FACTORS RELATED TO EXCESSIVE PATELLOFEMORAL LOADING IN REARFOOT RUNNING

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Running is recognized as one of the most popular exercise methods. Furthermore, running related injuries have been under the scope for the last few decades. Synchronous function between the segments of the lower limbs is necessary for efficient locomotion. Patellofemoral pain syndrome is a common exercise related syndrome and multifactorial in nature. The purpose of this study was to measure contact forces and frontal plane moments to detect the factors that are associated with atypically high patellofemoral joint loading in rearfoot striking (RFS) running pattern, and moreover, which could possibly contribute to development of the patellofemoral pain syndrome. 39 team sport female athletes with rearfoot striking pattern (confirmed with motion analysis by calculating the footstrike angle) were accepted to perform a shod running along a 15-m track at 4.0 m/s^2. An eight-camera system (Vicon T40, Vicon) recorded the marker positions and GRF data synchronously at 300 Hz and 1500 Hz, respectively. The results showed that patellofemoral contact force (PFCF) correlated significantly with hip extension and flexion moments as well as with knee extension moment (r= -0.576, -0.548, 0.967 respectively, $p<0.05$). Also spatiotemporal comparison showed that PFCF correlated negatively with the distance between the center of mass (COM) and heel contact (COM – heel distance, r= -0.350, $p<0.05$). Knee abduction moment correlated negatively with step frequency and positively with COM–heel distance (r= -0.329 and 0.355 respectively, $p<0.05$). In addition, knee abduction moment and maximum knee power had a significant positive correlation (r= 0.466, $p<0.05$). PFCF also correlated positively with knee flexion angle (r= 0.341), and a significant correlation existed between PFCF and maximum knee power (r= 0.467, $p<0.05$). It can be suggested that rearfoot striking pattern might contribute to development of the patellofemoral pain syndrome. It was shown that this running style stresses the patellofemoral joint and especially the medial side of the knee due to increased hip and knee extension moments, increased knee flexion angle and increased knee abduction moment. As the quadriceps' moment arm increased, the quadriceps was required to be more active to compensate the lack of activation of the hip musculature.

Keywords: Patellofemoral pain syndrome, rearfoot striking pattern (RFS), forefoot striking pattern (FFS), patellofemoral contact force, knee abduction moment, knee flexion angle

CONTENTS

1. INTRODUCTION

Human locomotion is a combination of complex synchronous actions of the joints and muscles for smooth outcome. Hip, knee and ankle joints and their ligamentous structures combined with the related bones and muscles need to be structurally certain for balanced movements. The main interest of this study is in the hip joint, patellofemoral joint and subtalar joint. If the movement between the lower limb segments is impaired, the forces acting upon the structures might become too high to withstand resulting in degenerative disorders or pain, and finally incapability to perform movement. Patellofemoral pain syndrome has been studied in the past decades, and the factors contributing to its development have tried to be mapped for preventing it to occur or rehabbing from the experienced pain. It has been shown that the nature of patellofemoral pain is multifactorial, and it is difficult to make conclusions what abnormality causes it, or is it the cause for abnormality. Some of the predisposing factors are structural anomalies in the hip, knee and/or ankle, like excessive anteversion, size and shape of the patella, subtalar joint and calcaneal orientation and increased Q-angle. Also leg length discrepancies, iliotibial band and/or calf musculature tightness, hip and thigh musculature weakness and/or their asynchronous activation might contribute to the development of the patellofemoral pain and degenerative disorders in the knee. Females tend to suffer from anterior knee pain more often than men, and in addition, physically active people are subjected to its development more often than sedentary people.

More recently, interest has risen upon the running techniques and their different loading patterns. The most of the runners use rearfoot striking pattern (RFS) in which the heel strikes the ground before the ball of the foot. On the contrary, the rest of runners habitually use forefoot striking pattern (FFS) including the ball of the foot hitting the ground before the heel. It is suggested that the first technique puts excessive load on the knee, whereas the latter one puts excessive stress on the ankle. In addition, step frequency and step length have been suggested to influence the loading pattern of the lower limb segments. It could be derived that the rearfoot striking pattern could, therefore, lead to patellofemoral pain, the development of degenerative disorders related to the knee, and finally increasing the risk for injuries.

For this reason this study was interested in the factors that might put body segments upon excessive forces during running especially when applying rearfoot running pattern.

2. ANATOMY OF HIP, KNEE AND ANKLE

Hip, knee and ankle have their own cruciate roles in human movement and locomotion. Lower extremities are connected to trunk via pelvic girdle, knee connects leg and thigh together, and ankle connects foot to leg. Movement in any of these parts influences actions elsewhere in the lower extremity, and this is why it is worth to examine the anatomy of these three main connecting parts of the lower body.

2.1 PELVIC GIRDLE

Pelvis, referred also as pelvic girdle (Figure 1), consists of the superior ilium, the postero-inferior ischium, and the antero-inferior pubis (Hamill & Knutzen 2009, 189). In other words, the pelvis is constructed by four main bones: the sacrum, two hip bones and the coccyx (Calais-Germain 2007, 43), and these bones are fully fused by the age of 20 to 25 years. Pubic symphysis connects the right and left sides of pelvis anteriorly, and this joint is supported by a pubic ligament. The pelvis is connected to the trunk and spinal column by the sacrolumbar joint, sometimes referred as lumbosacral joint, which is a very stable joint (Calais-Germain 2007, 56). Another joint within pelvis girdle is the sacroiliac joint, which connects the ilium and the sacrum, and also provides shock absorption for shear forces during the stance phase of gait (Hamill & Knutzen 2009, 190-191.) The pelvic girdle and the sacroiliac joint differ between females and males. Pelvic girdle of females is thinner, lighter and wider than that of males. (Hamill & Knutzen 2009, 188.) Females also have greater laxity in the sacroiliac ligaments which makes their pelvis more mobile compared to the pelvis of males, whose ligaments are thicker and stronger than females' (Hamill & Knutzen 2009, 191). The sacroiliac ligaments consist of sacrospinous and sacrotuberous ligaments inferiorly, and they act to oppose the adduction of the pelvis. The abduction of the pelvis is opposed by the posterior sacroiliac ligaments. (Calais-Germain 2007, 53).

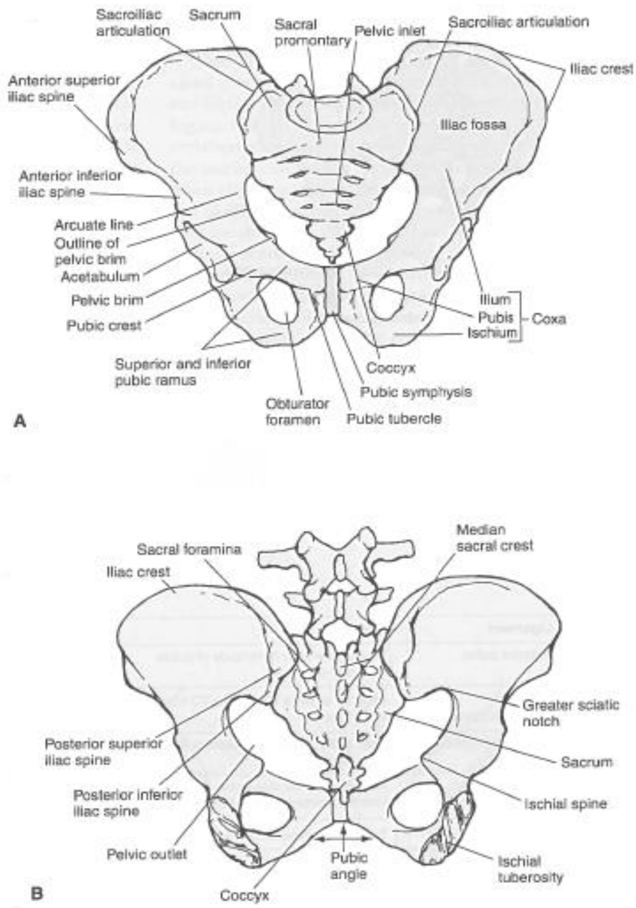


FIGURE 1. A. Posterior view and B. anterior view of the pelvic girdle. (Hamill & Knutzen 2009, 189.)

2.2 HIP JOINT

Hip is a stable joint (Figure 2) and mobile in three directions: flexion/extension, adduction/abduction, and medial/lateral rotation (Calais-Germain 2007, 194-197). Hip joint includes concave surface of acetabulum on the pelvis and head of femur. The articular surfaces are at the maximum contact when the pelvis and femur are at flexed and abducted position with respect to each other, and the femur is laterally rotating. (Calais-Germain 2007, 204.) The main ligaments of the hip joint are iliofemoral ligament, pubofemoral ligament and ischiofemoral ligament (Calais-Germain 2007, 206). Especially the femoral neck, which facilitates the congruence articulation within the hip joint and holds the femur away from the body, has a large contribution to balanced locomotion all the way through hip to ankle. The femoral neck is positioned at a specific angle with respect to the shaft of the femur in the frontal plane and is called the angle of inclination. This angle is approximately 125° within a range of 90° to 135° (Figure 3). (Hamill & Knutzen 2009, 195.)

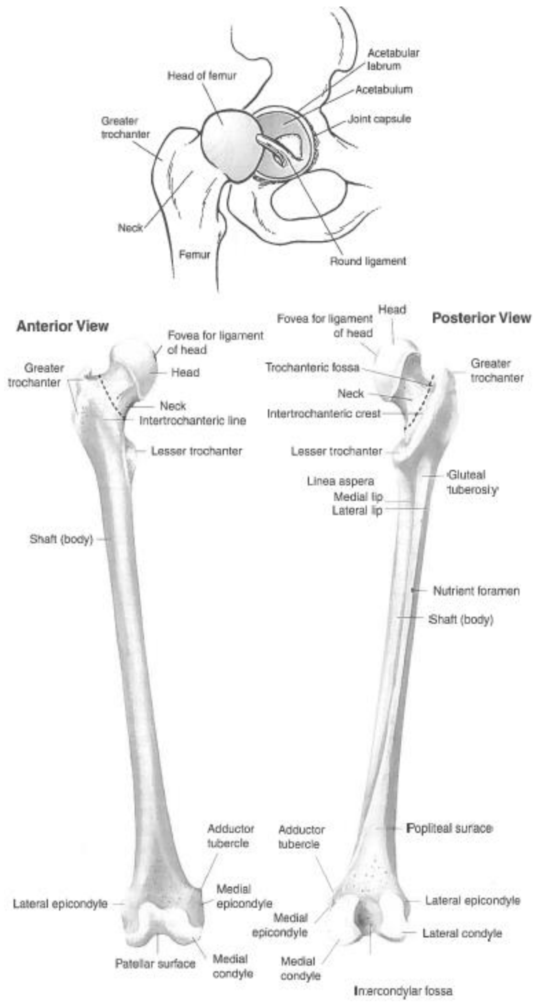


FIGURE 2. Anatomy of the hip joint consisting of the concave surface of the acetabulum and the head of the femur. (Hamill & Knutzen 2009, 194.)

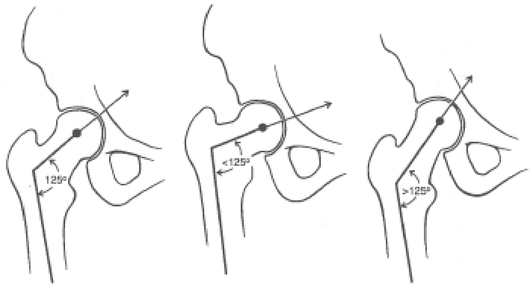


FIGURE 3.The angle of inclination. <125° means coxa vara and >125° means coxa valgus. (Hamill & Knutzen 2009, 195.)

If the angle is greater than 125°, the condition is called a coxa valga. This increased condition lengthens the limb, decreases the stress on the femoral neck, increases the stress on the femoral head and decreases the effectiveness of the hip abductors. The contrary occurs in the condition of coxa vara, when the angle of inclination is less than 125° (Figure 4). (Hamill & Knutzen 2009, 195.)

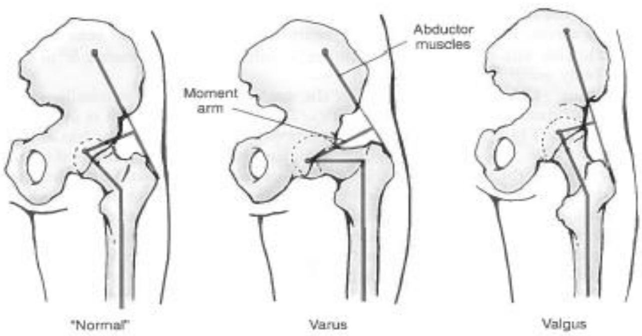


FIGURE 4. The length of the moment arm influences both the load on the femoral neck and the effectiveness of the hip abductors. Condition of coxa valga lengthens the limb, reduces the abductor activity, and increases the load on femoral neck at the same reducing the load on the neck. Vice versa happens in coxa varus. (Hamill & Knutzen 2009, 196.)

The angle of the femoral neck in the transverse plane is called the angle of anteversion (Figure 5). In a normal condition, the femoral neck is rotated anteriorly 12° to14° with respect to femur. Hip anteversion increases hip's effectiveness as an external rotator, and the mechanical advantage of the gluteus maximus is increased. In the case of the excessive anteversion in the hip joint the head of femur becomes uncovered, which leads to internally rotated posture or gait in order to keep the femoral head in the joint socket. This usually leads to increased Q-angle, which might lead to abnomalies of the lower body like genu varum or genu valgum (Figure 6), and finally to patellar malalignment and excessive pronation at the subtalar joint, which are explained in more detail later. (Hamill & Knutzen 2009, 195-196.)

