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Masterclass

A comparison of the spatiotemporal parameters, kinematics, and biomechanics between shod, unshod, and minimally supported running as compared to walking

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A B S T R A C T

Recreational running has many proven benefits which include increased cardiovascular, physical and mental health. It is no surprise that Running USA reported over 10 million individuals completed running road races in 2009 not to mention recreational joggers who do not wish to compete in organized events. Unfortunately there are numerous risks associated with running, the most common being musculoskeletal injuries attributed to incorrect shoe choice, training errors and excessive shoe wear or other biomechanical factors associated with ground reaction forces. Approximately 65% of chronic injuries in distance runners are related to routine high mileage, rapid increases in mileage, increased intensity, hills or irregular surface running, and surface firmness. Humans have been running barefooted or wearing minimally supportive footwear such as moccasins or sandals since the beginning of time while modernized running shoes were not invented until the 1970s. However, the current trend is that many runners are moving back to barefoot running or running in "minimal" shoes. The goal of this masterclass article is to examine the similarities and differences between shod and unshod (barefoot or minimally supportive running shoes) runners by examining spatiotemporal parameters, energetics, and biomechanics. These running parameters will be compared and contrasted with walking. The most obvious difference between the walking and running gait cycle is the elimination of the double limb support phase of walking gait in exchange for a float (no limb support) phase. The biggest difference between barefoot and shod runners is at the initial contact phase of gait where the barefoot and minimally supported runner initiates contact with their forefoot or midfoot instead of the rearfoot. As movement science experts, physical therapists are often called upon to assess the gait of a running athlete, their choice of footwear, and training regime. With a clearer understanding of running and its complexities, the physical therapist will be able to better identify faults and create informed treatment plans while rehabilitating patients who are experiencing musculoskeletal injuries due to running.

1. Introduction

The benefits attributed to running include cardiovascular and mental health, stress reduction, and enjoyment (Dugan & Bhat, 2005; Hafstad et al., 2009; Haskell et al., 1993; McWhorter et al., 2003). However, there are numerous risks associated with running as well (Bennell & Crossley, 1996). The most common risk factors related to running are musculoskeletal injuries which are often attributed to incorrect shoe choice, shoe wear, training errors, or other biomechanical factors associated with ground reaction forces. The typical runner makes choices related to shoe selection often based on personal preference, trend information, or a well intentioned running shoe store employee. Early man has been running barefoot or wearing minimally supportive footwear such as moccasins or sandals since the beginning of time (Bramble & Lieberman, 2004) while modernized running shoes were not invented until the 1970s (Lieberman et al., 2010). Since, Nike first revolutionized the running shoe in 1979; running shoes have gone through a major evolution with the most recent trend returning runners back to forefoot running with minimally supported running shoes (e.g., Vibram Fivefingers®, New Balance Minimus®, Nike Free®). Nike, re-revolutionized running shoes in 2001 with the Nike Free® "minimal" running shoe which helped spark the "minimalist movement". It is too early to accurately predict what impact this running trend will have on musculoskeletal related injuries; however, barefoot activities are natural to our bodies.

The goal of this master class article is to examine similarities and differences between shod and unshod (barefoot and minimally supported shoes) runners including spatiotemporal parameters, biomechanics, and running-related common musculoskeletal

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injuries. These running parameters will also be compared and contrasted to walking. The authors will use scientific recommendations and research articles to help inform the health care clinician to better empower them to make data informed treatment and running recommendations to the runners who sustain or hope to prevent musculoskeletal injuries.

2. Walking versus running

Human bipedal walking includes eight (8) phases of gait (Fig. 1). During all eight phases of walking gait at least one foot is in physical contact with the ground. During the initial contact, loading response, and preswing phases of walking gait both feet are in contact with the ground at the same time (Center, 2001). Running has similar gait sequences as compared to walking such as stance period absorption and propulsion (Dugan & Bhat, 2005) as well as the functional tasks of weight acceptance, single limb support, and swing limb advancement. Although running is a natural extension of walking, it has many dissimilarity that must be considered when treating the running athlete (Pink, 2010a,b). Fig. 1 compares the similarities and dissimilarities of walking and running gait.

Running differs from walking by certain characteristics such as the increased velocity or distance travelled per unit time and the presence of an airborne or float phase (Dugan & Bhat, 2005; Pink, 2010a,b). Even at the same speed, race walking is differentiated from running in that the later lacks double limb support and has a float phase (Dugan & Bhat, 2005). Runners have four distinct phases or events of running gait: 1) stance, 2) early swing or float, 3) middle swing, and 4) late swing or float (Fig. 1) (Pink, 2010a,b; Pink, Perry, Houglum, & Devine, 1994; Reber, Perry, & Pink, 1993). Running phases can be further delineated; the number of these subdivisions varies depending on the running speed. Seventy-eight (78) subdivisions have been identified during slower Level I running (Brody, 1987) during shod treadmill and over ground running (pace = slower than 8 min mile or speed = slower than 7.5 mph). These slow running subdivisions include: 1) stance = 29, 2) early swing = 10, 3) middle swing = 29, and 4) late swing = 10 subdivisions (Pink et al., 1994). During shod faster Level II running (Brody, 1987) (pace = greater than 7.5 min mile or speed = faster than 8.0 mph) in recreational runners, 72 subdivisions have been identified: 1) stance = 24, 2) early swing = 12, 3) middle swing = 24, and 4) late swing = 12 subdivisions (Pink et al., 1994). During running gait analysis it is impractical to focus on deviations by subdivision; the authors recommend assessment by running phase. Dugan and Bhat (2005) and others (Pink et al., 1994) have further demarcated these 4 phases of running gait, to match the 8 phases of walking gait (Dugan & Bhat, 2005). Although these phases of running gait occur in all runners, the subdivisions vary during

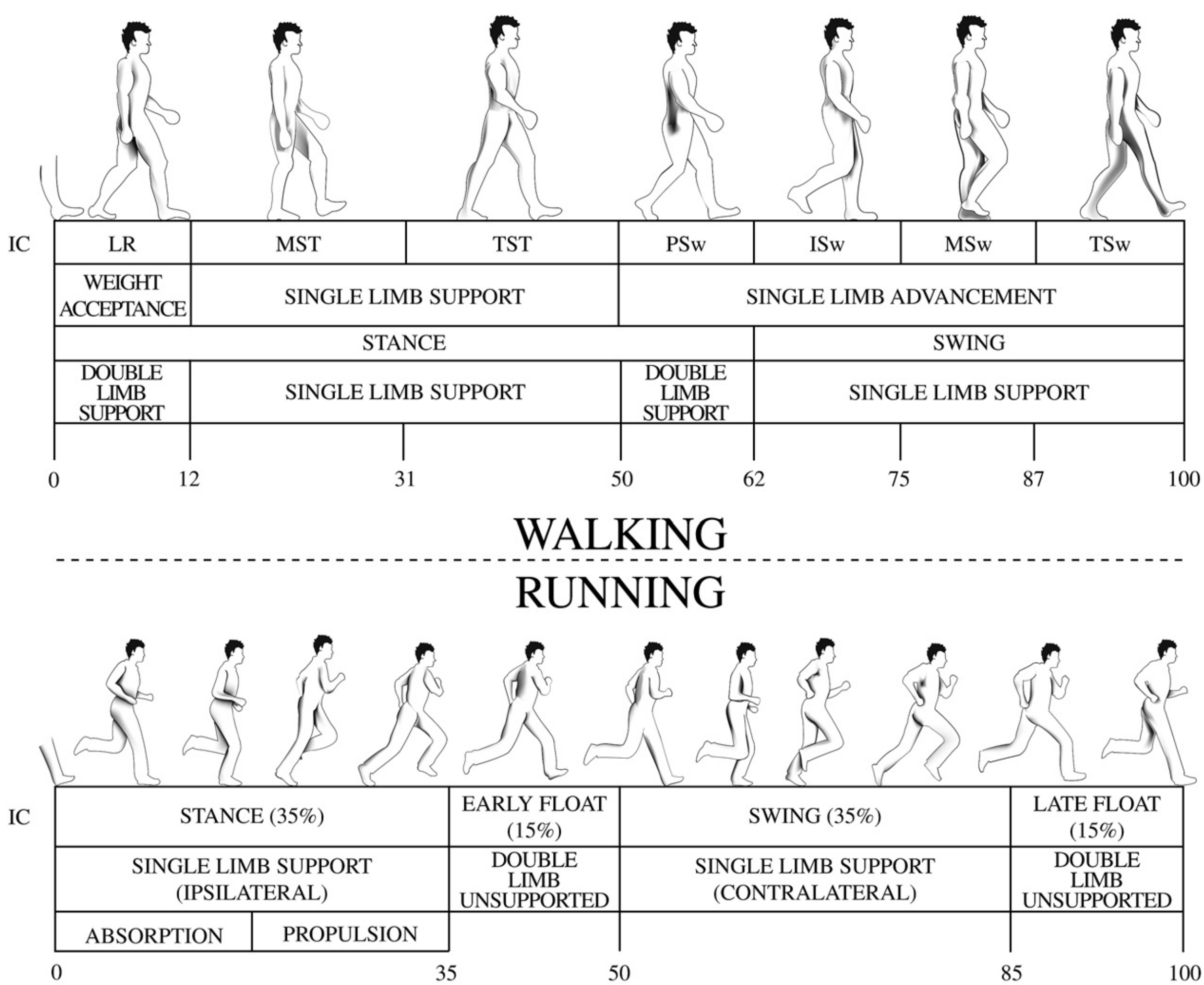

Fig. 1. Comparison of the phases of the walking and running cycles: Initial Contact (IC), Midstance (MST), Terminal Stance (TST), Preswing (PSw), Initial swing (ISw), Midswing (MSw), and Terminal swing (TSw).

shod and unshod running as well as running speed. An example is that all runners have a period of initial contact; however, this contact may occur at heel, midfoot, or forefoot.

