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Juha-Pekka Kulmala

The Effects of Locomotor Pattern Diversity and Ageing on the Lower Limb Joint Mechanics and Loading During Human Walking and Running

JYVÄSKYLÄN YLIOPISTO

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"Variation is a feature of all natural populations"
Charles Darwin (1809-1882)

ABSTRACT

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Many lower limb problems are increasingly seen as disorders where mechanical loading plays a central role in both the development and progression of the pathological condition. Although locomotor patterns during human walking and running demonstrate clear differences due to inter-individual variability and ageing, there is a substantial lack of research evaluating whether or not these factors expose certain musculoskeletal structures to higher than normal loading. Therefore, the aim of this thesis was to examine how locomotor pattern diversity and ageing affects lower limb joint mechanics and loading across different modes and intensities of locomotion. In line with previous studies, a clear inter-individual variability in both walking and running patterns were observed. Analysis of lower limb mechanics revealed that during walking, different gait strategies can considerably affect knee joint loading in that those individuals who demonstrated either knee extensor or flexor dominant gait patterns exhibited significantly higher medial knee joint loading compared with those who walked with a typical gait pattern. When running-induced joint loading profiles were compared between different running patterns, it was found that those who run with a forefoot striking pattern exhibit lower patellofemoral stress and medial knee joint loading than those who use a rearfoot striking pattern. However, parallel increases in ankle plantarflexor and Achilles tendon loading were observed among forefoot strikers. Finally, comparison of locomotor mechanics of different age-groups showed that the age-related propulsive deficit of the ankle joint becomes more severe as locomotion changes from walking to running to sprinting. As a result, old adults demonstrated higher muscular efforts of the more proximal lower limb muscles than their younger counterparts when walking and running at the same speed. During maximal sprinting, reduced muscular output of the hip and increased hip adduction but decreased trunk lateral flexion movements were also observed in older adults compared with their younger counterparts. Altogether, the findings of this work suggest that different walking and running patterns have considerable effects on the lower limb joint loading and thus may play a role in the development of stress-related joint injury. As a result, gait modification could be advisable for some individuals to manage their joint loading and consequent symptoms. In addition, the findings regarding age-related locomotor changes indicate that ankle propulsive deficit contributes most to the locomotor decline among older adults and therefore should be targeted when designing exercise interventions for this population.

Keywords: locomotion, gait pattern, joint kinetics, ageing, variability

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LIST OF ORIGINAL PUBLICATIONS

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II. Kulmala, J.P., Avela, J., Pasanen, K.., Parkkari, J. 2013. Forefoot strikers exhibit lower running-induced knee loading than rearfoot strikers. Medicine and Science in Sports and Exercise 45(12):2306-13.

III. Kulmala J.P., Korhonen M.T., Kuitunen S., Suominen H., Heinonen A., Mikkola A., Avela J. 2014. Which muscles compromise human locomotor performance with age? Journal of the Royal Society Interface. 11(100): 1-10.

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ABBREVIATIONS

ATF	Achilles tendon force
BW	Body weight
COM	Center of mass
EXT	Knee extensor dominant gait pattern
FFS	Forefoot strikers
FLX	Knee flexor dominant gait pattern
Fq	Quadriceps force
GPE	Gravitational potential energy
GRF	Ground reaction force
KE	Kinetic energy
La	Level arm of the Achilles tendon
Lq	Level arm of the quadriceps muscle
Ma	Ankle plantarflexor moment
MFS	Midfoot strikers
Mk	Knee extensor moment
OA	Osteoarthritis
PCA	principal component analysis
PFCF	Patellofemoral contact force
PFS	Patellofemoral stress
RFS	Rearfoot strikers
ROM	Range of motion
TYP	Typical gait pattern

CONTENTS

1 INTRODUCTION

Locomotion is the major function of the human musculoskeletal system. Although the basic patterns of human locomotion have remained essentially unchanged for centuries, a relatively large diversity in both walking and running patterns exist among the normal population. It has been long recognized that people walk with different muscular strategies (Pedotti 1977, Simonsen et al. 1997, Winter 1984) or run using either rearfoot or forefoot striking patterns during initial ground contact (Cavagna & Lafortune 1980, Nillson & Thorstensson 1989). In addition, previous studies have demonstrated age-related alterations in locomotion mechanics by showing that older adults rely less on ankle plantarflexors and more on knee and/or hip extensors compared with young adults (DeVita & Hortobagyi 2000, Cofre et al. 2011, Boyer, Andriacchi & Beaupre 2012, Monaco et al. 2009).

Musculoskeletal disorders are a major burden to the individual, society and the health care system. Of all musculoskeletal conditions, lower limb disorders constitute one of the largest proportions of problems for which people seek medical attention (Rathleff et al. 2013, Carnes et al. 2007). As the prevalence of lower limb disorders increases with age, rapid ageing of populations will further increase the societal costs of these problems worldwide (Urwin et al. 1998, Woolf & Pfleger 2003). Meanwhile, many lower limb problems are increasingly seen as conditions where joint mechanics play a central role in both development and progression of pathology and severity of clinical symptoms (Andriacchi et al. 2004, Myer et al. 2010, Powers 2003). For example, at the knee joint level, the site most commonly affected by the osteoarthritis (OA) is the medial tibiofemoral compartment, which is typically exposed to much higher mechanical loading compared with the lateral compartment (Schipplein & Andriacchi 1991).

Since the majority of the mechanical loading that the lower limbs are exposed to during daily living result from walking and running, analysis of gait mechanics is typically used as a model to study the dynamic loading environment of the lower limb structures (Pandy & Andriacchi 2010). However, although the biomechanics of human locomotion is widely studied, there is still a substantial lack of knowledge regarding how different walking and running patterns affect lower limb joint loading. In addition, relatively little is known about how ageing alters lower limb joint mechanics as locomotion changes from walking to running.

Therefore, the purpose of this thesis was to investigate how locomotor pattern diversity and ageing affects lower limb joint mechanics and loading during human walking and running. Information obtained from this study may be helpful for the prevention and management of stress-related lower limb disorders and will provide insights for preventing locomotor impairments among the elderly population.

2 REVIEW OF THE LITERATURE

2.1 Lower limb mechanics during human locomotion

Bipedal striding is a unique feature for humans and it has two modes, walking and running. Walking, the most common mode of locomotion always involves ground contact with one or two feet, whereas running involves either a contact-phase or a flight-phase (Farley & Ferris 1998). A normal gait pattern during walking and running depends on several biomechanical features. Previous studies have provided essential information for understanding these features. The following section will summarize some of the basic biomechanical principles of human walking and running that provide a framework for understanding not only normal locomotion mechanics but also mechanisms that may explain pathological conditions of the locomotor system.

2.1.1 Point-mass mechanics of locomotion

Already in the seventeenth century, Borelli (1680, 126-161) presented classic mechanical models that described walking as vaulting over stiff legs and running as rebounding on compliant legs. Ever since, walking and running have been treated as different mechanical paradigms.

Walking is characterized by a pendulum-like motion, where the body center of mass (COM) travels over a relatively extended leg very efficiently. The advantage of this pendulum-like motion is that during each step cycle it exchanges forward kinetic energy (KE) for gravitational potential energy (GPE) between heelstrike and mid-stance and thus requires minimal mechanical work to produce COM motion along an arc (Farley & Ferris 1998) (Fig. 1A). When humans increase locomotion speed over 2 m/s they typically switch to running (Minetti, Ardigo & Saibene 1994). In contrast to walking, running includes a flight phase where the body's COM reaches its highest position, whereas the lowest position of the COM occurs near the mid-stance resulting in very different pattern of the energy exchange (Cavagna, Thys & Zamboni 1976). As a result, the bouncing gait of running can be described as a mass-spring system, where exchange of kinetic and potential energies occurs in the same phase (Fig. 1B). A mass-spring system allows muscle-tendon units to store elastic energy during the braking phase (eccentric phase) and release it during the following propulsion phase (concentric phase) (Cavagna Saibine & Maganaria 1964). This stretch-shortening type of action provides more efficient production of positive work to accelerate the body during propulsion than attained by a concentric muscle action alone (Komi 1984, Cavagna Saibine & Maganaria 1964).

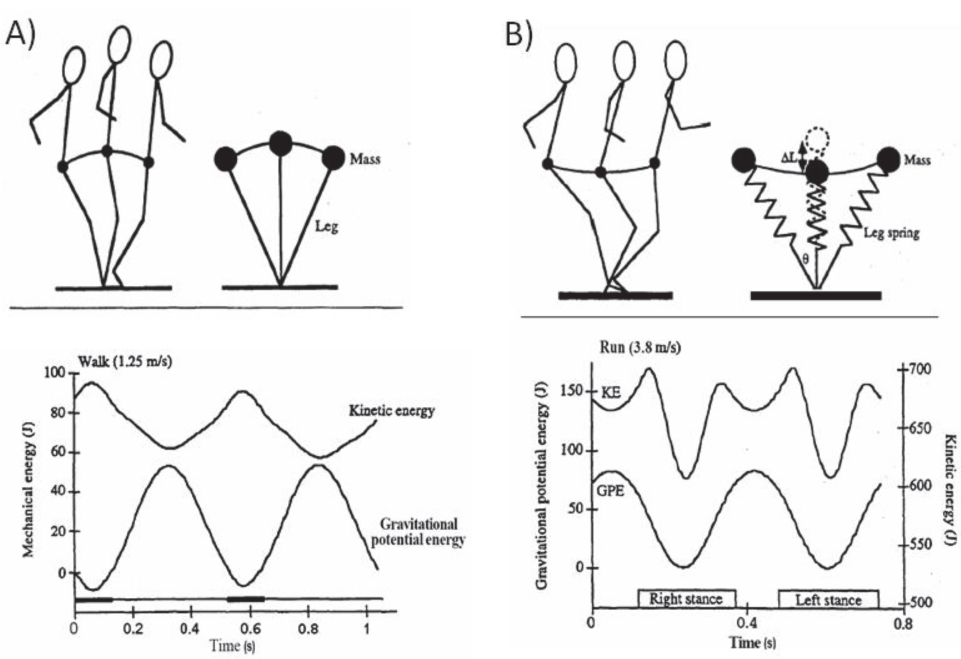


FIGURE 1. During walking (A) COM of the body and thus gravitational potential energy are highest during the mid-stance of gait. An inverted pendulum mechanism exchanges forward kinetic energy (KE) to gravitational potential energy (GPE) between heelstrike and mid-stance. During running (B), a mass-spring mechanism causes a decline in of both energies in the phase to a minimum between initial contact and mid-stance (modified from Farley & Ferris 1998).

2.1.2 Ground reaction forces

During ground contact, the lower limbs need to sustain the weight of the body against gravity while also providing enough force to move the body forward. The information about the forces generated by the lower limbs during locomotion can be obtained from the ground by mounted force plates. Due to gravity, the magnitude of the ground reaction forces (GRF) are greatest in the vertical direction where peak forces are about 1.2 times body weight (BW) during walking. The horizontal forward-backward GRF component is related to forward progression of the body's COM and reaches about 0.2-0.3 BW during walking (Winter 1984). As humans switch to slow running, peak vertical GRF is approximately twice as high as during walking (Nilsson & Thorstensson 1989) and further increased with higher running speeds (Weyand et al. 2000, Keller et al. 1996) (Fig. 2). Relatively high loading occurs also at the instant of initial contact, when the heel collides with the ground during rearfoot striking causing high vertical GRF impact peak and loading rates (Nigg 1997, Nilsson & Thorstensson 1989).