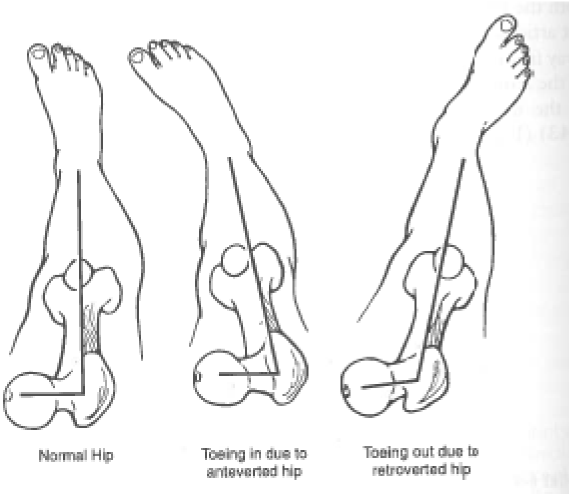


FIGURE 5. The angle of the femoral neck in the frontal plane is called the angle of anteversion. The angle being >14° the toe-in position occurs, and the angle being <12° the toe-out occurs (called retroversion). (Hamill & Knutzen 2009, 196.)

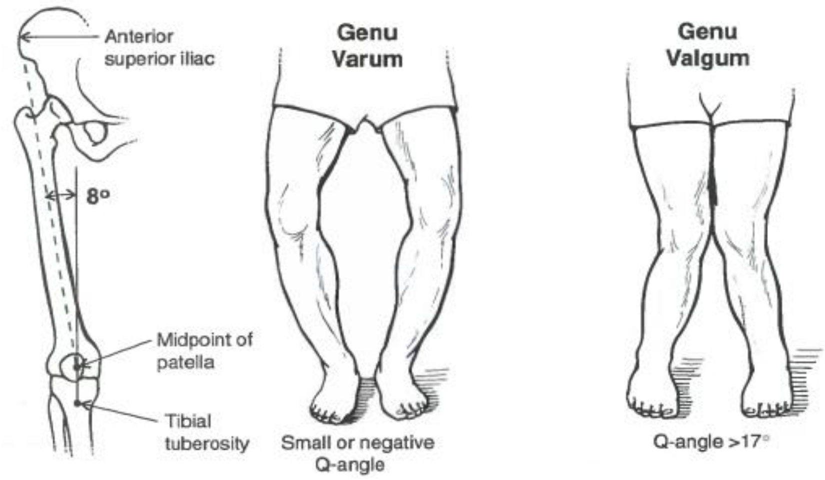


FIGURE 6. Q-angle and genu varum and genu valgum. (Hamill & Knutzen 2009, 213.)

2.3 KNEE JOINT

Knee locates between thigh and leg, and connects femur and tibia (Figures 7). The knee allows flexion and rotation of the leg and is designed to resist large loads to provide stability. Knee joint is a hinge joint consisting of three different articulations; tibiofemoral joint (sometimes referred as femorotibial joint), patellofemoral joint and tibiofibular joint (Figure 8 and 9). (Hamill & Knutzen 2009, 208.)

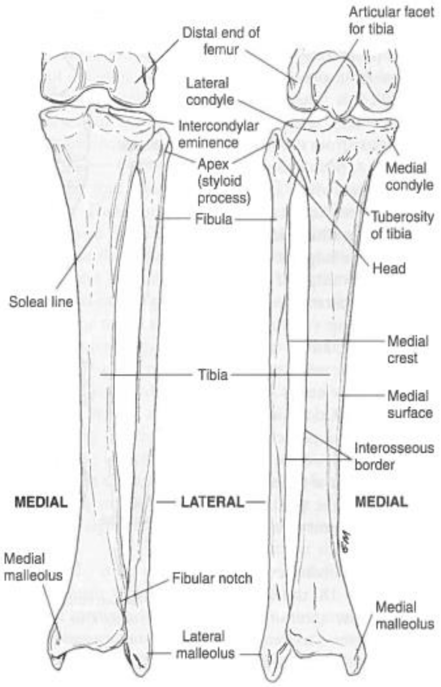


FIGURE 7.The anterior and the posterior views of the lower leg. (Hamill & Knutzen 2009, 209.)

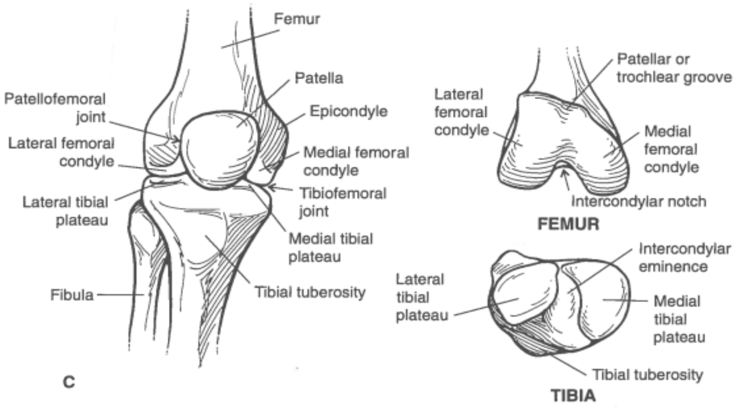


FIGURE 8. Anatomy of the knee. (Hamill & Knutzen 2009, 209.)

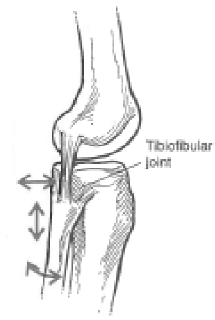


FIGURE 9. Tibiofibular joint. (Hamill et al. 2009, 214.)

The tibiofemoral joint is the largest and strongest joint of human body, and is often called a modified hinge joint that combines hinge and pivot joints. (Hamill & Knutzen 2009, 208.) Medial and lateral condyles of femur articulate medial and lateral tibial condyles, respectively (Goldblatt et al. 2003.) Above tibial condyles are epicondyles, which act as attachment sites of ligaments, muscles and capsules (Hamill et al. 2009, 210.) Also medial and lateral menisci act as intermediate between two joint plateaus (Goldblatt et al. 2003) and give stability by deepening contact surface of tibia. They are important in absorbing shocks and loads. Menisci are covered with lubracative material which decreases friction between femur and tibia. (Hamill & Knutzen 2009, 210.) Patellofemoral joint locates between femoral trochlea and patella, and it is responsible of the knee extension. Patella increases the moment arm by 15-30 %. (Goldblatt et al. 2003.) The patellar tendon connects patella to the tibial tuberosity. The tibiofibular joint is a small articulation joint between the head of the fibula and the tibial condyle. It rotates externally and superiorly in dorsiflexion of the foot. This joint is important in resisting static loads and to attenuate lateral tibial bending. (Hamill & Knutzen 2009, 214.)

The ligaments surrounding the knee are vital in giving the knee stability and resistance to pulling and rotating forces (Figure 10, 11, 12). First, the stability of the patella is provided, in addition to its location in the trochlear groove, with restrictive soft tissue elements as medial patellofemoral ligament, medial patellotibial ligament, medial patellomeniscal ligament, and medial and lateral retinaculum. Second, anterior and posterior cruciate ligaments (ACL and PCL) located in posterior part of the medial surface of the lateral femoral condyle and lateral surface of the medial femoral condyle, respectively, act as restraints to anterior and posterior translation of the tibia in relation to the femur. ACL also resists movements like internal rotation, varus, valgus and hyperextension, whereas PCL resists external rotation but also varus and valgus movements. Medial and lateral collateral ligaments (MCL and LCL) are located on the medial and lateral aspects of the knee. The MCL is important in resisting the external rotation and straight medial and lateral translation of the tibia. LCL resists varus movement in all flexion angles in addition to external rotation and posterior translation. (Hamill & Knutzen 2009, 211.) Also the iliotibial band (IT band), locating on the lateral side, has important role in acting against varus opening of knee. Finally, popliteus complex, locating in the posterior side of the knee, is a dynamic internal rotator of tibia. It acts against posterior tibial, translation, varus rotation and external rotation of tibia in relation to femur. (Goldblatt et al. 2003.) In summary, the bony structure of the femur, tibia and patella give the ability to resist high loads and pulling forces acting upon the knee, and ligaments, capsules and surrounding muscles crossing the knee joints stabilize the knee even further.

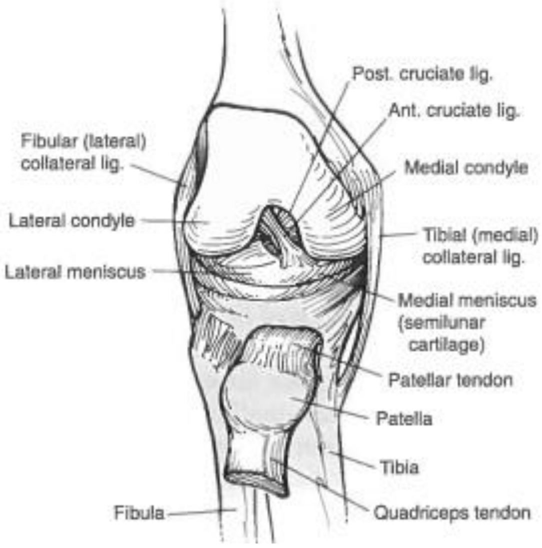


FIGURE 10. Anterior perspective of the knee. Ligaments and tendons of the knee. (Hamill & Knutzen 2009, 212.)

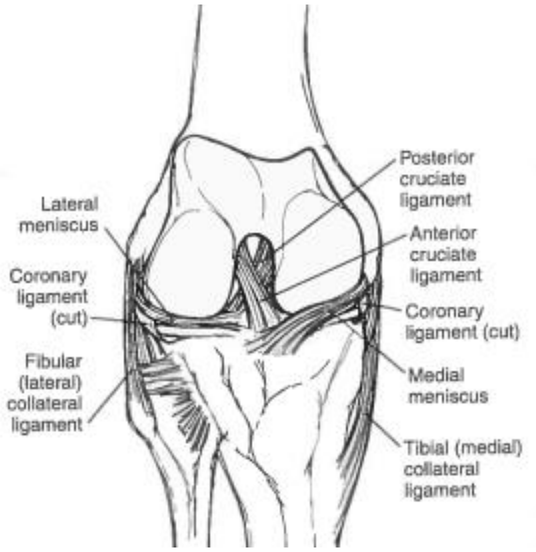


FIGURE 11. Posterior view of the knee. (Hamill & Knutzen 2009, 212.)

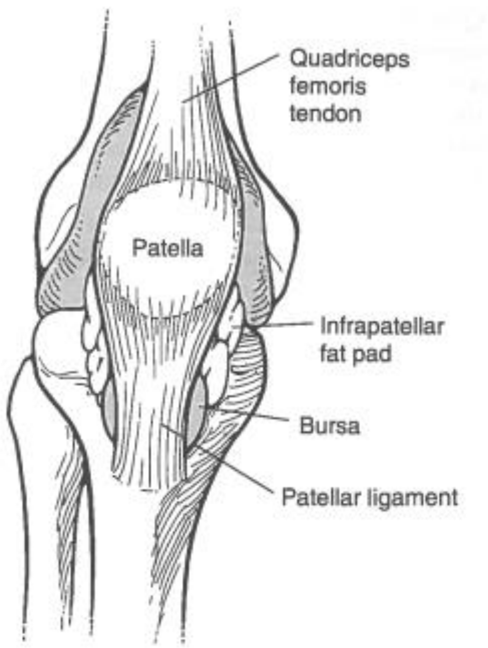


FIGURE 12. Anterior view of the patella. (Hamill & Knutzen 2009, 213.)

Figure 13 illustrates the range of movement of the knee. Like it has been noted above, the knee being a quite a stable anatomical construction it moves mainly in two planes: transverse and sagittal planes. Transverse plane consists of the flexion and extension of the knee, and sagittal plane of external and internal rotation of the knee.

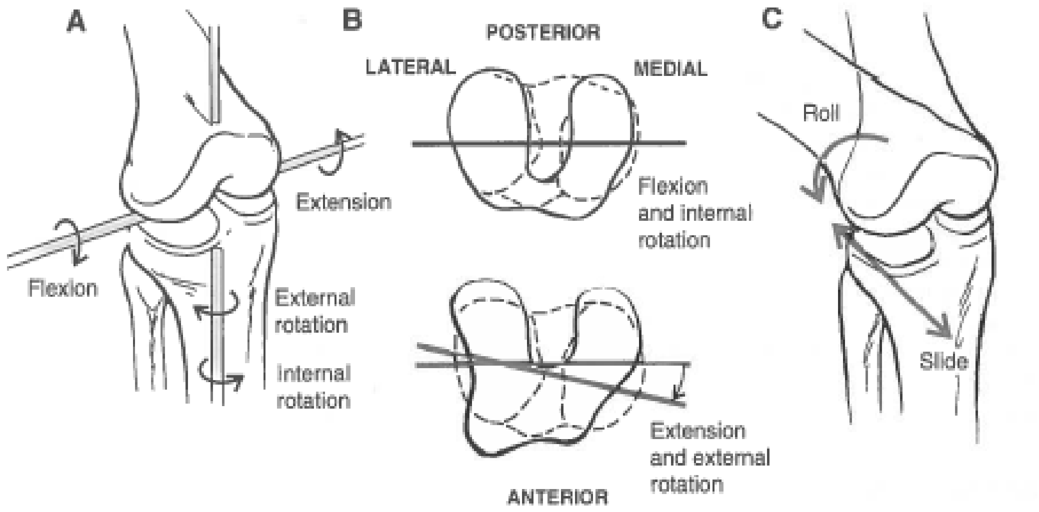


FIGURE 13. A. Flexion and extension, and internal and external rotation of the knee. B. In non-weight bearing condition flexion is accompanied with tibial internal rotation, and extension with tibial external rotation. C. Translatory movements of the femur on the tibial plateau surface. (Hamill & Knutzen 2009, 215.)