Float is the period when neither foot is in contact with the ground during the running gait cycle and occurs first between 35 and 50% (early swing) and 85–100% (late swing) of the running cycle (Figs. 1 and 2). These two float or airborne periods result in decreased stance time and increased swing time during the running cycle (Pink, 2010a,b). During walking, the ratio of stance to swing time is approximately 62:38 (Center, 2001) in contrast to the typical ratio of 35:65 while running at a "training pace" or Level I recreational running speeds (6.5 mph) (Pink, 2010a,b). These stance: swing ratios vary depending on running speed with faster speeds favouring more swing duration (Pink et al., 1994). During running the first 35% of the cycle is spent in single leg stance, while the remaining 65% of the cycle is spent in swing during slower-paced running. During slower-paced running, swing has three components that include early swing or float (15%), middle swing (35%), and late swing or float (15%) (Pink, 2010a,b) (Fig. 2). During faster-paced running, the stance to swing ratio changes to approximately 30:70. Shod runners spend approximately 33% more time airborne (float) while running at a fast pace as compared to shod running at a slower running pace (Pink et al., 1994). The actual duration of stance and swing are variable depending on the running speed; faster speeds results in decreased support and increased float periods. Barefoot runners typically spend a larger percentage in the float periods than do shod runners thus affecting the stance: swing ratio.

Running can be further sub-classified by velocity such as submaximal running or jogging typically occurring between the speeds of 5–10 mph in recreational runners (Mulligan, 2004); however jogging speed can overlap with both walking and running. Slow jogging ("slogging") differs from walking at the same speed by the absence of double limb support (Keller et al., 1996). Slogging differs from running at the same speed by its more characteristic vertical and "bouncy" running style (Keller et al., 1996). In contrast to running, the stance period is greater than the swing period during jogging. Elite endurance runners' speed can be as fast as 14.5 mph (Haile-Selassie, 2001). (Cavanagh & Kram, 1989) reported that the average speeds in recreational endurance runners typically vary from 7.2 mph to 9.4 mph (Cavanagh & Kram, 1989).

Much of the forward momentum during running is produced by the swing leg rather than the stance leg (Mann, 1982; Pink et al., 1994). During early swing there is an interaction between the knee and hip; knee flexion is immediately followed by hip flexion, both serving to promote forward body propulsion (Pink et al., 1994). Ankle motion does not differ as running speed increases suggesting that the ankle does not factor into running speed. This fact supports the concept that the power does not primarily come from the ankle (propulsion) thus suggesting that the term "push-off" is a misnomer (Pink et al., 1994). However, despite these factors it would be inaccurate to conclude that the ankle and corresponding musculotendinous structures (musculo-tendon springs) do not contribute to running energetics. Running energetics and mechanics do differ from walking. Thorpe et al (1999) reported that unlike the pendular mechanics of walking, running uses the mass-spring mechanics in the compliant lower limb in which muscles and tendons sequentially store and then release energy during the stance phase (Bramble & Lieberman, 2004).

Tendons and ligaments of the lower limb store energy during the loading response phase of the stance period of running (braking component of horizontal GRF) and then release this stored elastic strain energy through recoil at the end of stance (propulsive component of horizontal GRF) (Cavanagh & Kram, 1989; Ker, Bennett, Bibby, Kester, & Alexander, 1987). The human lower leg muscles consist of numerous long spring-like tendons attached to shorter muscles that can economically generate greater force when running. The Achilles tendon is the most important lower limb spring; however other structures such as the iliotibial band and the tendon of the fibularis longus muscle are also serving as valuable leg springs (Bramble & Lieberman, 2004). These muscle-tendon springs of the lower extremity are estimated to reduce the metabolic cost of running by approximately 50% (Alexander, 1991, 2005; Ker et al., 1987). The recoil from these springs, such as the Achilles tendon, support electromyographical and other running gait studies that suggest "propulsion" is not primarily a function of concentric muscle contraction (Pink et al., 1994). Unshod runners are better suited to utilize the elastic energy storage in the Achilles and arch of the feet as compared to shod runners (Lieberman et al., 2010). To most effectively utilize the spring mechanism, human runners flex the knees more in running than in walking (Farley, Glasheen, & McMahon, 1993). During weight acceptance during running, the leg spring compresses and the centre of mass moves inferiorly (Farley, Houdijk, Van Strien, & Louie, 1998). Running differs from walking in that the running utilizes more of a spring-mass model while the walking utilizes the inverted pendulum model (Farley et al., 1998). Understanding these important running energetics and mechanics will aid the physical therapist in making rehabilitation decisions in the injured runner.

Running produces greater joint stress and required motion as well as greater eccentric muscle contraction and activation than walking (Dugan & Bhat, 2005; Ounpuu, 1994; Pink, 2010a,b). Peak

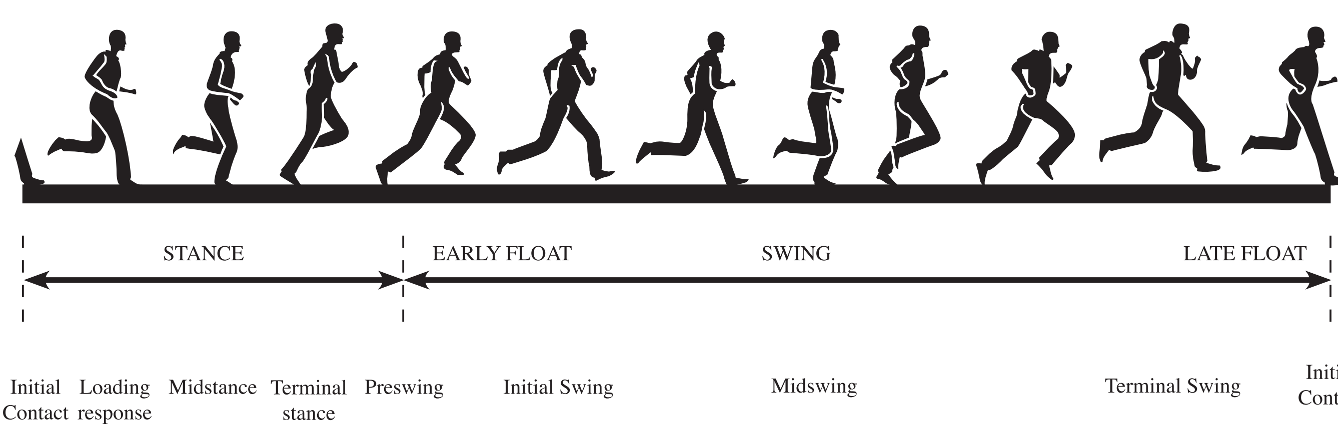

Fig. 2. Phases and periods of the shod running cycle.

vertical forces occur at the end of the stance period (Pink et al., 1994). Running produces greater joint excursion of trunk flexion, trunk and pelvic rotation, hip flexion, knee flexion, and ankle dorsiflexion as compared to walking (Ounpuu, 1994). During running, ankle range of motion did not differ significantly between slow and fast pace shod running where as knee flexion is significantly greater during the middle portion of the swing phase during fast running as compared to slow pace running for recreational runners (Pink et al., 1994). The hip requires more hyperextension in early swing and more hip and knee flexion during middle and late swings during fast running as compared to slow paced running (Pink et al., 1994). One study reported that hip flexion increases as the speed of the running increases (Mann & Hagy, 1980). Mann and Hagy (1980) reported that the sprinters demonstrated 10° to 15° more hip flexion motion as compared to runners. Runners demonstrated 20° greater hip flexion motion than the walkers. But, the data were applicable only to fast pace runners and lacks kinematic data for recreational runners (Mann & Hagy, 1980).

