The Effect of Banked-Curves on Running Mechanics: Plantar Foot Pressures

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Abstract

The purpose of this study was to investigate the dynamic patterns of pressure distribution of curved running with and without banked surfaces at two speeds. Seven male elite runners ran at 3.8 m/s and 7.0 m/s in three different conditions: (1) straight, and along a curve with either (2) no side inclination or (3) an inclination of 19%. Running speed significantly affected peak pressure for plantar foot regions ($p < 0.05$); however, few significant main effects were found for surface running condition or foot side. Center of pressure displacement showed similar patterns in both the medial-lateral and heel to toe excursion regardless the running condition, speed, or foot side. Future study needs to reduce measurement variability as well as to consider other dynamic foot-to-shoe components such as shear stresses.

Résumé

Le but de cette étude était de comparer la dynamique de distribution de la pression plantaire lors de la course en courbe sur piste relevée et plate à deux vitesses différentes. Sept coureurs d'élite de sexe masculin ont couru à 3,8 m/s, puis à 7,0 m/s, en trois conditions différentes : (1) en ligne droite et en courbe (2) plate ou (3) avec un dévers de 19 %. Tandis que la vitesse a significativement affecté le pic de pression plantaire ($p < 0,05$), la condition de course et le pied pris en considération ont eu peu d'incidence sur celui-ci. En outre, les déplacements du centre de pression selon les deux axes, médio-latéral et du talon à l'orteil, ont évolué de façon similaire, peu importe la condition de course, la vitesse ou le pied considéré. De futures recherches devront réduire la variabilité des mesures et considérer d'autres composantes dynamiques présentes dans le rapport pied-soulier, tel que le cisaillement.

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Chapter 1: Introduction and Literature Review

The Effect of Banked-Curves on Running Mechanics: Plantar Foot Pressures

1.1 INTRODUCTION

Running, as a recreational and competitive pursuit, is one of the most common forms of physical activity, in large part given its demonstrated health benefits (Hoberigs, 1992; Macera 1992, Van Mechelen 1992). However, where exterior conditions are prohibitive to running (e.g. densely populated urban centers; seasonal snow accumulation), indoor running facilities have evolved with various configurations and sizes. Often tracks may incorporate banked curves in an attempt to compensate for body lean presumably to enhance running performance and / or reduce lower limb injuries. More specifically, it is thought that a banked curve put less torque on the ankles and that it is easier to reach maximum speed without being injured (Greene 1987). However, the validity of this rationale has not been vetted with scientific verification; indeed, some evidence suggested that, conversely, indoor track running may adversely alter running symmetry and increase risks of injury (Beukeboom *et al.* 2000; Lanese *et al.* 1990). Hence, the intent of this study (presented in the subsequent chapter) was to investigate the ground contact plantar pressures of curved running with and without banked surfaces on an indoor track.

1.2 LITERATURE REVIEW

To provide an understanding of the pertinent issues related to the physical activity of running, a review of prior scientific studies, training practices, and epidemiological findings will be covered, as well as an exploration of previous work conducted related to the question at hand. More specifically, the following text will outline biomechanical factors related to running and related methods of analysis; a summary of issues associated with running injuries; and conclude with studies having specifically investigated curved and / or banked running conditions.

1.2.1 Running biomechanics

The analysis of running is not necessarily a new area of study. For example, in ancient Greece Aristotle theorized on the forces generating movement (Novacheck 1998). However, it has not been until recent decades with the development of sophisticate instrumentations capable of collecting copious amounts of information as well as expediting data analysis that researchers have been able to investigate running movement with precision. Our interest in human walking and running is obvious since they are fundamentally the primary means of body translation. Thus, the development of proficient gait movement patterns has important social and economical implications.

The following review of literature pertains to important concepts related to the mechanical nature of human locomotion, beginning with a description of kinematics, kinetics and other metrics commonly used in biomechanical analysis as well as identifying various measurement modalities. Further, the association of these parameters to running performance, particular with respect to our understanding of curved and banked will be address. Lastly, the risk of running related injuries will be presented and the implications of curved and banked conditions explored.

1.2.1.1 The running gait cycle

Running is cyclical, repeated gross movement pattern. This pattern or gait cycle begins with foot ground contact and ends when the same foot contacts the ground again. First foot contact is called "initial contact". The gait cycle may be subdivided in to two main phases: stance and swing. Stance finishes when the foot leaves the ground at "toe off" and marks the beginning of the swing phase. During walking, the stance phase is always longer than 50% of the gait cycle, which means that there are two periods of "double support", one at the beginning and the other at the end of the stance phase. The transition from walking to running occurs when periods of double support are no longer present, giving way to periods of double float when the swing begins and ends. In running, toe off always occurs before 50% of the gait cycle has been completed. The faster the running speed, the shorter the stance phase (Novacheck 1998, Paradisis and Cooke 2000).

Irrespective of the speed of the gait cycle, alternate periods of absorption and generation of energy occur during stance phases. During the absorption, the center of mass falls from its highest point during double float until about the middle of the stance phase. Then, during the second part of the stance phase the center of mass is moving upwards in which is known as generation phase (Novacheck 1998). In the current study, particular attention will be given to the stance phase.

1.2.1.2 Kinematics

Quantification of movement as a whole body or in body segments involves kinematic measures. Kinematics is the branch of mechanics that deals with the movement of the bodies or some points of the body without considering the forces that produce the motion (Millirion and Cavanagh 1990, Novacheck 1998). Kinematic description of the lower extremity's segment and joint motions during the running cycle provides insight into the movement's coordination. Typically, the greatest ranges of motion (ROM) in running are observed in the sagittal plane. Milliron and Cavanagh (1990) have described these movements for each joint of the lower limb during level running at a speed of 3.83 m/s. The following paragraphs summarize their observations:

Hip joint: at footstrike, the thigh has an inclination of about 25 degrees from the vertical because of the hip flexion. After maximum knee flexion during stance has been reached, hip and knee extension begin simultaneously at approximately the same rate. When toe-off occurs, the thigh is at its point of maximum backswing. During swing, the thigh reverses its direction almost immediately and the hip begins the flexion movement. This forward movement of the thigh is shifted to a backward motion (extension) well before footstrike.

Knee joint: at footstrike or initial contact, the knee is flexed about 10 to 20 degrees. Following initial contact and through mid stance the knee becomes more flexed during which is know as the cushioning or absorption phase. During late stance the knee extends, but it is still flexed at approximately the same degree it was during initial contact. After toe-off, the knee rapidly flexes reaching a maximum of about 110 degrees near mid swing. After this point, the knee extends at a similar rate it flexed during the early swing. The end point of extension is about 10 or 15 degrees shorter than full extension but right before footstrike an extra 5 degree flexion occurs.

Ankle joint: the ankle plantar flexed about 5 degrees immediately after initial contact. Due to knee flexion and the forward motion of the shank, there is a dorsiflexion at the ankle that finishes near mid stance. The ankle then begins to plantar flex, reaching its maximum angle of about 70 degrees (20 degrees from neutral) shortly after toe-off. During the greatest extent of the swing, the ankle comes back to a neutral position of about 90 degrees.

When comparing the ROM at the hip, knee and ankle in walking, running and sprinting it is evident that although the pattern of motion is fairly similar the extreme values and, as a consequence, the ROM are different. As an example, the knee flexion is about 55 degrees when walking at 1.2 m/s and it goes up to 130 degrees when sprinting. In the same way, the thigh flexion is not greater than 50 degrees when walking but it reaches about 100 degrees when sprinting (Novacheck 1998).

Kinematic patterns are sensitive to terrain profiles. For example, the ROM is not only affected by the speed but also by the incline of the running surface. Kuster *et al.* (1995) showed that when running downhill (19 degrees) some differences in lower extremity kinematics can be observed. At the ankle the major difference occurred from late stance to late swing. At toe-off the ankle was 8 degrees less plantarflexed when walking downhill. At the knee, the biggest differences were observed during stance phase and to early mid swing, with greater knee flexion in downhill walking. These differences were about 20 and 10 degrees during mid stance and mid swing, respectively. At initial contact the knee was found to be almost fully extended in both conditions. Finally, the primary differences for the hip were found during swing and early stance phase. In comparison, downhill walking showed a lower degree of hip flexion throughout the swing phase. Hence, lower limb kinematic movement patterns during running are pertinent measures to evaluate in assessing the gait adaptations to varied terrain conditions. As such, numerous research studies have used this as a parameter for investigation. More will be said about running kinematics in later sections (1.3).

1.2.1.3 Kinetics

In addition to kinematic description of running, kinetic estimates are also germane to running analysis. The study of kinetics helps to explain the factors underlying the movement. It begins to answer the "how and why" of motion. The following text will discuss two kinetic parameters (force and pressure) as observed between the ground and the runner's foot.

Ground reaction force (GRF) is the force that reacts to the push transmitted to the ground by the foot of the runner with identical magnitude and opposite direction (Miller 1990). For the analysis of the gait cycle, often this force is decomposed into three different directions: vertical, backward-forward (braking-propulsion), and side to side (medial-lateral). The vertical component is usually the largest and is related to the gravity. The other two components (known as shear forces) are present because of the friction between the ground and the foot. The vectoral sum of these forces acting around the foot at a given time represents the total GRF (Belli *et al.* 2002, Roy 1988). Ground reaction forces reflect the acceleration (or in some instances deceleration) of the total body and the support limb transmits the force to the ground (Miller 1990).

Related to force is the term pressure. In the context of the plantar foot surface, pressure is defined as the total perpendicular (or normal, as in right angles) force acting on a given surface divided by the area of that surface, and is expressed in units of force per area such as pounds per square inch (psi) or Newtons per square meter (Pascals, Pa) (Roy 1988). The center of pressure (COP) is the weighted average location point or the total pressure contact area. For this reason, its location changes along the stance phase (Miller 1990). In general, the COP is nearly equivalent to the center of GRF. During walking the COP corresponds to the forward progression of the whole body center of mass (Schmid *et al.* 2004). Various parameters such as running speed, hardness of the surface, cushioning of shoe and running style are known factors that will affect the location and movement of the COP. Further discussion of these factors will be presented subsequently.

To estimate GRF, pressure, and COP requires the use of dynamic measurement techniques (Graf, 1993). For example, force plates may be used to measure GRF between the shoe and the ground surface. Useful insights may be obtained from such measures. For instance, vertical force profiles can identify normal walking patterns that typically possess two GRF peaks: the first one corresponds to the rapid loading of the heel before the forefoot loading; and, the second peak represents the push-off phase. These two peaks are usually greater than the body weight (Roy 1988). Variation GRF profiles tend to correspond to variations in gait patterns itself. For instance, mid-foot strikers do not normally display the first peak. Similarly, when running a distinct GRF profile is produce: an initial, transient spike at foot contact and a second more prolonged GRF loading during weight acceptance and push-off.

Force plate devices are accurate but need to be anchored securely to a rigid concrete surface; thus, limiting their applicability outside a laboratory setting. Furthermore, these instruments cannot measure the interaction between the foot and the shoe.

In part for this reason, other instrumentations have been developed to study the foot-to-shoe. This would be relevant to the assessment of footwear design or the effect of orthoses, for examples. Typically, in-shoe devices will measure pressure, rather than force. Attention must be given to the thickness, size and position of the transducers so that the instrumentation does not in itself alter gait (Roy 1988).

Dynamic plantar foot surface pressure measurement corresponds to changes during the stance phase. Typically, a foot pressure mapping depicts an evident transfer of pressure from the heel to the toes during the gait cycle (Roy 1988). Dynamic pressure mappings often are receptive to subtle difference in foot placement. With regards to running, often the heel is the first portion of the foot to make contact (corresponding to the first spike in the GRF). At this point, pressure is concentrated mostly at the heel. Mid-foot strikers can be identified by a lack of peak heel pressure. The heel support persists for approximately 55 to 65% of the support phase, and reaches its peak at 15 to 20% of the stance. During the mid-stance, pressure is distributed between the heel and metatarsal heads. The latter sustain substantial pressures for about 80% of the cycle and reach their peak at about 75% of the stance. Usually, the first metatarsal accepts weight after the fifth metatarsal head, and pressure under the first metatarsal head ends later than under the fifth. The pressure under the mid-foot is low compared to the pressure under the heel and the toes (Roy 1988). Finally, the toes sustain pressure just prior the heel off, and they reach their peak late during the stance phase (around 85% of the support phase for the hallux). The pressure under the toes last for about 55 to 60% of the stance phase and the peak under the hallux is much greater than the peak under the other toes. These dynamic changes in pressure are reflected by the posterior-anterior migration of the COP from the heel to the toes. The medial-lateral COP path also has distinct and typical patterns that may give insight into the individual's supination-pronation balance during stance.

As noted with kinematics, these kinetic measures reflect changes in gait due to different factors such as running speed, midsole hardness, type of surface, body mass, area of contact, soft tissue under the foot, and so on (Hreljac 2004, Nigg *et al.* 1987).

The effect of running speed: When running, peak vertical forces are much bigger and they can reach values of two to five times body weight (Nigg *et al.* 1987, Nigg 2001, Roy 1988). Nigg *et al.* (1987) found that peak impact forces increased almost linearly from speeds of 3 m/s to 6 m/s. In addition, a nearly linear decrease in contact time was found running at different speeds from 3 m/s to 6 m/s (Nigg *et al.* 1987), and an inverse correlation was demonstrated between contact time duration and peak force (Roy 1988). Since contact time is shorter at higher speeds, running faster will produce larger peak forces.