2.4 ANKLE JOINTS

The foot supports the weight of the body when standing and during locomotion (Figure 14). It consists of the rearfoot (talus and calcaneus), midfoot (navicular, cuneiforms and cupoid) and forefoot (metatarsals and phalanges). Most of the motion of the foot is provided by three synovial joints, the talocrual (ankle joint), the subtalar (talocalcaneal) and the midtarsal joint (transverse tarsal). Also the ligaments provide a high range of motion of the foot, while supporting the foot and resisting pulling and shearing forces acting upon it during ambulation (Figure 15). (Hamill & Knutzen 2009, 223.)

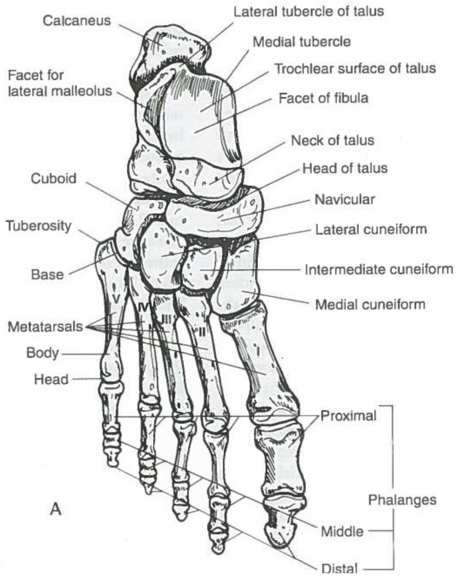


FIGURE 14. Superior view of the foot. (Hamill & Knutzen 2009, 224.)

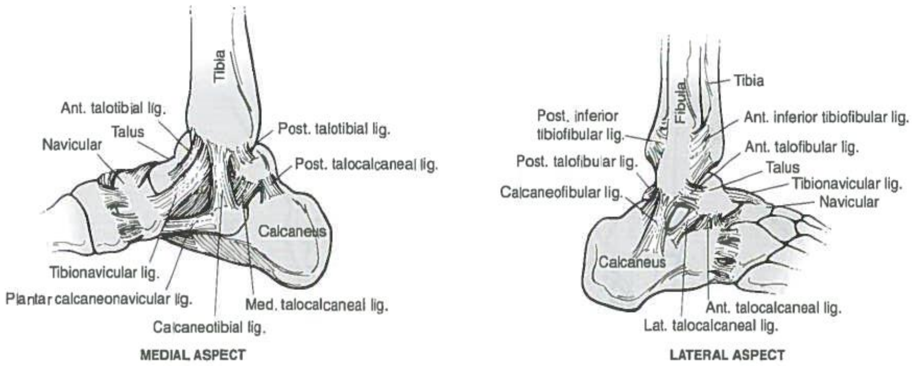


FIGURE 15. Ligaments of the foot. (Hamill & Knutzen 2009, 225.)

The talocrual joint is a combination of tibiofibular joint, formed by tibia and fibula, and tibiotalar joint, formed by tibia and talus. Talocrual joint is designed more to provide stability rather than for mobility, but it provides dorsiflexion as the foot moves toward the leg or vice versa. Subtalar joint exists distally from the talocrual joint, and is formed by the articulation between the talus and calcaneus. This joint provides the pronation and supination of the foot (Figure 16). (Hamill & Knutzen 2009, 223-224.)

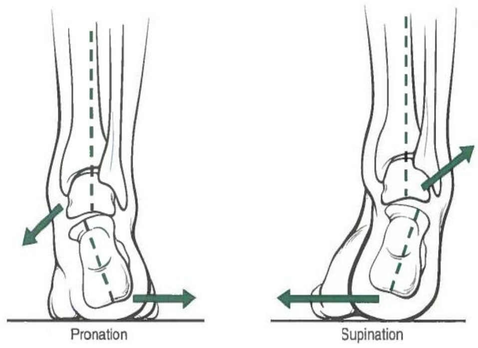


FIGURE 16. Pronation and supination when the foot is in contact with the ground. (Hamill & Knutzen 2009, 227.)

Pronation is defined by movements like abduction, dorsiflexion and eversion. (Donatelli et al. 1985.) During pronation subtalar joint responds to medial shear and internal rotation by allowing the calcaneus to move into valgus position (laterally). In addition, the talus moves medially in order to fully articulate with the middle facet of the calcaneus. (Root et al. 1977.) Supination can be defined as adduction, plantar flexion and inversion (Donatelli et al. 1985, Root et al. 1977). Supination is initiated by the inversion of the calcaneus and the talus is pushed into a lateral position. The main function of the subtalar joint is to absorb the internal rotation of the femur and tibia at the beginning of the stance phase and external rotation at the end of the stance by rotating through opposite actions of pronation and supination. Subtalar joint is important in shock absorption by pronation, and in addition, it allows the tibia rotate internally faster than femur and preventing the locking at the knee joint. (Root et al. 1977.)

The midtarsal joint, consisting of calcaneocuboid joint and talonavicular joint, contribute to inversion and eversion, abduction and adduction, and dorflexion and plantarflexion at the subtalar and talocrual joints. From the heel strike to flat foot the motion of the midtarsal joint is unrestricted as the subtalar joint allows pronation. The midtarsal joint locks up by the supination of the subtalar joint providing a rigid lever for push-off. (Root et al. 1977.) Range of movement of the subtalar joint is often measured by calcaneal eversion and inversion, which is considered to be a good indicator of the pronation (Donatelli et al. 1985). Normal ranges for inversion and varus orientation of the subtalar joint varies from 20° to 25.4° and for eversion and subtalar valgus orientation from 10° to 12°. (James et al. 1978, Tiberio et al. 1987.) High arch, also known as pes cavus, is thought to be inflexible, and in contrast the flat feet, pes planus, tends to be hybermobile and easily leads to pronation (Subotnick et al. 1985). Also rearfoot pronation angle is often measured when detecting the reasons for overpronation (Hamill et al. 2009). If the foot is not able to resupinate at the end of the stance phase coupled with subtalar locking providing balance and rigid push-off lever (Subotnick et al. 1985), the loss of stability could cause trauma to the foot. In addition the resultant forces are transmitted to the leg, knee and hips (Franco et al. 1987). When the pronated foot is unprepared to absorb the ground reaction forces, the muscles and ligaments are put upon greater stress and inefficiency causes loss of balance (Franco et al. 1987, Gurney et al. 2002).

2.5 MUSCLES OF LOWER EXTREMITY

Joint movements of lower limbs consist of flexion and extension, abduction and adduction, internal and external rotations, and at the very distal site ankle dorisflexion and plantarflexion, pronation and supination. The muscles are essential in production of these movements (Figure 17, 18, 19). Iliopsoas is the main hip flexor, but rectus femoris, sartorius, pectineus and tensor fascia latae also contribute to flexion of the thigh. The main extensor of the hip is hamstring complex. It consists of semimembranosus and semitendinosus, and lateral hamstring, the biceps femoris, where the former two muscles are not as active as the last one. However, in the case of more vigorous hip extension, the gluteus maximus is in the important role. The extensors are responsible of keeping the trunk in upright position. The main abductor of the thigh at the hip joint is gluteus medius, and secondary abductors are gluteus minimus, tensor fascia latae and piriformis. Abduction moves the thigh laterally in the horizontal plane, and foot being on the ground they move the pelvis on the femur in the frontal plane. In the contrary movement, the adductor muscles include gracilis, adductor longus, adductor brevis, adductor magnus and pectineus. Adductors bring the thigh across the body and are active in the swing phase. External rotation of the thigh reserves as power productor. Primary external rotators are gluteus maximus, obturator externus and quadratus femoris. Gluteus medius and minimus are two main muscles responsible for the contrary movement; internal rotation. (Hamill & Knutzen 2009, 201-202.)

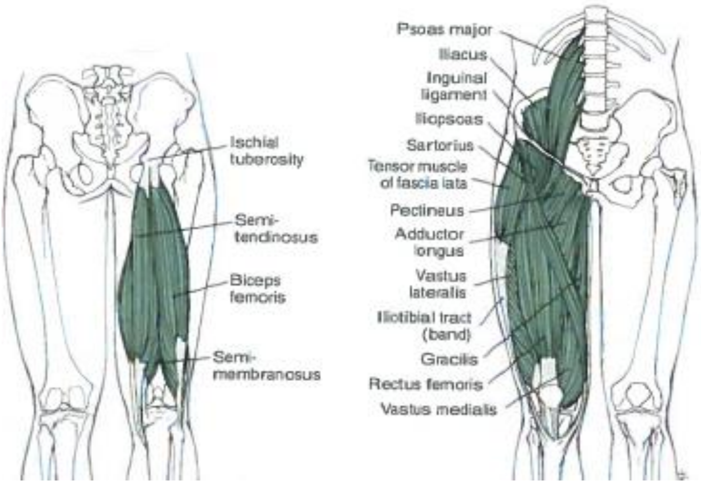


FIGURE 17. Muscles of the thigh. Flexors and extensors. (Hamill & Knutzen 2009, 200.)

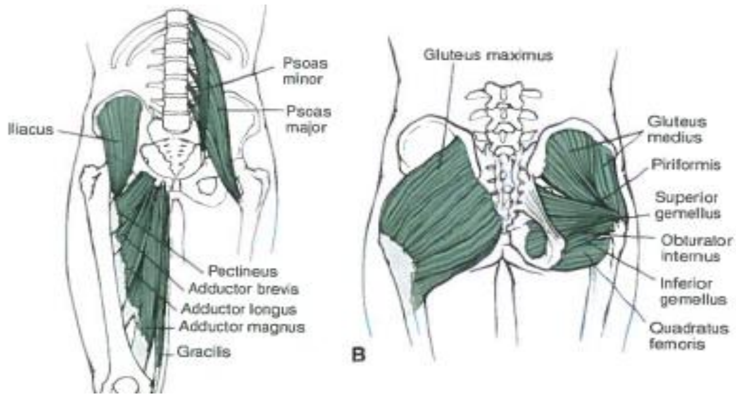


FIGURE 18. Adductors and abductors. (Hamill & Knutzen 2009, 200.)

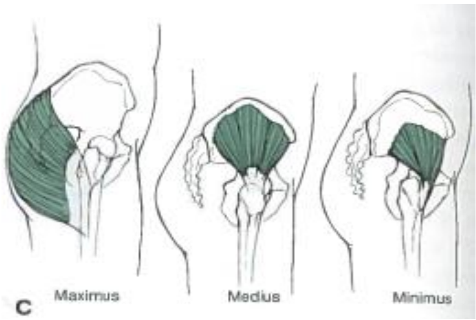


FIGURE 19. Gluteus muscles. (Hamill et al. 2009, 200.)

The movement of the knee is affected by the hip movement and position. The knee extension is in important role in any translation. The main muscle group responsible for knee extension is quadriceps femoris consisting of rectus femoris, vastus intermedius, vastus lateralis and vastus medialis. In addition, the latter consists of vastus medialis longus and vastus medialis obliquus. This great muscle complex acts also eccentrically as a decelerator of quickly flexing knee. Flexion of the knee occurs during the swing phase by slowing down the rapidly extending leg. During the support phase the knee flexion is controlled by the extensors to ensure stable downward movement. Hamstrings are responsible of the knee flexion. The greatest force is produced in 90° knee flexion and also when hip is in flexed position. Hamstrings also work with ACL to resist anterior displacement of tibia. External rotation of the lower leg is produced mainly by biceps femoris (responsible especially for external rotation of the tibia), and internal rotation mainly by semimembranosus and semitendinosus. (Hamill & Knutzen 2009, 216-217.)

The foot and ankle include 23 muscles and they have an important role in sustaining very high impact loads. Plantarflexion provides forward, translational and upward motions of the body, and occurs in heel-off and toe-off phases. It is often accompanied by supination and adduction. The main plantarflexion muscles are gastrocnemius and soleus (Figure 20). The reduction in plantarflexion is usually caused by weakness in these aforementioned muscles. The dorsiflexion is active in the swing and stance phases when the foot is lowered to the floor at the heel strike (Figure 21). It is mainly produced by tibialis anterior, but extensor digitorum longus and extensor halluces longus work in assistance. Eversion is produced by the peroneal muscle group, also known as pronators. Inversion is created by the tibialis anterior and tibialis posterior with assistive toe flexors. (Hamill & Knutzen 2009, 232-235.)

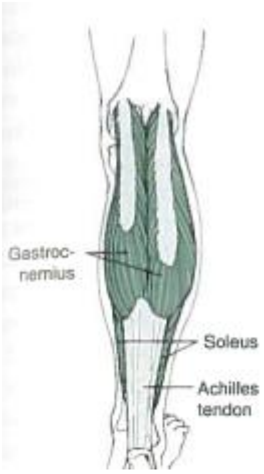


FIGURE 20. Posterior view of the muscles of lower leg. (Hamill & Knutzen 2009, 233.)

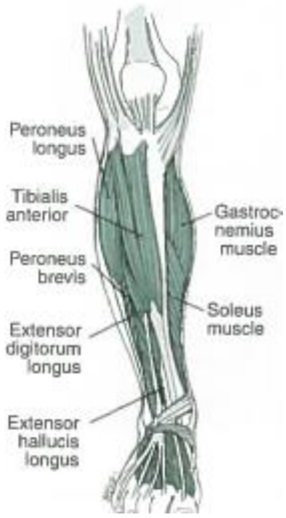


FIGURE 21. Anterior view of the muscles of lower leg. (Hamill & Knutzen 2009, 234.)