In addition to increasing joint excursion, running pace also affects force distribution at the foot. Running paces are divided into 1) jog pace at 8.5 min/mile, 2) train pace at 6.5 min/mile, 3) race pace at 5.4 min/mile (Reber et al., 1993). Increasing running paces challenges the ability of the tendons and aponeuroses of the foot to withstand the higher forces transferred onto them as the eccentric activity of the muscle increases (Pink, 2010a,b). Movement and strength impairments, as well as shoe selection, will alter running motion and timing. Joint excursion is similar during both treadmill running and over ground running so the treadmill is a useful clinical tool to assess sagittal plane lower extremity joint range during running (Pink et al., 1994). Running differs greatly from walking in regards to both spatial and temporal parameters as well.

3. Energy expenditure & spatiotemporal parameters of running

Running speed is determined by the spatial and temporal parameters of stride length and cadence. Spatiotemporal parameters during running are interrelated (Dugan & Bhat, 2005). In addition to speed, running is also referenced in a term of pace that is expressed in minutes per kilometre or mile such as a 6 min mile. It has been a long-standing theory that running had the same metabolic cost per unit of time regardless of the speed - simply put; the energy needed to run a set distance is the same regardless of speed. A recent study (Steudel-Numbers & Wall-Scheffler, 2009) is challenging this theory by reporting that humans may have an optimal running speed that has a lower metabolic energy cost; however, it should be noted that the difference between optimal and non-optimal running speed energy cost is relatively small. The authors report that energetic demands are higher at slower and faster speeds while intermediate speeds had the maximum energy efficiency. In regards to energy efficiency, although individualized, the mean optimal running speed in female amateur runners was 6.5 miles per hour (9.2 min mile) and 8.3 miles per hour (7.2 min mile) in their male counterparts. Much of the gender variability may be due to factors such as leg length and body size. (Steudel-Numbers & Wall-Scheffler, 2009)

Running unshod, barefoot, reduces energy expenditure by approximately 5% as compared to shod running at the same speed (Divert, Mornieux, Baur, Mayer, & Belli, 2005; Squadrone & Gallozzi, 2009). Studies have suggested that the higher oxygen consumption during shod running, with and without orthotics, may be more related to shoe mass rather than shod running patterns (Burkett, Kohrt, & Buchbinder, 1985; Divert et al., 2008). The new lightweight "minimal shoes" such as the Vibram Fivefingers® weighs only 5.7 ounces so may have similar oxygen consumption rates as barefoot running; however this has not been studied. Barefoot running on a treadmill at 8.0 mph in trained subjects demonstrated significantly higher stride frequency, anterior-posterior impulse, vertical stiffness, and leg stiffness as compared to shod runners; however, the net efficiency, which includes both metabolic and mechanical factors, still favoured barefoot running (Divert et al., 2008).

Running efficacy changes as running velocity increases. Taller runners generally have faster optimal running speeds. The least metabolically efficient running speed was the slowest speed tested of 4.5 miles per hour or a 13.3 min mile. The authors attributed this to the fact that this speed is the transition between walking and running gait (Steudel-Numbers & Wall-Scheffler, 2009). This is similar to findings reported by Usherwood and Bertram (2003) that the transition from walking to running is not energy efficient (Usherwood & Bertram, 2003). Several investigators have reported that even at the same locomotion speed, a fast walk is more metabolically efficient than a slow run despite the fact that participant's perceived exertion is lower while slow running (Brisswalter & Mottet, 1996; Hreljac, 1993; Tseh, Bennett, Caputo, & Morgan, 2002). However, running is more efficient than walking when velocity exceeds approximately 222.0 cm/s or 5 mph (Alexander, 2005; Falls & Humphrey, 1976). Humans typically switch from walking to running at approximately 5.1–5.6 mph, which correlates with the metabolic cost of transport (Alexander, 1991; Bramble & Lieberman, 2004; Margaria, Cerretelli, Aghemo, & Sassi, 1963). Running economy is more than assessing oxygen consumption at a given speed; it also should include running biomechanics (Dugan & Bhat, 2005).

Stride length is a function of the runner's height and leg length, corresponding with the runner's ability to extend their stride length to increase velocity (Ounpuu, 1994). During a 7-min mile (8.6 mph) run, the generalized optimal stride length is estimated to be 1.4 times the runner's leg length in shod runners (Youngren, 2005). Taller runners with longer leg lengths are reported to possess more optimal running strides; however, height and leg length are a poor determinant of optimal running stride on an individual basis (Youngren, 2005). Shod and unshod runners, on the average, self-select a running stride within 4 cm from their optimal running stride; however, this optimal stride is individualized and lacks a true predictor (Youngren, 2005). Elite runners tend to have a shorter stride length than less accomplished yet experienced runners (Youngren, 2005). The running stride is considerably longer than walking stride length. The running surface inclination also affects stride length. The stride length shortens while the stride rate (cadence) increases during uphill running as compared to level over ground running. During downhill running, the stride length typically lengthens while the cadence decreases.

The spatial parameter of step or stride width is much narrower during running than walking resulting in what is known as "runner's varus". This narrower-based gait in shod runners requires increased femoral adduction and internal rotation, greater tibial varum (not to be confused with genu varum at the knee), and greater rearfoot varus as the lateral calcaneus contacts at heel strike as compared to walking. This narrower heel-to-heel stride width occurs when the feet are planted more medially to minimize the lateral shift of the centre of gravity (CoG) due to the lack of double limb support in running. The CoG decreases or lowers as the running velocity increases with the step-to-step line of progression moving to or towards midline (Dugan & Bhat, 2005). During the stance phase of running gait, the lower limb is in a functional varus of 8°–14° (Dugan & Bhat, 2005). An example of "functional varus" would be femoral varus where the distal femur is inclined more towards midline than the proximal femur during running. At initial contact, the calcaneus of the typically shod runner is inverted to

4°–8° of varum (Cavanagh, 1982; Williams, 1985). This finding is more pronounced in shod than unshod runners. During maximal running, such as sprinting where speeds exceed 10 mph (Mulligan, 2004), the point of foot contact changes at initial contact from heel to the forefoot or midfoot contact in shod runners (Mann, Baxter, & Lutter, 1981). Also, running unshod (barefoot) or wearing minimally supported running shoes favour forefoot and midfoot initial contact.

In contrast to the spatial parameter of stride length, the temporal parameter of stride rate or cadence remains relatively consistent across different running event lengths (Hoffman, 1971; Rompottie, 1972). Stride rates range from 185 to 200 steps per minute or (93–100 strides per minute) across events and genders. When working with a client who desires to modify their running speed it is recommended to select the most efficient stride rate, and then adjust the stride length to obtain the desired speed (Youngren, 2005). This concept is further supported by others who suggest that a key to increasing running speed is to diminish stance time thus increasing swing or float time (Dugan & Bhat, 2005; Mann & Hagy, 1980; Pink et al., 1994). However, care should be taken to prevent overstriding at the end of late swing thus delaying initial contact as well as considering the optimal stride length for energy efficiency.

In addition to energy efficiency, running speed, cadence, and stride length variations affect stresses and demands on the musculoskeletal system of the lower kinetic chain. While running at 6.5 mph, the talocrural joint must dorsiflex to approximately 17° at the end of the stance period. Although not reported in literature, this range of motion demand at end of the stance phase is typically similar in both shod and unshod runners. While the maximum dorsiflexion requirement during walking is approximately 10°, these movement requirements can double during level running. Ankle dorsiflexion of less than 10° was significant in the development of morbus Osgood's Schlatter and Achilles tendinitis in shod runners (Sarcevic, 2008). Full knee extension is typically not achieved during running. At the end of the swing phase, late swing, the knee is at the maximum extension (approximately 11° short of full extension). This is approximately twice as much as knee flexion during terminal swing as compared to walking. During heel contact (initial contact), recreational runners landed in approximately 15° of knee flexion. This is approximately three times more than the amount of knee flexion at initial contact as compared to walking. During early stance (initial contact to loading response), the knee further flexes to accomplish weight acceptance (Pink et al., 1994). Overstriding will result in greater knee extension at the end of swing and the start of stance, adversely affecting the task of weight acceptance and increased tensile loads to the posterior knee and thigh structures. Overstriding can result in excessive "braking forces" in the anterioposterior (AP) force component at the weight acceptance task during running that can be injurious to muscles of the posterior lower extremity (Fig. 3) (Youngren, 2005). Of the three forces acting upon the body that can result in a running injury (i.e., tensile, compressive, and shear), excessive braking from over striding may result in an increase in repetitive tensile forces due to tissue elongation and increased intensity and duration of eccentric muscle contraction.