Midsole hardness: Another factor influencing the GRF is the cushioning of the shoes. Logical thinking lead us to assume that manipulating the material properties of the midsole under the heel will produce a change in the way impact forces act. Surprisingly, Nigg *et al.* (1987) demonstrated that the vertical impact forces do not necessarily vary with changing the midsole hardness. They also provided evidence that the midsole hardness do influence the point of application of the acting external GRF relative to the foot. The authors concluded that changing the midsole hardness is not related to a decreased peak force, but different cushioning materials can be used to influence the internal impact forces. Softer materials reduce ligamentous forces and increase joint forces, while harder materials produce the inverse results if nothing else is changed.

In another study, Hennig and Milani (1995) compared the influence of 19 different commercially available running shoes on foot pressure in a group of 22 individuals during overground running. Unlike from Nigg *et al.* (1987), they concluded that the different shoe model influenced peak foot pressures and altered foot mechanics.

Wearing shoes and barefoot: As it can be inferred from the last paragraph, it is not clear whether or not different types of shoes may influence the pressures under the foot. However, when the use of shoes was compared with walking barefoot, some differences have been found to be clear (Roy 1988). Shoes notably decreased the peak force under the posterior part of the heel. The initial spike seen in the vertical GRF when walking barefoot is lower and occurs later when wearing shoes. The use of shoes is also responsible for the decrease of the peak force under the lateral three metatarsal heads. In addition, the time to peak pressure for all metatarsal heads is longer in shoes (Roy 1988).

The effect of pronation: One of the factors regulating the effect that impact force has on an individual during running is the amount and rate of foot pronation, which is defined as a motion that combines dorsiflexion, abduction, and eversion of the foot with respect to the leg (Holden and Cavangh 1991). Pronation "is a protective mechanism during running since it allows impact forces to be attenuated over a longer period of time. Pronation is detrimental to a person only if the level of pronation falls outside of normal physiological limits (too low or too high), and if it continues after midstance. After midstance, it is necessary for the foot to become more rigid in preparation for toe off' (Hreljac 2004, p.847). Hence, numerous studies have focused on the relationship between pronation and foot/ankle injuries.

Pressure distribution and sensory input: According to Nurse and Nigg (2001), the patterns of human locomotion are central nervous system (CNS) sensory input dependent. That is feedback from the foot pressure provides constant information about the loading, joint kinematics, and the pressure distribution under the foot that in turn leads to modulation of locomotion patterns. Since the foot is usually the only part of the body in contact with the ground, input from cutaneous receptors in the foot must play an important role in the regulation of the gait cycle.

Nurse and Nigg (2001) also examined the plantar pressure changes that occur when altering sensation on the plantar surface of the foot by using ice. They found that, relative to the baseline condition, the COP showed a significant anterior shift in the first 15% of the step when the rearfoot was cooled. When the forefoot was cooled, the COP showed a posterior shift in the last 15% of the step cycle. They concluded that if sensory feedback is inhibited from a portion of the foot, the COP will shift towards an area of more sensitivity.

Taylor *et al.* (2004) have also experimented with plantar foot insensitivity using ice. The conclusion of their work is that experimentally induced plantar insensitivity significantly modifies the distribution, duration, and magnitude of the forces and pressures under the foot under walking conditions. The plantar pressure distributions in this study showed that the individuals adopted a more conservative gait with longer forefoot loading and less participation of the toes during propulsion.

The results obtained in the above mentioned studies indicate that the feedback input from the receptors under the foot plays a very important role in forces distribution and magnitude. It was seen that extremely low temperatures affect these receptors. Other factor such us type of surface, inclination of the track, radius of the curve, and so on could potentially influence the pattern of pressure distribution.

1.2.1.4 Electromyography (EMG)

A final item for discussion in terms of parameters used in gait analysis is electromyography (EMG), a measure that estimates the extent of dynamic muscle activity. Similar to the other measures, EMG patterns during running gait on level surfaces has been well described in several studies. During normal running the pattern of muscle activity may be summarized as follows:

Quadriceps: activity begins during the last half of swing preparing the limb for ground contact, and continues into the first half of the stance absorbing the impact during the phase of absorption (Novacheck 1998, McClay *et al.* 1990, Mann *et al.* 1986). During midswing only the rectus femoris is active, which is thought to control the posterior movement of the tibia during knee flexion. Further, this biarticular muscle is believed to play a role in energy transfer between segments (Novacheck 1998).

Hamstrings: (semitendinosus, semimembranosus and biceps femoris) active during the second half of swing and the first half of stance (McClay *et al.* 1990, Novacheck 1998) with no hamstring activity during the early mid swing, the flexion of the knee is thought to occur passively due to the rapid forward acceleration of the thigh. During late swing the main function of these muscles is to decelerate hip flexion and control knee extension. When the thigh begins moving backwards, these muscles work concentrically in order to extend the hip and flex the knee (McClay *et al.* 1990). It is believed that at contact hamstrings and quadriceps co-activate to provide increased joint stability during the impact (McClay *et al.* 1990).

Anterior tibialis: active during most of the gait cycle (McClay *et al.* 1990), the anterior tibialis first acts concentrically to produce ankle dorsi-flexion to permit foot clearance during swing and thus allow initial contact to occur with the hindfoot. Subsequently, tibialis anterior acts eccentrically during the first part of stance to control the lowering of the forefoot (Novacheck 1998).

Gastrocsoleus: (gastrocnemius and soleus) activity of the gastrocsoleus has been reported to begin during late swing to about 80% of the stance-phase. In late swing, these muscles stabilize the foot in preparation for the footstrike. During the stance it works actively as a plantar flexor (McClay *et al.* 1990).

In accordance with the above mentioned study, Karamanidis *et al.* (2003) showed that running at different speeds (from 2.5 to 3.5 m/s) and three different stride frequencies (preferred and ± 10% from preferred) the reproducibility of the EMG parameters was not influenced by these variations in running technique. However, in other circumstances EMG has proven to be a sensitive parameter to indicate changes in muscle activity as the running terrain changes. For instance, Swanson and Caldwell (2000) looked at the activity of the tibialis anterior, gastrocnemius, soleus, rectus femoris, vastus lateralis, medial hamstring, biceps femoris, and gluteus maximus while running on a treadmill in an incline and level state. Though the linear envelope EMG patterns were similar in both conditions, a large increase in activity during the stance phase when running uphill was detected for the gastrocnemius, soleus, rectus femoris, vastus lateralis, and gluteus maximus when running uphill.

In another study, Wank *et al.* (1998) compared lower limbs muscle EMG patterns while running over ground and on treadmill at two different speeds (4 and 6 m/s). The pattern of activity of the leg muscles was generally similar between these conditions, but some small differences could be systematically observed. For example, the biceps femoris showed greater magnitude and longer duration during foot contact and the first part of the swing phase in treadmill running. The rectus femoris also showed a larger activity on the treadmill during the hip flexion. Despite these minor changes, the overall muscular activity was almost the same comparing the two different running conditions and speeds.

In summary, walking and running gait is typically evaluated by use of kinematic, kinetic and / or EMG measures. For the current study, dynamic plantar foot surface measures were chosen as the dependant variables of focus. This technology allowed for testing to occur under the track conditions of interest. With the above review of measurement methodology and observation in mind, the following section will address our understanding of running on curves and cambers.

1.2.2 Running on curves and cambers

Human running gait has been analyzed in the past decades, with most previous studies being performed under controlled laboratory conditions. Arguably, this situation leads to obtain significantly different gait parameters than those obtained under natural conditions (Sun *et al.* 1996). Further, most of these studies exclusively related to forward progression only, and not with running along a curved path. In order to better understand the effects of running on different types of curves, research has been performed on the track or on the road, collecting useful kinematics and/or kinetics information to explain the changes in the gait cycle when dealing with curves. The following text will present those studies that have addressed the effects of curves and banks on running.

1.2.2.1 Effects of running on curves

Since the times of ancient Greece, athletes predominantly have been training and competing in a counterclockwise direction. Today's athletes continue to do so on the oval tracks. Modern athletes train almost every day in preparation for a competition. A substantial part of their training takes place on tracks, some of which are smaller radii indoor tracks (Beukeboom *et al.* 2000). This constant counterclockwise running training may affect the biomechanics of the gait cycle, especially in indoor tracks.

In a previous study, Greene (1985) attempted to explain the factors behind speed attenuation with decreased radii. He found that in the theoretical model "the basic mechanical phenomenon causing speed reduction is the increase in foot contact time necessary to maintain constant vertical impulse to compensate for the vectorial decrease of available vertical force as the individuals heels over into the turn". This increase in foot contact time generates a consequent decrease in the ballistic air time. The author proved that agreement between theory and experiment was good, but he pointed out that theoretical curves should be seen as the upper limit of attainable speeds at a particular radius. From a practical point of view, Green demonstrated that there will be a speed penalty when running on curved tracks relative to straight tracks.

Unbanked curves: Ryan and Harrison (2004) examined the effect of bend radius on lower limb kinematics and on the factors directly related to sprint speed: stride length (SL, defined as the length of a single step), stride rate or frequency (SR) and ground contact time (CT). In this study, 13 competitive sprinters were asked to sprint at maximum effort over a 70 meter section that included a curved section with 4 different radii from 10.5 to 45.04 m. The results indicated a clear curve radius related effects on running speed. SL and running speed increased with increasing the radius and CT generally decreased as the radius increased. This data showed that running on the inner lanes represent a disadvantage for the athlete. This is even more evident when running on indoor tracks, where the radius of the curve is smaller. The longer CT in the bends of smaller radius was found to be consistent with the reductions in running speed and SL. The authors explain these reductions by the augmented centripetal acceleration that is likely to occur when sprinting at maximum speed on curves with smaller radius. Under those conditions it will be essential for the athlete to generate greater centripetal force. Since maximum force output is restricted by leg strength, the athlete will spend more time in contact with the ground to maximize impulse generation and maintain flight distance.

The relation between indoor tracks and asymmetrical changes in muscle strength has recently been reported. Beukeboom *et al.* (2000) examined strength changes in the hindfoot invertor and evertor muscle groups of athletes training and competing primarily in the counterclockwise direction on an indoor flat curve. For this study invertor and evertor ankle strength was tested concentrically and eccentrically on both legs with 25 subjects prior to the onset of the indoor training season and at the end of the season. The data from this investigation showed that muscle imbalance was observed when comparing pre and post training results. The findings suggest that training and competing primarily in the counterclockwise direction on an indoor unbanked track resulted in muscle invertor-evertor imbalances that interacted with limb side. According to the authors, when running in curves a force component that is not present in linear running must be exerted by the athlete's support foot against the track in a parallel but opposite direction to the radius of the curve in order to accelerate in a curvilinear direction. During each stride "the evertors of the right (outside) foot and the invertors of the left (inside) foot would be primarily responsible for this additional force component in conjunction with the hip abductors and adductors" (Beukeboom *et al.* 2000). The authors summarize three differences between linear and curvilinear running as a result of the additional force component: a) there is a decrease in the available ballistic force with which athletes are able to push themselves off the ground against gravity. This phenomenon will produce a loss of speed when running on the curve; b) the dynamic stabilizer of the subtalar joint are exposed to greater forces running in curves and for that reason the evertors of the right foot and the invertors of the left foot are stressed to a greater degree; and c) the loss of running speed and the increase in subtalar joint stabilizing forces are inversely related to the radius of the curve.

Banked curves: The feeling of those running on banked running tracks is that a banked curve put less pressure on the ankles and that it is easier to reach maximum speed without being injured (Greene 1987). Running along a flat curve is, at the same speed and radius, more demanding than running on a banked turn. It seems that "stress is reduced most when the heel-over angle (defined as: heel-over angle = arctan (v^2/gR) where 'v' is the velocity, 'g' the gravitational acceleration, and 'R' the turn radius) of the runner is equal to the bank angle of the track" (Greene 1987). The optimal banked angle will be both speed and radius dependent. According to these conditions the track may be underbanked, optimally banked, or overbanked (Green 1987).

Greene (1987) investigated the effects of running on four different bank angles and with three different curve radii. He showed that bank angle effects can produce a 10% difference in the speed of the runner. A mismatch between the bank and the heel-over angle of 30 degrees will produce a decrease in speed of 11%. He concluded that for running tracks the turns should be banked and that each lane should be banked to a different degree. Greene proposed that the reason for speed attenuation with bank angle mismatch is that the available foot forces decrease because extra tendon force is required to maintain roll stability about the foot's anteroposterior pronation/supination axis (i.e. inversion/eversion). Decrease in foot force results in attenuated top speed because of decreased air time and increased ground time.

In summary, the works of Green (1987), Beukeboom *et al.* (2000) and Ryan & Harrison (2004) all indicate that curve running will create dynamic asymmetries in running gait patterns and may well promote morphological asymmetries under prolonged exposure. The next section will now discuss the effects to gait when the running surface possesses a camber (or side inclination or bank).

1.2.2.2 Effects of running on camber

Roads often possess varying degrees of camber to promote water drainage; thus, many runners must contend with running on unleveled terrain. Camber roads have been implicated in overuse injuries. Running repeatedly on the same side of the road places an uneven tension on the legs and can be responsible for alteration in kinematics and kinetics variables. For instance, Sussman et al. (2001) investigated the consequences of running at 0, 2.5, and 5 degrees camber at two running speeds: 6 and 7 mph. They found a significant main effect between right (downhill) and left (uphill) legs for the knee angles at initial contact and toe off. After the post-hoc analysis the differences were displayed between the right and left knees at toe off at 2.5 and 5 degrees running at 7 mph. It was also found that the downhill leg exhibited greater knee extension than the uphill leg, and the ankle showed greater values of plantar flexion as the level of inclination increased. The authors concluded that running on camber roads can be implicated as a possible cause of injury and that if this is the case, changes in training patterns should be applied in order to avoid this problem.

