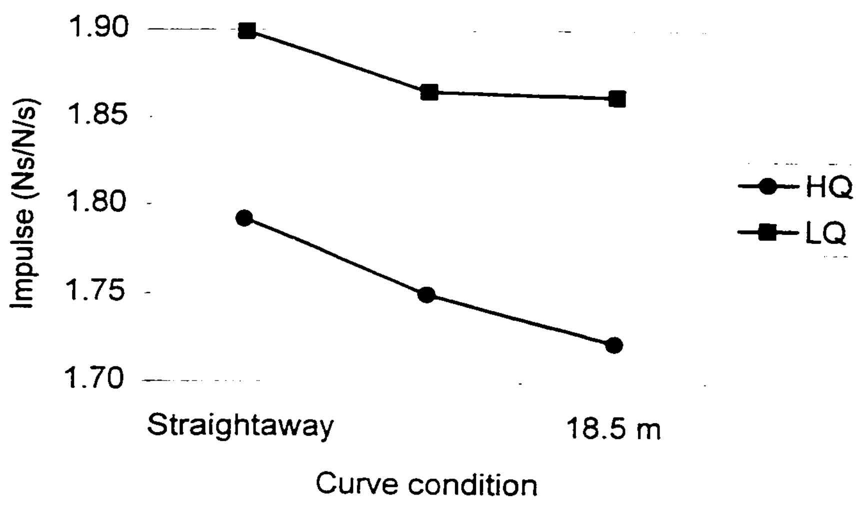


Figure 12. Mean normalized impulse for each subject group for each curve condition.



Continuous relative phase (CRP).

Tables 17, 18, and 19 show the mean values for CRP for group, curve condition, and interaction of curve and group, respectively. The relationships for CRP are illustrated in Figures 13 and 14. There was no significant difference for CRP between groups ($F = 0.000$, $p = 0.996$) or the interaction effects of groups and curve conditions ($F = 0.368$, $p = 0.705$). There was a significant difference in curve condition wherein the straightaway demonstrated a lower value for CRP than the 36.5 m curve ($F = 9.753$, $p < 0.05$). There was no significant difference between any other pairing of curve conditions. Perfect in-phase relation between segments is signified by a CRP value of 0° while perfect anti-phase is characterized by a CRP value of 180° (Hamill, Haddad, & McDermott, 2000). Therefore, it appears that the knee and ankle are more in-phase with joint flexion/extension while running a straightaway versus a wide curve but not so with a tight curve. There is not even a trend apparent in the statistical results to suggest that there may be a significant difference with larger subject groups so it is unlikely that in-phase and anti-phase characteristics are necessarily a function of track curvature in this study. Rather, it is possible that the wide curve existed closer to the transition point between coordination movement patterns than either the straightaway or tight curve and that the straightaway and tight curve, although both somewhat in-phase, actually utilize different movement patterns.

Lack of significant difference between the two groups would suggest that Q-angle does not influence the timing characteristics of ankle and knee extension/flexion. However, actual characterization of coordination of segments is the variability of CRP.

Table 17
Mean CRP values for group

Group	Mean CRP (°)
HQ	87.6
LQ	87.6

Table 18
Mean CRP values for curve condition

Curve	Mean CRP (°)	SD
Straightaway	83.3*	9.3
36.5 m	91.6*	11.1
18.5 m	87.8	14.2

* (F = 9.753, $p < 0.05$)

Table 19
Mean CRP values for group and curve condition interaction

Group – Curve condition	Mean CRP (°)	SD
HQ		
Straightaway	83.7	7.8
36.5 m	93.0	13.4
18.5 m	86.0	18.6
LQ		
Straightaway	83.0	11.6
36.5 m	90.3	9.7
18.5 m	89.5	10.0

Figure 13. Mean continuous relative phase for subject groups.