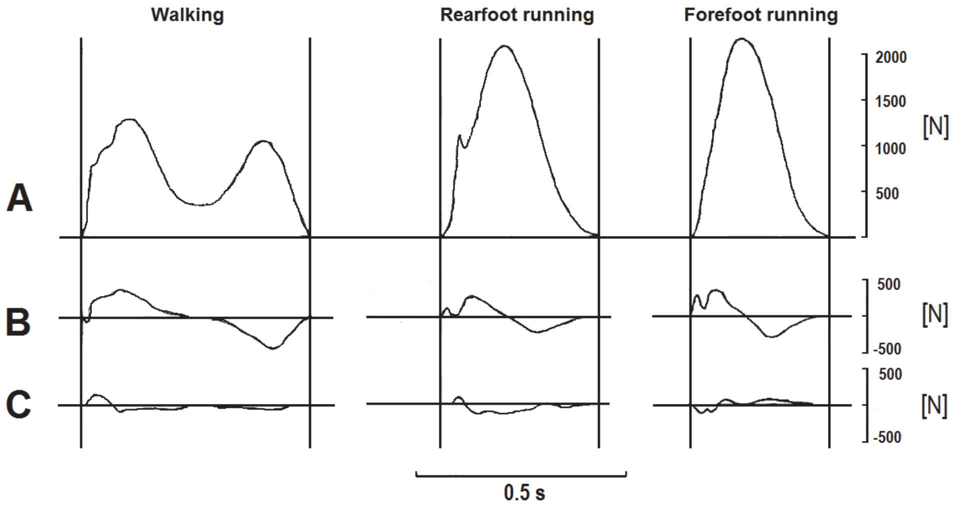


FIGURE 2. A) Vertical, B) anterior-posterior and C) medio-lateral GRF during walking and running with rearfoot and forefoot striking at 2.0 m/s (modified from Nilsson & Thorstensson 1989).

2.1.3 Joint dynamics

Human movement involves muscle-controlled rotations of body segments about their joint axis. Since early work of David Winter and colleagues in the late 1970's (Winter 1979, Winter & Robertson 1978), the analysis approach called inverse dynamics has become a standard method for determining joint dynamics. During legged locomotion, the analysis procedure involves acquisition of kinematic and GRF data by motion capture and force plate systems, respectively. Combination of these data sets via inverse dynamics computation provides information about the net joint moments and powers during the movement which reflect muscles' force generation and response to an external load due to gravity. During the stance phase, muscles must counteract external load (external joint moment) by generating internal joint moments in relation to the joint rotation axis (Winter 2009, 107-136). Besides knowledge of the muscle contributions to the movement, information about joint moments and forces are essential for understanding the amount of mechanical loads acting on the lower limb structures (Nigg & Herzog 1999).

Most of the motion during locomotion occurs in the sagittal plane. As a result, joint moments are largest in the sagittal plane for providing support and progression for the body. During walking, the knee and hip extensors are mainly responsible for weight acceptance during the early stance while ankle plantar flexors contribute most to the forward propulsion of the body's COM during the second half of the stance (Liu et al. 2006, Kepple, Siegel & Stanhope 1997, Neptune, Kautz & Zajac 2001). Therefore, a typical sagittal plane moment pattern during the first half of the stance phase of walking comprises the net hip and knee extensor moment driven by the gluteus maximus and quadriceps muscles, respectively, whereas during the latter part of the stance, a large ankle plantarflexor moment generated by the triceps surae muscle accelerates the body forward (Winter 1984). As locomotion mode change from walking to running, the knee is driven to a deeper flexion, associated with the spring-like function of the leg, which absorbs the impact of the body's weight during the first half of the ground contact (Novacheck 1998). This change in limb posture increases the moment arm of the ground reaction force in relation to the joint centers resulting in a decreased limb mechanical advantage, which, together with the greater ground reaction forces, substantially increases the amounts of joint moments during running when compared to walking (Biewener et al. 2004).

Modeling studies together with direct joint contact force measurements using instrumented hip and knee prostheses have revealed information about the loads acting on the lower limb joints. Even during walking, joint loads such as knee and hip contact forces exceed two to three times body weight, not only due to gravity, but also due to compression through muscle action (Paul 1966, Schipplein & Andriacchi 1991) (Fig. 3). Muscles crossing the joints comprise approximately two-thirds of the total joint contact forces while one-third is resulting from the GRF acting on the lower limb (Bergmann, Graichen & Rohlmann 1993, Bergmann et al. 2001). During running, higher external and internal forces acting on the lower limb may increase joint contact forces up 7-10 times BW (Edwards et al. 2008, Scott & Winter 1990).

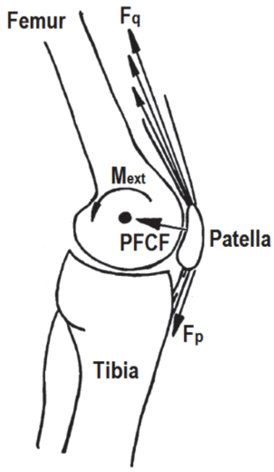


FIGURE 3. At the knee joint, the extensor moment (Mext) is a product of the quadriceps (Fq) and patella tendon (Fp) forces acting on the joint center. While these forces tend to extend the knee joint, they also cause internal joint loading such as patellofemoral contact force (PFCF) (modified from Scott & Winter 1990).

The secondary plane dynamics also play an important role in normal locomotion. Human bipedalism poses a major challenge to our locomotor system especially in the frontal plane, where the body's COM travels always medially from the base of support excluding a short double support phase during walking. To maintain lateral stability of the pelvis and the trunk, a sufficient internal frontal plane moment must be generated by the hip abductor muscles to counteract the force of gravity acting in the vertical direction (Winter 1995).

At the knee joint level, ground reaction forces pass medially at the joint center causing a moment (external knee adduction moment) that tends to move the knee towards varus alignment (Schipplein & Andriacchi 1991) (Fig. 4). As a result, a major percentage of the locomotion-related loading is directed to the medial compartment of the knee and therefore the magnitude of the frontal plane moment is frequently used as a surrogate measure for the knee joint loading (Miyazaki et al. 2002, Hinman et al. 2012a, Barrios et al. 2009, Baliunas et al. 2002). An increase of this moment indicates a shift of the axial tibiofemoral bone-on-bone contact force medially, which increases the loading of the medial compartment of the knee (Zhao et al. 2007, Schipplein & Andriacchi 1991). However, while the peak knee frontal plane moment is frequently used as a parameter to evaluate medial compartment loading, the magnitude of the knee extensor moment driven by the large quadriceps muscle also affects to the total joint loading, and should, thereby, be taken into account together with the frontal plane moment when gait-induced knee loads are evaluated (Walter et al. 2010).

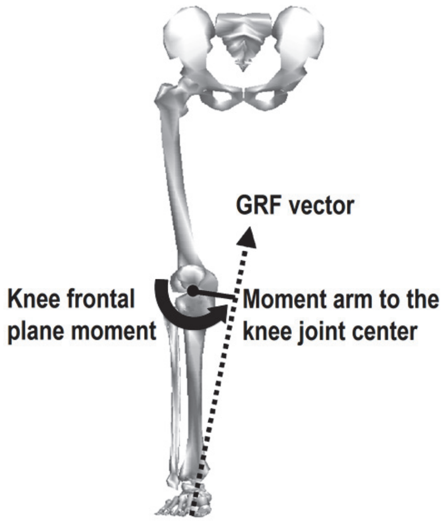


FIGURE 4. The GRF vector passes medial to the knee joint center resulting in frontal plane moment (internal abduction moment) that tends to adduct the knee joint. This leads to greater loading on the medial tibiofemoral compartment relative to the lateral (modified from Hinman & Bennell 2009).

2.2 Role of mechanical loading in the pathogenesis of stress-related injuries and disorders

Sufficient mechanical loading must be applied to the musculoskeletal system to maintain mechanobiological homeostasis and thus musculoskeletal health over a lifetime (Garber et al. 2011). If loads outside of the normal range are introduced, tissues of the body such as muscles, bones, and cartilages tend to adapt to accommodate the new loads (Khan & Scott 2009, Kannus et al. 1995). However, excessive loading can also have adverse effects for the musculoskeletal health. For example, there is a growing body of literature demonstrating that cumulative effects of repeated loading play a significant role in the development of stress-related injuries and disorders (Horisberger et al. 2013, Roemhildt et al. 2012, Scott & Winter 1990, Miyazaki et al. 2002, Chehab et al. 2014). Since the majority of the mechanical loading of daily living results from human locomotion, analysis of gait mechanics is typically used as a model to study the dynamic loading environment of the lower extremity during locomotion (Pandy & Andriacchi 2010). Traditionally, GRFs have been used to experimentally investigate the total mechanical loading that the lower limbs are exposed to during locomotion (Stacoff et al. 2007, Keller et al. 1996), while joint moments and forces reflect a local loading environment of the lower limb joints (Schipplein & Andriacchi 1991, Zhao et al. 2007).

2.2.1 Mechanical factors and the risk of osteoarthritis

Osteoarthritis (OA) is a common disease of the entire joint causing remarkable disability for the patient and extensive costs for the modern society (Dieppe & Lohmander 2005, Leskinen et al. 2012). In particular, osteoarthritis affects the articular cartilage, which covers synovial joint surfaces, and normally provides nearly frictionless sliding and good shock absorption capacity for the joints (Buckwalter & Mankin 1997). The risk of OA increases with age (Urwin et al 1998). It has been suggested that both biological and mechanical factors play a role in OA development, and that the progression of OA is a dynamic process in which the cumulative loads applied to the articular cartilage cause damage that can not be repaired by the biological cartilage regeneration mechanism (Andriacchi et al. 2004). Damaged cartilage loses its key biomechanical features, which leads to compromised joint function that includes pain as well as a decrease in muscle strength, joint range of motion (ROM), and stability (Dieppe & Lohmander 2005). Unfortunately, current treatment methods do not enable healing of this disease process, and thus joint replacement is often needed for the treatment of OA.

The large weight bearing joints, such as knee and hip, are commonly affected by the OA (Andriacchi et al. 2004). The most common site is the medial tibiofemoral compartment of the knee (McAlindon et al. 1992, Ahlbach 1968). Considerable evidence suggests that high gait-induced knee frontal plane moment (internal abduction moment), which increases medial compartment loading, plays an important role in OA initiation and progression (Schipplein & Andriacchi 1991, Baliunas et al. 2002, Roemhildt et al. 2012, Miyazaki et al. 2002, Chehab et al. 2014). Further support for a mechanical pathway of OA include observations that greater knee varus alignment strongly corresponds with higher knee abduction moment (Andrews et al. 1996) and subsequent medial knee OA progression (Cerejo et al. 2002, Sharma et al. 2001).

2.2.2 Mechanical loading and the risk of lower extremity overuse injuries

A most widely accepted theory regarding the pathomechanics of overuse injuries suggests that the symptoms are resulting from cumulative overloading of the musculoskeletal system, which causes damages to the structures when the healing period is not adequate for structural adaptation to take place (Horisberger et al. 2013, Winter & Bishop 1992, van Mechelen, Hlobil & Kemper 1992). In the lower extremities, overuse injuries are particularly common among sports such as running, where musculoskeletal structures are exposed to a high number of loading cycles. Previous studies have reported overall injury rate from 37 to 59 % among runners over 12 months period (Ristolainen et al. 2010, van Mechelen 1992).

It has been proposed that high vertical GRF impact peak and/or loading rate may increase the risk of running-related injuries such as bone stress fractures (Zadpoor & Nikooyan 2011, Milner et al. 2006). However, not all studies have found the association between injuries and the impact characteristics of the vertical GRF (Nigg 1997, Bredeweg & Buist 2011), which may be attributed to differences in subject characteristics and experimental conditions. Moreover, loading of the lower extremity structures does not depend solely on the amount of impact forces, but also on the internal loading such as joint moments and forces acting on the structure of interest. In fact, it has been reported that the peak vertical impact force had little effect on the peak internal loading forces acting on the chronic running injury sites (Scott & Winter 1990). Therefore, the amount of internal loading may play more important role in the development of running-related injuries than impact characteristics of the vertical GRF (Nigg & Enders 2013).