Gehlsen *et al.* (1989) measured the ROM of the right and left knee in 15 subjects running on a horizontal or laterally tilted treadmill. Inclination ranged from -10 to 10 degrees with increases of 5 degrees. They found that during the swing phase the ROM for flexion-extension was significantly different between 10 and -10 degrees camber and between 10 and -10 degrees compared to 0 degree camber. During the support phase statistical significance were found between 10 and -10 degrees camber as well as between 10 and -10 degrees compared to 0 degree camber. The internal-external rotation was significantly different for the swing and support phase when 10 and -10 degrees camber were compared. A significant difference was also found between 5 and -5 degrees camber during the swing phase. Finally, a significant difference between 10 and -10 degrees camber for both, the swing and support phase was found when analyzing valgus-varus motion. The authors concluded that these kinematics changes may be related to the point of application of joint forces. If the center of force is directed more medially or laterally than normal because of the camber, then care should be taken when running on those surfaces. Misalignment of the femorotibial angle in the knee can produce degeneration of one or other side of the knee depending on which one receives the higher load.

1.2.2.3 Similarity to leg length discrepancy

From the above studies, it may be postulated that running on a banked curve is likely to mimic the effects of running with a leg length discrepancy (LLD). The outside leg being the analogous to the short leg and the inside leg being like the long leg. LLD is defined as a condition in which lower extremities are noticeable unequal (Gurney 2002). Some authors believe that a LLD of 5 mm or more has significance in mechanical dysfunctions around the hip, pelvis and spine. Other researchers think that LLD lower than 12.7 mm have no pathological relevance (Woerman *et al.* 1984). Some studies have proven an association between LLD and some postural problems like lower back pain and scoliosis while other studies have related LLD (20 -30 mm) with changes in gait such as increased GRF, increased energy consumption, and increased lower extremity kinetic energy (Gurney 2002).

In his literature review on this topic, Gurney (2002) described the role of LLD on standing posture and on walking. When standing, the longer leg is compensated by pronation of the foot on the longer leg and supination and/or plantar flexion of the shorter leg. The knee and the hip can also compensate with an extension of the shorter leg and/or flexion of the longer limb. The COP is also affected by LLD in standing position. It has been reported that a 10 mm LLD produced a significant shift in the mediolateral position of the COP towards the longer leg (Gurney, 2002). During walking, some important gait characteristics with LLD include decreased stance time and step length on the shorter leg, decreased walking velocity, and increased walking cadence. Different compensatory mechanisms can contribute to lengthen the shorter leg including increasing downward pelvic obliquity, increasing knee extension in mid stance, toe walking, or a combination of any of these mechanisms. In addition, the longer leg may be shorten by increasing pelvic obliquity, circumduction, increasing hip and/or hip flexion, or any combination of these proposed mechanisms (Gurney 2002, Walsh *et al.* 2000). Furthermore, structural or functional leg length discrepancy has been found to produce asymmetry in the GRF. The longer leg was consistently found to have the larger GRF. Since a relationship has been established between GRF and economy during locomotion, GRF values can be useful to explain the higher energy expenditure observed in the gait cycle in individuals with LLD (Gurney 2002). In summary, it has been proposed that biomechanical abnormalities due to LLD are three times more significant in running compared to walking (Gurney 2002).

Hence, from the above studies evidence suggests that the mechanics of non-linear running will necessitate asymmetric lower limb coordination and unequal dynamic responses. This may have implications to performance and increased risks of injury. . To best appreciate running and running related injuries, a brief review of the forms and prevalence of injuries, as well as associated runner characteristics will be presented.

1.2.3 Running injuries

Several studies of running have examined the rate, sites and types of injuries providing, with more or less agreement, a complete description of this problem (Hoeberigs 1992, Hootman *et al.* 2002, Hreljac 2000 *et al.*, Hreljac 2004, Koplan *et al.* 1995, Lysholm and Wiklander 1987, Lanese *et al.* 1990, Macera 1992, Mechelen 1992, Messier and Pittala 1988, Messier *et al.* 1991, Messier *et al.* 1995, Orchard *et al.* 1996, Tauton *et al.* 2003, Taunton *et al.* 2001, Wen *et al.* 1998). It is theorized that there is an optimal level of stress for the body's biological structures. If an optimal, or close to optimal level, of stress is applied (number of repetitions, intensity, frequency, duration) with an adequate recovery time, the structure being stimulated should have a positive adaptation (Hreljac 2004). If any of these factors are not appropriate, the risk of injury is present. However, unknown is exactly what the optimal level of stress is and where the threshold between positive and negative tissue and structural adaptations lies. Given the multifactorial nature of this activity, identifying the etiology of running injuries is not precise.

In general according to the literature, runners are classified as having an injury when

(Messier *et al.* 1995, Tauton *et al.* 2002, Wen *et al.* 1998):

- They had pain or symptoms during or immediately after a run
- They had pain or symptoms related to the beginning of a running program
- The injury forced them to stop running or significantly decrease the distance run
- The injury made them visit a medical specialist

1.2.3.1 Prevalence of running injuries

With regards to the prevalence and nature of running related injuries, there are substantial discordances in the literature regarding the injury-incidence rate in runners varying from 24 to 77%; however, when considering only studies with samples size of more than 500 participants the yearly incidence rate fluctuates between 37 and 56% (Van Mechelen 1992). As an alternate way of measuring injuries prevalence, differences in the rate of injuries per exposure have also been obtained. For instance, Lysholm and Wiklander (1987) found an injury incidence of 3.6 injuries per 1000 hours of running, and an incidence between 2.5 to 5.8 injuries per 1000 hours of running for competitive athletes. When comparing outdoor and indoor tracks the number of injuries per 1000 hours of running was 1.65 for the indoor season and 1.25 for the outdoor season (Lanese *et al.* 1990). This latter point is particular relevant to the research question at hand.

1.2.3.2 Site of injury

Most of the studies indicated that in runners the knee is the site of most frequent injuries (Lysholm and Wiklander 1987, Mechelen 1992, Messier *et al.* 1991, Tauton *et al.* 2002). According to Van Mechelen (1992) knee injuries account for 25% of the injuries.

Messier *et al.* (Messier 1991) reported a rate injury for the knee between 23.2 and 41.7%. Tauton *et al.* (2002) showed that injuries at the knee are about 42.1% of the total injuries. Other common sites for injuries are foot/ankle (16.9%), lower leg (12.8%), hip/pelvis (10.9%), Achilles/calf (6.4%), upper leg (5.2%), and lower back (3.4%). In general it is accepted that more that 66% of the running injuries occur in the lower extremities (Hootman *et al.* 2002) and 70 to 80% of those injuries are at the knee and below. As a consequence of these injuries runners have to stop or decrease their training in 30 to 90% of the cases, they have to go over a medical treatment in 20 to 70% of the cases, and they have to be absent from work in 5% of the situations (Van Mechelen 1992).

1.2.3.3 Characteristics of the runners

Numerous factors may influence the location, severity and prevalence of injury; some of these include: age, gender, anthropometrics, training regime, flexibility as well as the often noted question of shoe type and running surface. However, on further examination, some of these have been shown to have little correspondence to injury. For instance, though suspect, the factor of age has been shown not to be related with running injuries. In fact some of the studies showed that younger runners were more susceptible to be injured than older runners. However, these results should be taken cautiously because of the so called "healthy runner effect" by which the subjects that remain free of injuries over the years continue to run (Macera 1992, Van Mechelen 1992). Similarly with regards gender, Hootman *et al.* (2002), Lysholm and Wiklander (1987), Macera (1992), and Van Mechelen (1992) found that gender does not seem to be an important risk factor related to running injuries. For instance, the latter authors reported that women tended to experience more injuries than men, but the difference found was not statistically significant.

In regard to the anthropometric characteristics of the runners, Van Mechelen (1992) speculated that taller and/or overweight individuals could be more vulnerable to running injuries due to the greater forces applied on the bones and muscles. However, those higher forces to overcome could be compensated by stronger muscles and larger bone areas. According to this author no study has shown a correlation between bodyweight and running injuries. Some authors looked at lower extremity alignment abnormalities such as excessive Q-angle or leg length discrepancies (LLD); however, contradictory results were found in this matter. Some studies have shown a link between these abnormalities and the rate of running injuries, while others were not able to demonstrate such a relationship (Hreljac 2004).

1.2.3.4 Volume, intensity and density

In addition to the above mentioned variables, the nature of running exposures has been implicated as an injury risk factor. For instance, weekly running frequency, distance, and duration have been related to running injuries for both men and women (Hoeberigs 1992, Hreljac 2004, Macera 1992, Van Mechelen 1992,). For long distance runners Lysholm and Wiklander (1987) found a significant relation between the distances covered in a month of training and the number of injuries in the following month. Further, Hoeberigs (1992) and Macera (1992) noted from a review of numerous injury reports that distance run per week seems to be the primary predictor for running injuries. The authors based their conclusion in data obtained from 10 studies proving this point. Distance run remains a strong predictor of injury even after adjusting the results for other running-related activities.

In regard to the running intensity, Van Mechelen (1992) explained that the rate of injuries is related to performance. High performance runners seem to be more exposed to injuries than middle to low level runners. On the contrary, Macera (1992) commented that even the high running pace is indicative of physical strain and thus potentially an increased risk of injuries, running speed is not necessarily an important factor after adjusting for the total distances.

A third component of training schedule is density: the ratio between work and rest. Appropriate recovery periods are always necessary in order to avoid injuries. Hreljac (2004) indicated that adequate resting intervals are important to the body structures to handle the stress at which they are being exposed. In order to prevent injuries care must be taken in this regard.

1.2.3.5 Range of motion and flexibility

Yet another factor that is speculated to be related to running injuries has been a lack of flexibility (Van Mechelen 1992). Poor flexibility may increase the stiffness of a muscle, possibly placing more stress on the adjacent joint. It can also be indicative of muscle fatigue, thereby leading to improper mechanics. These are some of the main reasons why stretching before and after running is widely recommended. However, Hreljac *et al.* (2000) showed that stretching the hamstrings as a mean of warm up or cool down is not an effective strategy to avoid injuries. Nevertheless, the author suggested that having good flexibility may be an important factor preventing problems. This would mean that even though an adequate range of motion and flexibility are desirable, the use of stretching exercise pre and after running should not be seen as an essential requirement.

1.2.3.6 Running shoes and surface

Lastly, much attention has been given in the literature to understanding the mechanisms of running and the optimal design of footwear to both improve performance and reduce running related injuries.

Landing forces from consecutive foot falls has long been suspected as an important dynamic parameter corresponding to running injuries. The forces involved in landing are slightly higher than the own bodyweight when walking and they go up to 1.5 to 5 times bodyweight when running (Hreljac 2004). The higher the speed the bigger these forces are. The shock during contact is mainly absorbed by the lower extremities and the foot is the link between the floor and the body. It seems to be evident that increasing the cushion in the shoe should play an important role in shock absorption, and therefore prevent muscle and joints from injuries (Van Mechelen 1992). In the same way, running on softer surfaces should produce a lower load in the lower extremities. However, Nigg *et al.* (1987) found that increasing the hardness in the midsole of the shoe did not produce a decrease in foot pressure at any speed.

Given the well known positive effects of running, this activity is chosen by a number of people as a way of having a healthy lifestyle, competing, performing and relaxing. However, as seen earlier, some risks are associated with this sport. People should organize their running training in such a way that the beneficial effects are maximized and the risks of injuries minimized.

1.3 SUMMARY

Running, by itself or as part of other sport contexts, is one of the most practiced physical activities around the world. Several reasons such as health benefits, relaxation, competition, performance, fitness status improvements, low cost, no needs of a schedule or specific facilities, and so on have been important in order to make this activity so popular. Because of its popularity running has been extensively examined by researchers in the past decades, and several benefits and potential risks have been identified. Even though prior investigations have provided useful insights, the full extent of the effects of running in curves and cambers is still unknown. Future research in this field would provide runners, trainers, and doctors a better understanding of the benefits and problems likely to occur when running on curves. Hence, this was in large part the rationale for the current study.

Accordingly, the objective of this study was to investigate the physical effects of curved running with and without banked surfaces. More specifically, dynamic patterns of pressure distribution while running at two speeds were investigated to compare different flat and banked curved conditions.

Chapter 2: Research Article

The Effect of Banked-Curves on Running Mechanics: Plantar Foot Pressures

2.1 INTRODUCTION

Running has become one of the most accepted forms of physical activity given its demonstrated health benefits (Macera 1992, Van Mechelen 1992). Where exterior conditions are prohibitive to running (e.g. densely populated urban centers; seasonal snow accumulation), indoor running facilities have evolved with various configurations and sizes. Often tracks may incorporate banked curves in an attempt to compensate for body lean presumably to enhance running performance and / or reduce lower limb injuries. More specifically, it is thought that a banked curve put less torque on the ankles and that it is easier to reach maximum speed without being injured (Greene 1987). However, the validity of this rationale has not been vetted with scientific verification; indeed, some evidence suggested that bank curve running may increase injury risk.