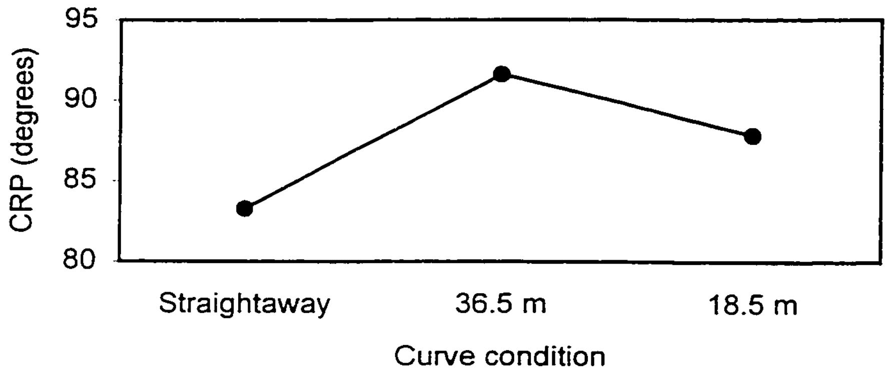
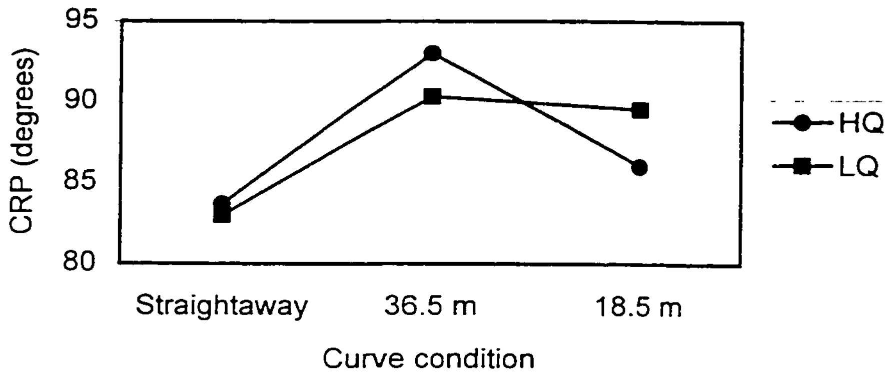


Figure 14. Mean continuous relative phase for each subject group for each curve condition.



Variability of continuous relative phase (vCRP).

Tables 20, 21, and 22 show the values of vCRP for group, curve condition, and the interaction of group and curve condition for the coupling of ankle and knee extension/flexion. Illustrations of the vCRP are shown in Figures 15 and 16. No significant differences were measured for group ($F = 0.041$, $p = 0.845$), curve condition ($F = 0.839$, $p = 0.471$), or the interaction of group and curve condition ($F = 0.342$, $p = 0.721$).

Table 20
Mean vCRP values for groups

Group	vCRP (°)
HQ	38.4
LQ	37.7

Table 21
Mean vCRP values for curve conditions

Curve condition	Mean vCRP (°)
Straightaway	41.9
36.5 m	35.6
18.5 m	36.8

Table 22
Mean vCRP values for interaction of group and curve condition

Group – Curve condition		Mean vCRP (°)
HQ		
	Straightaway	44.5
	36.5 m	34.1
	18.5 m	36.6
LQ		
	Straightaway	39.3
	36.5 m	37.0
	18.5 m	36.9

The contrast of finding significant differences between curve conditions for CRP and not finding any significant differences for vCRP is interesting. Although current research is concentrating on pronation of the subtalar joint and does not involve ankle joint extension/flexion mechanics or phase dynamics, it would seem to be a related factor that should be included in studies with STJ pronation. The variability of CRP demonstrates the stability and adaptability of the coordination pattern of a segment system (van Emmerik, & van Wegen, 2000). Therefore, it appears that the knee and ankle as a system have a similar level of stability and adaptability for both the HQ and LQ. As well, this stability and adaptability appears to be maintained across curves of differing radii when compared to running a straightaway. However, the fact that the coordination patterns are stable does not necessarily mean that they are optimal for performance or for decreasing the probability of injury. The fact that there is a difference between the straightaway and 36.5 m curve condition for CRP, but not vCRP, suggests that some structures of the leg, the knee in particular, may be receiving impact shock forces at inappropriate times due to the difference in timing of the segment movements. As well, if the segments are not in-phase with each other then the tissues that normally attenuate the impact shock of GRF may be transferring some force to tissues that are not adapted to attenuate these forces, or attenuating it in inappropriate positions themselves. This in turn may be a factor in the etiology of running injuries.

Figure 15. Mean variability of continuous relative phase for curve conditions.