3. BIOMECHANICS OF LOWER EXTREMITY AND LOCOMOTION

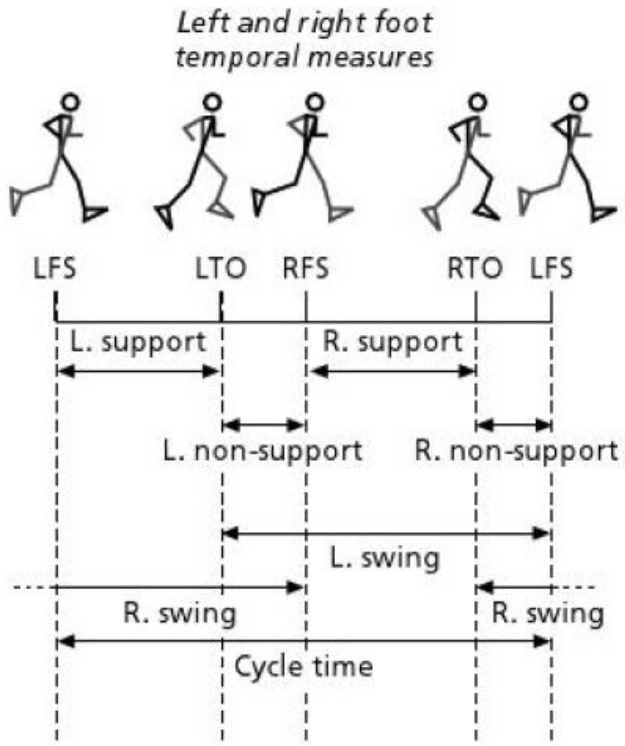
Normal locomotion consists of two phases; swing and stance phases that construct gait cycle. Stance phase is now under the scope, because at this time the foot is in contact with the ground and put upon external forces. The stance phase can further be divided into three divisions; contact, midstance and propulsion phases. First, the contact phase starts with heel strike and continues until the whole foot is on the ground (foot is flat). In the second phase the midstance extends from here until the heel lifts off the ground. Finally, the propulsion phase begins when the heel rises and ends when toes are off the ground. (Tiberio et al. 1987.) At the moment of the heel strike the subtalar joint is slightly supinated, which is followed by pronation (Root et al. 1977) and the leg slightly flexes (Donatelli et al. 1985).

3.1 WALKING

Being more specific about the range of movement of the hip, knee and ankle during walking, the angles can also be specified. During the swing phase of walking the hip has 8°-10° of external rotation, 12° of abduction and 35°-40° of flexion later in the phase. The knee has 60°-88° of flexion, 12°-17° of rotation, and 8°-11° of valgus during the swing phase, and the ankle is in neutral position. Just before the heel strike and throughout the support phase the hip has 4°-6° of internal rotation, 12° of adduction in stance. At the heel strikethe knee is flexed by 5°-8°, and the ankle and has 10° of plantarflexion and 2°-3° of supination. During the support phase the knee is flexed 17°-20°, has 5°-7° of internal and 7°-14° of external rotation, and 3°-7° of varus. In the mid stance the ankle is dorsiflexed by 5°-10°, has 3°-10° of pronation and 6°-7° of calcaneal eversion. Until the heel-off the ankle has 3°-10° of supination. At the toe-off the ankle is plantarflexed 20° and has 4° of calcaneal eversion, while knee and hip are fully extended. In other words, when the foot makes contact with the ground, the knee flexes and the foot is in a slight supination and plantarflexed. Subtalar pronation starts immediately accompanying internal rotation of the hip and knee joints. The stress is put upon the medial side when the maximum pronation is reached at 35°-50° of the stance phase. After this phase the knee joint begins to extend and externally rotate forcing the subtalar joint to start resupinate. (Hamill & Knutzen 2009, 198.)

3.2 RUNNING

Running differs from walking in various ways, even though it includes similar phases in gait cycle compared to walking (Figure 22). First, the stride length and rate increase compared to walking (Luhtanen et al. 1978). On the contrary the cycle time and time spent in support decreases in running. Before the foot strike, extension of the hip has begun. However, there is a slight period of flexion after the foot has made contact with the ground, and the hip movement quickly resumes extending. (Nilson et al. 1985.) In addition, also the knee joint performs two periods of flexion; first during the swing phase and second during the support phase. The first happens so that the leg moment of inertia can be reduced, which helps the swinging of the leg. The range of motion can be specified in the same way as was done with walking. During the swing phase the knee has 80° of flexion and 8° of valgus. During the support phase the flexion angle increases till 36°, internal and external rotations are 8° and 11°, respectively, and varus angle is 8°. During running like in walking the ankle stays in neutral position during the swing phase, but prior to contact it is dorsiflexed by 10°. In the midstance the ankle has 50° of dorsiflexion and 8°-15° of pronation. At the toe-off phase the ankle switches from 50° of dorsiflexion to 25° of plantarflexion. (Hamill & Knutzen 2009, 198.)



FRIGURE 22. Human locomotion in running. LFS=left foot strike, LTO=left toe off, RFS=right foot strike, RTO=right toe off. (Zatsiorsky 2000, 163.)

3.3 RUNNING STYLES

Runners can roughly be divided into two or three groups: rearfoot strikers (RFS), who contact the ground with the heel, midfoot strikers or forefoot strikers (MFS or FFS), who contact the ground with the heel and ball simultaneously, and forefoot strikers, who contact the ground first with the ball of the foot. Ground reaction forces and center of pressure vary in three different directions during the support phase; vertical, antero-posterior and medio-lateral, when changing the running speed and the type of running. The vertical peaks tend to vary the most among the rearfoot strikers and forefoot strikers. Rearfoot runners demonstrate two force peaks during support phase, in the contrary to rearfoot runners who exhibit just one force peak (Figure 23). The first peak is called an impact peak that is present when the foot hits the ground. The second peak is called an active peak when the muscle activity is present. (Nigg 1986.) It has been shown that the first vertical force peak is 2,2 (+/- 0.4) BW for rearfoot strikers and the second vertical peak had a mean of 2,8 (+/- 0,3) BW for rearfoot runners and

2,7 (0+/- 0,2) BW for midfoot runners at running speed of 4,5 m/s (Lafortune 1982). The first vertical force peak has been shown to correlate with faster running speed. In general it has been noticed that when the running speed increases, also the vertical ground reaction force increases, as well as the loading rate, and the initial contact with the ground moving forward within the foot. Also antero-posterior impulse increases with faster running speed. (Frederick and Hagy 1986.)

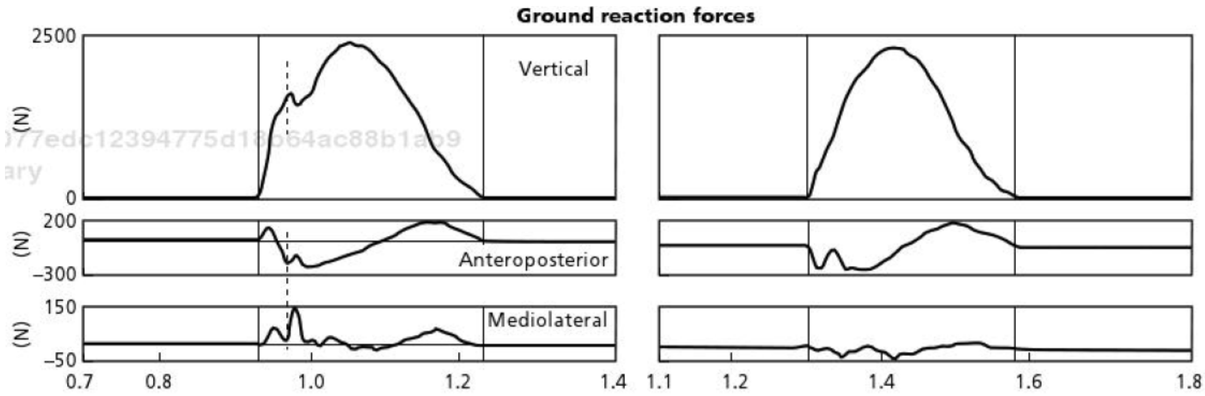


FIGURE 23. Ground reaction forces when running with rearfoot striking pattern (left) and with forefoot striking pattern (right). (Zatsiorsky 2000, 168.)

In general, it has been said that 75% of the runners naturally adopt the rearfoot technique. The forefoot pattern has been shown to decrease the risk of knee injuries, but the superiority of this technique is not self-evident. The forefoot strikers usually demonstrate higher risk for developing ankle-related injuries like Achilles tendinopathy and triceps surae eccentric loading injuries. (Strearne et al. 2014.) During the first 30 ms of the foot strike rearfoot strikers demonstrate an extensor moment of the hip and flexor moment of the knee. However, ankle net moment stays near to zero. Also these runners usually exhibit a sudden unloading of the Achilles tendon. Midfoot strikers (usually also counts for forefoot strikers) demonstrate an immediate drop in the hip extensor moment and knee flexor moment towards the zero, but the ankle shows an extensor moment in the beginning of the stance. (Zatsiorsky, 2000, 169.)

Stearne et al. (2014) studied the running mechanics of rearfoot or forefoot patterns in competitive runners who were using either of the techniques or were imposed to a contrary technique they naturally exhibit. The results showed that the habitual rearfoot strikers demonstrated greater stride and stance time, and smaller stride frequency than habitual forefoot strikers. The first two already refer to higher loading pattern and ground reaction forces in the rearfoot running pattern. Peak instantaneous power absorption at the knee was greater in RFS runners compared to FFS runners, and also when the FFS runners were imposed to RFS pattern. Peak knee extension moments did not differ between habitual conditions, but significantly increased when habitual FFS runners were imposed to RFS. Knee abduction moment was significantly greater in habitual RFS runners compared to FFR. Moments and powers at the hip joint were no different in any condition between the two techniques. RFS runners had significantly greater percentage contribution of the knee joint to total negative lower limb work and average power during stance phase than FRS runners. Peak instantaneous ankle power absorption was significantly greater in FFS runners compared to RFS. It increased when RFS runners were imposed to FFS technique and decreased when the FFS runners were imposed to RFS. Even though, there was no difference in ankle internal rotation between the two habitual patterns, it increased when habitual RFS runners were imposed to FFS. Peak plantarflexion moment was greater in all conditions with respect to FFS pattern. Ankle negative average power was greater in habitual FFS runners compared to habitual RFS, and increased when the RFS runners were imposed to FFS. Total positive and negative average powers of the lower limb were greater both in habitual and imposed FFS conditions compared to habitual or imposed RFS. In summary, the rearfoot striking pattern demonstrates greater knee abduction and extension moments, but produces less power than forefoot striking pattern. Forefoot striking pattern includes greater plantarflexion moment and ankle power, and requires more energy in total than RFS pattern. In other words, RFS technique puts stress on the knee, but FFS puts more stress on the ankle and can be more energy consuming overall compared to RFS technique. (Strearne et al. 2014.)

In addition, Kulmala et al. (2013) also compared forefoot and rearfoot strikers and whether these runners would exhibit different lower limb loading profiles. Firstly, rearfoot and forefoot strikers did not demonstrate any differences in the hip abductor strength, navicular drop or tibiofemoral angle. However, during the stance phase FFS had shorter contact time and COM (center of mass) – heel distance than RFS. FFS strikers demonstrated lower peak hip adduction and hip abduction moment than RFS. RFS runners had greater knee flexion angle than FFS, which led to increased quadriceps moment arm. This in turn, led to increased eccentric muscle force requirement in order to resist the braking phase. This is suggested to cause greater patellofemoral contact force (PFCF) and joint stress (PFJS), which were inevitably greater in RFS runners compared to FFS. In addition, RFS runners had greater knee extension moment than FFS, but at the ankle level the FFS had less dorsiflexion at the initial contact, greater plantarflexion moment and greater Achilles tendon force than RFS during the stance phase. Impact peak and loading rate were lower in the FFS runners than in RFS. (Kulmala et al. 2013.)

4. DEFINITION AND DEVELOPMENT OF PATELLOFEMORAL PAIN SYNDROME

Patellofemoral pain syndrome (PFPS) is defined as anterior knee pain that refers to the pain within the anterior aspect of the knee that includes peripatellar area, distal patella pole region and central part of patella (Brushøj et al. 2007). This symptom is also sometimes referred as chondromalacia patellae, intra-articular patellar chondropathy, patellar arthralgia, patellar subluxation, jumper's knee, runner's knee, and pain from the hip and or the saphenous nerve (Witvrouw et al. 2005, Malek et al. 1981). Patellofemoral pain syndrome is often also referred as an overuse and overload injury where large forces repetitively load the knee. Pain is usually felt after prolonged sitting knees flexed, during stair climbing and/or descending, squatting and kneeling. (Lankforst et al. 2012, McConnell et al. 1996, Brushøj et al. 2007.)

4.1 CAUSES AND DEVELOPMENT

4.1.1 Predisposing structures

Some of the predisposing factors are acute trauma, overuse (increased retinacular, subchondral and cartilage stress), knee ligament injury and surgery, instability, overweight, genetic predisposition, malalignment, dysfunction of the knee extensor mechanism, recurrent hemorrhage into the joint, anomalies of patella, prolonged synovitis, and immobilization (Natri et al. 1998, Keller et al. 2007). Knee anomalies are considered to contribute to the development of patellofemoral pain. The patella is the largest sesamoid bone in human body providing a fulcrum for static and dynamic stabilization forces. The posterior articular cartilage of patella is divided by vertical ridge into two parts; a large lateral facet and smaller medial facet. According to Wiberg and Baumgartl the patellar shape of 1 and 2 are the most stable and the rest; 3, ¾ , 4 and 'Jägerhut' are shapes that signify more unstable conditions (Malek et al. 1981), meaning that if the shape of patella or the articular cartilage differ from these two most stable configurations, the stability would be diminished. Wiberg variations can show a loss of the medial facet of the patella, leaving the lateral facet responsible for the contact with the femur (Amis et al. 2007). However, 60% of the joint load is normally sustained by the lateral facet, which is coped with a larger contact area from lateral to median ridge. In full extension the patella is not in contact with the trochlear groove, and patella is dependent on the soft tissues on the lateral aspect of the knee to maintain stability. When the knee flexes three forces act upon it; patellar tendon force, quadriceps force and the contact force. These forces pull the patella posteriorly, which provides the stability. As the knee flexes, the contact force increases with the patella tension. This stability factor, however, counts more than the depth of the trochlear groove. (Amis et al. 2007.)