Various biomechanical measures may be used to study running mechanics (Nigg *et al.* 1987, Nigg 2001, Roy 1988). One such measure has been to record plantar surface pressures between the runner's foot and shoe. Pressure may be defined as the total normal (perpendicular) force acting on a given surface area (Roy 1988) with the center or pressure (COP) being a representative point of balance. Further, COP changes during the stance phase (Miller 1990) of walking and running corresponds to the forward progression of the whole body center of mass (Schmid *et al.* 2004). Typically, the foot pressure transition displays an evident transfer of weight from the heel to the toes over a gait cycle (Roy 1988). However, various parameters such as running speed, midsole hardness, type of surface, body mass, area of contact, soft tissue under the foot and running style have been shown to affect the location and movement of the COP (Hreljac 2004, Nigg *et al.* 1987). Other factor such us type of surface, inclination of the track, radius of the curve, and so on could potentially influence the pattern of pressure distribution.

Hence, the intent of this study was to investigate the physical effects of curved running with and without banked surfaces. More specifically, dynamic patterns of pressure distribution while running at two speeds were investigated to compare different flat and banked curved conditions.

2.2 METHODS

2.2.1 Participants

Seven elite middle to long-distance male runners provided their written consent to participate in this investigation after being fully informed of the nature and demands of the study. Each runner completed a questionnaire regarding training habits and injury history before participation. Participants were excluded if they presented any recent lower limb injury, previous lower limb surgery, and/or a leg length inequality greater than 5 mm. All runners had access to the McGill Fieldhouse where the evaluations took place. The study had the approval of McGill University's Faculty of Education Ethics Review Board. The mean (±SD) age, height and body mass of the participants were as follows: 20.6 ± 3.0 years, 67.8 ± 3.9 kg and 178.2 ± 4.7 cm.

2.2.2 Experimental protocol

Anthropometric measurements

Body weight was measured to the nearest 0.1 kg using a Tanita BF 350 scale plus body fat monitor (Tanita Corp., Tokyo, Japan). Height was measured to the nearest 0.1 cm using a wall mounted measuring tape. Leg length was measured during standing to the nearest 0.1 cm by means of a measuring tape from the ASIS of each leg to the medial malleolus.

Participants preparation

A total of 8 piezo-resistive sensors (Force Sensor Array (FSA), Verg Inc., Winnipeg, Canada) were placed under the plantar surface of each foot using double sided tape (figure 2.1). For each foot, the pressure sensors were positioned under the following anatomical sites: Lateral heel (Lheel), medial heel (Mheel), lateral mid-foot (Lmid), medial mid-foot (Mmid), first, third and fourth metatarsal heads (met1, met3, and met4), and hallux. Each flexible sensor was 1.27 cm long, 1.27 cm width, and 0.35 cm thick, and the cables connecting the sensors had a diameter of approximately 1mm. The pressure values in all sensors were recorded to a portable data logger at a sampling rate of 100 Hz. Pressure values were measured by the system up to 679 kPa (100 psi). FSA system calibration was conducted prior to each test session. The calibration was conducted according to the manufacturer's instructions using a pressure bladder (Tekscan West, Boston, USA).

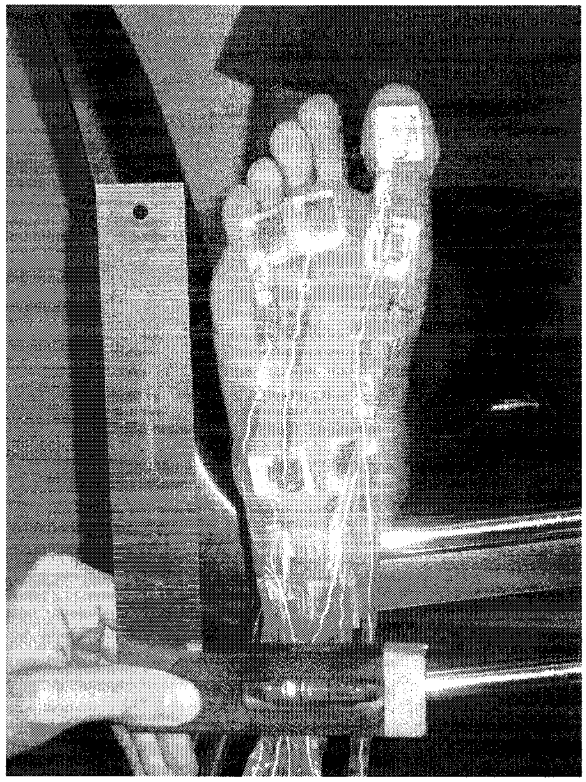

Figure 2.1: Sensor positioning under the plantar surface of the right foot

After the sensors were applied, the plantar surface of each foot was photographed with a digital camera Cannon Power shot S30 (Cannon Inc., Tokyo, Japan) for later determination of centre of pressure (COP) distribution. Then, a new pair of Hansen polyester-cotton ankle socks (Sara Lee Corp., Indianapolis, USA) was used to cover the feet. The sensors of each cluster that were not used were carefully wrapped together and taped to the calf using 3M hypo-allergenic transpore transparent plastic tape (3M Corp., Minnesota, USA). The cables from each cluster of sensors were secured to the legs by means of the above described transpore tape. Then, a pair of M, L or XL Under Armor Lycra-Spandex sport tights (Under Armor, Baltimore, USA) was worn by the participants in order to avoid the cables from moving during the test. Once the tights were on, runners were asked to wear a Running Room polyester fiber backpack (Infinity Sports Inc., Natick, USA) in which the FSA data logger was secured. Participants wore their own shoes. Preparation for the test took about 20 minutes (Figure 2.2).

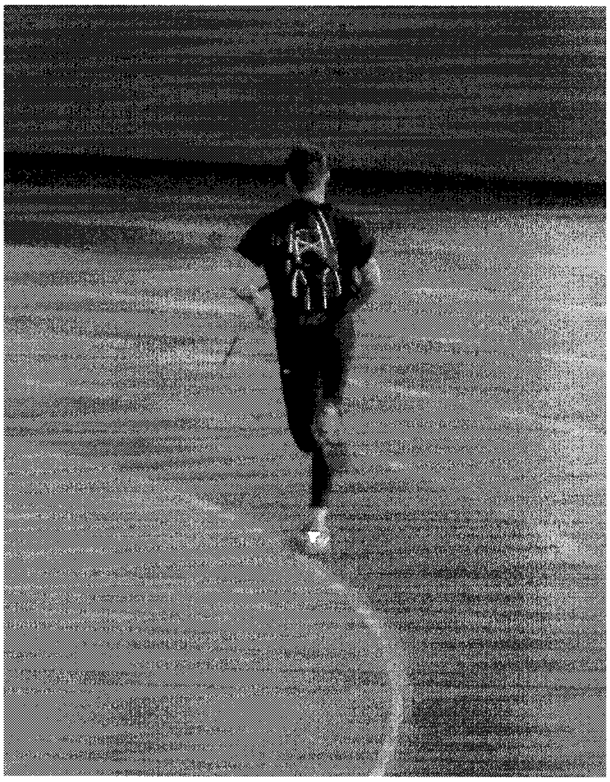

Figure 2.2: Runners' set-up during the evaluations

2.2.3 Testing protocol

All tests were conducted in the McGill Fieldhouse's on a 200 m, 6 lane natural rubber material Mondo Super X Performance Indoor track (Mondo America Inc., Laval, Canada). Prior to testing, baseline standing plantar foot pressure was recorded from each one of the sensors. Then, participants were asked to run at the speeds of 3.8 m/s (13.7 km/h) and 7.0 m/s (25.2 km/h) in three different conditions: Straight (1) and along the curve with either no side inclination (2) or an inclination of 19% (3) . For the 0% inclination, subjects ran on lane A, while for the 19% inclination, subjects ran on lane 1 (figure 2.3). By using two adjacent lanes, differences in radius would be minimized. The radius for lane A is 24.8 m and the radius for lane 1 is 25.7 m, which results in a radius difference of 3.5%.

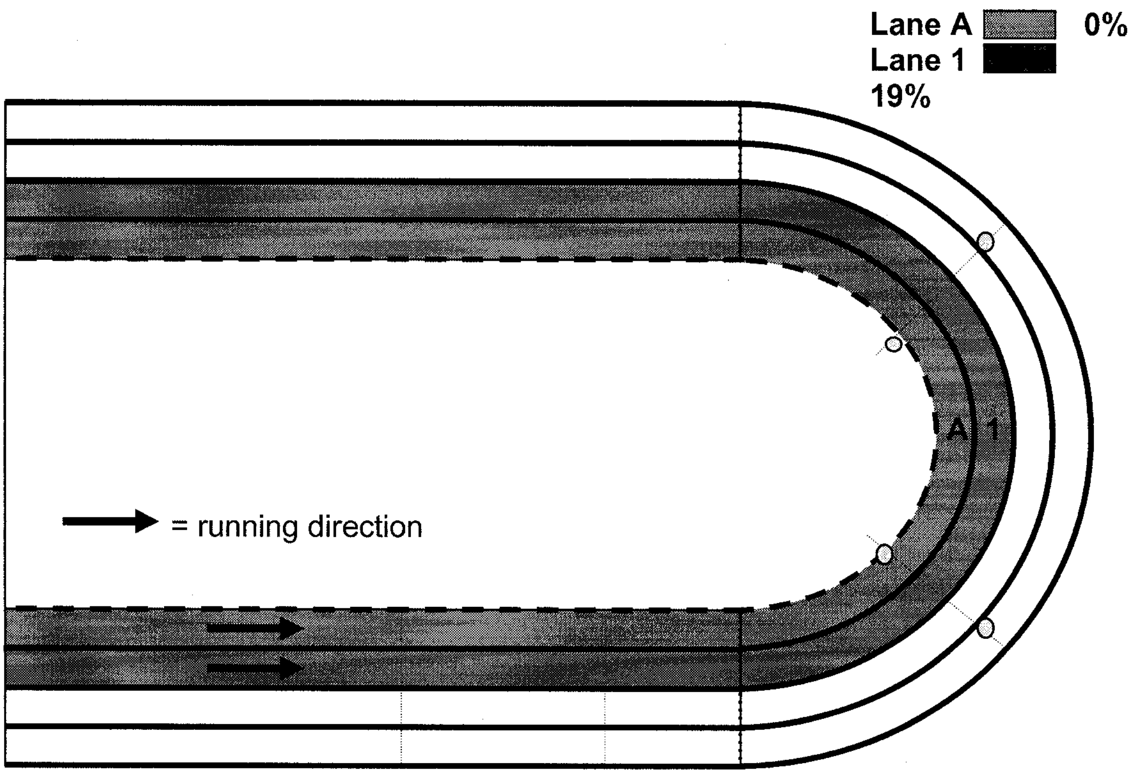

Figure 2.3: Plan of the indoor track were the evaluations took place

The running speed of 3.8 m/s corresponded to a jogging speed for these runners and was within typical range from previous studies (Milliron and Cavanagh, 1990). The speed of 7.0 m/s was calculated as the optimal speed for this indoor track. The optimal speed (v_{opt}) being the theoretical speed to both 1) enable the body lean in order to follow the curved path and 2) for the body to be oriented perpendicular to the track surface. Optimal speed was calculated as follows:

$$v_{opt} = \sqrt{\tan\theta * r * g}$$

where θ is the slope angle of the curve in radian, r is the radius of the curve and g is gravity. For lane 1, θ is 11°/57.3 (θ = 0.19 radians), r is 25.7 m and g is 9.81 m/s^2; thus, the optimal running speed was 7.0 m/s (Green, 1987).

Before the test began, runners warmed-up for a period of 10 min to become accustomed to the pace of the two running speeds. Trials at the different speeds (i.e. 3.8 and 7.0 m/s) and conditions (i.e. running 1) Straight; 2) On a curve with 0 % incline; 3) On a curve with 19 % incline) were randomized in order to exclude order effect from this study. For each speed and condition, two trials run over 20 m were performed. The 20 m running zone was delimited with white tape that was clearly visible for the participants. Subjects started running approximately 10 m before that zone in order to reach the desired speed before entering in the target area. Running speed was checked by means of a stop-watch and trials that exceeded ±0.5 m/s of the required speed were repeated. The FSA system was manually initialized before the trial and stopped at the end of that trial by means of a trigger connected to the data logger. After completion of a trial, data collected from the FSA was downloaded to a Toshiba Libretto 110 CT laptop computer (Toshiba Corp., Tokyo, Japan). The time needed to download the data was about 3 minutes and that time was used by the runners for recovery. Each session on the track consisted of 12 tests (i.e. 3 conditions x 2 speed x 2 trials) which took about 30 min to be completed.

When running on the curve, runners were video taped by means of a Sony DCR-TRV17 mini DV Camcorder (Sony Corp., Tokyo, Japan) in order to determine runners' trunk leaning at different speeds and cambers.

2.2.4 Data analysis

Data collected from the FSA system was treated and analyzed using MATLAB R12 (MathWorks Inc., Natick, USA). Three representative strides from each trial were partitioned from the full record of data for later statistical analysis. For each one of those strides three local events were determined: Peak value, initial contact and toe-off (Figure 2.4). Stance phase was determined as the time between initial contact and toe off.

The centre of pressure (COP) was derived from the following equations:

$$COP_X = \frac{\sum (\text{pressure} * \text{Xcoordinate})}{\sum \text{pressure}}$$

$$COP_Y = \frac{\sum (\text{pressure} * \text{Ycoordinate})}{\sum \text{pressure}}$$

Where: pressure = measured pressure from individual sensors (kPa);

Xcoordinate = medial-lateral position of an individual sensor (centimeters);

Ycoordinate = antero-posterior position of an individual sensor (centimeters);

COP_X = plantar foot surface COP in the X (medial-lateral) direction;

COP_Y = plantar foot surface COP in the Y (antero-posterior) direction.