4. Biomechanical considerations

In addition to runner's varus, other hip region related biomechanical dysfunctions have been attributed to walking and running-related injuries. Abnormal hip kinetics due to diminished hip-muscle performance resulting in excessive hip internal rotation and/or adduction have been reported to be responsible for, or at least contributes to, common running injuries such as acetabular labral pathology (Austin, Souza, Meyer, & Powers, 2008), iliotibial friction syndrome (Ferber, Noehren, Hamill, & Davis, 2010; Fredericson et al., 2000; Fredericson & Wolf, 2005), patellofemoral syndrome (Dierks, Manal, Hamill, & Davis, 2008; Powers, 2003; Souza & Powers, 2009), chronic ankle sprains (Gribble & Hertel, 2004; Gribble, Hertel, Denegar, & Buckley, 2004; Miller & Bird, 1976), and even low back pain (Childs et al., 2004; Iverson et al., 2008).

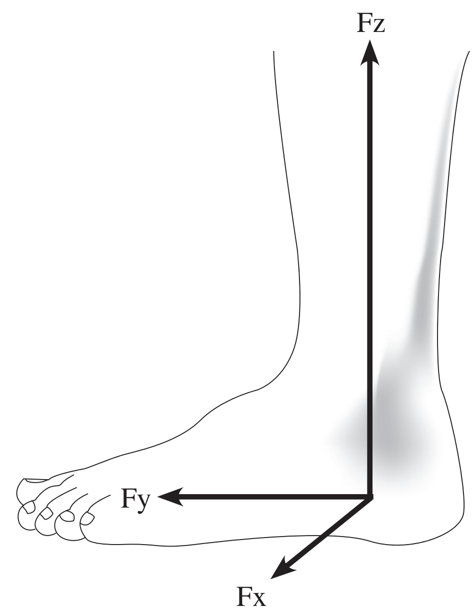

Fig. 3. The vectors represent the direction of forces acting on the foot during the stance period of running. Abbreviations: Fz, the vertical force; Fy, the antero-posterior horizontal sheer force; Fx, the mediolateral horizontal shear force.

Of these pathologies, patellofemoral pain syndrome is the most prevalent, representing approximately 20% of all running-related injuries (Taunton et al., 2002). Weakness of hip abductors and external rotators has been associated with PFPS, iliotibial band syndrome, and non-contact anterior cruciate ligament injuries (Fredericson et al., 2000; Hewett et al., 2005; Ireland, Willson, Ballantyne, & Davis, 2003; Powers, 2003; Rabin & Kozol, 2010; Robinson & Nee, 2007; Meira & Brumitt, 2011). The hip abductors and external rotators function eccentrically during the first half of the stance phase to control hip adduction and internal rotation, respectively (Perry, 1992). Runners with patellofemoral pain syndrome (PFPS) displayed weaker hip abductor muscles as compared to asymptomatic runners (Dierks et al., 2008). This weakness is more pronounced at the end of prolonged runs when the runner was in an exerted state. This abductor weakness is closely associated with increased hip adduction or hip varus (Dierks et al., 2008). This weakness associated with peak hip adduction angles is further increased at the end of the run (Dierks et al., 2008). These movement and muscle performance impairments are more prevalent in shod runners than unshod or minimalist runners that utilize a non-heel strike strategy during weight acceptance. In addition to the proximal influence of the hip and knee kinematics in runners, distal factors also influence the knee joint.

Increased rearfoot eversion is also associated with increased genu valgum and lateral patellar force vectors. Numerous studies have identified an association between genu valgum and lower-arched feet (McClay & Manal, 1998; Nawoczenski, Saltzman, & Cook, 1998; Nigg, Cole, & Nachbauer, 1993; Powers, 2003). Although it has been speculated that foot intrinsic muscles are weakened with continued use of shoes with arch supports or stiffened stoles leading to increased pronation and collapse of the medial longitudinal arch (Lieberman et al., 2010; Robbins & Hanna, 1987); currently there is no evidence to support this theory. To date, no studies have shown any difference in foot muscle strength in shod versus unshod runners. Mayer, Hirschmuller, Muller, Schuberth, and Baur (2007) proposed that activities performed in shoes with rigid orthotics would result in decreased intrinsic muscle strength; however, they reported no change in intrinsic strength but realized an increase in calf strength with orthotic therapy. The authors also reported a reduction of pain in runners receiving orthotic intervention (Mayer et al., 2007).

Limited ankle dorsiflexion has been reported to cause increased subtalar joint pronation, increased knee valgus, and altered lower extremity movement patterns during functional tasks such as step down activities and drop-jump landing (Gross, 1995; Rabin & Kozol, 2010; Sigward, Ota, & Powers, 2008) found increased dynamic knee valgus during drop-jump landing in females associated with limited hip external rotation and ankle dorsiflexion. In addition, increased dynamic knee valgus in the frontal plane has been associated with activities that require simultaneous ankle dorsiflexion and knee flexion (Rabin & Kozol, 2010). Such activities include running in individuals with tight heel cords and a heel contact running pattern. Physical therapists should assess ankle dorsiflexion range when they observe aberrant lower extremity movement patterns (Rabin & Kozol, 2010). In conclusion, clinicians should assess proximal (hip adductors and hip internal rotators) and distal structures (ankle plantar flexors) for tightness and movement impairments as well as for proximal structure (hip extensors, hip abductors, and hip external rotators) weakness in clients with altered movement patterns such as lower extremity "medial collapse" during running.

5. Impact forces

Running is potentially most injurious when the foot makes contact with the ground due to striking impact of the foot (resultant ground reaction forces) being transferred up the lower kinetic chain (Lieberman et al., 2010). There are three primary types of foot contacts during running: 1) Rearfoot strike where the calcaneus contacts the ground first, 2) midfoot strike in which the rearfoot and forefoot meets the ground simultaneously, and 3) forefoot strike where the forefoot lands on the ground first followed by the heel (Lieberman et al., 2010). Unshod runners commonly land on their forefoot and less commonly on their midfoot or rearfoot (Fig. 5); however, some barefoot or minimally supported runners do land on their heels. Conversely, shod runners usually run with the rearfoot ground contact which may be facilitated by the elevated and cushioned heel of the modern running shoes (Lieberman et al., 2010); however, initial contact varies depending on numerous factors including running speed (Dugan & Bhat, 2005).

Sprinters contact the ground with their forefoot while shod distance runners (75–90%) commonly land on their rear foot (Hasegawa, Yamauchi, & Kraemew, 2007; Kerr, Beauchamp, Fisher, & Neil, 1983; Lieberman et al., 2010; Pink, 2010a,b). Rearfoot strike runners have to repeatedly absorb impacts up to 3.0 times the runner's body weight (BW) (Keller et al., 1996; Lieberman et al., 2010). These sudden and high rate magnitude forces will travel rapidly up the lower kinetic chain and may contribute to the high incidence of running-related injuries such as tibial stress fractures (compressive injury) and plantar fasciitis (tensile injury) (van Gent et al., 2007; Milner, Ferber, Pollard, Hamill, & Davis, 2006; Pohl, Hamill, & Davis, 2009). All forms of running produces greater ground reaction (GRF) or impact forces than walking. Vertical GRF increases in a linear manner as running speed increases up to approximately 60% of maximum speed at which point the forces plateau (Keller et al., 1996).

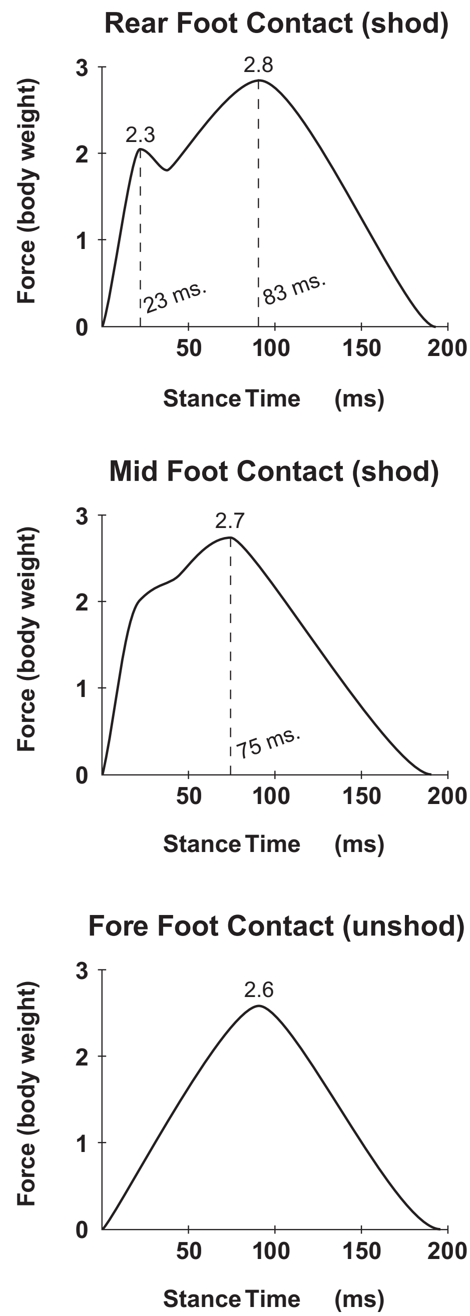

Fig. 4. Ground reaction forces for different strike patterns in shod and unshod runners.