Some of the signs of malaligment are detected via abnormal sulcus angle, congruence angle, lateral patellofemoral angle and patellofemoral index (Keller et al. 2007). Malalignments like patella alta is a condition where patella is located too high. It can be measured using Insall-Salvati Index or Blackburne-Peel Index. Subluxation can be detected by several tests, and for example, an abnormal sulcus angle has its contribution in patellofemoral pain. (Malek et al. 1981.) Sulcus angle is defined as an angle posterior to patella and locates between the slopes of articular facets (Amis et al. 2007), more specifically the angle formed by the highest points of the lateral and medial femoral condyles and the lowest point of intercondylar sulcus (Powers et al. 2000). In normal knee the deepest portion of the intercondylar groove typically overlies the midpoint of the posterior condyle interval (Powers et al. 2000). Both the steepness of the slopes and depth of trochlear groove have their effect on stability (Amis et al. 2007). It has been suggested that a large sulcus angle, greater than 150° would correlate with the patellofemoral symptoms (Amis et al. 2007), when the normal range is 138° +/- 6° (Tecklenburg et al. 2006).

Medial and lateral patellar displacement can be determined by the “bisect offset” measurement. In these method the bisect offset is measured by drawing a line connecting the posterior femoral condyles and projecting another perpendicular line anteriorly through the apex of the trochlear groove. Lateral patellar alignment compared to perpendicular line is expressed as a percentage of the total patellar width (Powers et al. 2000), or by congruence angle which normally range from -6° to +6° (Keller et al. 2007). In general, the patellar displacement increases with a larger sulcus angle especially in terminal extension (Powers et al. 2000). Also medial and lateral patellar tilt can be measured by defining the lateral patellofemoral angle which is formed between a line drawn over the highest part of both condyles and a line drawn along the lateral patellar facet. An angle that is laterally open is considered as normal, whereas parallel lines refer to subluxation of patella. (Keller et al.

2007.) The patellar tilt is also usually reported as degrees (Powers et al. 2000). It has been suggested that subjects suffering from the patellofemoral syndrome have a greater patellar tilt angle in flexion compared to people without patellofemoral syndrome, and as in case of patellar tilt in relation to sulcus angle mentioned above, greater sulcus angle is a predictor of greater amounts of patellar tilt (Powers et al. 2000).

Finally, patellofemoral index, defined as a ratio between the shortest distance between the medial condyle and articular ridge and the shortest distance between the lateral condyle and lateral facet, is a type of measurements that can be used when detecting subluxation. The ratio of 1.6 or less is normal and otherwise it indicates lateral patellar compression. (Keller et al. 2007.) In addition, the medial patellofemoral ligament and tibial tubercle lateralization can be assessed (Keller et al. 2007). Also patellar deviation and tibial tubercle deviation can be observed and usually they are associated with patellofemoral pain (Keller et al. 2007).

Pain is also examined with a few widely known tests. ‘Dynamic patellar tracking test’ tests active instability and tracking when the patella in normal condition is supposed to glide into trochlear groove in extension. ‘Patellar compression test’ detects pain and abnormal grinding when patella is moved superiorly and inferiorly, against the trochlear groove. ‘Patellar tilt test’ detects the movement of lateral edge of the patellafrom the lateral femoral condyle. If the border does not move over the horizontal line the test is considered as positive for tightness of the lateral retinaculum. ‘Apprehension test’ is positive when the patient becomes resistive due to pressure exerted from medial to lateral direction of the patella. Also quadriceps and iliotibial band flexibility are often tested, in addition to the strength of the quadriceps and hip. (Keller et al. 2007.)

4.1.2 Q-angle

The contribution of quadriceps-angle, Q-angle (Figure 24), to the development of patellofemoral pain syndrome is debated. Q-angle is defined as an angle measured between the line connecting the anterior superior iliac spine to the center of patella, and the extension of a line from the tibial tubercle to the same reference point of the patella (Park et al. 2011). In other words, it is an angle between the quadriceps load vector and the patellar tendon load vector, and it refers to the orientation of quadriceps muscle force acting on the patella

(Mizuno et al. 2006). The normal mean value for Q-angle is 15° (Park et al. 2011, Mizuno et al. 2006), and angles smaller or bigger than 15° are usually considered as abnormal and could contribute to the development of patellofemoral disorders via malaligment and maltracking of the patellofemoral joint. In the case of the normal alignment the contraction of quadriceps creates a lateral force vector on the patella. This occurs because the resultant quadriceps force vector and the patella tendon force vector are not collinear. (Powers et al, 2003.)

In addition, it has been shown that increases in Q-angle (when the knee is flexed beyond 20°) can result in increased lateral facet pressure when the patella is being forced against the lateral femoral condyle (Hestroni et al. 2006). It is hypothesized that this lateral maltracking increases the risk of the subluxation and dislocation of the patella, in addition to patellofemoral pain and degeneration of articular cartilage (Mizuno et al. 2006). An in vitro study with six cadaver knees showed that when the Q-angle was increased the patella shifted laterally and rotated medially at low flexion angles of the knee. It was suggested that medial retinaculum acted to limit the patellar shift and rotation. As the flexion angle increased, the patellar shift and rotation decreased due to articular constraints within the trochlear, in other words in large flexion angles patella sits onto the trochlear groove. Medial tilt increase was consistent at all flexion angles due to patellar riding along the lateral ridge of the trochlea. It was also suggested that lateral retinaculum limit the magnitude of medial tilt throughout flexion. It could be concluded at some extent that this altered tracking pattern increase the lateral patellofemoral contact pressure. The same study also noticed that as the Q-angle was decreased, patella tilted laterally due to tracking along the medial trochlear ridge, but the magnitude of lateral tilt increase was smaller than the medial tilt increase with larger Q-angles. In addition, the medial patellofemoral contact pressure was smaller than lateral patellofemoral contact pressure when Q-angle was decreased and increased (Mizuno et al. 2006).

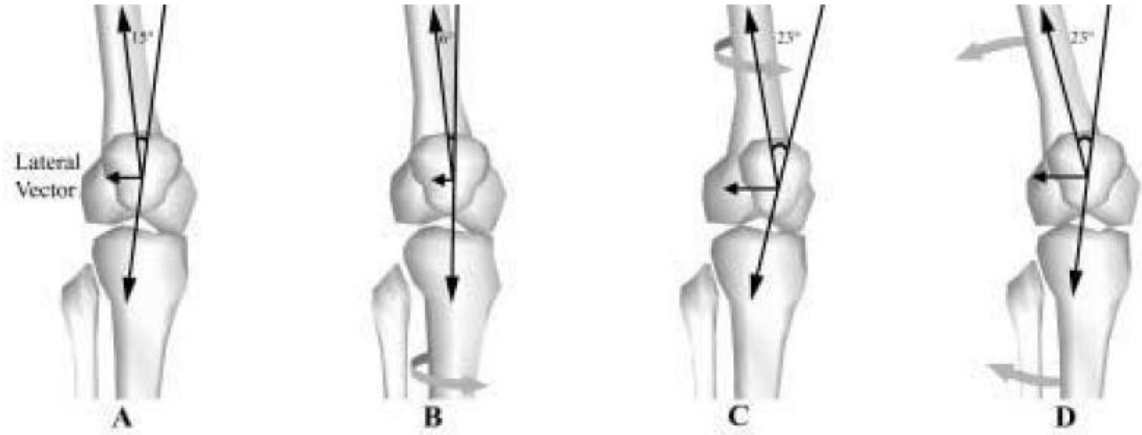


Figure 24. **(A)** Q-angle measured as an angle between the quadriceps load vector and the patellar tendon load vector. Normal alignment of the tibia and femur creates an offset in the quadriceps force vector and patellar tendon force vector which results in a lateral force vector acting on the patella. **(B)** The Q-angle and lateral force vector decrease as the tibia internally rotates. **(C)** Femoral external rotation increases the Q-angle and the lateral vector. **(D)** Q-angle and lateral vector increase by knee valgus. (Powers 2003.)

4.1.3 Valgus and varus orientation

Valgus and varus orientations (Figure 6) of lower limbs are also associated with changes in Q-angle. Knee valgus is also called medial knee displacement, and knee varus is called lateral knee placement. In knee valgus the knee is twisted outward and vice versa in knee varus. When the Q-angle is decreased the tibiofemoral valgus angle also decreases significantly. This orientation referring to more varus alignment could actually increase the medial contact pressure in the tibiofemoral joint. (Mizuno et al. 2006.) Because females tend to have a greater Q-angle by nature, they also tend to suffer from patellofemoral pain syndrome more often than men. In general, it has been suggested that females with patellofemoral pain syndrome have abnormal hip and knee mechanics which refer to excessive contralateral pelvic drop, increased (peak) hip internal rotation and (peak) hip adduction, and decreased (peak) knee adduction (in other words, increased knee abduction). (Willy et al. 2012, Noehren et al 2012, Souza et al. 2009.) Especially knee valgus is seen more with females because a wider pelvis would move the center of mass of the body more medial to the hip joint center, which increases the adduction moment. Knee valgus angle refers to malalignment in which femur is adducted and tibia abducted, this is usually combined with a quite large Q-angle.

Also often the valgus orientation of the knee is a result of femoral adduction when tibia stays relatively vertical. If the tibia rotates internally, it is usually a coping mechanism when hip is already rotating internally. Excessive femoral adduction has suggested occurring during dynamic tasks also due to weakness of the hip abductors. (Powers et al, 2003.)

In addition, contralateral pelvic drop (Figure 25) is often companied with increased external moment (Powers et al, 2003). However, a recent study by Noehren et al. (2012) including females with and without patellofemoral pain did not display contralateral pelvic drop or trunk lean, but ipsilateral trunk lean over the leg during the stance phase. This was suggested to occur due to weak muscles in the core and weak hip abductors, and in order to minimize the energy demands of lateral displacement of the trunk. In addition, in another study (Dierks et al. 2008) females and males with patellofemoral pain displayed decreased hip abduction and external rotation strength at the end of the prolonged running, and interestingly half of the patellofemoral pain syndrome group displayed hip abduction during the first half of the stance which was most likely a compensation mechanism to avoid pain. It was also suggested that hip abduction was a result of an ipsilateral trunk lean and pelvic elevation. Bolgla et al. (2008) also suggest that it is a compensatory strategy to avoid pain when the female subjects with patellofemoral pain demonstrated less internal hip rotation than asymptomatic subjects.

Hip abduction can be related to varus alignment (Andriacchi et al. 1994), and in addition, knee varus is created by hip and tibial external rotations and decreased Q-angle. Also varus gonarthrosis is usually associated with increased knee adduction moment, angular impulse and increased load put upon the medial aspect of the knee (Landry et al. 2007). It is generally accepted that external knee adduction moment is associated with an increased loading on the medial compartment of the knee and has an effect on the progression of the degenerating diseases (Miyazaki et al. 2002). Studies have shown that actually males with patellofemoral pain often display excessive dynamic varus angle (Willy et al. 2012, Carter et al. 2002). Willy et al. (2012) showed that male subjects with patellofemoral pain (PFP) squatted and ran with greater knee adduction, had a greater knee external adduction moment and contralateral pelvic drop than asymptomatic males. Males with PFP also squatted and ran with lower hip adduction than females with PFP, and in addition, these symptomatic females had smaller knee adduction moment than symptomatic males. Dierks' (2008) sex-mixed cohort study confirmed that females with patellofemoral pain ran with increased hip adduction and internal rotation, whereas the most males with patellofemoral pain ran with decreased hip adduction, meaning a greater knee varus orientation (Dierks et al. 2008). It has been also reported that runners who developed PFPS had greater knee abduction impulses than subjects who remains asymptomatic.

A general assumption, however, is that dynamic hip adduction leads to increased external adduction moment at the knee. This external adduction moment can be quite large, which would lead to a compensatory strategy in which internal abduction moment would be increased. However, the compensation mechanism mentioned above might again put stress more on medial side of the knee and also on the lateral ligaments resulting in degenerative joint diseases like osteoarthritis and/or patellofemoral pain. (Stefanyshun et al. 2006, Stefanyshyn et al.1999, Sharma et al. 2010.) In other words, varus gonarthrosis is usually associated with increased knee adduction moment, angular impulse and increased load put upon the medial aspect of the knee (Landry et al. 2007).

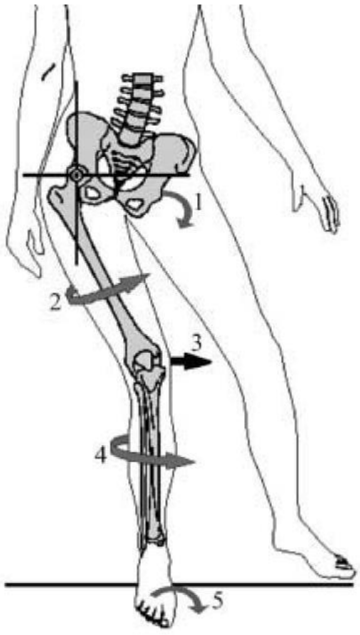


Figure 25. **(1)** Contralateral pelvic drop. **(2)** Internal rotation of the femur. **(3)** The knee valgus. **(4)** Internal rotation of the tibia. **(5)** Pronation of the foot. (Powers 2003.)