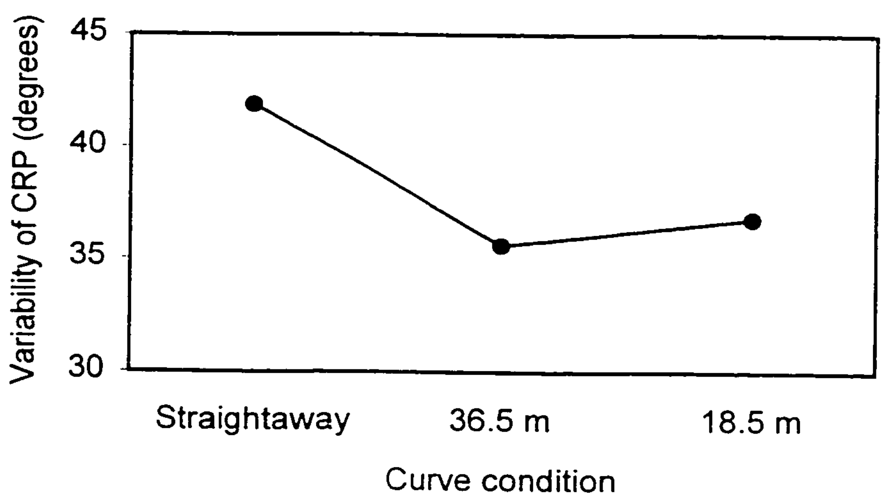
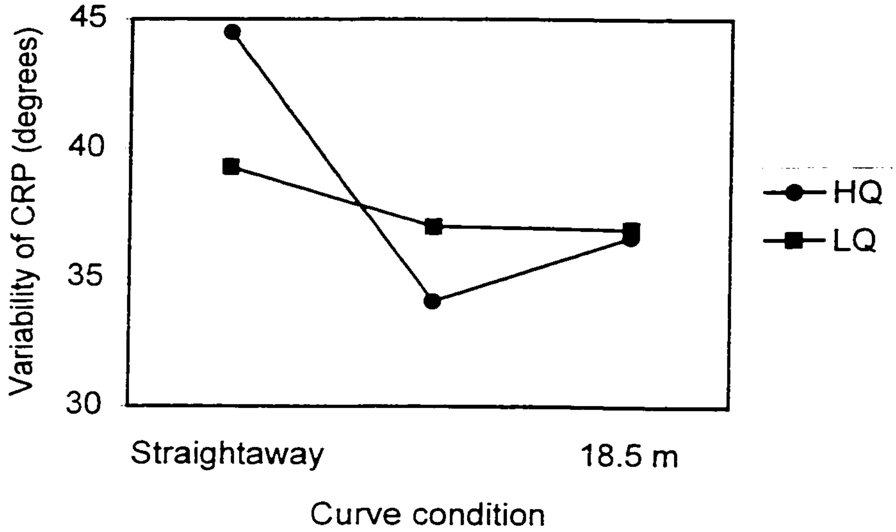


Figure 16. Mean variability of continuous relative phase for each subject group for each curve condition.



Unfortunately, no research exists on the phase relationship between pronation/supination of the STJ and the extension/flexion of the ankle. However, since the STJ is supinated at impact while the ankle is flexed, and the STJ is pronated at toe off while the ankle is extended, one can assume that there is some phase relationship between the two joint movements. Therefore, based upon this assumption, there may be a similar phase relationship between the STJ and knee while running track curves. While studying groups of differing Q-angle Heiderscheit et al. (1999) did not find a significant difference in continuous relative phase of any joint couplings, including STJ pronation/supination coupled with knee extension/flexion. The subjects repeated trials running the straightaway and did not run any curves. The results of the present study support the results from the study by Heiderscheit and colleagues for straightaway running since no significant differences were present between HQ and LQ groups for vCRP and CRP while running the straightaway. No other research exists involving CRP for running track curves and therefore, the present study results may not be validated for the curve conditions.

CHAPTER V

SUMMARY AND CONCLUSIONS

The purpose of this study was to examine the effects of Q-angle on running curve kinematics and kinetics. Developing a better understanding of the functioning of each body segment and the interaction of all body segments during running will allow for better prevention and treatment of running injuries. Approximately 40 million people run in North America and injury occurrence in runners ranges between 24 and 85%. These numbers correspond to between 9.6 and 34 million people injured in North America a year due to running injuries. Therefore, determining the etiology of running injuries and developing preventative measures based on the etiology would seem to be of great importance.

There were three general null hypotheses in this study. Results of this study failed to reject null hypotheses H_01 and H_03, while H_02 was rejected by this study:

1. H1: there are no differences between runners with high Q-angles and runners with low Q-angles: $X_1=X_2$.
2. H2: there are differences in running a straightaway, a 36.5 m curve, and an 18.5 m curve: $Y_1 \neq Y_2 \neq Y_3$.
3. H3: there are no significant interaction effects between groups of varying Q-angle and curves of varying radii: $Z_1=Z_2=Z_n$.

The significant difference between running velocities for the 36.5 m and 18.5 m curves presents some interesting possibilities regarding lower extremity coordination.