During shod jogging or running, the vertical ground reaction force may reach 2.0–3.0 times the body weight (Keller et al., 1996). Slow jogging (between 3.3 and 6.7 mph) produces vertical GRF up to 1.6 times more than walking at the same speed, thus making jogging a potentially more injurious activity as compared to walking (Keller et al., 1996). The magnitude of the ground reaction force is affected by a number of variables including running style (rearfoot, midfoot or forefoot strike), speed, stride length, jogging versus running, footwear, ground surface, and inclination of the running surface.

Cavanagh and Lafortune (1980) examined the force components (Fig. 3) in shod runners that were either rearfoot strikers (RFS) or midfoot strikers (MFS). In this study, seventeen subjects including 10 males and 7 females were participated with a mean age of 24 years. Twelve subjects were recreational runners and the

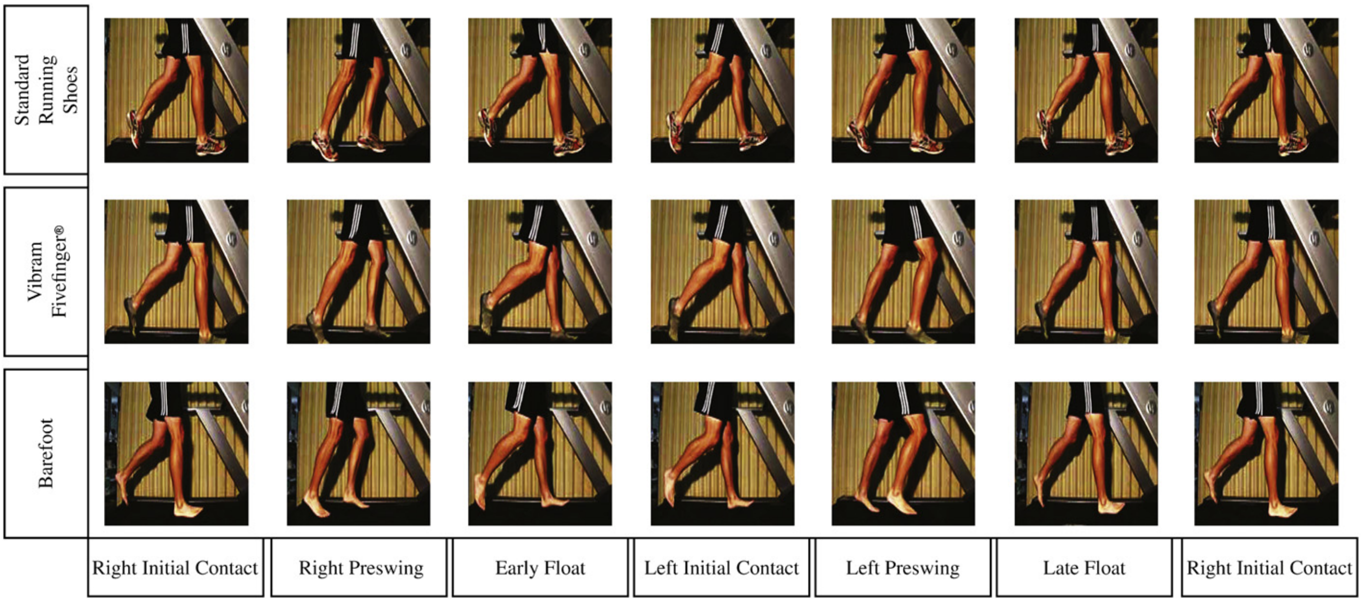

Fig. 5. Comparison of the periods and phases of the running cycle comparing and contrasting foot-ground contact while running with standard running shoes, minimal footwear (Vibram Fivefinger®), and barefoot.

remaining five were varsity athletes. They found similar vertical GRF or vertical force component (Fz) of 2.8 BW and 2.7 BW for RFS and MFS, respectively; however unlike the MFS, the RFS had two impact peaks during stance rather than one. The RFS first peak (2.2 BW) termed as impact peak occurred just after initial contact during loading response, the weight acceptance period, and followed by a second peak (2.8 BW) known as thrust peak occurred at midstance (Cavanagh & Lafortune, 1980; Keller et al., 1996). Lieberman et al. (2010) reported similar findings when they compared habitually shod and barefoot adult runners from America during RFS and forefoot strike (FFS) running (Lieberman et al., 2010). Shod and unshod RFS running produced a double peaked Fz vertical GRFs or an impact transition while the unshod toe-heel-toe FFS running style generated no impact transition (a smooth single peak) (Fig. 4).

The anterioposterior (AP) force component (Fy) in the sagittal plane (Fig. 3) has a braking and propulsive phase with the transition between the phases occurring at approximate 48% of the stance period or midstance for both rearfoot and midfoot groups. The MFS had two impact peaks during stance rather than one as compared to the RFS. In mechanics, impact force is defined as a high force or shock that occurs when two or more bodies collide into each other during a short period of time. The effect is dependent on the relative velocity of the bodies to one another. The horizontal force (Fy) is also an impact force (since foot and ground collides each other). The first peak (0.45 BW) of MFS (braking) occurs after initial contact during loading response at 11ms and then returns to zero before it reaches the second peak (propulsive) of 0.45 BW (same magnitude) at 38ms. The MFS had two abrupt AP forces of approximately 0.45 BW while the rearfoot strikers had a more gradual AP force throughout the stance period reaching a peak of 0.5 WB at 139ms.

Finally, the mediolateral (ML) force component (Fx) in the frontal plane (Fig. 3) was three times greater in the MFS than the RFS; however, it should be noted that the Fx was relatively small at 0.35 BW and 0.12 BW respectively. This force component (Fx) is comparatively smaller than vertical (Fz) and anterior-posterior components (Fy). The RFS demonstrated a continuous anterior movement of the centre of pressure (COP) during the stance period of running while MFS had a posteriorly directed COP during the first 20 ms of weight acceptance phase of stance (Fig. 6) (Cavanagh & Lafortune, 1980; Munro, Miller, & Fuglevand, 1987). The human body attempts to attenuate high frequency impact forces in all planes. Inadequate attenuation of these forces may result in microtruma to soft tissue and bone (Nigg, Denoth, & Neukomm, 1981).

Immerging research over the last decade is focussing on the "soft tissue compartment vibrations" in the lower extremities produced during the stance phase of running (Nigg, 2001; Wakeling & Nigg, 2001). This concept of muscle tuning postulates that the body attempts to minimize soft tissue vibration initiated at impact by muscle adaptation or activity prior to heel strike during heel-toe running to change the mechanical properties of the soft tissue compartment (Boyer & Nigg, 2006, 2007). The impact portion of the GRF produced during running is primarily due to rapid deceleration of the leg after initial contact (Boyer & Nigg, 2007). The constant-force hypothesis proposes that the central nervous system (CNS) uses muscle tuning to maintain GRF relatively constant regardless of the sole firmness (Zadpoor & Nikooyan, 2010). The GRF increases as the shoe hardness increases in experimental mechanical leg model. But, the GRF is constant in the actual human runner. The reason is the human body has the ability to predict the GRF input signal and accordingly modulates the muscle activity to compensate the changes in the shoe hardness (Zadpoor & Nikooyan, 2010). Muscle activation within 50ms prior to impact serves as a preparatory mechanism to minimize vibration from the anticipated impact by generating joint or leg stiffness and/or movement of the leg segment while running at a constant speed (Nigg, 2001). As the speed of running increases so does the impact energy (Boyer & Nigg, 2004). In accordance with energy conservation laws, increased muscle activation occurs at fast speeds to increase stiffness of the soft-tissue package to dampen the vibration frequency (Boyer & Nigg, 2004). These theories propose that the body attempts to minimize these vibrations through *a priori* CNS regulated muscle contractions in order to preserve impact forces at a constant level or frequency at the next running stride. These theories continue to be investigated and perhaps future soft-tissue vibration studies will enable researchers to assess individual runner's unique resonate frequencies to improve