4.1.4 Subtalar orientation

All of these deformities mentioned above might contribute to subtalar deformities or vice versa. Segments and joints of lower extremity are in interaction with each other, and this way they compensate one another if there are malaligments or other deficiencies which alter the normal function. Subtalar pronation usually leads to tibial internal rotation, and vice versa with the subtalar supination which leads to external rotation of the tibia (Tiberio et al. 1987, Subotnick et al. 1985). It is assumed that an abnormal subtalar joint pronation would affect the tibia to excessively rotate internally, and this increased tibial internal rotation could actually decrease the Q-angle. In contrast, abnormal subtalar supination would result in external tibial rotation which would increase the Q-angle. (Nawoczenski et al. 2005.) It is generally hypothesized that excessive subtalar pronation would contribute to patellofemoral joint dysfunction (Tiberio et al. 1987). Even though, abnormal subtalar pronation and excessive tibial internal rotation might result in a rotatory strain on soft tissues of the lower limb, and this way affect the tibiofemoral joint, this does not necessarily (always) apply to patellofemoral joint. Excessive internal rotation of the femur caused by tibial internal rotation would accommodate the tibiofemoral joint by decreasing the forces acting upon it. Unfortunately this ‘moves’ the stress onto the patellofemoral joint. In addition, excessive subtalar pronation coupled with increased internal tibial rotation actually decreases the Q-angle, which in turn decreases the stress put upon the lateral aspect of the knee. (Tiberio et al. 1987.)

Subtalar alignment and patellofemoral joint function have been studied and some conclusions made about their relationship. One scenario has been introduced about the increased forces acting upon the patellofemoral tendon due to increased subtalar pronation angle. It is suggested that to achieve knee extension in midstance, the tibia should externally rotate in relation to femur to ensure a normal limb motion during gait. If the foot fails to resupinate and prevents the tibia to externally rotate, the femur must compensate the situation by rotating internally relative to tibia to ensure the knee extension. This internal rotation of the femur moves the patella medially, which increases the Q-angle and the lateral component of the quadriceps muscle vector. In addition, the femoral internal rotation results in the patella to glide in the femoral trochlear when the quadriceps are still contracting, and this leads to increased pressure between lateral articular surface of the patella and the lateral femoral condyle. Finally this results in an increased lateral tracking of the patella when the tibiofemoral joint is in full extension. (Tennant et al. 2001, Bohnsack et al. 2009, Tiberio et al. 1987.)

A study of Moss et al. (1992) showed that female athletes with patellofemoral pain had a longer time to maximum pronation and a lower mean velocity to maximum pronation than their asymptomatic counter partners. Also varus alignment of the forefoot has been shown to increase the risk of the tibial stress syndrome development (Sommer et al. 1995). Duffey et al. (2000) studied long-distance runners and the factors associated with anterior knee pain. They noticed that symptomatic runners pronated the feet less than asymptomatic runners in the first 10% of the stance phase. It was concluded that the lack of pronation caused the lack of shock absorption, meaning that the runners suffered from increased impact shock (Hestroni et al. 2006)**.** However, according to Hestroni et al. (2006) any significant correlations between maximum foot pronation and ankle range of motion did not exist.

4.1.5 Leg length discrepancy

In addition to these above mentioned possible explanations for the development of patellofemoral pain and about their interdependence, abnormal subtalar motion and its effect to the PFPS can also be further explained by leg length discrepancy. Leg length discrepancy (LLD) is defined as a condition in which the legs are noticeably unequal in length, and could cause pain and early appearance of osteoarthritis in the lower extremities when uncorrected (Kujala et al 1987). It can be divided into two etiological groups; structural and functional. Structural leg length discrepancy is usually associated with shortening of bony structures, and functional LLD is defined as a result of altered biomechanics of lower extremity. Gurney (2001) summarizes that “Functional, or apparent LLD is a result of muscle or joint tightness across any joint in lower extremity or spine.” Some of the more common causes can be pronation or supination of one foot in relation to other, hip abduction/adduction tightness/contracture, knee hyperextension due to quadriceps femoris weakness, and lumbar scoliosis. The coping methods that are usually seen in subjects with LLD are steppage gait (increased hip and knee flexion), circumduction (increased hip abduction at swing phase), vaulting (increased plantar flexion at step phase) and hiphiking (increased ipsilateral lumbar side flexion at swing). (Gurney et al. 2001.) Altered gait characteristics also include decreased stance time and decreased step length on the shorter leg, decreased walking velocity and increased cadence (Bhave et al. 1999). Perttunen et al. (2004) confirmed as well that the shorter limb bore the weight for less time than the longer limb, and the pressure was higher on the long than upon the short limb. In addition, ground reaction force (GRF) has been reported being larger on the longer than shorter leg. (Perttunen et al. 2004.)

Gurney (2001) studied older adults and showed that when the leg length discrepancy was artificially increased, a significant increase in muscle recruitment was seen. Especially increased quadriceps activity could be due to slight flexion of the longer limb in trying to balance the leg length difference and prevent tilting of the pelvis too much over the shorter leg during the stance phase. (Gurney et al. 2001, Perttunen et al. 2004.) There are variations in what is the length difference between the legs that produces complications. It has been proposed that limb length difference of 6.4 mm in athletes and 19 mm (1.9 cm) in non-athletes could produce pathology. Also LLD could have three times more significant effects during running when compared to walking. (Subotnick et al. 1981.) Song et al. (1997) found that the mechanical work performed by the longer leg was larger compared to shorter one when the length discrepancy was 16.4 mm. This 5.5 % length difference led to greater vertical displacement of the center of gravity of the body. (Song et al. 1997.) Perttunen et al. concluded that the mean value of LLD of over 2.5 cm could cause an asymmetrical gait (Perttunen et al. 2004). It is proposed that, in the case of LLD, the shorter limb undergoes supination and longer limb is forced into pronation in order to ensure the balanced gait (Gurney et al. 2002). These movements affect normal biomechanics when applied in abnormal way, and could lead to anterior knee pain and the development of other overuse injuries.

In addition, when considering normal gait parameters, abnormal dorsiflexion is believed to have its part in symptom development in the lower extremity. Soleus-gastrocnemius complex has been studied when trying to find explanations for this relationship. (Hughes et al. 1985.) Hughes (1985) concluded that tight soleus and gastrocnemius muscles prevent the talocrual joint from reaching the required 10° for normal ankle and lower extremity motion. To compensate this deficit the foot excessively pronates thus enhancing the dorsiflexion. However, this overpronation causes the subtalar joint to remain unlocked during the midstance (potentially also during the push-off phase) and further prevents the midtarsal joint and first ray becoming rigid lever. This, in turn, obliges the lateral metatarsals to absorb the forces. In summary, if the dorsiflexion is not available or it is deficient, and in addition, fixing the foot a rigid lever is hindered for push-off phase, the more force is required to propel the body. (Hughes et al. 1985). Lastly, as the leg length discrepancy exists and especially as the longer leg being under the risk of harmful kinematics due to overpronation, also the iliotibial band (ITB) tightness can develop. Excessive pronation usually causing excessive tibial internal rotation might stretch the iliotibial band, which would in turn move the patella laterally and increase the stress on the lateral side of the knee. It is also suggested that weak gluteus medius and this way poor control of medial hip rotation could contribute to the tightening of the ITB. Subjects with PFPS have shown having tighter iliotibial band. (Hudson et al. 2009.)

4.1.6 Muscular imbalance

It is debated whether or not the muscular imbalance between the lower limbs or their asynchronous activation contributes to the development of the patellofemoral pain. Patellar tracking has been studied comparing symptomatic PFPS group and asymptomatic group. A study by Ling (2010) patellar tracking was induced by selective activation of individual quadriceps components, vastus medialis obliguus (VMO), vastus medialis longus (VML) and vastus lateralis (VL). The results showed that oblique orientation of the VMO muscle fibers and medial location of the tendon insertion of the patella produces medial patellar tilt and lateral rotation with VMO contraction in healthy subjects. In contrast, subjects with PFPS demonstrated medial patellar rotation with VMO contraction. They suggested that PFPS group might have more superiorly oriented VMO muscle fibers or a different insertion site, in addition to weakness in the VMO fibers compared to healthy subjects. VML activation produced medial tilt and lateral rotation of patella as well as patellar extension for both the PFPS group and healthy subjects. However, VML activation generated an anterior translation in PFPS group contrary to the posterior translation in the control group. This suggests that the anterior translation, in other words, decreased posterior stabilization might cause a moment arm dysfunction as the patella is not as firmly gliding to the trochlear groove as in the posterior translation. In healthy subjects activation of the VL produced lateral patellar translation at full extension and medial translation at 20-degree flexion. The VL activation in PFPS group produced lateral patellar translation in both conditions mentioned before. This again refers to weakness of the VMO muscle and imbalance in extensor mechanisms, which can further overload the lateral compartment of the patellofemoral joint. (Ling et al. 2010, Fulkerson et al. 2002.)

Also Duvigneau et al. (2008) showed that the relative peak torque of the quadriceps muscles per body weight and body mass index had lower values in the subjects with PFPS than in controls. Another study (Kaya et al. 2010) showed that women with patellofemoral pain syndrome had significant decrease in the quadriceps peak torque, volume, cross sectional area (CSA) and hop test on the affected side compared to controls. Also in the study of Witvrouw et al. (2000), females with PFPS had decreased quadriceps and gastrocnemius muscular flexibility, decreased vertical jump performance and increased medial patellar mobility compared to controls. In addition to decreased strength values of quadriceps Powers et al. (1996) noticed that the decrement was associated with decreased external knee flexion moment which could be related to avoidance pattern when knee pain is present. Studies have shown that when the quadriceps strength was measured at 60°/s and 240°/s the values were lower in the PFPS group compared to control group (Witvrouw et al. 2000, Duvigneau et al. 2008). In addition, mean intensities of all vastus muscles have been shown to be lower in the PFPS group than in controls (Powers et al. 1996). However, the hamstrings' strength was lower in the PFPS group only at higher velocities compared to controls (Witvrouw et al. 2000).

In addition to strength measurements, also firing patterns of lower limb muscles have been studied. According to Witvrouw et al. (2000) VMO and VL muscles' reflex response times were faster in the PFPS group compared to controls. In addition, in the PFPS group the VMO's reflex response time was faster than VL's response time. The reflex response times could suggest the anticipatory mechanism ensuring stable kinematics, when, for example, pain is present at the knee. The same study stated also that the firing sequence between these muscles (VMO and VL) were almost the same in the subjects with PFPS, but differed in the controls where the VMO was firing earlier than VL. In the contrary, Cowan et al. (2002) demonstrated that VL fired earlier than VMO in the subjects with PFPS compared to controls. Also DeFare et al. (2007) has suggested that a 5-ms delay in the VMO can significantly increase the lateral patellofemoral joint loading. Alterations have been shown to occur at the knee joint kinematics when the VMO function was lost by nerve branch block. This resulted in patellofemoral and tibiofemoral lateral shift, and tibiofemoral external rotation. Patellofemoral joint also trended laterally. This suggests that tibiofemoral joint is directly affected by the VMO. Lower VMO activity compared to VL activity has been reported by Mariani (1979), but in the contrary Powers (1996) concluded that there was no difference in the activities between the VMO and VL (Mariani et al. 1979, Powers et al. 1996). In summary, delayed and decreased activation of the VMO might direct the patellofemoral and tibiofemoral joints laterally, increase lateral stress on the patellofemoral joint and lead to degeneration of the joint resulting in pain and syndrome development. In addition, the decreased CSA and strength might put lower limbs under the risk of imbalanced movements and excessive compressing and shearing forces.

Not only are the muscles of the thigh and leg studied, but also the hip muscles have their contribution to lower limp kinematics. The gluteus maximus (GMAX) and gluteus medius (GMED) have been examined for their influence to lower limb kinematics. Hip muscles contribute to the hip internal and external rotations, and to hip adduction and abduction. GMAX acts eccentrically at the weight bearing phase during running controlling the hip flexion and internal rotation, and also has an influence on the knee extension during the late stance phase. (Swanson et al. 2000.)

Wilson et al. (2011) examined the timing and activation of GMED and GMAX among females with and without PFPS during running. The results showed that the activation of GMED was delayed and also the duration of activation was shorter in the PFPS group than in controls. The GMAX activation and timing were similar between the groups. In the contrary, the study of O'Sullivan et al. (2012) stated that there was no difference in the GMED activation between PFPS and asymptomatic group. In addition, Wilson et al. (2011) noticed that there was a moderate correlation between GMED onset time and hip adduction, as well as between GMAX onset time and both hip adduction and internal rotation. One possible explanation for altered kinematics and development of PFPS would be the anticipatory muscle recruitment of hip muscles due to experienced pain. This delayed activation pattern could facilitate the knee abduction, which is believed to excessively load the patellofemoral joint (Willson et al. 2011).

Association of PFPS and weak hip musculature is also supported by Souza et al. (2009), who demonstrated that females with PFPS exhibited weaker hip muscles and increased hip internal rotation than controls during running. They concluded that the weak hip musculature caused the internal rotation of the hip. Also Powers et al. (1996) showed using MRI that lateral patellar tilt and lateral patellar displacement was a result of femoral internal rotation, which can possibly be caused by hip internal rotation. In addition, weak gluteus medius (GMED) and poor control of hip medial rotation is believed to contribute to tightening of the iliotibial band, and further, like mentioned earlier, to the lateral patellar tracking, lateral patellar tilt, and lateral patellar compression (Hudson et al. 2009). In the contrary to the aforementioned studies, Bolgla et al. (2008) actually showed less hip internal rotation in the PFPS group than in controls during running, hopping and single-leg squat due to pain (Bolgla et al. 2008). In spite of the varying results, Ferber et al. (2011) showed that a 3-week hip-abductor strengthening protocol not only increased the strength of the hip muscles, but also decreased the patellofemoral pain in the subjects with PFPS.