One third of all running-related injuries involve the knee joint and the patellofemoral joint accounts for approximately 60 % of these injuries (van Mechelen 1992, Clement et al. 1981). It has been suggested that patellofemoral pain (PFP) results from repetitive patellofemoral joint loading, which leads to pathological changes in the subchondral bone, including bone microfractures (Radin & Rose 1986, Horisberger et al. 2013), increased subchondral bone metabolic activity (Draper et al. 2012), and elevated bone water content (Ho et al. 2013). Furhermore, these changes are believed to predispose to the initiation of patellofemoral joint OA (Utting, Davies & Newman 2005, Hinman et al. 2013, Farrokhi, Colletti & Powers 2011).

As humans switch from walking to running, a substantial increase in patellofemoral joint loading occurs, which may explain why the patellofemoral joint is so susceptible to injury due to running (Scott & Winter 1990). Modelling studies of human locomotion have reported that a peak contact force in walking is less than one times BW (Heino Brechter & Powers 2002, Mason et al. 2008) but increases up to 5-7 BW during running (Flynn & Soutas-Little 1995, Roos, Barton & van Deursen 2012). Furthermore, a previous study of Stefanyshyn et al. (2006) examining lower extremity mechanics of runners found a link between high running-induced frontal plane moment and the development of patellofermoral pain (PFP). The researchers suggested that PFP potentially resulted from elevated patellofemoral joint loading due to higher knee frontal plane moment. In addition, recent evidences suggest that altered hip and/or knee movements such as excessive hip adduction and rotation or large knee abduction are associated with the development of PFP (Myer et al. 2010, Noehren, Hamill & Davis 2012, Boling et al. 2009, Giddings et al. 2000). It has been shown that these movement patterns tend to increase forces on the lateral aspect of the patellofemoral joint, which may thus contribute to the pathomechanics of the PFP (Lee et al. 1994, Huberti & Hayes 1984, Souza et al. 2010).

The Achilles tendon is another commonly injured site in the lower extremity. Mechanical loading of the Achilles tendon may reach tensile forces of 2-4 BW during walking (Finni, Komi & Lukkariniemi 1998, Giddings et al. 2000) and 6-8 BW during running (Scott & Winter 1990, Giddings et al. 2000). Repeated exposure to high tensile forces are thought to be a major precipitating factor of Achilles tendinopathy, which is a common overuse condition among physically active people (Asplund & Best 2013). The support for a mechanical pathway of Achilles tendinopathy includes observations that a high strain-induced loading of the tendon typically causes microruptures of the tendon's collagen fibrils leading to local stress deprivation (Riley 2008, Järvinen et al. 2005).

2.3 The effects of subject-specific locomotor patterns and aging on the lower limb mechanics and loading during walking and running

Results from direct joint contact force measurements using instrumented knee and hip prostheses suggest that there is a considerable inter-individual variability in the musculoskeletal loading conditions during locomotion (Heinlein et al. 2009, Bergmann et al. 2001). Because excessive mechanical loading is known to play a critical role in several musculoskeletal pathologies, it is important to understand the underlying factors associated with increased loading conditions. While much of the research to date have investigated how factors such as overweight, leg alignment, footwear, or injuries and disorders affect walking and running mechanics (Browning & Kram 2007, Baliunas et al. 2002, Shin et al. 2009, Devita, Hortobagyi & Barrier 1998, Besier et al. 2009, Erhart et al. 2008), only little is known about the effects of different walking and running patterns on the lower limb joint loading. Also, it remain unknown how ageing alters lower limb mechanics as locomotion changes from walking to running to sprinting.

2.3.1 Inter-individual variability in waking mechanics

Already early studies of human locomotion observed a large inter-individual difference in walking mechanics. Pedotti (1977) was one of the first researchers who found that, despite similar kinematics, the joint moment patterns together with muscle activation profiles showed large differences between subjects during walking (Pedotti 1977). Another early demonstration of inter-individual variability was carried out by Winter (1984), who observed that the highest variability in joint kinetics was present at the knee joint level. This finding together with more resent observations (Simonsen et al. 1997, Simonsen & Alkjaer 2012) shows that the main variations in knee kinetics involves either clear dominance of the knee extensor moment during the whole stance phase of gait carried out by the greater quadriceps effort, or alternatively, lower extensor moment during the loading response and a clear dominance of knee flexor moment during the terminal stance driven by an increased contribution of the hamstring muscles (Fig. 5). Interestingly, different walking patterns may lead to clearly different lower limb loading profiles. A study of Simonsen and Alkjaer (2011) found that those who walk with knee extensor dominant gait patterns exhibit a clearly higher tibiofemoral joint contact force compared to people with knee flexor dominant gait patterns.

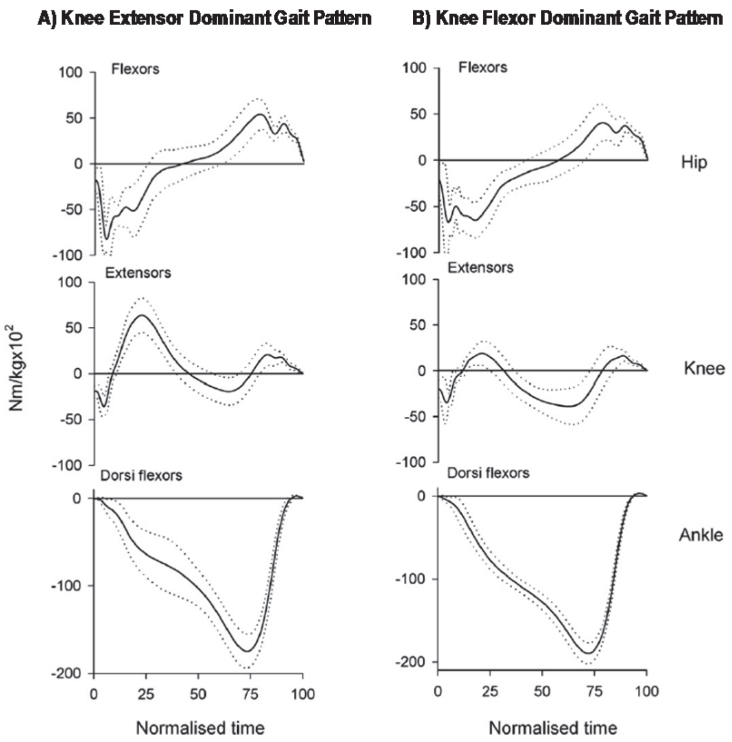


FIGURE 5. The main variations in the joint moment patterns during walking typically involve either A) clear dominance of the knee extensor moment during the whole stance phase, or alternatively, B) low knee extensor moment during early stance phase following dominance of knee flexor moment (adapted from Simonsen and Alkjaer 2012 with permission from Elsevier LTD).

2.3.2 Inter-individual variability in running mechanics

In addition to distinct walking patterns, people demonstrate clear differences in running patterns, especially in foot placement during initial ground contact. As a result, previous studies have typically categorized runners into rearfoot strikers (RFS), mid-foot strikers (MFS) or forefoot strikers (FFS) based on their landing strategy (Altman & Davis 2012, Cavanagh & Lafortune 1980, Lieberman et al. 2010). The reported presence of RFS pattern varies between 75-90% among long distance shod runners while the rest of the runners use either MFS or FFS patterns (Hasegawa, Yamauchi & Kraemer 2007, Larson et al. 2011).

Biomechanical comparison of the different foot strike patterns have shown that runners with FFS demonstrate lower vertical GFR impact peak and reduced vertical GRF loading rate (Lieberman et al. 2010, Cavanagh & Lafortune 1980). High impact forces and/or loading rate are often suggested to play a role in the development running-related injuries such as bone stress-fractures (Zadpoor & Nikooyan 2011, Milner et al. 2006). Reductions in the impact force and loading rate in FFS are due to a plantarflexed foot at initial contact, followed by a dorsiflexion movement driven by eccentric work of the ankle plantarflexors (Cavanagh & Lafortune 1980, Lieberman et al. 2010). This movement strategy increases activation of the calf muscles (Giandolini et al. 2013) because of higher plantarflexor moment (Williams, McClay & Manal 2000) and Achilles tendon strain (Perl, Daoud & Lieberman 2012) compared with those who use RFS pattern. At the knee joint level, running with the FFS pattern has been associated with lower eccentric quadriceps work during the braking phase of the ground contact when compared with running with the RFS pattern (Arendse et al. 2004). Based on these findings, it has been speculated that increased ankle contribution during FFS could potentially decrease efforts of the quadriceps muscle, resulting in lower knee joint loading. Furthermore, recent studies examining the effects of foot striking pattern on running-related injuries suggest lower injury incidence in the knee and hip joints for the runners with FFS strategy compared with RFS runners (Daoud et al. 2012) while the incidence of the foot and ankle injuries have been found to increase (Ridge et al. 2013) or stay essentially similar (Daoud et al. 2012).

2.3.3 The effects of ageing on the locomotion mechanics

It is well known that aging of the human body leads to progressive impairments in physical performance. Cross-sectional (Danneskiold-Samsoe et al. 2009) and longitudinal (Frontera et al. 2000) studies have demonstrated significant reductions in muscle force generation capacity in older age, primarily originating from a loss of muscle fibres and motor-units as well as muscle quality deterioration (i.e. loss of strength per unit of muscle mass) (Doherty 2003). Moreover, neural adaptations such as reduced central drive (Stevens et al. 2003) and reflex sensitivity (Kallio et al. 2010) may also contribute to performance impairments with age.

These detrimental effects of the aging process on the neuromuscular system lead to locomotor decline (Rantakokko, Mänty & Rantanen 2013) and are associated with changes in walking mechanics (Winter 1990). In general, older adults exhibit a slower self-selected walking speed with a shorter step length, shorter relative swing phase, and less ROM at the ankle and hip compared with younger adults (Winter et al. 1990, Judge, Davis & Ounpuu 1996, Kerrigan et al. 2001, Kerrigan et al. 2001). The underlying kinetic changes comprise lower horizontal ground reaction force and reduced muscle power generation especially from the ankle plantarflexors (Winter et al. 1990, Judge, Davis & Ounpuu 1996). However, because many of these parameters are speed dependent (Winter 1984), changes in walking mechanics can be attributed to slower walking speed and shorter steps among older adults. Studies comparing walking mechanics in older versus younger people at the same walking speed have demonstrated a reduction in ankle plantarflexor power generation but an increased power generation from the hip and/or knee extensors among older adults (Cofre et al. 2011, Boyer, Andriacchi & Beaupre 2012, Monaco et al. 2009, DeVita & Hortobagyi 2000, Kerrigan et al. 1998) (Fig. 6). This so-called distal to proximal shift in joint powers in walking is suggested to be a compensation strategy for the elderly to accommodate diminished force generation capacity of the lower limbs (Beijersbergen et al. 2013).