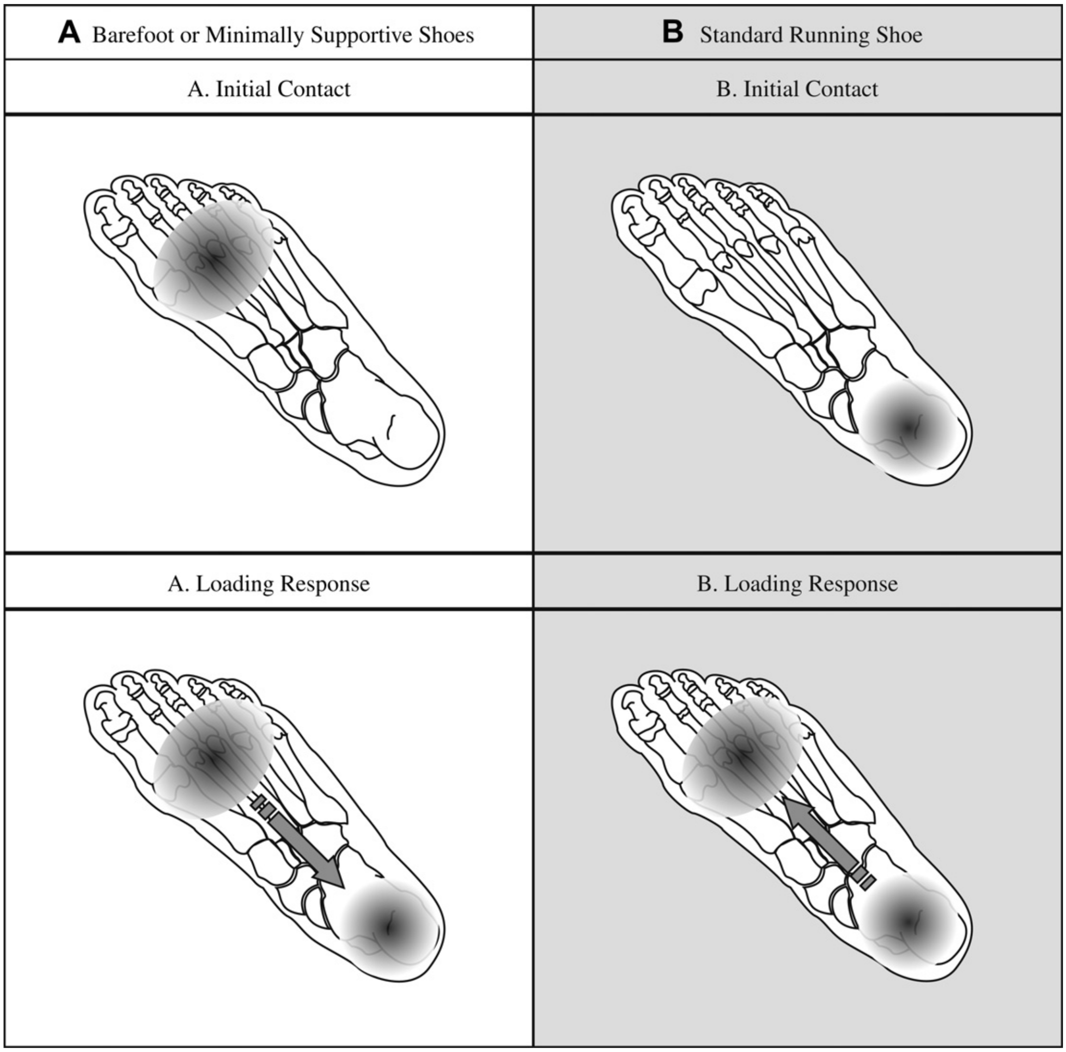

Fig. 6. The center of pressure (CoP) during running: A. While running barefoot or with minimally supportive shoes the CoP is located at the forefoot or midfoot at initial contact and then travels in an anterioposterior horizontal shear force (Fy) direction during loading response. B. While running with standard running shoes the typical CoP is located at the rearfoot at initial contact and then travels in a posteroanterior (Fy) direction during loading response.

performance, modify running patterns, modify running shoes, and even minimize injuries.

6. Running surface firmness & slope

The type of surface that the athlete's foot comes into contact during running also affects GRF acting on the body as well as muscle activation. The reader is alerted to a common misconception regarding impact forces related injuries proposed to be a result of running on hard, non-compliant surfaces. Shod runners typically adjust their "leg stiffness" thus experiencing similar impact forces on either hard or soft surfaces (Dixon, Collop, & Batt, 2000). Unshod runners also adjust their leg stiffness but will recruit muscles in a slightly differing pattern and intensity due to the different foot striking pattern. Leg stiffness is the ratio between the peak force and maximum leg compression during ground contact. During running leg stiffness decreases as the height of the elevation increases. This allows the runner to maintain their stability without concentrating on the irregularities of the ground while running on the uneven ground (Grimmer, Ernst, Gunther, & Blickhan, 2008).

Even though a firm surface has potentially high impact forces as compared to more complaint running surfaces; runners subconsciously adjust their lower extremity stiffness in order to maintain a consistent vertical stiffness. Runners adjust by reducing stiffness on hard surfaces and increasing stiffness on compliant surfaces (Ferris, Louie, & Farley, 1998). Runners quickly adjust their leg stiffness on their first step when they encounter a new surface such as the transition from a soft to hard surface (Ferris, Liang, & Farley, 1999). Vertical stiffness is composed of leg and surface stiffness (Tillman, Fiolkowski, Bauer, & Reisinger, 2002). By reducing leg stiffness, runners can maintain constant stiffness to offset increased surface stiffness (Ferris et al., 1998). Humans modify leg stiffness in order to maintain similar stride frequency, contact time, and peak ground reaction forces on various surfaces (Ferris et al., 1999; Ferris et al., 1998); however, these adjustments may place the runner at risk for injury (Tillman et al., 2002). Runners with smaller feet, relative to body size, are required to make greater leg stiffness adjustments (Ferris et al., 1998). Leg stiffness adjustments are accompanied by kinematic and kinetic adjustments as well. Tillman et al. (2002) found no significant differences in shoe reaction forces among the different running surfaces (asphalt, concrete, grass, and a synthetic track) (Tillman et al., 2002).

These findings are consistent with Dixon et al. (2000) that reported similar impact forces across three synthetic running surfaces (Dixon et al., 2000). The similar running impact forces are maintained through kinematic adjustments, including changing knee flexion patterns, to changing leg stiffness in order to compensate for changes in running surfaces with high stiffness (Dixon et al., 2000; Feehery, 1986; Nigg & Yeadon, 1987). When running on a compliant surface, runners will assume a more extended leg posture during ground contact (Farley & Gonzalez, 1996). Although these changes in lower extremity joint angles are relatively small ($<7°$); however, even small angle change affects muscle forces (Biewener, 1989; Dixon et al., 2000; Tillman et al., 2002). Running on non-complaint surfaces do not expose shod runners to increased impact forces due to compensatory biomechanical changes; however, these changes in joint moments and muscle forces (active force) stress the musculoskeletal system potentially resulting in sprains and strains. Impact force is a collision force while active force is generated by muscle contraction. Active force is measured on a force plate but not through single muscle needle electromyography. Runners can also alter forces by changing foot strike biomechanics while running. Forefoot or midfoot initial contact generates relatively low collision forces even on non-compliant, firm surfaces thus allowing for smaller active force production (Lieberman et al., 2010).

Roadways typically have a camber or canter to promote proper drainage. The centre or crown of the road typically has a higher elevation than the edge of the roadway. When running on a camber, runners will attempt to right themselves in the frontal plane to maintain a vertical trunk. Even running trails are designed with a cross slope, which is sloped perpendicular to the direction of running to promote drainage. Again, the human body attempts to regulate leg stiffness to adapt to the running environment. The human body maintains its upright stability by adjusting the "leg stiffness" without focussing on the irregularities that is encountered while running on the uneven ground (Grimmer et al., 2008). A study performed by Muller and Blickhan (2010) with runners running along a runway that consisted of sections that had a step up of 10 cm and also sections of a step down of 10 cm. The leg stiffness decreased by about 20.4% during the step up section and also similar findings (decrease in leg stiffness 18.8%) when running on the step down section of the lowered track (Muller & Blickhan, 2010). Runners decrease their leg stiffness regardless if the running surface transition is either elevated or declined (Grimmer et al., 2008; Muller, Grimmer, & Blickhan, 2010). During road running it is typically advisable to run against the flow of traffic; however, this may be mandated by local traffic laws. When running against traffic on a straight stretch of road, the leg closest to the centre line of the roadway will have a shorter distance to the ground at foot strike than does the opposite limb resulting in an environmental leg length discrepancy. Running on a cantered surface has been identified as a potential risk factor for such running-related injuries as iliotibial band friction syndrome where the leg closest to the centreline was potentially at a greater risk of injury. Repetitively running in the same direction on a running track can result in similar asymmetrical stresses to the human body from the runner leaning into the curve of the track.