4.1.7 Step length and rate

Step length has been under the scope when trying to figure out how to decrease the symptoms and especially the pain in the knee. A few studies have examined the effects of different step lengths and step rates on the patellofemoral joint stress (PFJS). Wilson et al. (2014) found that a 10% reduction in the preferred step length in the female runners with and without PFPS caused a decrease in the patellofemoral joint stress and peak knee extension moment. However, the PFPS group demonstrated less reduction in these variables than the controls. In the contrary, 10% increase in the step length showed an increase in stance time, greater peak knee flexion angle and extensor moments, and increased patellofemoral joint stress for both groups.

Lenhart et al. (2014) and Heidescheit et al. (2011) compared the step rate manipulation and joint kinematics during running. The first study used step rates +/- 10% of preferred cadence during running and concluded that patellofemoral force magnitude, loading rate and impulse are all diminished with 10 % increase in cadence. In the latter study the results showed that when the step rate was increased 10%, less mechanical energy was observed in then knee and hip. Furthermore, vertical excursion of center of mass, braking impulse, peak knee flexion angle, peak hip adduction and its angle, and hip internal rotation moment were also decreased. All this suggests that less force and stress are put upon the knee joint.

Interestingly, the Noehren et al. (2012) examined the effect of the real-time gait retraining on the hip mechanics in the subjects with PFPS. This treadmill study done in females with PFPS gave promising results. The hip adduction, hip internal rotation, contralateral pelvic drop and pain experienced reduced due to feedback. Especially reduction in hip adduction and pelvic drop would reduce the stress and iliotibial tension on the lateral aspect of the patella, respectively. In addition, 86% reduction in pain was seen in patients with PFPS.

4.1.8 Patellofemoral joint stress and reaction forces

Like discussed above subjects with PFPS are believed to demonstrate excessive stress and forces on the knee joint which may lead to degenerative disorders due to anatomical or/and functional reasons. Studies have been done to directly or indirectly estimate the forces that are suggested to differ between the subjects with and without PFPS. Usually the parameters of interest are peak patellofemoral joint stress (PFJS), patellofemoral joint reaction force (PFJRF), PFJ reaction force-time integral, patellofemoral contact force, peak knee extensor moment and knee joint angles. Wirtz et al. (2012) studied females with and without PFPS during running and compared the PFJS, PFJS-time integral as well as hip and knee transverse plane kinematics between the two groups. The results did not give any significant difference in the PFJS and PFJS - time integral between the groups. However, they believe that PFPS group demonstrated some compensatory mechanisms which are generally believed to exist. They also believe that there was co-contraction between the flexors and extensors at any given extension moment, which affected the results. In addition to these indifferent results, excessive hip internal rotation and knee external rotations were seen in the PFPS group.

The peak patellofemoral joint stress was also examined by Heino-Brechter et al. (2002), who compared the PFJS between the subjects with and without PFPS (patellofemoral pain syndrome) during stair ascent and descent. Like in Wirtz's study, either in this study significant group differences in the PFJS and PFJS - time integral were not found. However, when looking at the results closer, the PFPS group demonstrated reduction in both the patellofemoral joint reaction force (PFJRF) and PFJRF-time integral compared to asymptomatic group during the stair ascent condition. Because there were no differences in the knee kinematics between the groups the reduction was suggested to be attributed to quadriceps avoidance strategy. Quadriceps inactivation leads to reduced extensor moment which was found in PFPS group compared to controls. This was created by lowering the cadence and increasing the forward trunk lean. Leaning the trunk forward is often seen in avoidance strategy because it decreases the moment about the joint by bringing the center of mass closer to the knee joint. Interestingly, even though, the PFJRF was reduced, the PFJS was still similar between the groups like mentioned earlier. This was explained by reduced patellofemoral contact area which increased the stress on the knee.

During the stair descent condition the PFJRF and PFJRF-time integral were similar between the groups, however, there was a difference in the PFJS pattern between the groups. The PFJS occurred earlier in the PFPS group compared to the control group, which could refer to possible malalignment in the PFPS group. In general, during stair descent the first 60% of the descending phase is associated with decreased flexion angle of the knee which is in turn leads to reduced patellofemoral contact area, and in the end to increased stress upon the knee joint. Lastly, PFJRF-time integral was 1.5 times greater during stair descent compared to stair ascent condition, which would lead to greater overall stress on the knee, although, PFJS did not significantly differ between the tasks. All in all, in spite of the indifference in the PFJS between the two groups, it could be suggested that subjects with PFPS would demonstrate compensatory strategies to reduce the stress acting on the knee and avoid pain.

Partly contrary results were found in the study of Chen and Powers (2014) as it was shown that there was a significant reduction in the peak resultant PFJRF also in the stair descent condition in the PFP (patellofemoral pain) group compared to asymptomatic group. In addition, the same was observed in the stair ascent condition supporting earlier findings, and during walking. This study showed that posterior and superior components of PFJRF were lower in the PFP group compared to the controls. This refers to the fact that the activation of muscles responsible for posterior force component, vastus medialis and lateralis, and for superior component, rectus femoris and vastus intermedius, was decreased due to avoidance strategy, which is again in line with previous studies. On the contrary, subjects with PFP demonstrated higher lateral force component than asymptomatic subjects, which is not surprising due to general notion that subjects with PFP often demonstrate subluxation of the patella.

5. PURPOSE OF THE STUDY

Purpose of the study was to determine which biomechanical factors in the rearfoot running technique could be associated with atypically high patellofemoral joint loading. Rearfoot striking pattern (RFS) has been suggested to load the knee more than forefoot striking pattern (FFS), which in turn is generally associated with higher loading pattern within ankle. Especially females, who tend to have increased Q-angle (for anatomical reasons), also often demonstrate increased knee abduction moments. This in turn could lead to development of knee related degenerative disorders, and for example, to experienced patellofemoral pain.

The aim of the study was to detect the patellofemoral contact forces and joint moments of 39 female subjects by analyzing their running by applying an eight-camera system and inverse dynamics methods. It was hypothesized that females, who applied RFS, would demonstrate a relationship between forces acting upon the knee and the applied rearfoot running pattern during running.

6. METHODS

6.1 Participants

In the beginning of the experiment the total number of participants was 286 including males and females. However, 39 female floorball and basketball players (age: 17.2 years +/- 3.4, height: 169.0 cm +/- 5.7, weight: 60.6 kg +/- 8.7) running with RFS pattern were chosen for 3-D running analysis in this study. Subjects' background information was gathered by a questionnaire to confirm that they did not have history of any musculoskeletal problems meaning injuries or surgeries that could influence the running. The study was approved by the ethics committee of the Pirkanmaa Hospital District, Tampere, Finland (ETL code R10169), and was conducted in accordance with the Declaration of Helsinki.

6.2 Experimental protocol

The study experiment was conducted in UKK-institute, Tampere, Finland. The participants were selected by the gender, in this case only females were accepted, and by visually identifying the running pattern. Only subjects that ran with rearfoot running pattern were further analyzed. Tibiofemoral angle was measured from a static standing position by motion analysis measurement. The mechanical axis in the frontal plane was calculated using Plug-in Gait model (Vicon Nexus v1.7; Oxford Metrics, Oxford, UK), and the angle was defined between joint centers of ankle, knee and hip. 34 reflective markers were set bilaterally on the subjects (on the shoe over the second metatarsal head and over the posterior calcaneus, lateral malleolus, lateral shank, lateral knee, lateral thigh, anteriorsuperior iliac spine, posterior superior iliac spine, clavicula, sternum, seventh cervical vertebra, 10th thoracic vertebra, shoulder, elbow, two wrist markers, finger, and four head markers), and the Plug -in Gait full body model was used in detection. Also height and weight were measured.

Participants were instructed to do a warm-up by biking on exercise bicycle for 5 minutes. After this they were asked to perform a shod running along a 15-meter long track at 4 m/s. The participants were instructed to run normally towards the mark on a wall ahead. Two photocells were used to detect and control the speed of the running (+/-0.2 m/s), and if it did not hit the target, instructions were given to change the speed. Also the starting mark position was changed if necessary. Five successful force platform contacts with the left leg were analyzed.

6.3 Running analysis

To record the positions of the markers (at 300 Hz) an eight-camera system (Vicon T40, Vicon) was used, and a force platform (AMTI BP6001200; AMTI, Watertown, MA) was used in data gathering (at 1500 Hz) of ground reaction forces (GRF). Trajectories of the reflective markers and data of GRF were low-pass filtered using a 4^{th}-order Butterworth filter with cutoff frequencies of 12 Hz for the markers and 50 Hz for the GRF. In analysis, GRF data were analyzed by the Signal software (v.4.1; Cambridge Electronic Design, Cambridge, UK) to detect the contact times using 20-N thresholds for GRF. Average vertical loading rate was calculated by defining the total change in force which was divided by the total change in time. In the calculation a period of 20% and 80% between the ground contact and vertical impact peak was used (Milner et al. 2006). Kinematic and kinetic analysis, and a COM calculation were performed using the Plug-in Gait model (Vicon Nexus v1.7, Oxford Metrics). Foot contact with the ground and toe-off from the ground were detected when calculating cadence and step length. Distance between COM and heel contact with the ground (heel marker) was defined. Joint angles and internal joint moments (Nmkg^{-1}) during the stance phase of running were calculated by averaging five force plate contacts. Finally, the contact force of patellofemoral joint was calculated as a function of knee extensor moment (M_k) and knee flexion angle (x) by applying the biomechanical model described by Ho et al. (2012). Quadriceps muscle's effective moment arm (L_q) was calculated by using the knee flexion angle and applying a nonlinear equation (Van et al. 1986):

$L_q = 8.0E^{-5}x^3 - 0.013x^2 + 0.28x + 0.046$ (Altman et al. 2012)

Quadriceps force (F_q) was calculated:

$F_q = M_k / L_q$ (Arendse et al. 2004)

PFCF was calculated:

$PFCF = F_q k$ (Baliunas et al. 2002)

The constant k was estimated for knee joint angle position(x) using the following nonlinear equation on the basis of the curve fitting to the data of van Eijden et al. (35):

$k(x) = \frac{(4.62E^{-1} + 1.47E^{-3}x^2 - 3.84E^{-5}x^2)}{1 - 1.62E^{-2}x + 1.55E^{-4}x^2 - 6.98E^{-7}x^3}$ (Bredeweg et al. 2011)

6.4 Statistical analysis

Kinematic and kinetic data were time normalized (0%–100%) and averaged across five trials to get individual mean curves for the parameters of interest. Running-induced knee loading was evaluated by defining the magnitude of the peak knee extensor and abduction moment, and peak PFCF. First, the averages and standard deviations were calculated for the parameters of interest. Second, two-tailed (bivariate) Pearson correlations were executed. 95% confidence intervals were calculated and reported for comparisons (SPSS 18.0; SPSS, Chicago, IL). P values less than 0.05 were accepted and considered as significant.

7. RESULTS

Table 1 demonstrates the mean values of the parameters of interest in this study. The parameters include the spatiotemporal parameters: COM-heel distance, step length, frequency (or cadence), contact time with the ground, and other parameters of interest: joint moments and angles, joint powers and force acting on the patellofemoral joint (patellofemoral contact force, PFCF).

TABLE 1. Mean values (average) and standard deviations (SD) of the parameters of interest.

Parameters of interest	Average	SD
COM-Heel distance (cm)	142,952	32,828
Step length (m)	1,433	0,091
Frequency (steps/min)	176,501	14,244
Contact time (s)	0,219	0,018
Patellofemoral contact force (PFCF, BW, N)	4,988	0,945
Knee abduction moment (Nm/kg)	1,931	0,605
Knee extensor moment (Nm/kg)	3,510	0,611
Hip extensor moment (Nm/kg)	2,196	0,750
Hip flexor moment (Nm/kg)	-2,824	0,683
Hip abduction moment (Nm/kg)	2,321	0,416
Ankle plantarflexor moment (Nm/kg)	2,453	0,330
Ankle dorsiflexor moment (Nm/kg)	-0,583	0,220
Knee flexion angle at heel contact (°)	20,409	3,820
Knee flexion angle max (°)	49,987	3,104
Knee abduction angle at heel contact (°)	-1,537	4,744
Hip flexion angle at heel contact (°)	-5,370	2,917
Hip extension angle at toe off (°)	45,327	5,252
Hip abduction angle at heel contact (°)	-8,895	4,109
Hip abduction angle max (°)	16,737	3,648
Hip power max (W)	5,163	2,422
Hip power min (W)	-17,224	5,767
Knee power max (W)	16,895	3,969
Knee power min (W)	-22,407	3,806
Ankle power max (W)	12,458	2,369
Ankle power min (W)	-6,212	1,405

Correlations were made about the parameters (Table 2). Patellofemoral contact force (PFCF) normalized with body weight (BW) significantly correlated with hip extensor (Figure 26) and flexor moments (Figure 26), and with knee extensor moment (Figure 26), r= -0.576, -0.548, 0.967, respectively ($p<0.05$). Other significant correlations existed between maximum knee power and PFCF (Figure 26), and knee abduction moment (Figure 27), r= 0.467 and 0.466, respectively. Table 2 demonstrates that negative correlations existed between the PFCF and maximum hip power and minimum hip power, r= -0.336 and -0.377, respectively. Spatiotemporal comparison showed that PFCF positively correlated with COM-heel distance, r= 0.290. Knee abduction moment negatively correlated with step frequency and positively correlated with COM–heel distance, r= -0.329 and 0.355 respectively. PFCF positively correlated with knee flexion angle, r= 0.341. In addition, PFCF showed a negative correlation with maximum hip power, r= -0.336, but positive correlation with minimum hip and knee powers, r= -0.377 and -0.406, respectively, when taking into a consideration the eccentric nature of the movement.