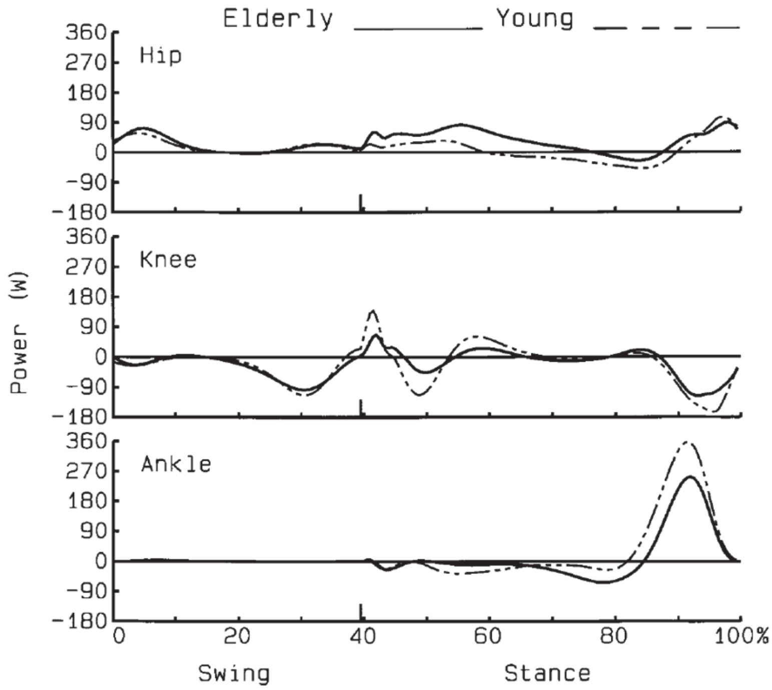


FIGURE 6. Lower limb joint powers during walking exhibit a distal to proximal shift in older age (adapted from DeVita and Hortobagyi 2000 with permission from the American Physiological Society).

Compared with walking, running poses even greater challenges for aging humans. Previous research have shown that at a given speed, older runners exhibit lower GRF and take shorter steps at a higher frequency compared with younger runners (Bus 2003, Cavagna, Legramandi & Peyre-Tartaruga 2008). Other findings include a larger knee flexion angle observed at initial ground contact but a reduction in the knee flexion excursion during the first half of the stance phase in ageing adults (Bus 2003, Fukuchi & Duarte 2008). To date, only two studies have examined lower limb joint kinetics in older versus younger runners. The results of these studies show that older runners exhibit lower ankle plantarflexor moment and power generation, but exhibit no differences in the kinetics of the knee and hip joints when running at 2.7 m/s. At maximal sprinting, where greater GRF generation is required to support and propel the body (Weyand et al. 2000, Keller et al. 1996), older adults demonstrate reduced GRF generation, decreased step length and increased ground contact time when compared with younger counterparts (Korhonen et al. 2009). So far, no studies have examined the effects of ageing on lower limb joint kinetics at higher running speeds.

In addition to sagittal plane changes, studies have reported age-related differences in the secondary plane mechanics during locomotion. Results from walking studies suggest that older adults exhibit a reduction in hip adduction movement during the stance phase, which may represent a compensatory strategy to maintain lateral balance (Ko, Hausdorff & Ferrucci 2010, Ko et al. 2009). Recent studies have also reported age-related changes in the knee frontal plane mechanics during running. Fukuchi et al (2013) observed greater knee frontal plane moment impulse in old compared with young male runners. An increased frontal plane moment has also been reported in mature female runners (Lilley, Dixon & Stiles 2011). Since high knee frontal plane moment has been considered a predictor of both patellofemoral pain (Stefanyshyn et al. 2006, Stefanyshyn et al. 1999) and knee OA (Miyazaki et al. 2002) increased knee frontal plane moment may potentially increase the risk of these disorders. However, underlying mechanisms explaining higher knee frontal plane moment among elderly runners remain unclear.

3 PURPOSE OF THE STUDY

Many lower limb disorders, such as knee OA, patellofemoral pain and Achilles tendinopathies, are increasingly seen as disorders where mechanical loading plays a major role in both initiation and progression of the pathology and severity of clinical symptoms. Although previous studies have shown that locomotor patterns during walking and running exhibit large differences due to inter-individual variability and ageing, there is still a substantial lack of research evaluating whether these differences expose certain musculoskeletal structures to higher than normal loading. Better understanding of the overloading mechanisms and age-related alterations in locomotion could advise us to recognize those persons who potentially have a higher risk of initiation and progression of stress-related lower limb disorders and thus may offer valuable information for prevention purposes. Therefore, the overall goal of this work was to investigate the effects of inter-individual variability and aging on the lower limb joint mechanics and loading profiles during walking and running. The thesis is based on the four original research papers, which are referred to in the text by Roman numerals. The specific aims of these papers can be characterized as follows:

1) Walking patterns across the normal healthy population exhibit a large inter-individual variability, especially at the knee joint, where movement during the stance phase of gait can be driven with either knee extensor or flexor dominant gait strategies between the knee extensors and flexors. However, the effects of different walking patterns on medial knee loading remain unclear. The aim of study one was to examine whether distinct walking patterns are associated with different medial loading profiles at the knee joint (paper I).

2) Runners can be categorized into RFS and MFS of FFS based on the landing strategy at the instant of initial ground contact. However, there is a lack of understanding regarding the effects of a runner's foot strike pattern on the Achilles tendon and especially patellofemoral joint

loading, which are two of the most commonly injured sites of the body among runners. Therefore, study two compared loading profile of the knee and ankle joints of the runners with a RFS versus FFS pattern (paper II).

3) Structural and functional alterations due to aging lead to locomotor decline and predispose elderly people to overloading of the musculoskeletal system. Despite previous work emphasizing age-related biomechanical alterations during low intensity locomotion, it remains unclear how lower limb mechanics change during more intensive locomotion with age. Therefore, study three was set up to examine: i) how ageing affects sagittal (paper III) and ii) secondary plane mechanics (paper IV) across different modes and intensities of locomotion.

4 RESEARCH METHODS

4.1 Subjects and design

Altogether 349 healthy subjects participated in these studies. Subjects in papers I, III, and IV were males, and in paper II females. A questionnaire on participants' background information was used to confirm that they did not have a previous history of any musculoskeletal problems, such as a recent injury or surgery, which could have an effect on the locomotion pattern of the subject. The study was approved by the local ethics committee and was performed in the accordance with the Declaration of Helsinki.

Cross-sectional study designs were used to examine biomechanical differences between selected groups during locomotion. Selected characteristics of the study subjects are described in table 1. In paper I, a cluster analysis was used to divide subjects into different groups based on their knee joint moment profiles (see details p.33). As a result, three groups of clearly different knee joint moment profiles were identified. The typical gait pattern group (TYP) consisted of 11 subjects and the knee extensor (EXT) and flexor (FLX) dominant gait pattern groups consisted seven and six subjects, respectively. In paper II, 19 subjects with FFS technique were recognized among 286 measured athletes by determining foot strike angle during initial ground contact according to Altman and Davis (2012). A pair-matched group of rearfoot strikers (RFS) included 19 females with similar weight and height. In paper III and IV, three different age groups of competitive healthy male athletes (sprinters, long jumpers) with several years of training background participated in the experiments.

TABLE 1 Mean (SD) characteristics of the study subjects.

	N	Age (yr)	Height (m)	Weight (kg)
Paper I				
Group TYP	11	42 (12)	1.80 (0.05)	84.0 (13.3)
Group EXT	7	35 (8)	1.81 (0.04)	83.8 (5.6)
Group FLX	6	43 (11)	1.82 (0.06)	79.2 (11.0)
Paper II				
All subjects	286			
FFS runners	19	19 (5)	1.69 (0.05)	63.2 (9.2)
Matched RFR runners	19	18 (4)	1.69 (0.05)	62.8 (8.6)
Papers III & IV				
Young athletes	13	26 (6)	1.81 (0.04)	73.3 (8.0)
Middle-aged athletes	13	61 (5)	1.78 (0.06)	79.6 (9.6)
Old athletes	13	78 (4)	1.72 (0.06)	69.7 (7.8)

Measurements were conducted in a motion analysis laboratory (papers I and II) and in an indoor sports hall (papers III and IV, Fig. 7). After familiarization and a warm-up period, subjects performed several walking trials at a self-selected speed (papers I, III and IV), running trials at 4.0 ± 0.2 m/s (papers II-IV) and two 60 m sprinting trials at maximal effort (III-IV). The speed was monitored with photocells in the middle section of the runway, which was also used as a capture area for the motion analysis. Subjects used their own running shoes during experiments.

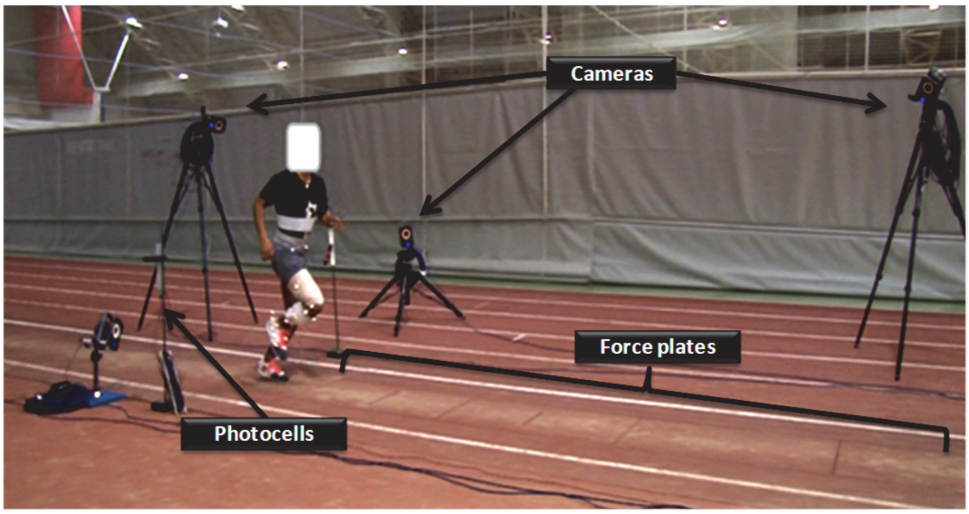
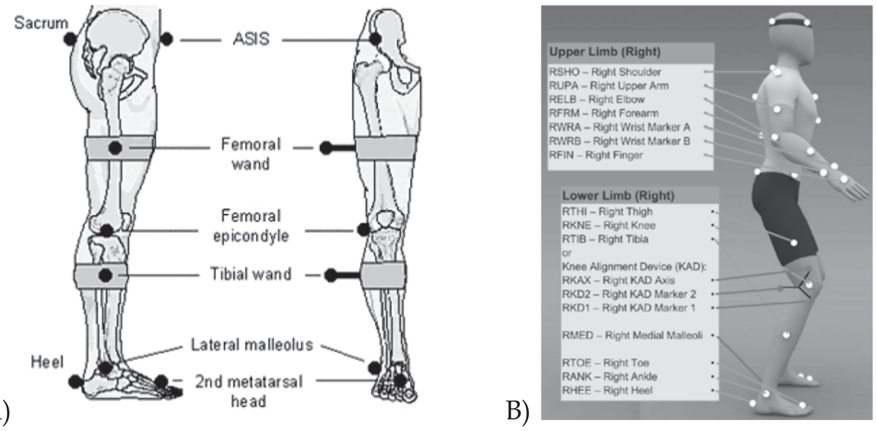


FIGURE 7. Measurement setup in an indoor sports hall.

4.1.1 Analysis of walking and running mechanics

In walking measurements (paper I), a five camera system (Cohu CCD camera, San Diego, CA) and a force platform (9861A, Kistler Instrument AG, Winterthur, Switzerland) were used to collect marker positions and ground reaction force data synchronously at 50 and 2,000 Hz, respectively. Motion (300 Hz) and GRF (1500 Hz) data for papers II-IV were collected by using an eight-camera system (Vicon T40, Oxford, UK) and force platforms (AMTI, Watertown, MA, USA). Anthropometric measurements and placement of retro-reflective markers were performed according to a modified Helen Hayes model (Vaughan, Davis & O'Connor 1999, 15-43), paper I) and full body Plugin gait model (Vicon Nexus v1.7; Oxford Metrics, Oxford, UK, papers II-V, Figure 8).