The trail slope or the inclination of the road surface parallel to the direction of running also changes joint reaction forces, joint range of motion, muscle length-tension requirements, and the type and intensity of muscle contractions. Movement scientists use GRF data to measure impact forces and loading rate, to understand braking and propulsion, and to calculate muscle forces (Gottschall & Kram, 2005). (Gottschall & Kram, 2005) quantified GRF during uphill ($+3°$, $+6°$, and $+9°$ incline) and downhill running ($-3°$, $-6°$, and $-9°$ incline). When compared to level shod running, data impact force peaks were considerably greater for downhill running and smaller for uphill running. All subjects performed rearfoot contact for all three downhill inclinations, level, and $+3°$ uphill running. All subjects converted to midfoot contact by $+9°$ uphill running. Peak impact forces and load rates were highest at $-9°$ declined running and lowest at $+9°$ inclined running where the impulse force was absent. Parallel braking force peaks (Fig. 4) were highest during downhill running and lowest during uphill running.

Gottschall and Kram (2005) combined their data with those of (Hreljac, Marshall, & Hume, 2000) to recommend that: "Persons trying to recover from impact injuries would benefit by avoiding downhill running and possibly incorporating purely uphill treadmill running" (Gottschall & Kram, 2005). During level walking the majority of muscles function essentially during at least a portion of the gait cycle. This eccentric contraction, controlled lengthening under tension, is accentuated during downhill running; this especially is the case for the knee and hip extensors as well as the anterior (pretibial muscles) and posterior tibial muscles (Eston, Mickleborough, & Baltzopoulos, 1995). Uphill running primarily utilizes concentric muscle contractions especially the hip extensors (gluteals and hamstring muscles) and knee extensors as well as the posterior tibial muscles.

In general, running downhill increases lower extremity joint compressive forces and increases eccentric muscle contractions as well as increasing stride length and decreasing stride rate. In general, running uphill requires more concentric muscle contractions, reduces joint reaction forces, and yet requires greater range of motion demands such as ankle dorsiflexion and trunk flexion. Uphill running increases the stride rate while reducing the stride length and promotes FFS or MFS contact.

In summary, as the foot strikes, the GRF is transmitted through the runner's body. To minimize the vertical GRF, the human body will alter leg stiffness through altering muscular activity and joint angles to maintain impulse forces relatively equal across different running surfaces in order to minimize injuries. Increased lower extremity joint flexion requires greater muscle activation and fatigue and altered joint reaction forces, which could potentially lead to injury. While many running coaches encourage runners to train on a variety of surfaces, it may be wise to change running surfaces gradually to allow the body to acclimate. The physical therapist will need to consider all of these factors when prescribing a return to running regimen for a recovering runner or attempting to prevent a reoccurring injury. Individuals trying to recover from impact injuries may benefit from avoiding or limiting downhill running and consider utilizing low-impact, concentric uphill treadmill running. To overcome the potential risks of repetitive injuries from running unidirectional on cantered surfaces; the physical therapist may recommend reversing the running direction.

7. Shod and unshod running

In spite of technological advancement, as many as 6 out of 10 runners were estimated to get injured every year. Lieberman et al. (2010) explains that modern running shoes with large, flared, elevated heels, inflexible soles, and stiff arch supports promotes heel-to-toe running. These cushioned high heeled running shoes limit proprioception as well (Lieberman et al., 2010). Two studies even suggest that running shoes can increase the risk of ankle sprains while running due to either reduced proprioception or somatosensory information (Robbins, Waked, & Rappel, 1995) or the increased leverage arm (as a consequence of the moment about the subtalor joint) from the elevated heel (Stacoff, Steger, Stussi, & Reinschmidt, 1996). A systematic review published in 2009 concluded that there is no evidence to support the current practice of prescribing elevated running shoes with cushioned heels and pronation control systems tailored for pronators to prevent injuries (Richards, Magin, & Callister, 2009); however, an improperly fitting running shoe can be potentially injurious especially in the older runner (McWhorter et al., 2003).

A recent study was also unable to support the utility of prescribed running shoes tailored to the runner's foot type for injury prevention (Knapik et al., 2010). Military recruits ($n = 2676$) were randomly assigned to either an experimental or control group (Knapik et al., 2010). The experimental group received running shoes matched to arch type (motion control shoes for low arches, stability shoes for medium arches, and cushioned shoes for high arches) while the control group received stability shoes regardless of arch type (plantar foot shape). The results of the study demonstrated that assigning running shoes based on arch type showed little difference in injury risk for male or female recruits as compared to the control group (Knapik et al., 2010). A recent study suggests that minimally supported shoes might actually improve rehabilitation outcomes as compared to conventional running shoes (Ryan, Fraser, McDonald, & Taunton, 2009).

Twenty-one subjects with chronic plantar fasciitis ($\geq$6 months) completed a 12-week multi-element exercise regimen while wearing either a minimal shoe with ultraflexible midsole shoes (Nike Free 5.0) or conventional running shoes (Ryan et al., 2009). Although both groups' pain reduction outcomes were similar by the 6-month follow-up, the minimal shoe group reported an overall reduced pain level throughout the study as compared to the conventional running shoe group. The authors concluded that minimal footwear (Nike Free 5.0) may result in reductions of plantar foot pain earlier while performing an exercise regimen as compared to conventional running shoes. This may be because many modern running shoes have stiff soles and arch supports that can potentially promote weakening of the foot intrinsic muscles and reduced arch strength (Lieberman et al., 2010). These factors have been purported to contribute to considerable demands on the plantar fascia and promote excessive foot pronation that can cause or delay recovery of plantar fasciitis (Lieberman et al., 2010). It is important to note at this time that although these negative connotations related to conventional running shoes is gaining support, it is far from being globally accepted by runners, scientists, and health care providers. As an example, the ASICS Corporation, that is now producing minimal shoes, but is also promoting a new elevated running shoe for women to help prevent injuries. This 13 mm elevated heel is proposed to help prevent Achilles tendon injuries around the fourteenth day of the menstrual cycle when they are more prone to injury. Research supports the theory that women are significantly more prone to injury around the midcycle or the ovulatory phase and had a lower than expected injury rate during the luteal phase of the menstrual cycle (Wojtys, Huston, Boynton, Spindler, & Lindenfeld, 2002). In addition to the trend of minimal running shoes, there is also a trend towards maximal, heavily cushioned rocker sole walking shoes. The benefits and claims that have been purported from these walking shoes have included: Improved posture, reduced pain, increased strength of core muscles, improved muscle activity of the lower extremity, weight loss, reduced cellulites, and improved circulation; however, this claims have not be supported by large sample, independent research.

Despite the proposed negative aspects associated with heel-to-toe running; currently there is only anecdotal evidence that forefoot or midfoot striking patterns may help prevent or allay many lower quarter repetitive stress injuries. To date, there are no studies directly examining the efficiency of forefoot or midfoot strike patterns on running injuries as compared to rearfoot contact. Presently there is also a lack of peer-reviewed comparisons of injury rates between barefoot, minimally shod, and shod runners. Despite the presence of comparative injury data, numerous articles suggest that wearing conventional running footwear may not be essential or may even have adverse affects (Bishop, Fiolkowski, Conrad, Brunt, & Horodyski, 2006; Clement, Taunton, & Smart, 1984; Clement, Taunton, Smart, & Mcnicol, 1981; Cook, Brinker, & Poche, 1990; van Mechelen, 1992; Robbins & Hanna, 1987; Shakoor & Block, 2006; Siff & Verkhoshansky, 1999; Squadrone & Gallozzi, 2009); however, running barefoot or with minimal footwear may not be risk free either.

Web logs (blogs) and other websites have posted personal case studies or anecdotal theories that claim minimally supportive shoe or barefoot running may accelerate the development of other injuries such as sesamoiditis, metatarsal stress fractures, metarsalgia, and fat pad syndrome (Burge, 2001). It is likely that as minimally supported shoe use continues to become more popular, the prevalence of certain FFS and MFS related running injuries will begin to be studied. Additionally, running surfaces with stones, pieces of glass, nails, and debris on roadways or ungroomed trails are not suitable for barefoot running (Squadrone & Gallozzi, 2009) and running on ungroomed or irregular surfaces may also require greater range of motion, especially in the foot and ankle joints.