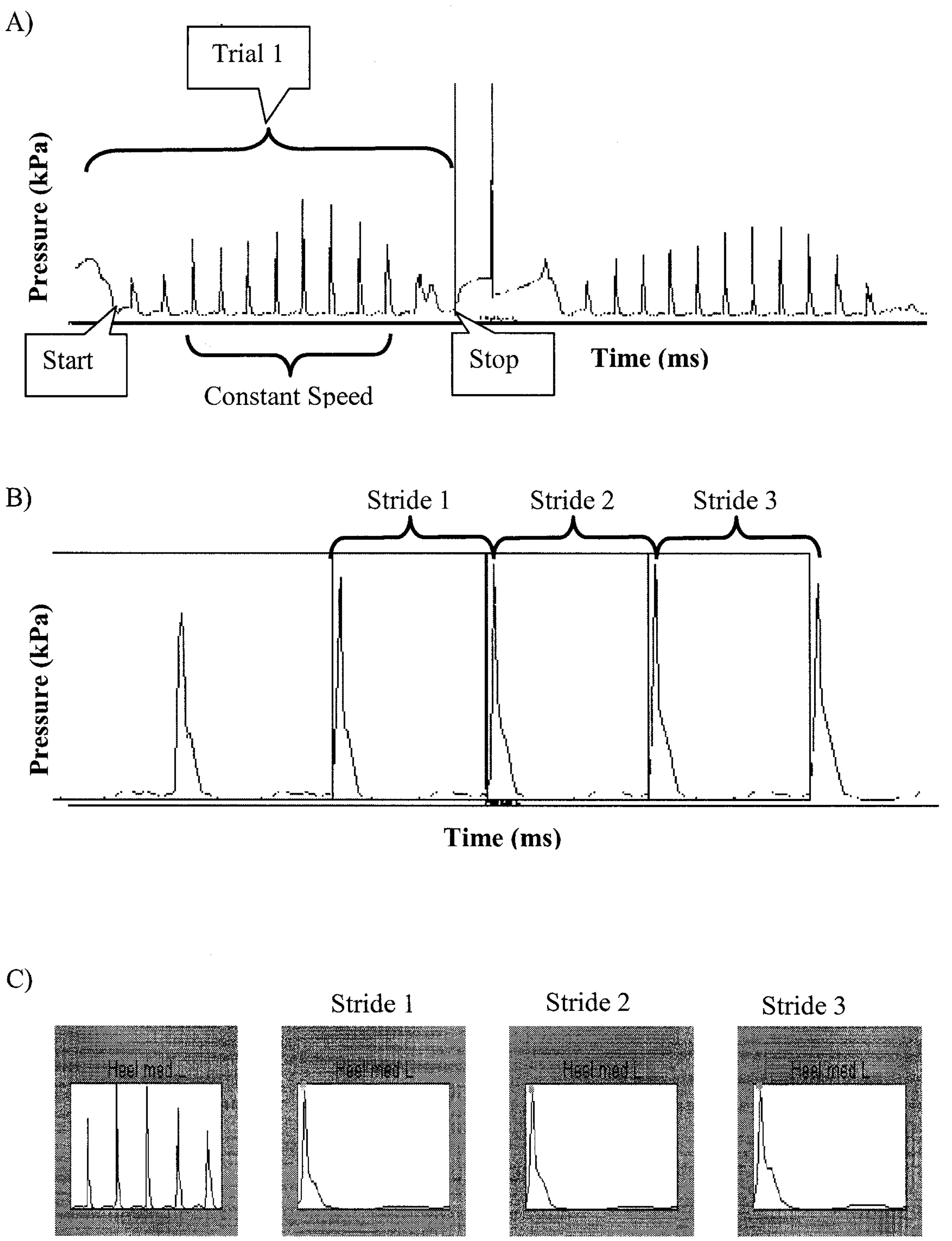

Figure 2.4: Graphical representation of the pressure-time data obtained with MATLAB R12. A) Pressure over time for two consecutive running trials; B) Three strides selected for statistical analysis; C) Peak pressure events for 3 consecutive strides for one sensor.

In order to locate COP pictures from the plantar surface of the foot were taken with the sensors positioned in the corresponding sites. A ruler was set alongside the foot for sizing. The pictures were downloaded from the digital camera, opened and treated using Microsoft Photo Editor 3.0.2.3 (Microsoft Corp., Mississauga, Canada). Each picture was rotated in such a way that middle point of the heel and the second toe were lined up. The line between these two points represented the foot's midline. An X,Y pixel coordinate was determined for each one of the sensors and for the point located in the middle of the heel. This last mentioned point was considered to be the new origin (0,0). Then the X,Y pixel coordinate of each sensor was compared to the origin, generating a set of new X,Y coordinates.

The new X,Y coordinates were then multiplied with their associated pressure values. The results from each one of the sensors of the plantar foot were summated and then divided by the mean pressure, resulting in the centre of pressure. This procedure was done using an Excel worksheet from Microsoft Office XP (Microsoft Corp., Mississauga, Canada).

Body's lean angle was determined when running on the curve. For each condition three trials were captured for each foot using Pinnacle Studio Version 8 (Pinnacle Systems Inc., Mountain View, USA). Body's absolute angle was calculated with HU-M-AN (HMA Technology Inc., King City, Canada) from the mid-line of the ankle to the middle of the head when the supporting foot was in the middle of the stance phase (white traces in Figure 2.5). The mean value from the three captured trials was used as the body's lean angle for that condition. Since the camera was placed on the exterior side of the curve a slight leaning from the perpendicular was expected. In order to diminish error in body's lean angle calculation the results obtained were corrected by using two fixed structures perpendicular to the floor (black traces in Figure 2.5). The values obtained for the absolute angle from those structures were averaged and subtracted or added to the body's lean angle (Figure 2.5).

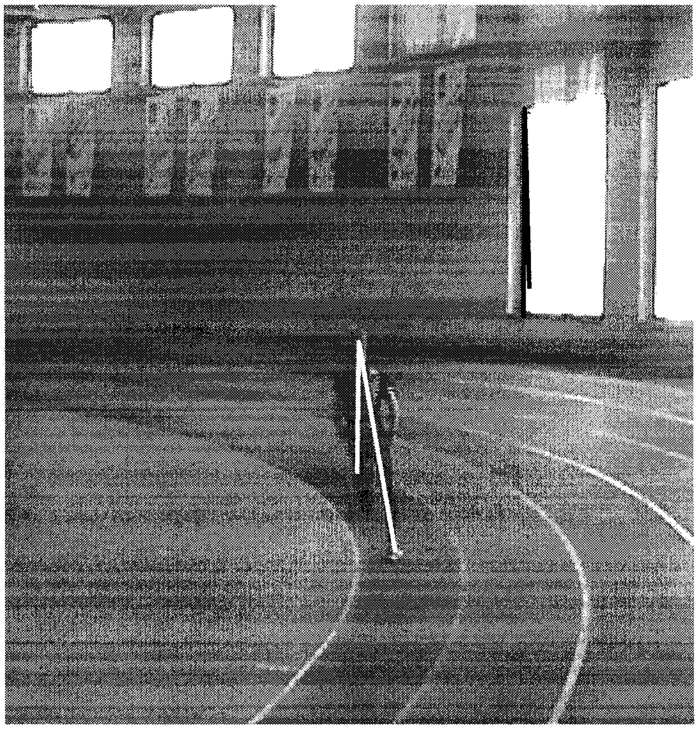

Figure 2.5: Representative figure for body's lean angle calculation

2.2.5 Statistical analysis

A within-subject repeated measures analysis of variance (ANOVA) was used to test the main effects and interactions of running condition, speed, and foot side (Statistica 5.1) of peak pressures for each sensor location. Differences were declared significant at a p<

0.05. Post hoc analysis (Tukey Honest Significant Difference Test) was performed on significant main effects.

2.3 RESULTS

Although few significant main effects were found for condition and foot side, running speed was seen to significantly affect peak pressure for most of the sensors (Table 2.1, Figure 2.6). Peak pressure under met4 was statistically different according to the running condition. Post hoc Tukey HSD test showed higher pressures on Curve 0 condition compared to Straight condition ($F_{2,24}= 3.61$; $p < 0.05$). Further, peak pressures under the right foot were significantly higher than under the left foot for met4 ($F_{1,12}= 13.23$; $p < 0.01$). No other significant main effects were observed for condition and foot side. At a speed of 7.0 m/s peak pressures were significantly higher than at a speed of 3.8 m/s for Mheel ($F_{1,12}= 7.19$; $p < 0.05$), Lheel ($F_{1,12}= 13.13$; $p < 0.01$), Lmid ($F_{1,12}= 9.46$; $p < 0.01$), met3 ($F_{1,12}= 9.56$; $p < 0.01$), met4 ($F_{1,12}= 6.58$; $p < 0.05$), and Hallux ($F_{1,12}= 5.62$; $p < 0.05$).

Table 2.1. Mean (SD) peak pressure values for each condition, speed, and foot side

	Condition	Pressure (kPa)	Speed (m/s)	Pressure (kPa)	Foot Side	Pressure (kPa)
Medial Heel	Curve 0	282.2 (167.7)	3.8	255.0 (165.7)	R	265.0 (163.0)
	Curve 19	293.4 (192.0)	7.0	312.3† (175.1)	L	302.3 (180.4)
	Straight	275.4 (160.2)				
Lateral Heel	Curve 0	262.0 (205.4)	3.8	263.1 (213.6)	R	337.7 (261.2)
	Curve 19	286.5 (203.6)	7.0	318.5† (220.6)	L	243.8 (152.0)
	Straight	323.8 (244.8)				
Medial Midfoot	Curve 0	125.7 (142.8)	3.8	102.2 (92.9)	R	170.1 (160.0)
	Curve 19	120.6 (146.2)	7.0	131.3 (151.9)	L	63.4 (26.1)
	Straight	103.9 (82.0)				
Lateral Midfoot	Curve 0	102.0 (76.7)	3.8	80.1 (46.1)	R	96.7 (66.3)
	Curve 19	94.5 (62.2)	7.0	111.1† (77.1)	L	94.5 (64.5)
	Straight	90.2 (56.5)				
Metatarsal 1	Curve 0	287.7 (151.6)	3.8	287.3 (166.3)	R	289.4 (140.1)
	Curve 19	304.3 (179.4)	7.0	303.0 (173.1)	L	300.8 (195.1)
	Straight	293.4 (180.4)				
Metatarsal 3	Curve 0	301.0 (170.2)	3.8	282.6 (161.3)	R	303.2 (171.7)
	Curve 19	318.7 (186.3)	7.0	332.4† (188.6)	L	311.8 (182.6)
	Straight	302.8 (177.7)				
Metatarsal 4	Curve 0	388.1* (227.4)	3.8	314.6 (198.2)	R	473.9 ‡ (230.2)
	Curve 19	358.9 (216.7)	7.0	378.9† (230.0)	L	219.6 (90.7)
	Straight	293.2* (199.1)				
Hallux	Curve 0	322.6 (209.3)	3.8	297.5 (208.1)	R	316.1 (208.1)
	Curve 19	318.0 (221.8)	7.0	350.9† (227.2)	L	332.3 (230.1)
	Straight	332.1 (231.5)				

* condition difference; † speed difference; ‡ side difference ($p<0.05$)

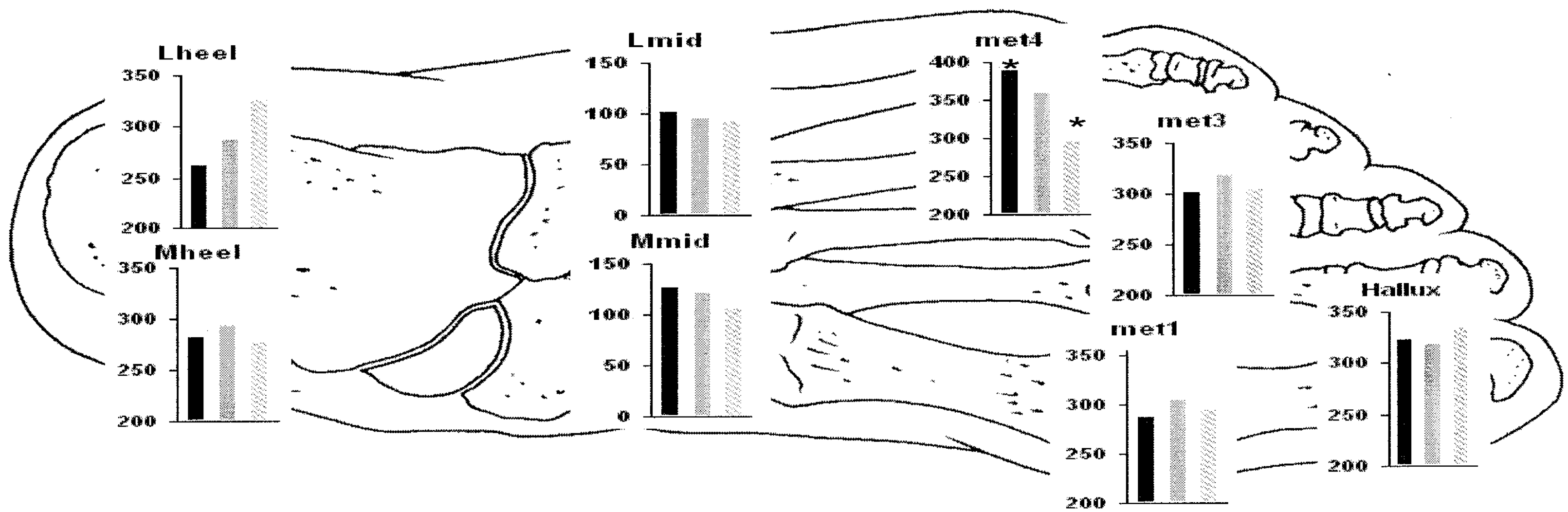

Figure 2.6. Mean peak pressure values (kPa) for each sensor in each condition. ■ Curve 0; ▒ Curve 19; ░ Straight

* Statistically significant (p<0.05) for condition

The foot side-speed interaction showed that for met4 sensor:

- peak pressures under the right foot at 3.8 m/s were significantly higher than peak pressures under the left foot at 3.8 m/s and 7.0 m/s;
- peak pressures under the right foot at 7.0 m/s were significantly higher than peak pressures under the left foot at 3.8 m/s and 7.0 m/s and peak pressures under the right foot at 3.8 m/s.