Further studies regarding running track curves should include various velocities imposed upon subjects as they run track curves of differing radii and have the effects on lower extremity coordination compared to the lower extremity coordination while running at self selected pace.

It was not expected that subjects between groups would exhibit differences in peak vertical ground reaction force since there was not any previous research that has suggested that Q-angle is related to vertical force production. As well, since previous studies have not found any differences in peak vertical force between track curve conditions, it was expected that this study would fail to reject $H_0 2$. Considering that runners lean in to the curve in order to successfully navigate it, more medial force would be expected to be applied than normal, as shown by Harvie (1998), and less vertical would be applied. However, this study supported previous results that peak vertical force is consistent across track curvature.

There was a trend evident for the time of peak vertical force occurrence. This trend may signify a difference in timing of peak force attenuation for runners with high Q-angles compared to runners with low Q-angles. This may be an important factor distinguishing an increased injury probability for runners with high Q-angles. Future studies involving larger subject numbers will be able to clarify this point and distinguish whether the trend is meaningful or not. Similar time of peak vertical force occurrence between track radii is important since it may have additive effects to the misalignment already occurring while the subject navigates a curve. Assuming the misalignment to change how the tissues absorb the forces, then having them absorb them at a different time may have an increasingly detrimental effect with decreasing curve radii. However, there were not any significant differences between curve conditions or groups, and no significant effects for the interaction of groups and curve conditions.

Similarly, there were no significant effects or differences involving average vertical force or total vertical impulse. Therefore, there is not an increased probability of injury due to increased force absorption while running curves of different radii or for runners with high Q-angles.

Continuous relative phase represents the relative timing of segments movements relative to each other. It was expected that groups would exhibit significant differences for CRP and the variability of CRP. Coordination of the movement pattern is described by the calculation of the variability of CRP. There were no significant differences in the vCRP or CRP between groups. There were no significant differences in the vCRP of curve conditions, but there was a significant difference between curve conditions for CRP. This presents an interesting concept: that segments can demonstrate a stable coordination pattern yet be out of phase with each other. The vCRP being calculated as the standard deviation of CRP at each percent of stance phase allows for analysis of stability of the movement pattern. Therefore, the movement pattern for both groups on each curve demonstrated similar stability. However, the phase difference between the ankle and knee joint may change how impact shock is attenuated and possibly lead to injury. Further investigation into the coordination between the ankle and knee are needed to clarify where the difference in coordination while running track curves lies. Specifically, studies should investigate the phase relationship between the knee and subtalar joint.

Research examining curve running is extremely limited. This study certainly demonstrates the need for further investigations involving running curves and the effects that curves may have on lower extremity coordination, and therefore, the implications that running curves may have on injury.

The conclusions of this study are that runners with high Q-angles are not at greater risk for injury than runners with low Q-angles. Furthermore, all runners, regardless of individual Q-angle measurement, may be at increased risk of injury while repetitively running curves.

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APPENDIX A

INFORMATION AND CONSENT FORM

Faculty:

Human Kinetics

Project Title:

Influence of Q-angle on lower extremity coordination during curve running.

Project Outline:

Male subjects will be required to run and strike a force platform with the left foot while running along a straightaway and two curves of different radii. Subjects will be filmed for kinematic data and kinetic data will be collected from the force platform. Velocity of running will be self-selected by the subject and they will wear their own shoes. Quadriceps angle will be measured on the left leg of each subject prior to data collection and subjects will be divided into two groups based on magnitude of the angle. Measurement will involve use of a goniometer and palpation of the subject's anterior superior iliac spine and tibial tuberosity.

Participation and Confidentiality:

Subjects will be volunteers from the local running community. Withdrawal from testing will be permitted without question and may occur at any time during the study. Subject identity and individual results will be kept confidential. Any data released will be grouped, thereby preserving subject confidentiality.

Researcher:	Jason Leach
Faculty Advisor:	Dr. Wayne Marino

APPENDIX B

SUBJECT CONSENT FORM

Subject Consent:

I, ______________________ (name of participant - please print), have willingly volunteered to participate in this study. I understand that I will be required to run along a straightaway and two flat turns of different radii. Furthermore, I understand that my identity and individual results will be kept in confidence. As well, I am able to discontinue participation in this study without explanation at any time. I realize that the running trials will not require me to perform different than I am accustomed to in my regular training and that I will not become injured in any way due to my participation in this study.

I have had the experimental protocol explained to me and consent to participation.

Signature of participant:_________________

Date: ________________________________

Signature of researcher: _________________

VITA AUCTORIS

NAME:	Jason Leach
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YEAR OF BIRTH:	1972
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