A) B)

FIGURE 8. Marker placement was carried out according to A) a modified Helen Hayes model (Vaughan, Davis & O'Connor 1999) and B) full body Plugin gait model (Vicon, Oxford, UK).

Data analysis consisted of low-pass filtering of the marker trajectories and GRF data with a fourth order Butterworth filter. Cut-off frequencies for marker and GRF data were 6 and 50 Hz in paper I and 12 and 50 Hz in papers II. In papers III and IV, cut-off frequency of 18 Hz for both GRF and motion data was used to avoid impact artefacts during sprinting (Bisseling & Hof 2006, Bezodis, Salo & Trewartha 2013). Inverse dynamics approaches according to a modified Helen Hayes model (paper I) and Plug in Gait model (papers II-IV) were used to calculate joint moments and powers. Knee frontal plane moment (internal knee abduction) together with knee extensor moment plane were used to evaluate total medial knee joint loading (Zhao et al. 2007, Walter et al. 2010). Kinematic and kinetic data during the stance phase were time normalized (0-100%) and averaged across several contacts during walking and running of the selected leg. Foot contact and toe-off events were used to calculate cadence, step length, and width.

In paper II, impact loading characteristics during running was determined by analyzing vertical GRF impact peak and loading rate using Signal software (v.4.1; Cambridge Electronic Design, Cambridge, UK). First, mean time from ground contact (vertical force threshold 20 N) to impact peak was determined for the RFS (0.0282 s). This time point was then used to determine the magnitude of the vertical impact force for FFS (Lieberman et al. 2010). Average vertical loading rate was calculated as the total change in force divided by the total change in time between 20% and 80% of the period between ground contact and vertical impact peak (Milner et al. 2006). Calculation of the position of the COM was performed using the Plug-in Gait model. COM–heel distance in the anterior–posterior direction was determined during initial ground contact as the difference between heel marker and COM.

To estimate knee and ankle loading in paper II, inverse dynamics driven biomechanical models were used to calculated patellofemoral contact force (PFCF) and stress (PFS) and Achilles tendon force (ATF) during running. PFCF was calculated as a function of knee flexion angle (x) and knee extensor moment (Mk) according to Ho et al (2012). First, an effective moment arm of the quadriceps muscle (L_q) was calculated as a function of knee flexion angle using a nonlinear equation, which is based on the cadaver data reported by van Eijden et al. (1986):

$$L_q = 8.0E^{-5}\,x^3 - 0.013x^2 + 0.28x + 0.046 \qquad [1]$$

Second, quadriceps force (Fq) was calculated as follows:

$$Fq = Mk\;/\;L_q \qquad [2]$$

Finally, PFCF was calculated as the product of the quadriceps force (Fq) and a constant (k):

$$PFCF = F_q{*}k \qquad [3]$$

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. (1985):

$$k(x) = (4.62E^{-1} + 1.47E^{-3}\,x - 3.84E^{-5}\,x^2)\;/\;(1\text{-}1.62E^{-2}\,x + 1.55E^{-4}\,x^2 - 6.98E^{-7}\,x^3) \qquad [4]$$

PFS was then calculated as the PFCF divided by the patellofemoral contact area. Contact area was estimated according to Ho et al. (Ho, Blanchette & Powers 2012) by fitting a second-order polynomial curve to the data of Powers et al. (Powers, Lilley & Lee 1998) (83 mm^2 at 0°; 140 mm^2 at 15°; 227 mm^2 at 30°; 236 mm^2 at 45°; 235 mm^2 at 60°; 211 mm^2 at 75° of knee flexion).

$$PFS = PFCF\;/\;\text{contact area} \qquad [5]$$

ATF was determined by dividing the plantarflexion moment (calculated by inverse dynamics) by the estimated Achilles tendon lever arm (L_a) as described by Self and Paine (2001):

$$ATF = M_a / L_a \quad [6]$$

$$L_a = -0.5910 + 0.08297a - 0.0002606\,a^2 \quad [7]$$

where a = ankle angle

4.1.2 Cluster analysis

According to previous studies, the largest inter-individual variability in joint moment patterns exists in the knee joint level (Pedotti 1977, Simonsen et al. 1997, Winter 1984). Therefore, in paper I, subjects were categorized into different gait pattern groups using K-means algorithm on data consisting of five knee joint moment parameters (peak values during the stance phase): knee frontal plane moment (maximum abduction moment of first or second peak), knee extensor moment, knee flexor moment, and knee external rotation and internal rotation moments. Prior to data clustering, the variables were standardized by linear scaling into the closed interval [0, 1]. The relevant dimensions were chosen by a principal component analysis (PCA) method. The first three principal components, which accounted for >90% of the total variance, were used as the input variables to the clustering algorithms. Three clustering models, for K = {2, 3, 4}, were analyzed. To avoid locally optimal clustering solutions, 100 restarts from random initial prototypes points were applied for each model. The models were validated visually through silhouette plots and comparing average silhouette indices: 0.41 (K = 2), 0.42 (K = 3), and 0.46 (K = 4) (Rousseeuw 1987). Regardless of the silhouette indices, the three-cluster model was preferred over the four-cluster model, because the group sizes in the four-cluster model were too small for further analysis. For the three-cluster model, the numbers of subjects assigned into the clusters were 6, 7, and 11. The obtained K-means model was evaluated by relative validation using the K-spatial medians clustering method, (Äyrämö, Kärkkäinen & Majava 2007), which is based on non-parametric multivariate statistics that are less sensitive to outliers and deviating observations. No differences were observed between the K-means and K-spatial medians based partitions.

4.1.3 Hip abductor strength and navicular drop

In paper II, hip strength and anthropometric parameters were measured to exclude their effects on between-group running mechanics. Isometric strength of the hip abductor muscles was tested with a hand-held dynamometer (Baseline; Fabrication Enterprises, Elmsford, NY) based on the previously published method (Krause et al. 2007, Thorborg et al. 2010). Subjects were in a supine position, with the legs extended and the ankles dorsiflexed on the testing table. Two straps were used to stabilize the pelvis and trunk against the testing table. The examiner positioned the dynamometer approximately 2 cm proximal from the lateral malleolus, after which, the subject performed one submaximal practice trial and two test trials (at least 10-s rest period between trials) against the dynamometer fixed by the examiner. A 3-s isometric maximum voluntary contraction was conducted, and the average of two successfully completed test trials was selected for the analysis. Two research physical therapists carried out the tests. The method has been shown to have intratester and intertester reliabilities ranging from 0.73 to 0.97 (Krause et al. 2007, Thorborg et al. 2010).

Navicular drop was used to assess foot structure (Mueller, Host & Norton 1993). Navicular drop was defined as the difference (mm) between navicular height in barefoot standing with the subtalar joint in a neutral position and in a relaxed stance. Navicular tuberosities were palpated and marked with a pen. To determine navicular height in the subtalar joint in a neutral position, the examiner palpated the medial and lateral prominence of the talus with the thumb and forefinger during pronation and supination of the foot. The neutral position of the subtalar joint was determined when the talar prominences were congruent medially and laterally. From this position, the distance between the ground and the navicular mark was measured. The subject was then instructed to perform walking in place and stop in relaxed stance, then the distance between navicular mark and ground was measured again. One research physical therapist carried out the tests. This method has been found to have intratester and intertester reliabilities ranging from 0.73 to 0.96 (Mueller, Host & Norton 1993, Sell et al. 1994).

4.1.4 Tibiofemoral angle

In papers II-IV, tibiofemoral angle was determined from a static standing position using motion analysis. Knee joint mechanical axis (varus alignment was when angle was >0° and valgus when angle was <0°) in the frontal plane was defined as an angle between ankle, knee, and hip joint centers calculated by the Plug-in Gait model. This procedure has been shown to estimate mechanical axis alignment similar to full-limb weight-bearing radiographs ($R^2 = 0.54$) (Mundermann, Dyrby & Andriacchi 2008). In paper I, the dynamic knee adduction angle, which relates directly with static alignment (Barrios et al. 2009), was used to determine varus-valgus alignment in the subjects. The lack of between group differences in the tibiofemoral angle suggests that similar knee alignments ware present among the compared groups in this thesis.

4.1.5 Statistical analysis

All statistical comparisons were performed with SPSS software (Version 18.0, SPSS, Chicago, Illinois, USA). Means and standard deviations (SD) were determined for the parameters of interest. Shapiro-Wilk and Levene's tests were used to assess the normality of distribution and the equality of variances, respectively. P values less than 0.05 were considered significant.

In paper I, The univariate between-cluster differences of selected parameters, including subject, spatio-temporal, kinetic and kinematic variables were tested using nonparametric Kruskal–Wallis test. When a significant difference was found, the Mann–Whitney U-test with Bonferroni adjustment was performed to compare two clusters.

In paper II, subject, spatiotemporal, and kinematic and kinetic parameters were compared between FFS and RFS with two-tailed independent t-tests.

In papers III and IV, differences in subject, spatiotemporal, and kinematic and kinetic parameters between three age groups were compared using one way ANOVA with Bonferroni adjustment.

5 RESULTS

The main findings of the thesis are presented in this chapter. Original papers (I-IV) should be consulted for additional details.

5.1 Different walking patterns and medial knee joint loading

The cluster analysis identified three groups with clearly distinct knee joint moment profiles (Fig 8). The largest cluster consisted of 11 subjects and showed a typical knee extensor moment profile during the stance phase. Therefore, this cluster was categorized as the typical gait pattern (TYP). For a cluster of seven subjects, the knee extensor moment was apparent during almost the entire stance phase and was thus categorized as the knee extensor dominant gait pattern (EXT). A cluster of six subjects showed the lowest knee extensor moment and a clear dominance of the knee flexor moment during terminal stance and was thus identified as the knee flexor dominant gait pattern (FLX).

Knee sagittal plane kinetics were statistically different among the three clusters. EXT produced 33% higher extensor moment compared with TYP (0.86 ± 0.11 vs. 0.58 ± 0.17 Nm/kg, $P<0.05$) and 59% higher compared with FLX (0.35 ± 0.10 Nm/kg, $P<0.01$, Fig. 1). The opposite pattern was present in the knee flexor moment where FLX (0.55 ± 0.13 Nm/kg) produced 76% and 44% higher peak values than EXT (0.13 ± 0.18 Nm/kg, $P<0.01$) and TYP (0.31 ± 0.13 Nm/kg, $P<0.01$), respectively (Fig. 1). In the frontal plane, TYP cluster demonstrated significantly lower knee frontal plane moment (0.35 ± 0.11 Nm/kg) showing a 43% lower first peak of knee abduction moment compared with EXT (0.61 ± 0.16 Nm/kg, $P<0.01$) and a 44% lower second peak of knee abduction moment compared with FLX (0.50 ± 0.11 Nm/kg, $P<0.05$, Fig. 1). In addition, consistent differences existed among clusters in the knee transversal plane kinetics, which showed significantly lower external rotation moment in the TYP (0.12 ± 0.04 Nm/kg) compared with EXT (0.17 ± 0.03 Nm/kg, $P<0.05$) and FLX (0.20 ± 0.02 Nm/kg, $P<0.01$, Fig. 1).