For met4 there was also a foot side-condition interaction that showed that:

- peak pressures under the right foot on C 0 condition were significantly higher than peak pressures under the right foot on Straight condition and the left foot on every condition.
- peak pressures under the right foot on C 19 condition were significantly higher than peak pressures under the left foot on every running condition.

Under the Mheel sensor the following condition-speed interactions were seen:

- peak pressures on C 19 condition at the speed of 7.0 m/s were significantly higher than peak pressures on C 19 and Straight conditions at the speed of 3.8 m/s.
- peak pressures on Straight condition at the speed of 7.0 m/s were significantly higher than peak pressures on Straight condition at the speed of 3.8 m/s.

There was a condition-speed-foot side interaction ($F_{2,20} = 4.38$; $p < 0.05$) for the Hallux sensor where the peak pressures under the left foot on Curve 0 at 7.0 m/s were significantly higher than the peak pressures under the left foot on Curve 0 at 3.8 m/s.

Table 2.2 depicts the mean values and main effect for Stance Time, Stance Time %, and Stride Rate. A main effect for speed was found for each one of these variables:

- Stance Time ($F_{1,12} = 53.06$; $p < 0.01$) and Stance Time % ($F_{1,12} = 15.38$; $p < 0.01$) were significantly shorter at 7.0 m/s than at 3.8 m/s.
- Stride Rate ($F_{1,12} = 31.24$; $p < 0.01$) was significantly higher at 7.0 m/s than at 3.8 m/s.

The leaning of the trunk was 15.0 (± 2.3) degrees and 5.8 (± 1.9) degrees at 7.0 m/s and 3.8 m/s respectively. This produced a significant main effect for speed (F 1,12= 323.37; $p < 0.01$). There was a significant main effect for foot side ($F_{1,12} = 19.29$; $p < 0.01$) with the body being more leaned when the left foot was in the support phase(left foot 11.8 vs. right foot 9.1). When comparing conditions, the absolute angle of the trunk was 10.7 (± 5.2) degrees and 10.2 (± 5.0) degrees for Curve 0 and Curve 19 conditions. These values were not statistically significant ($F_{1,12} = 1.93$; $p > 0.05$). Figure 2.7 represents the trunk leaning for each condition and speed.

The COP displacement showed a similar pattern in both the medial-lateral and heel to toe excursion regardless the running condition, speed, or foot side (Figures 2.8 and 2.9).

Table 2.2. Mean (SD) stance time, stance time percentage, and stride rate values for each condition, speed, and foot side

	Condition		Speed (m/s)		Foot Side	
Stance Time (s)	Curve 0	0.15 (0.03)	3.8	0.17 (0.03)	R	0.14 (0.04)
	Curve 19	0.15 (0.04)	7.0	0.14* (0.03)	L	0.16 (0.03)
	Straight	0.15 (0.04)				
Stance Time (%)	Curve 0	23.3 (4.0)	3.8	24.1 (5.1)	R	22.0 (5.9)
	Curve 19	23.0 (5.5)	7.0	21.8* (4.8)	L	23.9 (3.8)
	Straight	22.6 (5.7)				
Stride Rate (Strides · s^{-1})	Curve 0	1.55 (0.16)	3.8	1.45 (0.05)	R	1.53 (0.14)
	Curve 19	1.53 (0.13)	7.0	1.61* (0.15)	L	1.53 (0.14)
	Straight	1.51 (0.12)				

* Statistically significant for speed (p<0.05)

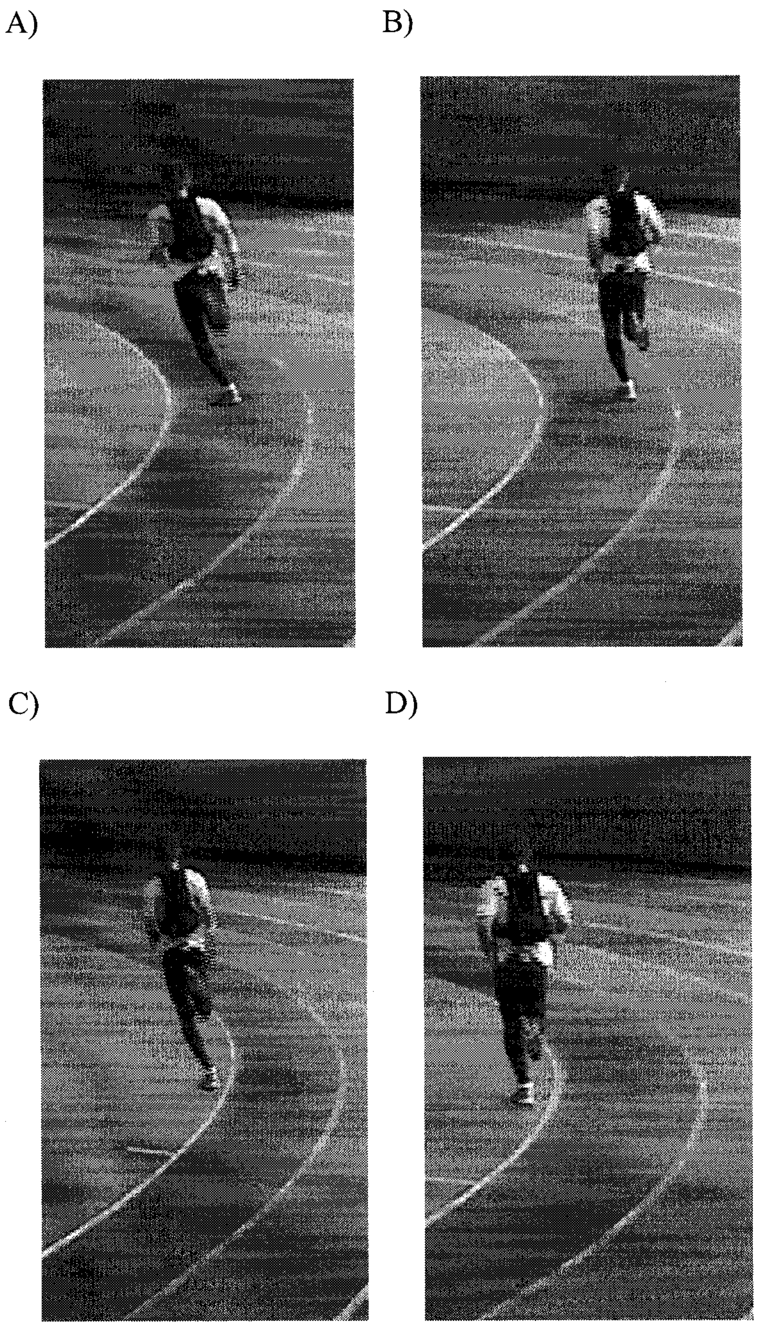

Figure 2.7. Trunk leaning for each condition and speed. A) Curve 19 at 7.0 m/s; B) Curve 19 at 3.8 m/s; C) Curve 0 at 7.0 m/s; D) Curve 0 at 3.8 m/s

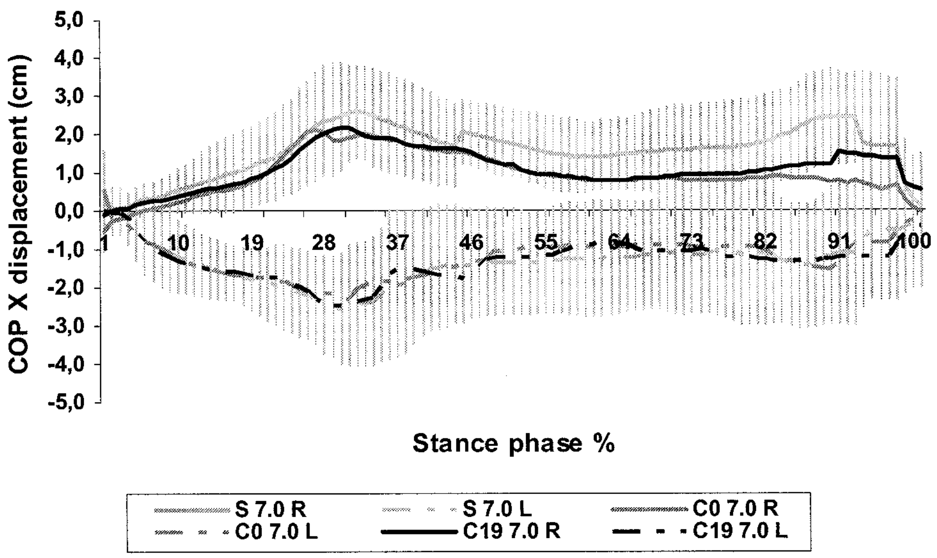

Figure 2.8. Medial-lateral centre of pressure displacement (group average) shows a similar pattern in the COP excursion regardless the running condition, speed, or foot side. Grey vertical bars represent standard error for Straight condition at 7.0 m/s

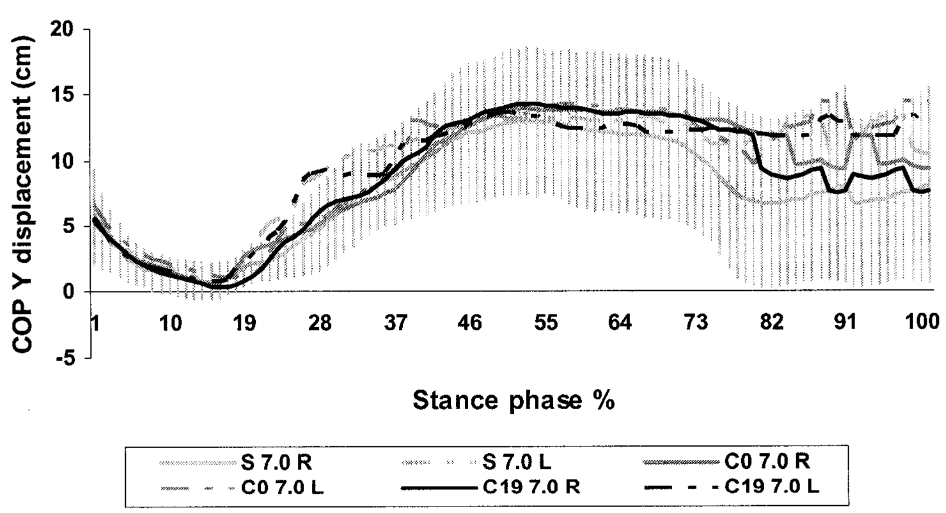

Figure 2.9. Heel to toe centre of pressure displacement (group average) shows a similar pattern in the COP excursion regardless the running condition, speed, or foot side. Grey vertical bars represent standard error for Straight condition at 7.0 m/s

2.4 DISCUSSION

2.4.1 Body leaning

For this study, two running speeds were selected: slow run at 3.8 m/s and a fast run (sprinting) of 7.0 m/s. As was expected, the lean angle of the body was greater when running at 7.0 m/s than at 3.8 m/s. You will recall that the 7.0 m/s speed represented the optimal speed (v_{opt}) for the indoor track. The v_{opt} theoretically enables both 1) the body to lean into the curve to produce the necessary centripetal force to follow the circumferential path, and 2) the feet to be oriented perpendicular (or in a typical flat surface alignment) to the track surface. According to the v_{opt} equation by Green (1987), running on lane A or Lane 1 of the curve at a speed of 7.0 m/s should produce a body leaning of 11 degrees. However, results from this investigation showed that the leaning of the body was 15.0 (± 2.3) degrees from the vertical (note this represented the average of both left and right leg support during mid-stance). The lean angles recorded were similar for both flat and banked conditions at v_{opt}. Despite that Green showed from his empirical testing a good agreement between theoretical and actual results, lean angle observed during current testing exceeded by approximately 4° the theoretical angle from the calculation. Differences in ground surface texture and rigidity may account for some of the discrepancy; that is, in Green's study, he used raised wooden platforms to vary bank angle in comparison to the track's solid concrete base. More importantly though, the lean angle disagreement may be explained in part by the method of its calculation. In the current study, lean angle was calculated from mid-head to mid-ankle of the supporting limb. In Green's study the lean angle with respect to the vertical was estimated from the body's center of mass (approximately mid-hip level) to a mid-foot ground contact position. Hence, the current method of calculation used to capture the "whole" body lean may have overestimated with respect to Green's method based on the different anatomical locations used. For reasons of accuracy the present study chose to use the actual location of the supporting limb's ankle (easily identified in the video image) as the definitive base of support as opposed to vague estimation of a mid-foot ground contact position. Interestingly enough, when comparing between right and left foot contacts during the support phase, significant differences were observed such that the lean angle was greater during left foot contact with the surface (by 2.7°). The effective step width taken and oscillation during alternate ipsi-contralateral was not considered in Green's equation. Finally, the use of the higher mid-head location rather than the mid-trunk may have accounted for greater lean angles as the torso and head may themselves lean in farther than the whole body's center of mass.