Although barefoot or forefoot running may reduce the risk of repetitive stress injuries such as medial tibial stress syndrome or shin splints; it could theoretically increase the risk of Achilles tendon-related injuries. Barefoot and minimal footwear running as well as forefoot and midfoot striking patterns are not the panacea for all running injuries. Clients seeking advice regarding transitioning from a rearfoot contact to a forefoot or midfoot contact running style should be cautioned to progress slowly to avoid lower extremity soreness or injury. All runners should be considered individually; consider the case report of a forefoot contact runner with shin splints who improved after the physical therapist changed the strike pattern to rearfoot contact (Cibulka, Sinacore, & Mueller, 1994).

Squadrone and Gallozzi (2009) compared the spatiotemporal, kinetic and kinematic variables between barefoot, Vibram Fivefingers® (VF) shoes, and traditional running shoes in the experienced barefoot runners (Squadrone & Gallozzi, 2009). Running in Vibram Fivefingers® shoes resulted in increased stride length, higher pressure under the metatarsal heads, higher thrust peak forces with decreased step rate from barefoot running. Barefoot runners tend to actively adopt a flatter foot placement at foot strike to reduce the local pressure under the heel (De Wit, De Clercq, & Aerts, 2000; Squadrone & Gallozzi, 2009). In regards to kinetics and pressure parameters, peak local pressure was significantly lower under the heel, midfoot, and hallux during barefoot and VF running as compared to conventional running shoes (Squadrone & Gallozzi, 2009). Surprisingly, peak pressure under the toes was significantly higher while running in VF as compared to running barefoot. Wearing VF while running mimics barefoot running and changes running patterns. Vibram Fivefingers® closely resembles barefoot running (Fig. 5); however, VF spatiotemporal parameters more closely matched standard running shoes than barefoot running (Squadrone & Gallozzi, 2009). Differences were found in kinetics, the vertical GRF during weight acceptance was significantly lower in VF as compared to standard running shoes, 1.59 BW and 1.72 BW respectively. This is more closely matched to barefoot running with vertical GRF of 1.62 BW.

Maximum oxygen consumption (VO_2) was significantly lower when running in Vibram Fivefingers® shoes compared with standard running shoes. Although the difference in running economy between shod and barefoot is not significantly different (Squadrone & Gallozzi, 2009); energy efficiency slightly favoured barefoot running (Divert et al., 2005; Squadrone & Gallozzi, 2009). In regards to kinematics, there was a significant difference in total range of motion at the ankle with more joint excursion occurring while running with VF than with standard running shoes, but, no significant differences were found at the knee joint. There is increased plantar flexion with significant increase in the stride frequency and significant decrease in stride length, and contact time in barefoot runners versus standard shod runners (De Wit et al., 2000). Subjects tend to dorsiflex more when landing with the standard running shoes compared to the VF and barefoot running (Fig. 5). Thus impact forces were significantly higher in shod runners as compared to running barefoot or in VF (Squadrone & Gallozzi, 2009). De Wit et al. (2000) reported greater knee flexion angles at initial contact during barefoot running in trained subjects at 7.8, 10.0, and 12.3 mph which contributed to "touchdown geometry". The angle of ankle, knee, and hip (Geometry of the leg) when the foot contacts the ground (touchdown geometry) influences the leg stiffness due to changes in the alignment of the ground reaction force vector relative to the joints. It also affects the muscle-tendon length and the level of required muscle activation when counteracting the ground reaction force (Agarwal & Gottlieb, 1977; Farley et al., 1998; Gottlieb & Agarwal, 1978; Greene & McMahon, 1979; Hunter & Kearney, 1982; Nielsen, Sinkjaer, Toft, & Kagamihara, 1994; Sinkjaer, Toft, Andreassen, & Hornemann, 1988; Weiss, Hunter, & Kearney, 1988; Weiss, Kearney, & Hunter, 1986a, 1986b). Hennig, Valiant, and Liu (1996) reported that runners tend to change their landing style to reduce lower impact forces when running with harder shoe soles (Hennig et al., 1996). One study reported that barefoot runners demonstrated greater leg stiffness as compared to shod runners throughout the stance phase of running. (De Wit et al., 2000; Divert et al., 2005).

Shod runners typically run with heel-to-toe gait patterns while barefoot or minimally supported runners tend to run with a toe to heel gait pattern. Barefoot runners tend to make foot contact in the sagittal plane with greater ankle plantar flexion (Lieberman et al., 2010) and knee flexion (De Wit et al., 2000). Despite these considerable differences, frontal and transverse plane movements may not be significantly different when comparing barefoot to shod running. One study concluded that tibiocalcaneal bone movements were not significantly different within subjects between shod and unshod running with the mean effect being less than 2° while the between subjects differences were as great as 10° (Stacoff, Nigg, Reinschmidt, van den Bogert, & Lundberg, 2000). The authors reported that calcaneotibial movement coupling were similar and concluded that calcaneal eversion and tibial internal rotation typically occurred together during initial contact and loading responses while calcaneal inversion and tibial external rotation occurred from midstance through take off (initial swing) in all runners regardless of shoe selection or lack of shoes while running (Stacoff et al., 2000). In this study it should be noted that the calcaneotibial movement coupling was only minimally affected by wearing running shoes, running shoes with modifications to the sole (including changes to the shape or size of the flare), or orthotics to support the arch; only the orthotic with calcaneal support altered the movement coupling (Stacoff et al., 2000).

8. Physical therapy

The physical therapist is a movement science expert and is the health professional most skilled at running gait assessment. The physical therapist will examine the runner's gait over ground or on a treadmill and may even use video recording to assess biomechanical faults and joint angles (Pink et al., 1994). The physical therapist should also assess the runner's shoe wear patterns and shoe fit (McWhorter et al., 2003). To properly treat and potentially prevent running injuries, physical therapists need a thorough understanding of running gait (Dugan & Bhat, 2005).

Supplementary video related to this article can be found at doi: 10.1016/j.ptsp.2011.09.004.

Running injuries are multi-factorial in origin and are numerous. Approximately 65% of chronic injuries in distance runners are related to routine high mileage, rapid increases in mileage, increased intensity, and hills or irregular surface running (Keller et al., 1996). A detailed history of the running and training background is essential while assessing runners. Consideration of resumption or a return to running may begin when normal day-to-day ambulation is pain free and must be individualized according to symptoms and physical findings; however, there are no studies that have compared different return to running regimens (Bennell & Brukner, 2005).

As running shoe trends change towards "minimalist" shoes and shoe manufactures scramble to bring their new products to the market, the physical therapists will be asked to weigh in on shoe recommendations. The typical conventional running shoe with the elevated heel has a 10–12 mm drop while the minimal support shoes may have a reduced drop of ≤4 mm. If the adaptation period is rushed, a habitually shod runner may potentially experience soreness and even injury when transitioning from traditional to minimalist running shoes. Runners will eventually adapt to the change in footwear; however, it is worth considering that the greater the change in the shoe drop, the longer duration of the acclamation period. In addition, habitual shod runners will be conditioned to perform a heel-to-toe running gait pattern and may have difficulty transitioning to a toe to heel pattern since approximately 75% of shod runners heel strike (Hasegawa, Yamauchi, & Kraemer, 2007). In the Lieberman et al. (2010) study, habitually shod runners tended to continue rearfoot striking at initial contact during barefoot running even on hard surfaces; eventually adapting a flatter foot (more plantar flexed foot) landing placement (Lieberman et al., 2010). If a physical therapist is consulting with a client that is transitioning to minimally supported shoes or to barefoot running, it is appropriate to discuss strike pattern (forefoot or midfoot) options as well as the duration of the transition period.

9. Conclusion

A thorough understanding of normal walking and running gait is integral in the prevention and proper treatment of running-related injuries (Dugan & Bhat, 2005). Although there are similarity between walking and running; there exists more dissimilarity. The main difference between barefoot and shod running is that the initial contact during barefoot running occurs on the forefoot or midfoot instead of the rear foot. Vibram Fivefingers® have similar properties as barefoot running but provides a thin protective interface between the runner's foot and the running surface. Despite some adverse factors, running shoes serve a protective role in certain lower chain pathologies and in aberrant environments as well as being able to accommodate a corrective orthotics. To date, there is no scientific evidence directly linking running shoes to injury; conversely, nor that minimally supported or barefoot running prevents injuries or enhances running performance. The knowledgeable physical therapist will assess running gait as well as the runner's training regime that could perpetuate an injury. With a clearer understanding of running and its complexities, the physical therapist will be able to better identify faults and create informed treatment plans while rehabilitating patients who are experiencing musculoskeletal injuries due to running.

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