Kinematics among the clusters showed a difference especially at the knee joint. The subjects in the EXT cluster demonstrated a significantly larger knee flexion angle at the instant of heel contact compared with FLX (9.3 ± 4.7 vs. 0.7 ± 3.4°, P<0.01, Fig. 2). Similarly, during loading response, EXT showed a significantly higher knee flexion angle (24.5 ± 6.3°) than FLX (10.5 ± 3.8°) and TYP (16.8 ± 5.6°) clusters (P<0.01, P<0.05), respectively (Fig. 2). Different kinematic profiles were also observed at the ankle joint where a significantly larger peak ankle plantarflexion angle during the loading response was present in the FLX compared with EXT (12.6 ± 4.7 vs. 5.6 ± 4.3°, P<0.05, Fig. 9). No statistically significant differences among the clusters were found for subject (age, height and weight) or spatio-temporal parameters (speed, step length and width).

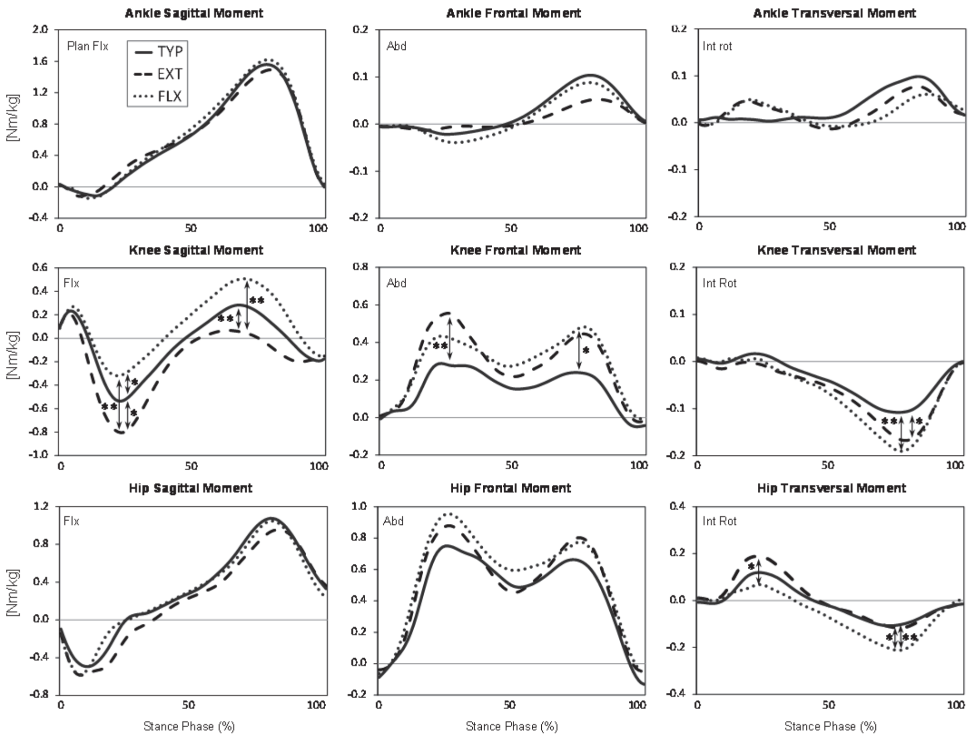


FIGURE 8. Mean internal joint moment curves over the stance phase of gait for the typical gait pattern (TYP), knee extensor dominant gait pattern (EXT), and knee flexor dominant gait pattern (FLX) (*P<0.05, **P<0.01).

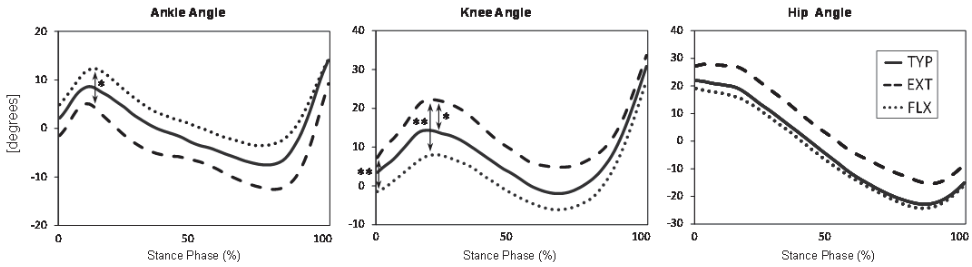


FIGURE 9. Mean joint angle curves over the stance phase of gait for the typical gait pattern (TYP), knee extensor dominant gait pattern (EXT), and knee flexor dominant gait pattern (FLX) (*P<0.05, **P<0.01).

5.2 Lower limb loading during rearfoot and forefoot running

Knee kinematic and kinetic parameters for RFS and FFS runners are described in figure 10. FFS demonstrated 16% lower PFCF (4.3 ± 1.2 vs. 5.1 ± 1.1 BW, P<0.05) and 15% lower PFS compared with heel strikers (11.1 ± 2.9 vs. 13.0 ± 2.8 MPa, P<0.05). Peak knee extensor moments showed a trend towards lower values in FFS compared with RFS (3.54 ± 0.69 vs. 3.13 ± 0.77, P=0.09). In the frontal plane, FFS exhibited a 24% lower knee abduction moment compared with RFS (1.49±0.51 vs. 1.97±0.66 Nm/kg, P<0.05). In addition, FFS showed a lower peak knee flexion angle during the stance phase of running (46.9 ± 4.5 vs. 51.9 ± 3.1°, P<0.01).

Ankle kinematic and kinetic parameters and vertical GRF for RFS and FFS are described in figure 11. FFS showed 19% higher plantarflexor moment (3.12 ± 0.40 vs. 2.54 ± 0.37 Nm/kg, P<0.001) and Achilles tendon force (6.3 ± 0.8 vs. 5.1 ± 1.3 BW, P<0.01) compared with RFS. During initial ground contact, FFS exhibited less dorsiflexion compared with RFS (2.3 ± 9.2 vs. 24.8 ± 4.3°, P<0.001). Comparison of GRF parameters showed 26% lower impact peak values for FFS compared with RFS (1.23 ± 0.35 vs. 1.93 ± 0.21 BW, P<0.001). Similarly, FFS demonstrated a 47% lower vertical GRF loading rate than RFS (98.5 ± 21.2 vs. 51.9 ± 16.7 BW/s, P<0.001, Fig. 4D).

FFS also demonstrated lower peak hip adduction during the stance phase than RFS runners (13.1 ± 3.9 vs. 17.0 ± 4.8°, P<0.01). Spatiotemporal comparison showed significantly shorter contact time for FFS than RFS (0.196 ± 0.015 vs. 0.219 ± 0.019 s, P<0.001). In addition, FFS runners demonstrated shorter COM–heel distance at initial ground contact (0.085 ± 0.026 vs. 0.142 ± 0.030 m, P<0.001). No differences were present in the cadence, step length and width or any of the subject's background characteristics between groups.

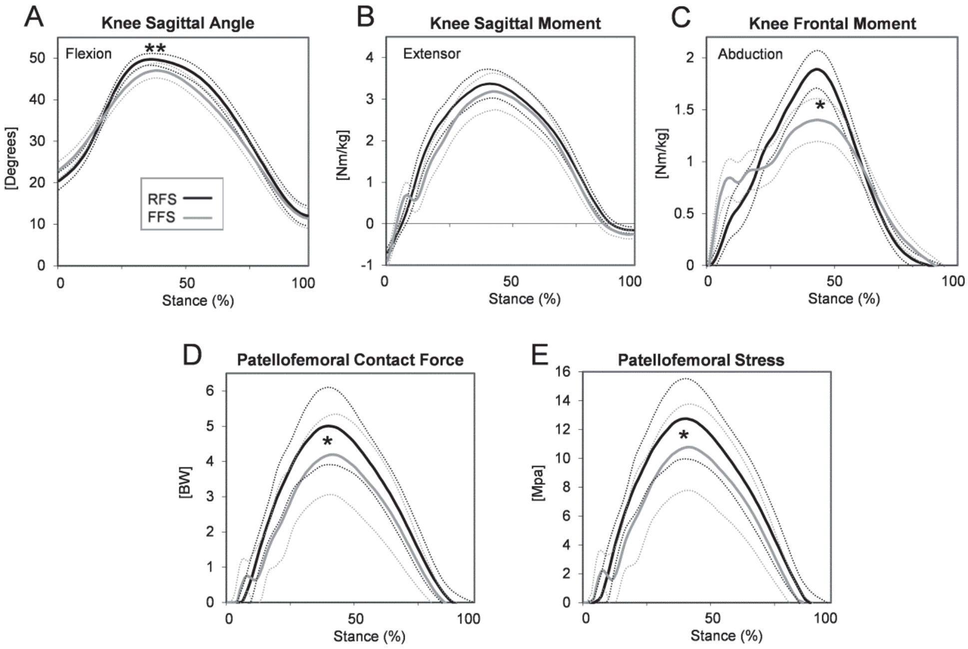


FIGURE 10. Knee kinematics and kinetics between RFS and FFS (*P<0.05, **P<0.01).

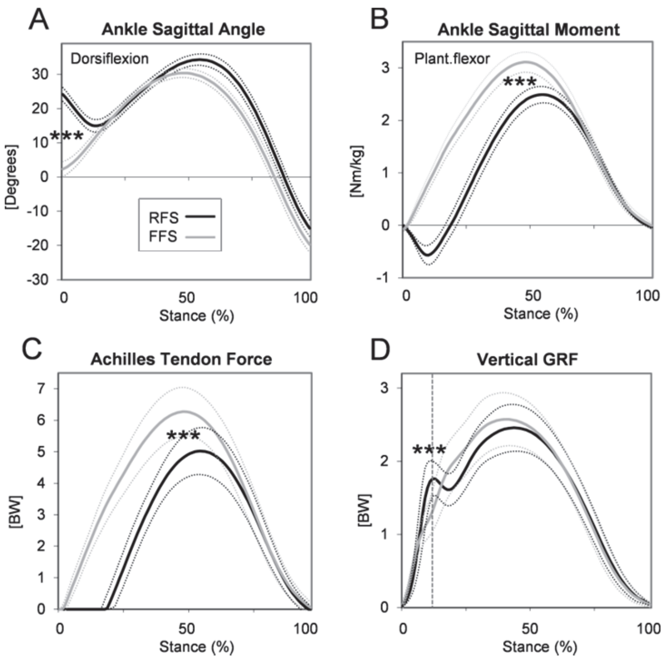


FIGURE 11. Ankle kinematics and kinetics and vertical GRF between RFS and FFS (***P<0.001).

5.3 Mechanics of walking, running and sprinting in different age-groups

5.3.1 Spatio-temporal parameters

Table 2 describes spatiotemporal parameters for the age-groups. No differences were present during the walking (1.6 m/s) and running (4 m/s) speeds. The young group (9.3 m/s) had a higher maximum sprinting speed compared with the middle-aged (7.9 m/s, P<0.001) and old (6.6 m/s, P<0.001) groups. Step parameters did not differ between groups during walking. During running, the contact time was similar between groups but the old athletes had shorter step length and higher step frequency compared with the young (P<0.001, P<0.001) and middle-aged (P<0.05, P<0.001) athletes. During sprinting, the old and middle-aged groups showed shorter step length (P<0.01, P<0.001), greater contact time (P<0.001, P<0.001) and lower step frequency (P<0.001, P<0.001) than the young group.

TABLE 2. Spatiotemporal parameters of young, middle-aged and old groups during walking, running and sprinting.