2.4.2 Stance time and stance rate

As expected, stance time decreased (and inversely stride rate increased with increased running speed. This is consistent with Greene (1985) and Ryan and Harrison (2004) findings of curve running. Furthermore, as running speed increased, step frequency increased and stance time decreased (Mann *et al.*, 1986). However, with the combination of slower speed, curved path and banking, it was conjectured that the symmetry of stance time and stance rate would be lost. In large part, it was speculated that this combination would induce a functional gait asymmetry similar to that observed in individuals with leg length discrepancy (Gurney, 2002). Further, it was thought that symmetry could be restored by running the banked curve at the higher speed, v_{opt}. However, neither condition of flat curve or banked curve was found to affect the ratio left-right stance time and stride rate. Thus, it is evident that both the degree of curvature ($\approx$ 25 m) and the degree of banking (11°) were not severe enough to perturb running gait. Potentially, the smaller step width (e.g. ≈ 10 cm) associated with running could have compensated for the different running speeds and conditions; that is, with a narrow ipsi-contralateral step-to-step distance, the unleveled surface would be minimal.

2.4.3 Plantar foot pressure

As expected, there was a main effect of speed such that peak pressures were significantly higher at 7.0 m/s than 3.8 m/s including the Mheel ($p<0.05$), Lheel ($p<0.01$), Lmid ($p<0.01$), met3 ($p<0.01$), met4 ($p<0.05$), and Hallux ($p<0.05$). Mmid and met1 showed similar trends with higher peak pressures at 7.0 m/s. These observations are supported by prior studies; for instance, Nigg *et al.* (1987) and Belli *et al.* (2001) concluded that increasing running speed increased impact forces and impact loading rate. As well, Van Mechelen (1992) showed that at higher speeds, impact forces were greater. The present findings are consistent with these observations.

Surprisingly, few significant differences in the peak pressures were observed for most of the sensors between curved and straight running or between flat and banked 19% (11 degrees) curves, as well as between foot side. The only significant main effects were with respect to met4 location such that greater peak pressures were found (1) for the flat curve in comparison to straight running ($p< 0.01$), and (2) for the right foot in comparison to the left ($p<0.01$). Though significant interactions were found for the Mheel and Hallux, these reflected primarily speed differences. Therefore, in general, the null hypotheses were found to be true; that is, dynamic plantar foot pressures were not perturbed by the curved or banked conditions; further, no asymmetry between left and right feet was observed.

This contrasts with expectations inferred from prior studies (Beukeboom *et al.* 2000; Gehlsen *et al.* 1989; Greene 1987; Sussman *et al.* 2001). Given that meticulous attention was taken with the instrumentation (i.e. involving pre- and post-testing calibration and verification of pressure signal for each subject), there is confidence in the accuracy of the pressure measurement obtained.

So what can explain this apparent contradiction? To accommodate for the acute medial-lateral body lean during curved running and in particular the acute foot-to-ground contact angle produced in the frontal plane, the fore-foot varus-valgus orientations during stance must be altered (i.e. more fore-foot varus on the right and valgus on the left) to provide a regular base of support for locomotion. The greater pressures under metatarsal four of the right foot during flat curve running are in part evidence for this. Potentially, by this adjustment foot loading distribution (as represented by our pressure measures) is maintained such that the established motor pattern is not disturbed. Indeed, this is suggested by the similar COP patterns in both the medial-lateral and heel-to-toe excursions regardless of the running condition, speed, or foot side.

According to Nigg (2001), the patterns of human locomotion are central nervous system (CNS) sensory input dependent. Feedback from the foot pressure provides constant information about the loading, joint kinematics, and the pressure distribution under the foot. Since the foot is usually the only part of the body in contact with the ground, input from cutaneous receptors in the foot must play an important role in the regulation of the gait cycle. In Addition, Nurse and Nigg (2001) and Taylor *et al.* (2003) demonstrated that plantar pressure changes occur when altering sensation on the plantar surface of the foot. Prior to this study, it was thought that the COP would shift towards the lateral side of the left foot and the medial side of the right foot when running on the curve. If the foot were rigid then this may have been true; however, the dynamic malleable properties of the foot would appear to nullify this outcome and permit similar plantar pressure distribution and thus similar cutaneous inputs necessary for gait.

Alternatively, the full extent of the differences in pressure may not have been captured. Given that the sensors used could only measure the normal or perpendicular pressures, it is possible that transverse (shear) pressure differences existed but were not detected. Though shear sensors for in-shoe testing are not technologically available at this point in time, potentially other test environments could be constructed to capture this directional component (e.g. force plate mounted at varying bank angles within a curved platform). If this is possible then the vectorial orientation of the foot loading would be different; that is, the combined normal and shear loads may result in greater varus loading on the left (inside) foot and greater valgus loading on the right (outside) foot during flat curve running. The banked curve could offset this altered foot loading but only for one specific speed, that is, optimal speed. Hence the observed invertor-evertor muscle imbalance between limb sides as observed by Beukeboom *et al.* (2000), thought to result from habitually training and competing in the counterclockwise running direction, may well still be due to asymmetric varus and valgus loadings (about the anterior-posterior supination-pronation axes) of the feet (Green, 1987). Further, this asymmetric loading may be exasperated on indoor unbanked track (with smaller radius of curvature) thereby corresponding to the noted greater injury rates for indoor versus outdoor seasons.

2.5 CONCLUSION

In conclusion, this study demonstrated that even though running speed significantly affected peak pressures, stance time and stride rate, these parameters were not affected by the running conditions (curves and slopes), foot side, or interactions between foot side and conditions. In addition, COP patterns were not affected by either running speed or conditions. Several factors such as the large standard deviations obtained about the mean estimates, the limited subject cohort to elite runners whom actively train on the test track conditions, and the fact that only normal (perpendicular) pressures were measured may have influenced these observations. Further studies should look at other surface stress components such as shear force as well as examine more proximal body segment kinematics to better understand the mechanics of running on curved and sloped track surfaces.

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Chapter 3: Conclusion and Summary

The Effect of Banked-Curves on Running Mechanics: Plantar Foot Pressures

In summary, though running speed was observed to significantly affect peak pressures, stance time and stride rate, these parameters were not affected by the running conditions (curves and slopes) or interactions between foot side and conditions. Furthermore, COP patterns were not affected by either running speed or conditions. This conclusion is supported indirectly from similar non-significant findings in a parallel study using electromyographic measures of lower limb muscles under the same research design (conducted by Jen Gow, Laboratory of Biomechanics, Department of Kinesiology and Physical Education, McGill University).

Several factors may have influenced these observations. Foremost, given the large standard deviations obtained about the mean estimates, the ability to discriminate significant differences between measures was much reduced. Hence, future study should contain a larger sample size to improve statistical power as well as consider standardizing footwear to reduce confounding factors. Another aspect that may have influenced the results was the delimited subject cohort to elite runners whom actively train on the test track conditions. It may be advantageous to use subjects non-habituated to the test conditions to eliminate potential long-term running adaptations made by the elites.

Regarding the method of pressure measurement, alternate devices may be considered that could measure full plantar surface contact (Emed-Pedar, Novel Electronics Inc., MN, USA) with higher spatial resolution and sampling as a means to improve measurement accuracy and avoid signal alias, respectively. Furthermore, it may well be that the normal (as in perpendicular) pressures per se were not affected by these conditions; however, other surface stress components such as shear force should be examined as logically these may account for the transverse plane direction changes. In addition, further studies should examine more proximal body segment kinematics to better understand the mechanics of running on curved and sloped track surfaces.

Chapter 4: References Literature Review

The Effect of Banked-Curves on Running Mechanics: Plantar Foot Pressures

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Appendices

Appendix A: Information and Consent Form

Information and Consent document

The Effect of Banked-Curves on Knee and Ankle Mechanics: EMG and Plantar Foot Pressure

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Introduction

Extensive research has been conducted to evaluate running patterns on levelled surfaces, but little is known about how we adapt to inclined or irregular terrains and its potential association to injury. Furthermore, the rate of running injuries has been shown to be different in indoor versus outdoor track running, so there may be a relationship with the configuration of indoor tracks and the prevalence of injuries.

Purpose of the Study

The purpose of this study is to compare the muscle activity of the knee and ankle and center of pressure when running on an indoor track with different curve inclinations.

Your participation in this study involves:

1. Providing informed consent prior to the experimental session.
2. Completing a brief medical history and training habits questionnaire.
3. Perform the following tasks during the experimental session:
    a. Run at 3.8 m/s, and sprint at 7.0 m/s on a 10-meter distance. Once on a straightaway, levelled curve, and an inclined curve.
    b. Each condition will be performed twice, once for EMG measurements and once for plantar foot pressure. Three trials will be performed for each condition. A total of 36 trials will be required.
    c. The surface electrodes will be held in place on the muscles at the lower leg with adhesive tape and the wires will be secured to the body using 3M surgical tape and/or athletic tape.
    d. Pressure sensors will be taped underneath the foot at specific points to measure pressure during contact with the ground.
    e. Foot switches will be taped on the insoles of your shoes to measure foot contact on the ground.

f. Two reflective markers will be secured on your back and neck to measure trunk inclination.

g. During the running trials, a backpack will be secured to carry the data acquisition box.

Potential Risks

This research project involves no greater risks than present in your everyday life, mainly because you are already comfortable with the testing environment and the surface electrodes will not interfere with your normal running technique. The track will be free from any obstacles and the number of trials performed should not take you to exhaustion.

Benefits

There are no personal benefits to be derived from participating in this study. The information that we will obtain will help us increase our understanding of the effects of curved inclination on running gait patterns.

Subject Rights

Your participation in this study is voluntary. You are free to withdraw from the study at anytime and for any reason without prejudice with regards to your training involvement with the McGill Track and Field Team. You are also free to ask questions to the experimenter at any time.

Privacy and Confidentiality

The confidentiality of your results will be maintained by substituting your name by a number assigned to this particular research project. This list of subject names and coordinates will be locked in the physical education biomechanics laboratory of McGill University and only the main investigator and his supervisors will have access to this list.

Contacts

In the event of adverse effects or if you need additional information, you can contact the investigators' supervisor:

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CONSENT

I, ______________________________________, AGREE TO VOLUNTARILY PARTICIPATE IN THE STUDY DESCRIBED ABOVE ABOUT THE EFFECT OF BANKED-CURVES ON KNEE AND ANKLE MECHANICS: EMG AND PLANTAR FOOT PRESSURE.

I have received and read a detailed description of the experimental protocol. I am fully satisfied with the explanations that were given to me regarding the nature of this research project, including the potential risks and discomforts related to my participation in this study.

I am aware that I have the right to withdraw my consent and discontinue my participation at any time without any prejudices.

Signatures

SUBJECT

____
(signature)

WITNESS

______________________________________ ______________________________________
(signature) (print name)

Date: ___________________________

Appendix B: Subject Information and Medical History Questionnaire

Subject Information and Medical History Questionnaire

SUBJECT IDENTIFICATION

Name: ____ ID code: _________

Age: ________ years Sex: M or F Telephone number: ___________

MEDICAL HISTORY

1. Have you ever been affected by joint disorders? Yes or No

   If yes, specify ____

2. Have you recently complained of pain in the lower limbs, hips, or back? Yes or No

   If yes, specify ____

3. Are you currently taking any medication? Yes or No

   If yes, specify ____

4. Do you have other medical conditions that should be mentioned? Yes or No

   If yes, specify ____

Training history

1. How many training sessions do you perform each week? _________
2. For how long do you normally run?_________ hours
3. On average, in a training session, what is the distance that you run? _______ km
4. How many years have you been running for?_______ years
5. How many years have you been running indoors?______ years

Anthropometric measurements

Height: _________ m Weight: _________ kg

Leg length: Right ________ cm Left: ___________ cm

Appendix D: COP individual data

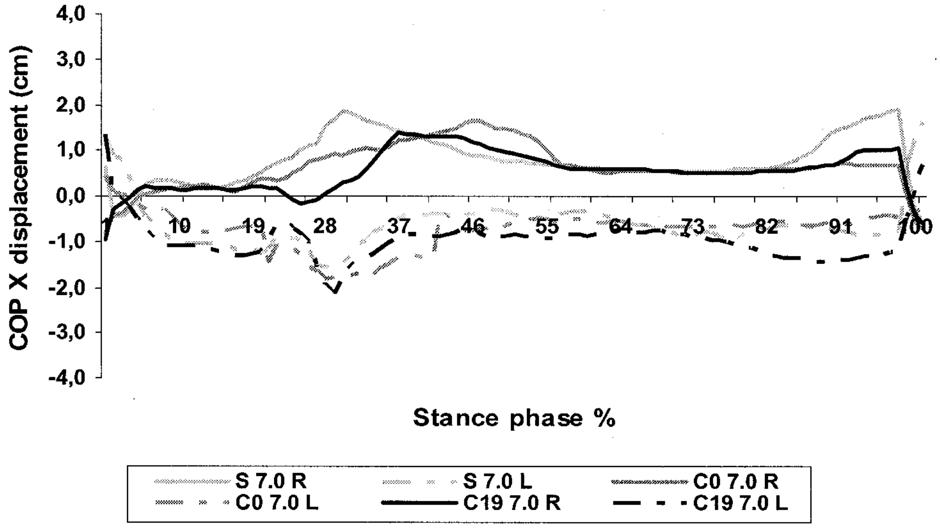

Figure x. Medial-lateral centre of pressure displacement for one representative subject shows the COP excursion in different running conditions for both foot sides at 7.0 m/s