TABLE 2. Correlations between patellofemoral contact force (PFPC) and the parameters, and between knee abduction moment and the parameters. **Significant correlation, *correlation (r), $p<0.05$.

Paraemetrs of interest	Patellofemoral contact force (PFCF, BW, N)		Knee abduction moment (Nm/kg)	
	r	p	r	p
COM-Heel distance (cm)	-0.350*	0,029	0.355*	0,026
Step length (m)	0.090	0,584	0,193	0,239
Step frequency (steps/min)	-0,065	0,695	-0.329*	0,041
Contact time (s)	-0,191	0,245	0,169	0,303
Patellofemoral contact force (PFCF, BW, N)	1		-0.127	0,442
Knee abduction moment (Nm/kg)	-0.127	0,442	1	
Knee extensor moment (Nm/kg)	0.967**	0,000	-0.143	0,386
Hip extensor moment (Nm/kg)	-0.576**	0,000	0.174	0,290
Hip flexor moment (Nm/kg)	-0.548**	0,000	-0.210	0,200
Hip abductor moment (Nm/kg)	0.141	0.391	0.243	0.136
Ankle plantarflexor moment (Nm/kg)	-0.200	0,221	0.160	0,329
Ankle dorsiflexor moment (Nm/kg)	-0.147	0,372	-0.067	0,684
Knee flexion angle at heel contact (°)	0.170	0,302	-0.154	0.349
Knee flexion angle max (°)	0.341*	0,034	0.001	0.994
Knee abduction angle at heel contact (°)	-0.308	0.056	0.416**	0.008
Hip flexion angle at heel contact (°)	0.069	0,676	-0.264	0.105
Hip extension angle at toe off (°)	-0.005	0,978	-0.116	0.482
Hip abduction angle at heel contact (°)	0.272	0,094	-0.107	0.516
Hip abduction angle max (°)	0.145	0,377	-0.128	0.437
Hip power max (W)	-0.336*	0,016	-0.229	0,161
Hip power min (W)	-0.377*	0,018	-0.158	0,336
Knee power max (W)	0.476**	0,003	0.466**	0,003
Knee power min (W)	-0.406*	0,010	-0.234	0,152
Ankle power max (W)	0.195	0,235	-0.060	0,715
Ankle power min (W)	0.027	0,872	0.165	0,316

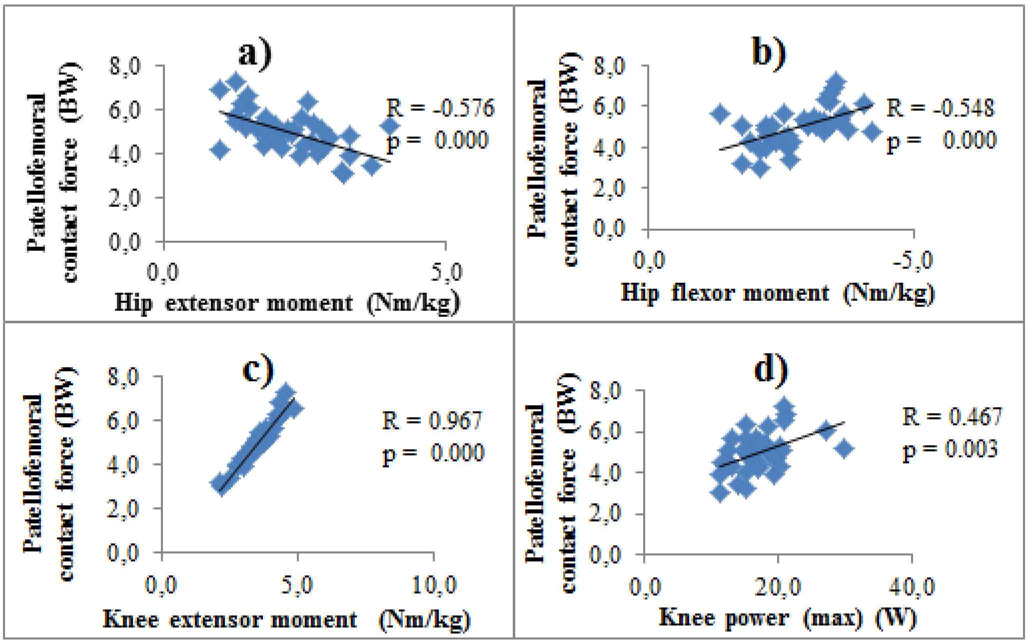


FIGURE 26. **a)** A significant negative correlation between PFCF (BW) and hip extensor moment. **b)** A significant positive correlation between PFCF (BW) and hip flexor moment. **c)** A significant positive correlation between PFCF (BW) and knee extensor moment. **d)** A significant positive correlation between PFCF and maximum knee power.

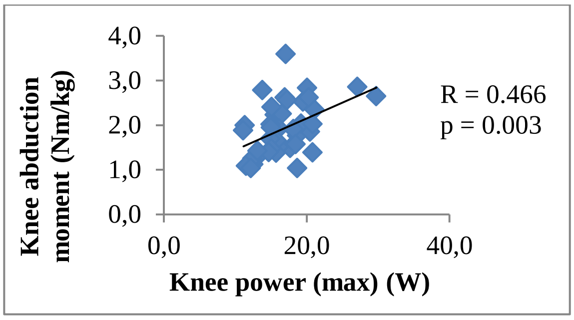


FIGURE 27. A significant positive correlation between knee abduction moment and maximum knee power.

8. DISCUSSION

Patellofemoral pain syndrome is multifactorial. Various factors and their combinations might contribute to its development. The matter of interest in this study was to determine which biomechanical factors in the rearfoot running technique could be associated with atypically high patellofemoral joint loading. The results showed that female team players experimented in this study running with rearfoot striking pattern (RFS) demonstrated high loading pattern acting upon the knee which was supported by the significant correlations found between patellofemoral contact force (PFCF) and hip extensor moment (r= -0.576), PFCF and flexor moments (r= -0.548), and PFCF and knee extensor moment (r= 0.967). In addition, support is given by the significant correlations between maximum knee power and PCFC (r= 0.467) and maximum knee power and knee abduction moment (r= 0.466).

Generally it has been stated that ground reaction force curves include two vertical peaks for rearfoot striking (RFS) pattern whereas forefoot striking (FFS) pattern include just one. In addition, previous studies have shown that RFS usually have greater stance time, longer stride length and greater cadence than FFS. It has also been shown that knee abduction moment, and hip adduction and abduction moments are greater in RFS pattern compared to FFS pattern. (Strearne et al. 2014, Kulmala et al. 2013.)

According to our results the hip flexor moment and extensor moment correlated significantly with the patellofemoral contact force, and they can be concluded to be in line with the earlier notions made by Kulmala et al. (2013). It could be suggested that RFS pattern might lead to less active hip flexion and extension musculature. In other words, the activation of hip flexors like iliopsoas, rectus femoris and secondary hip flexors sartorius, tensor fascia latae and pectineus could be decreased. Hip flexion is also controlled by the hamstrings and with increasing running speed by the gluteus maximus (Hamill and Knutzen 2009, 241). These muscles have their part in acting eccentrically to break down the vertical and horizontal velocities ensuring a smooth and balanced contact with the ground and prevent the collapsing of the hip and also prevent the trunk from pitching forward as the trunk flexes (Hamill and Knutzen 2009, 201). Their greater activation correlates positively with the stress put upon the knee.

Like mentioned earlier the hip extensor gluteus maximus eccentrically controls the trunk and hip flexion in combination with hamstrings on the stance side while running, but especially gluteus maximus contributes to hip extension in more vigorous activities like in running. Gluteus maximus, in addition to obturator externus and guadratus femoris, also contributes to external rotation of the thigh during the late swing until the toe-off. (Hamill and Knutzen 2009, 202.) Taking into a consideration that there was a significant negative correlation between the PFCF and the hip flexor moment, it could be suggested that the hip extensors, especially hamstrings and gluteus maximus, could be less activated in the RFS pattern.

Greater hip adduction and abduction moments seen in RFS runners (Stearne et al. 2014, Kulmala et al. 2013) could lead to a suggestion that hip abductors, mainly gluteus medius, which is also an internal rotator of the thigh with gluteus minimus, might be less activated or weak. These muscles, gluteus medius and minimus, assist in hip flexion as well, and this further supports the conclusion that the weak hip musculature could be associated with RFS pattern. Also Wilson et al. (2011) showed that if the gluteus maximus and medius activations were delayed, it could lead to increased hip adduction and internal rotation. This was demonstrated in the subjects with patellofemoral pain syndrome (Wilson et al. 2011). In addition, in the study of Souza et al. (2009) females with patellofemoral pain had weaker hip musculature and greater hip internal rotation than asymptomatic females. This is in line with the results of Powers et al. (2003) who showed that lateral patellar tracking was a result of increased internal femoral rotation, which in turn was possibly caused by increased internal rotation of the hip. Lastly, in line are also results that PFPS subjects demonstrate increased hip adduction and contralateral pelvic drop in addition to decreased knee adduction (Willy et al. 2012, Noehren et al 2012, Souza et al. 2009.) It has been shown that a hip strengthening rehabilitation program reduced anterior knee pain in patients with patellofemoral pain syndrome, which could have occurred due to decreased stress on the knee. (Ferber et al. 2011).

Knee extension moment has shown to increase when habitual FFS runners switched to RFS pattern (Strearne et al. 2014) and also RFS runners tend to have greater knee flexion angle than FFS (Kulmala et al. 2013). Our results show that there was a significant positive correlation with the patellofemoral contact force and knee extension moment in the females running with the RFS pattern. In addition, the PFCF and knee flexion angle positively correlated. It has been stated that the subjects with PFPS have decreased quadriceps strength and CSA area, which would lead to decreased knee extension moment (Duvigneau et al. 2008, Witvrouw et al. 2005, Powers et al. 2003, Kaya et al. 2010, Powers et al. 1996), as taking into a consideration that quadriceps are the main knee extensors. In addition, quadriceps and gastrocnemius contribute to the knee flexion at heel strike and midstance by acting eccentrically to slow down the braking impulse. Even though, it has been suggested that the knee extension moment and knee flexion angle decrease in subjects with PFPS due to pain avoidance (Wirtz et al. 2012, Heino-Brechter et al. 2002, Chen and Powers 2014), the results of this study suggest that the increases in these variables are more likely caused by the lack of activation of hip and ankle musculature associated with RFS pattern. Hughes et al. (1985), for example, stated that subjects with forefoot varus have decreased dorsiflexion, which impairs the pronation and further the supination at the toe-off phase. When the foot is unable to be formed into a rigid level for push-off, more energy is required from other force creating aspects of the lower limb. Our results could be thought to support this notion when the activity of the quadriceps was indeed increased along the RFS pattern.

In this study a significant correlation was seen between patellofemoral contact force and maximum power production of the knee, as well as between knee abduction moment and maximum power of the knee. This refers to the same suggestion mentioned above that the greater quadriceps activation and reduced activation of the hip and ankle might be associated with RFS pattern leading to greater stress put upon the knee. Maximum power production of the knee demonstrates the concentric work of the quadriceps, which occurs at the propulsion phase when the knee extends. The same conclusion can be made as above that especially hip extension might be less active which requires more work from the quadriceps. However, maximum knee power is calculated as a function of knee extension times joint angle velocity which could mean that the power result can be due to an increase in one variable or another. In addition, negative correlation was found between the patellofemoral contact force and maximum power production of the hip, which refers to same conclusion. Lastly, there were positive correlations between the patellofemoral contact force and minimum hip and knee powers, which refer to eccentric work done by hip and knee during the stance phase. This negative work, typical also for RFS pattern, increased the loading on the knee.

Lastly, there was a positive correlation between the knee abduction moment and COM-heel distance, which could mean that with increasing step length the control of the hip decreases. On the contrary, the increasing step frequency decreases the knee abduction moment referring more balanced lower limb function. The negative correlation between the COM-heel distance and patellofemoral contact force is against the expectations, because it could be believed that patellofemoral contact force would increase with increasing step length. Support to this is given by Wilson et al. (2014) who found that 10% increase in step length resulted in increased stance time, peak knee flexion angle and extensor moment, and patellofemoral joint stress.

There are some limitations in this study. First, the study included only young female athletes, which means that generalizations should be made with caution. Second, the values for patellofemoral contact force (PFCF) might be underestimated because they were calculated from the average of five trials (time normalized data) during the stance phase. In addition, the antagonistic joint forces were not taken into account, which also might lead to an error in PFCF. Lastly, longitudinal running studies on the rearfoot striking pattern and associated biomechanics would give a more in depth view of the mechanisms contributing to actual degenerative disorders and patellofemoral pain syndrome.

9. CONCLUSIONS

The aim of this study was to examine if there are excessive forces put upon and acting on the knee when running with rearfoot running pattern. Our results suggest that as patellofemoral contact force (PFCF) was found to significantly correlate with flexor and extensor moments of the hip, extensor moment of the knee, and maximum knee power in addition to significant correlation found between maximum knee power and knee abduction moment, the RFS pattern could be associated with an inactivation of the hip musculature, which in turn is often compensated with increased activation of the quadriceps and hamstrings. Also this supports previous studies and results which suggest that RFS technique might lead to excessive loading of the knee. Lastly, our results are in line with previous conclusions and suggestions that the RFS pattern might also lead more varus type of condition meaning that knee abduction increases as well as the stress on the medial aspect of the knee, which could eventually result in degenerative disorders within the knee like chondromalacia patellae and patellofemoral pain syndrome.

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