	Speed (m/s)	Step length (m)	Step width (m)	Contact time (ms)	Swing/flight time (ms)	Step frequency (steps/min)
Walking						
Young	1.6 ± 0.1	0.81 ± 0.04	0.11 ± 0.03	633 ± 20	109 ± 11	115 ± 4
Middle-aged	1.6 ± 0.2	0.85 ± 0.08	0.12 ± 0.03	647 ± 50	110 ± 11	111 ± 7
Old	1.6 ± 0.2	0.81 ± 0.07	0.11 ± 0.03	624 ± 50	100 ± 10	115 ± 7
Running						
Young	4.1 ± 0.2	1.49 ± 0.13***	0.04 ± 0.04	207 ± 17	162 ± 25***	163 ± 10***
Middle-aged	4.0 ± 0.2	1.45 ± 0.10	0.06 ± 0.04	212 ± 22	145 ± 27	168 ± 11
Old	4.0 ± 0.2	1.32 ± 0.10^{+}	0.07 ± 0.04	210 ± 16	117 ± 19^{++}	184 ± 7^{+++}
Sprinting						
Young	9.3 ± 0.4***	2.09 ± 0.10***	0.04 ± 0.04	118 ± 13***	109 ± 11	265 ± 15***
Middle-aged	7.9 ± 0.5###	1.94 ± 0.13##	0.03 ± 0.05	140 ± 10###	113 ± 15	237 ± 10###
Old	6.6 ± 0.7^{+++}	1.67 ± 0.12^{+++}	0.05 ± 0.03	156 ± 17^{+}	102 ± 7	234 ± 16

Statistical significance between Young and Old (*), Young and Middle-aged (#), and Middle-aged and Old (+), respectively (*#+P<0.05, ** ## ++P<0.01, *** ### +++ P<0.001).

5.3.2 Ground reaction forces

GRF results of different age-groups are shown in the figure 11 and table 3. During walking, GRF comparisons showed no differences between age-groups. During running, peak vertical force of the young and middle-aged groups was 19% (P<0.001) and 12% (P<0.001) higher than in the old group, respectively. In addition, peak propulsion force of the young group was 25% higher compared with the old group (P<0.001). During maximal sprinting, the peak vertical forces of the young and middle-aged groups were 19% (P<0.001) and 12% (P<0.001) higher than in the old group, respectively. Compared with the old group, the peak braking forces were 28% and 18% greater (P<0.001, P<0.05) and peak propulsion forces 36% and 20% greater among the young and middle-aged groups (P<0.001, P<0.001). In addition, the young group showed 39% greater peak medial GRF during sprinting than the old group (P<0.01).

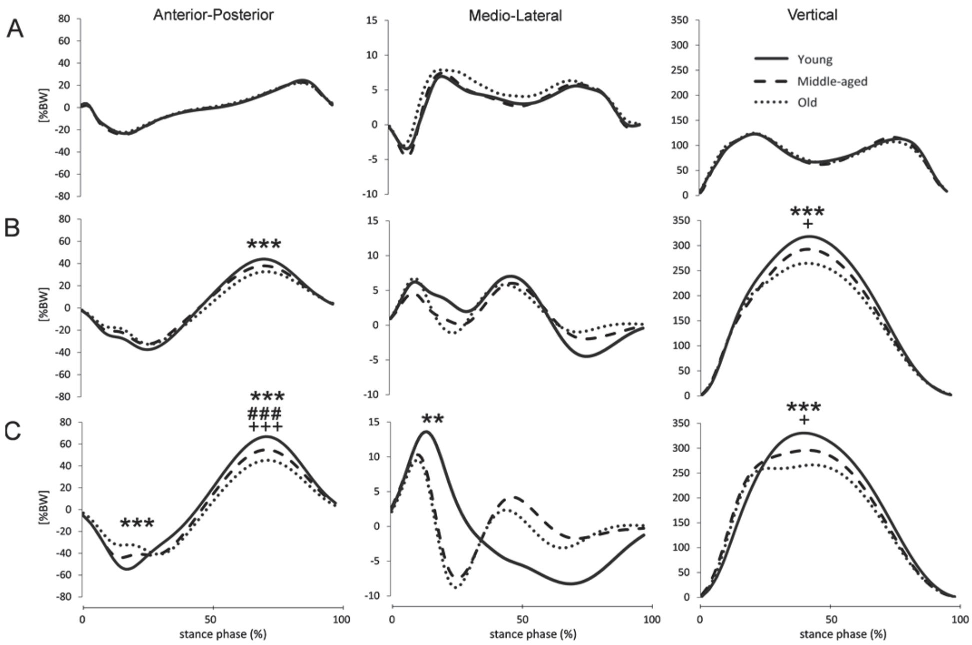


FIGURE 12. Mean GRF curves for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively (*#+P<0.05, ** ## ++P<0.01, *** ### +++ P<0.001).

5.3.3 Sagittal plane mechanics

Walking. No differences were found in the lower limb joint moments (Fig. 13) during walking, but in joint powers, the old group demonstrated 22% less power generation of the ankle plantarflexors (P<0.05) but 31% more power absorption of the knee extensors (P<0.05) compared with the young group (Fig. 14, Table 3). In addition, the athletes in middle-aged and old groups showed trends towards lower ankle plantarflexor moment (5% and 10%, respectively) but higher hip extensor moment (16% and 22%) and power (27% and 41%) than the athletes in young group (Fig. 13 and 14, Table 3). Results of kinematic analysis showed that during the first half of the stance, the old athletes flexed at the knee more compared with the young (P<0.01) and middle-aged (P<0.05) athletes (Fig. 15, Table 3). Additionally, the old group demonstrated more hip flexion than the young group (P<0.05, Fig. 15, Table 3).

Running. Joint kinetics differed between groups at the ankle joint level where the young group demonstrated 25% higher ankle plantarflexor moment (P<0.001, Fig. 13, Table 3) and 31% more power absorption (P<0.05, Fig. 14, Table 3) than the old group. Furthermore, the ankle joint power generation was 41% (P<0.001) and 22% (P<0.001) greater in the young and middle-aged athletes, respectively, compared with the old athletes (Fig. 14, Table 3). In the knee kinetics and kinematics there were no significant group differences. At the hip joint level, the athletes in the old group showed more flexion compared with the young (P<0.01) and middle-aged (P<0.05) athletes (Fig. 15, Table 3). In addition, the middle-aged and old groups showed a tendency towards greater hip kinetics, with significant difference observed in the hip extensor power, where the old athletes had 41% higher values compared with the young athletes (P<0.05, Fig. 14, Table 3).

Sprinting. Joint kinetics differed significantly between groups at the ankle and hip joints, but not at the knee joint. The young athletes produced 14% and 27% more ankle plantarflexor moment compared with the middle-aged (P<0.05) and old (P < 0.001) athletes (Fig. 13, Table 3), respectively. Ankle power absorption of the young group was 32% (P<0.01) and 53% (P<0.001) greater and power generation 30% (P<0.001) and 47% (P<0.001) greater compared with the middle-aged and old groups, respectively (Fig. 14, Table 3). At the hip joint level, the young group demonstrated 25% greater extensor moment and 34% greater flexor moment than the old group (Fig. 13, Table 3). Hip power generation of the athletes in the young group was 31% and 38% greater compared with middle-aged (P<0.01) and old (P<0.001) athletes, respectively (Fig. 14, Table 3). Hip power absorption of the young and middle-aged groups was 38% (P<0.001) and 29% (P<0.05) higher compared with the old group, respectively (Figure 14, Table 3). In addition, kinematic comparison showed more hip flexion in the old compared with the middle-aged athletes (P<0.05) and a trend towards higher ankle dorsiflexion angle in the old group (Figure 15, Table 3).

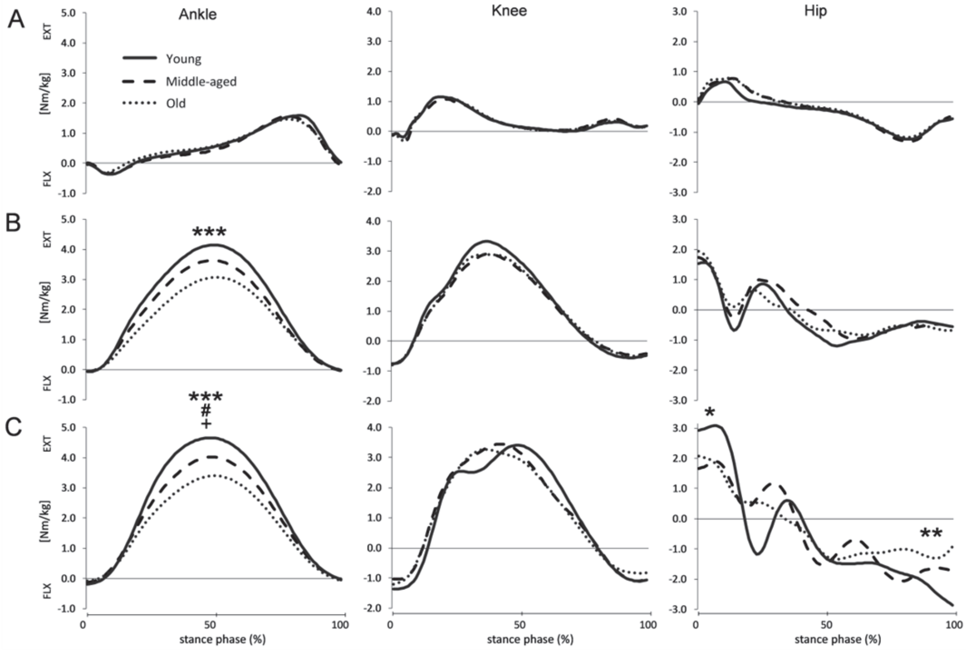


FIGURE 13. Mean joint moment curves in the sagittal plane for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively ($^{*\#+}P<0.05$, $^{**\ \#\#\ ++}P<0.01$, $^{***\ \#\#\#\ +++}P<0.001$).

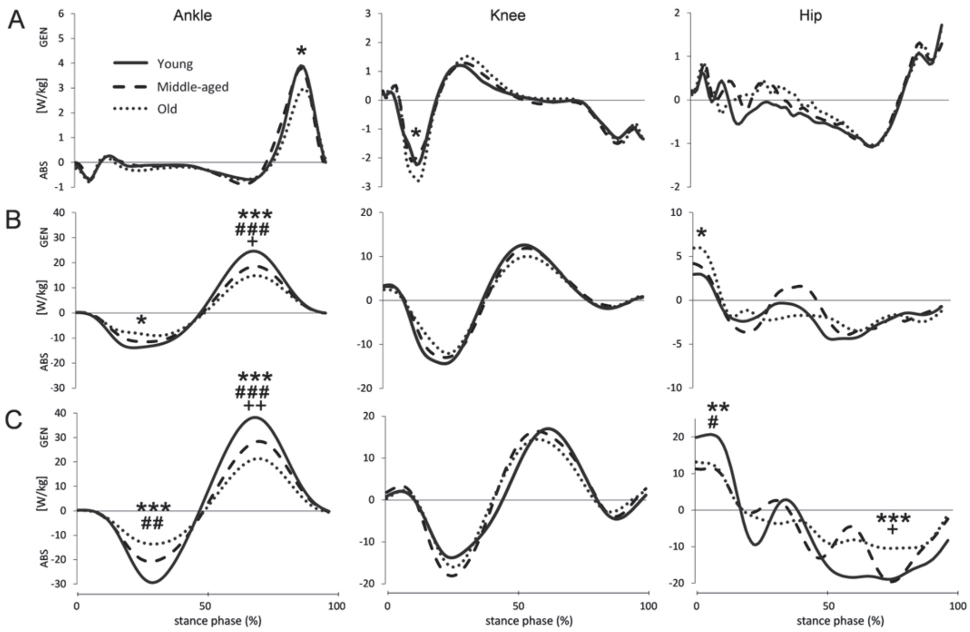


FIGURE 14. Mean joint power curves in the sagittal plane for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively (*#+ $P<0.05$, ** ## ++ $P<0.01$, *** ### +++ $P<0.001$).