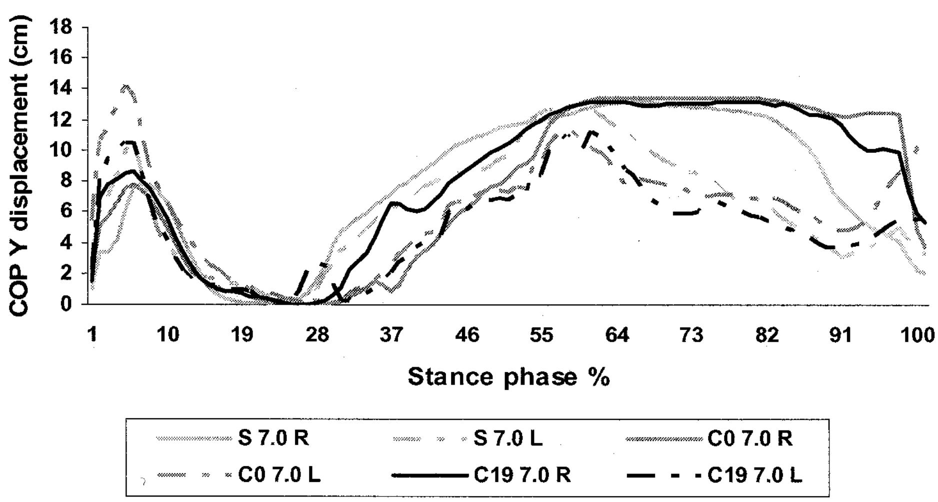

Figure x. Heel to toe centre of pressure displacement for one representative subject shows the COP excursion in different running conditions for both foot sides at 7.0 m/s. In this instance, the left and right COP excursion was similar for the first 60% of stance but varied towards push-off

Appendix E: ANOVA tables data

Table E.1. Main effects and interactions p values from ANOVA table for each sensor location

	Mheel	Lheel	Mmid	Lmid	met1	met3	met4	Hallux
	p value	p value	p value	p value	p value	p value	p value	p value
Foot side	0.6752	0.4156	0.0783	0.9460	0.9093	0.9308	**0.0034**	0.9006
Condition	0.6510	0.1781	0.4998	0.3299	0.7372	0.3374	**0.0426**	0.8718
Speed	**0.0200**	**0.0035**	0.1562	**0.0096**	0.5396	**0.0093**	**0.0247**	**0.0392**
Foot side * Condition	0.6939	0.9541	0.3687	0.1604	0.3853	0.3797	**0.0184**	0.5366
Foot side * Speed	0.4753	0.8838	0.2401	0.7583	0.8453	0.3344	**0.0175**	0.7224
Condition * Speed	**0.0153**	0.0552	0.7665	0.4221	0.5369	0.4365	0.1934	0.0548
Foot side * Condition * Speed	0.3851	0.7283	0.6389	0.5595	0.2294	0.5588	0.7303	**0.0265**

Mheel, medial heel; Lheel, lateral heel; Mmid, medial midfoot; Lmid, lateral midfoot; met1, metatarsal 1; met3, metatarsal 3; met4, metatarsal 4.

Table E.2. Main effects and interactions F values from ANOVA table for each sensor location

		Mheel	Lheel	Mmid	Lmid	met1	met3	met4	Hallux
	df	F value	F value	F value	F value	F value	F value	F value	F value
Foot side	12	0.1845	0.7110	3.7041	0.0048	0.0137	0.0079	**13.2258**	0.0164
Condition	24	0.4370	1.8558	0.7140	1.1619	0.3096	1.1371	**3.6112**	0.1382
Speed	12	**7.1943**	**13.1310**	2.2884	**9.4587**	0.4033	**9.5564**	**6.5834**	**5.6198**
Foot side * Condition	24	0.3711	0.0471	1.0405	1.9766	1.0007	1.0085	**4.7446**	0.6422
Foot side * Speed	12	0.5432	0.0223	1.5278	0.0991	0.0401	1.0115	**7.5847**	0.1335
Condition * Speed	24	**5.0043**	3.2772	0.2689	0.8941	0.6418	0.8582	1.7608	3.3695
Foot side * Condition * Speed	24	0.9934	0.3213	0.4564	0.5949	1.5860	0.5963	0.3185	**4.3750**

df, degrees of freedom; Mheel, medial heel; Lheel, lateral heel; Mmid, medial midfoot; Lmid, lateral midfoot; met1, metatarsal 1; met3, metatarsal 3; met4, metatarsal 4.

Table E.3. Main effects and interactions p values from ANOVA table for stance time, stance time percentage, stride time, stride rate, and trunk leaning

	ST	ST %	St time	St rate	Trunk leaning
	p value	p value	p value	p value	p value
Foot side	0.3583	0.4393	0.9503	0.9326	**0.0009**
Condition	0.9988	0.7711	0.1350	0.0998	0.1903
Speed	**0.0000**	**0.0020**	**0.0000**	**0.0001**	**0.0000**
Foot side * Condition	0.5680	0.5712	0.7437	0.7454	0.4008
Foot side * Speed	0.2119	0.3410	0.7606	0.7991	0.4600
Condition * Speed	0.6314	0.6152	0.5107	0.5649	0.1430
Foot side * Condition * Speed	0.4163	0.4277	0.7237	0.7185	0.2255

ST, stance time; ST %, stance time percentage of the gait cycle; St time, stride time; St rate, stride rate.

Table E.4. Main effects and interactions F values from ANOVA table for stance time, stance time percentage, stride time, stride rate, and trunk leaning

		ST	ST %	St time	St rate	Trunk leaning
	df	F value	F value	F value	F value	F value
Foot side	12	0.9126	0.6400	0.0041	0.0075	**19.2857**
Condition	24	0.0012	0.2628	2.1792	2.5412	1.9275
Speed	12	**53.0634**	**15.3753**	**41.2735**	**31.2386**	**323.3704**
Foot side * Condition	24	0.5791	0.5733	0.2999	0.2974	0.7589
Foot side * Speed	12	1.7391	0.9831	0.0972	0.0677	0.5826
Condition * Speed	24	0.4687	0.4959	0.6912	0.5849	2.4571
Foot side * Condition * Speed	24	0.9092	0.8800	0.3277	0.3352	1.6330

df, degrees of freedom; ST, stance time; ST %, stance time percentage of the gait cycle; St time, stride time; St rate, stride rate.

Table E.5. Post hoc test for the condition by speed interaction in the medial-heel sensor

	C0 3.8	C0 7.0	C19 3.8	C19 7.0	S 3.8	S 7.0
Mean pressure	289.2	275.3	244.3	342.4	231.5	319.2
	p value	p value	p value	p value	p value	p value
C0 3.8	-	0.9956	0.5926	0.4131	0.3278	0.8833
C0 7.0	0.9956	-	0.8689	0.1867	0.6178	0.6145
C19 3.8	0.5926	0.8689	-	**0.0182**	0.9970	0.1110
C19 7.0	0.4131	0.1867	**0.0182**	-	**0.0062**	0.9568
S 3.8	0.3278	0.6178	0.9970	**0.0062**	-	**0.0425**
S 7.0	0.8833	0.6145	0.1110	0.9568	**0.0425**	-

C0 3.8, curve 0 condition at the speed of 3.8 m/s; C0 7.0 curve 0 condition at the speed of 7.0 m/s; C19 3.8, curve 19 condition at the speed of 3.8 m/s; C19 7.0, curve 19 condition at the speed of 7.0 m/s; S 3.8, Straight condition at the speed of 3.8 m/s; S 7.0, Straight condition at the speed of 7.0 m/s.

Table E.6. Post hoc test for the main effect condition in the metatarsal 4 sensor

	Curve 0	Curve 19	Straight
Mean pressure	388.1	358.9	293.2
	p value	p value	p value
Curve 0	-	0.7028	**0.0383**
Curve 19	0.7028	-	0.1856
Straight	**0.0383**	0.1856	-

Table E.7. Post hoc test for the foot side by condition interaction in the metatarsal 4 sensor

	R C0	R C19	R S	L C0	L C19	L S
Mean pressure	556.0	508.7	356.9	220.2	209.2	229.6
	p value	p value	p value	p value	p value	p value
R C0	-	0.9361	**0.0081**	**0.0001**	**0.0001**	**0.0001**
R C19	0.9361	-	0.0651	**0.0002**	**0.0002**	**0.0003**
R S	**0.0081**	0.0651	-	0.1177	0.0765	0.1663
L C0	**0.0001**	**0.0002**	0.1177	-	0.9999	1.0000
L C19	**0.0001**	**0.0002**	0.0765	0.9999	-	0.9986
L S	**0.0001**	**0.0003**	0.1663	1.0000	0.9986	-

R C0, right foot running on Curve 0 condition; R C19, right foot running on Curve 19 condition; R S, right foot running on Straight condition; L C0, left foot running on Curve 0 condition; L C19, left foot running on Curve 19 condition; L S, left foot running on Straight condition.

Table E.8. Post hoc test for the foot side by speed interaction in the metatarsal 4 sensor

	R 3.8	R 7.0	L 3.8	L 7.0
Mean pressure	407.2	540.5	222.0	217.3
	p value	p value	p value	p value
R 3.8	-	**0.0126**	**0.0012**	**0.0010**
R 7.0	**0.0126**	-	**0.0002**	**0.0002**
L 3.8	**0.0012**	**0.0002**	-	0.9992
L 7.0	**0.0010**	**0.0002**	0.9992	-

R 3.8, right foot running at the speed of 3.8 m/s; R 7.0, right foot running at the speed of 7.0 m/s; L 3.8, left foot running at the speed of 3.8 m/s; L 7.0, left foot running at the speed of 7.0 m/s.

Table E.9. Post hoc test for the foot side by running condition by speed interaction in the Hallux sensor

	R C0 3.8	R C0 7.0	R C19 3.8	R C19 7.0	R S 3.8	R S 7.0	L C0 3.8	L C0 7.0	L C19 3.8	L C19 7.0	L S 3.8	L S 7.0
Mean pressure	320.2	344.6	259.3	344.8	301.1	326.9	223.7	401.8	315.3	352.6	365.4	335.0
	p value	p value	p value	p value	p value	p value	p value	p value	p value	p value	p value	p value
R C0 3.8	-	1.0000	0.9197	1.0000	1.0000	1.0000	0.4480	0.6720	1.0000	0.9994	0.9895	1.0000
R C0 7.0	1.0000	-	0.6170	1.0000	0.9923	1.0000	0.1783	0.9448	0.9998	1.0000	1.0000	1.0000
R C19 3.8	0.9197	0.6170	-	0.6141	0.9944	0.8573	0.9986	0.0654	0.9525	0.4955	0.3223	0.7576
R C19 7.0	1.0000	1.0000	0.6141	-	0.9920	1.0000	0.1769	0.9459	0.9997	1.0000	1.0000	1.0000
R S 3.8	1.0000	0.9923	0.9944	0.9920	-	0.9999	0.7336	0.3900	1.0000	0.9726	0.8902	0.9991
R S 7.0	1.0000	1.0000	0.8573	1.0000	0.9999	-	0.3577	0.7681	1.0000	0.9999	0.9972	1.0000
L C0 3.8	0.4480	0.1783	0.9986	0.1769	0.7336	0.3577	-	**0.0105**	0.5211	0.1248	0.0679	0.2647
L C0 7.0	0.6720	0.9448	0.0654	0.9459	0.3900	0.7681	**0.0105**	-	0.5966	0.9802	0.9983	0.8655
L C19 3.8	1.0000	0.9998	0.9525	0.9997	1.0000	1.0000	0.5211	0.5966	-	0.9978	0.9773	1.0000
L C19 7.0	0.9994	1.0000	0.4955	1.0000	0.9726	0.9999	0.1248	0.9802	0.9978	-	1.0000	1.0000
L S 3.8	0.9895	1.0000	0.3223	1.0000	0.8902	0.9972	0.0679	0.9983	0.9773	1.0000	-	0.9997
L S 7.0	1.0000	1.0000	0.7576	1.0000	0.9991	1.0000	0.2647	0.8655	1.0000	1.0000	0.9997	-

R C0 3.8, right foot running on curve 0 condition at the speed of 3.8 m/s; R C0 7.0, right foot running on curve 0 condition at the speed of 7.0 m/s; R C19 3.8, right foot running on curve 19 condition at the speed of 3.8 m/s; R C19 7.0, right foot running on curve 19 condition at the speed of 7.0 m/s; R S 3.8, right foot running on Straight condition at the speed of 3.8 m/s; R S 7.0, right foot running on Straight condition at the speed of 7.0 m/s; L C0 3.8, left foot running on curve 0 condition at the speed of 3.8 m/s; L C0 7.0, left foot running on curve 0 condition at the speed of 7.0 m/s; L C19 3.8, left foot running on curve 19 condition at the speed of 3.8 m/s; L C19 7.0, left foot running on curve 19 condition at the speed of 7.0 m/s; L S 3.8, left foot running on Straight condition at the speed of 3.8 m/s; L S 7.0, left foot running on Straight condition at the speed of 7.0 m/s.