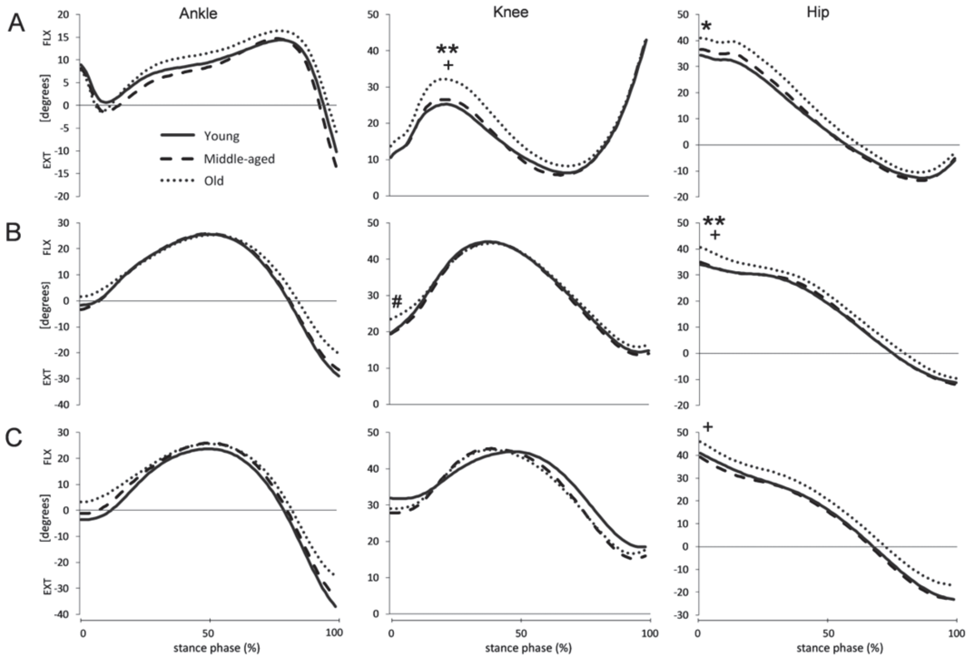


FIGURE 15. Mean joint angle curves in the sagittal plane for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively (*#+P<0.05, ** ## ++P<0.01, *** ### +++ P<0.001).

TABLE 3. Ground reaction force and joint kinematic and kinetic parameters in the sagittal plane of young, middle-aged and old groups during walking, running and sprinting.

Parameters	Walking			Running			Sprinting		
	Young	Middle-aged	Old	Young	Middle-aged	Old	Young	Middle-aged	Old
Ground reaction forces									
Peak braking (% BW)	-24 ± 3	-25 ± 3	-24 ± 4	-38 ± 4	-33 ± 6	-33 ± 5	-58 ± 11***	-51 ± 5	-42 ± 7^{+}
Average braking (% BW)	-11 ± 2	-12 ± 2	-12 ± 2	-23 ± 2**	-20 ± 3$^{\#}$	-19 ± 3	-33 ± 4***	-31 ± 3	-26 ± 3^{++}
Peak propulsion (% BW)	25 ± 3	23 ± 3	23 ± 4	44 ± 4***	39 ± 7	33 ± 5	70 ± 5***	56 ± 6$^{\#\#\#}$	45 ± 6^{+++}
Average propulsion (% BW)	11 ± 1	11 ± 2	11 ± 2	26 ± 3***	22 ± 4	19 ± 3	41 ± 3***	34 ± 4$^{\#\#\#}$	27 ± 4^{+++}
Peak vertical (% BW)	124 ± 7	127 ± 10	126 ± 9	319 ± 33***	295 ± 31	266 ± 19^{+}	337 ± 33***	313 ± 40	274 ± 15^{++}
Average vertical (% BW)	83 ± 2	82 ± 3	83 ± 3	177 ± 14***	164 ± 14	154 ± 11	188 ± 11***	178 ± 13	163 ± 7^{++}
Ankle									
Foot strike angle (°)	25.8 ± 4.3	27.1 ± 4.4	24.5 ± 5.6	1.6 ± 11.0	1.8 ± 7.7	5.4 ± 13.2	-1.7 ± 7.9*	3.7 ± 7.3	7.4 ±9.8
Angle at intial contact (°)	9.0 ± 3.4	8.1 ± 4.1	7.8 ± 4.4	-1.6 ± 8.9	-3.3 ± 6.6	1.5 ± 11.3	-3.4 ± 5.9	-1.2 ± 7.6	3.2 ± 8.4
Plantarflexor moment (Nm/kg)	1.66 ± 0.31	1.57 ± 0.16	1.49 ± 0.17	4.17 ± 0.70***	3.66 ± 0.63	3.11 ± 0.40	4.69 ± 0.65***	4.05 ± 0.59$^{\#}$	3.42 ± 0.28^{+}
Power absorbtion (W/kg)	-1.0 ± 0.4	-1.1 ± 0.4	-1.3 ± 0.4	-15.0 ± 4.9*	-12.9 ± 3.7	-10.3 ± 2.6	-31.9 ± 8.6***	-21.6 ± 8.2$^{\#\#}$	-15.0 ± 3.9
Power generation (W/kg)	4.1 ± 0.7*	3.9 ± 0.8	3.2 ± 0.8	25.3 ± 4.1***	19.2 ± 3.9$^{\#\#\#}$	15.0 ± 3.2^{+}	41.0 ± 4.6***	29.0 ± 4.9$^{\#\#\#}$	21.8 ± 4.9^{++}
Knee									
Angle at intial contact (°)	10.5 ± 3.7	10.5 ± 4.9	13.8 ± 4.8	19.6 ± 4.3	19.4 ± 3.7	23.5 ± 5.4	31.9 ± 6.0	27.8 ± 4.9	29.0 ± 4.4
Flexion excursion (°)	25.4 ± 4.6*	27.0 ± 5.6	32.4 ± 4.5^{+}	25.3 ± 4.3	25.7 ± 5.8	21.1 ± 6.5	13.2 ± 6.2	18.6 ± 6.0	16.6 ± 4.7
Externsor moment (Nm/kg)	1.22 ± 0.22	1.13 ± 0.32	1.21 ± 0.27	3.35 ± 0.50	2.95 ± 0.59	2.99 ± 0.56	3.59 ± 0.96	3.58 ± 0.50	3.42 ± 0.50
Power absorbtion (W/kg)	-2.4 ± 0.7*	-2.6 ± 0.8	-3.5 ± 1.4	-15.5 ± 6.5	-14.8 ± 5.5	-12.7 ± 5.7	-13.9 ± 10.0	-19.3 ± 6.9	-16.0 ± 5.8
Power generation (W/kg)	1.2 ± 0.3	1.5 ± 0.6	1.7 ± 0.5	12.4 ± 3.5	12.3 ± 3.3	10.2 ± 2.9	17.6 ± 8.2	18.2 ± 6.0	14.2 ± 3.2
Hip									
Angle at intial contact (°)	34.3 ± 6.0*	36.6 ± 3.6	41.0 ± 7.3	34.0 ± 5.7**	34.9 ± 5.7	40.6 ± 4.1^{+}	40.9 ± 7.1	39.2 ± 7.4	46.0 ± 5.7^{+}
Extensor moment (Nm/kg)	1.30 ± 0.34	1.54 ± 0.47	1.66 ± 0.49	1.85 ± 0.53	2.28 ± 0.79	2.11 ± 0.39	3.44 ± 0.70*	3.10 ± 0.91	2.59 ± 0.73
Flexor moment (Nm/kg)	-1.31 ± 0.21	-1.41 ± 0.21	-1.26 ± 0.18	-1.57 ± 0.52	-1.58 ± 0.43	-1.33 ± 0.51	-3.36 ± 1.03**	-2.99 ± 0.82	-2.21 ± 0.70
Power generation 1 (W/kg)	1.0 ± 0.8	1.6 ± 0.7	1.7 ± 0.7	4.1 ± 1.5*	5.6 ± 1.9	6.9 ± 3.0	25.1 ± 10.3**	17.2 ± 5.2$^{\#\#}$	15.5 ± 6.8
Power absorbtion (W/kg)	-1.3 ± 0.5	-1.3 ± 0.4	-1.3 ± 0.4	-7.8 ± 4.1	-8.2 ± 2.0	-7.3 ± 4.3	-29.6 ± 7.7***	-25.8 ± 8.1	-18.3 ± 5.1^{+}
Power generation 2 (W/kg)	1.7 ± 0.4	1.8 ± 0.5	1.8 ± 0.9	-	-	-	-	-	-

Data shown as mean ± SD; BW = body weight; Power generation 1 = first half of the stance; Power generation 2 = second half of the stance during walking.
$^{*\,\#\,+}$ statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively.
$^{*\,\#\,+}$ $P < 0.05$; $^{**\,\#\#\,++}$ $P < 0.01$; $^{***\,\#\#\#\,+++}$ $P < 0.001$.

5.3.4 Secondary plane mechanics

Walking. No differences between age-groups were present in the secondary plane kinetics (Fig. 16) or kinematics (Fig. 17-18, Table 4).

Running. No significant differences were present in the knee frontal plane moments. Kinematic analysis showed that the old group had reduced frontal plane pelvic upward angle during toe-off compared with the middle-aged ($P <$ 0.01) and young groups (P<0.05, Fig. 17. Table 4). In the transversal plane, the old group demonstrated larger pelvis internal rotation during initial contact compared with the young group (P<0.05, Fig. 18. Table 4).

Sprinting. Kinetic analysis showed that the middle-aged group had a 37% higher knee abductor moment than the old group (P<0.05, Fig. 16, Table 4). However, no differences were found in the hip abductor moment. Kinematic comparison showed that the old group demonstrated larger frontal plane hip adduction and pelvis drop ROM when compared with the young group (P<0.05, P<0.01, Fig. 17, Table 4). The old group also showed less trunk lateral flexion during the stance when compared with the middle-aged (P<0.05) and young groups (P<0.05, Fig. 17, Table 4). In the transversal plane, the old group demonstrated more pelvis internal rotation during the initial contact when compared with the middle-aged (P<0.05) and young groups (P<0.05, Fig.18, Table 4). Instead, during toe off, the pelvic internal rotation was larger in the young group when compared with the middle-aged (P<0.05) and old (P<0.05, Fig.18, Table 4) groups.

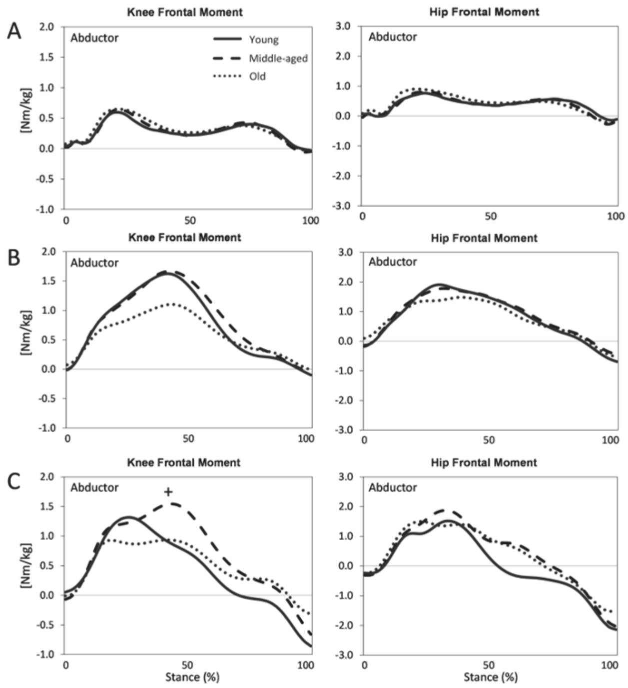


FIGURE 16. Mean frontal plane moment curves for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively (*#+P<0.05, ** ## ++P<0.01, *** ### +++ P<0.001).

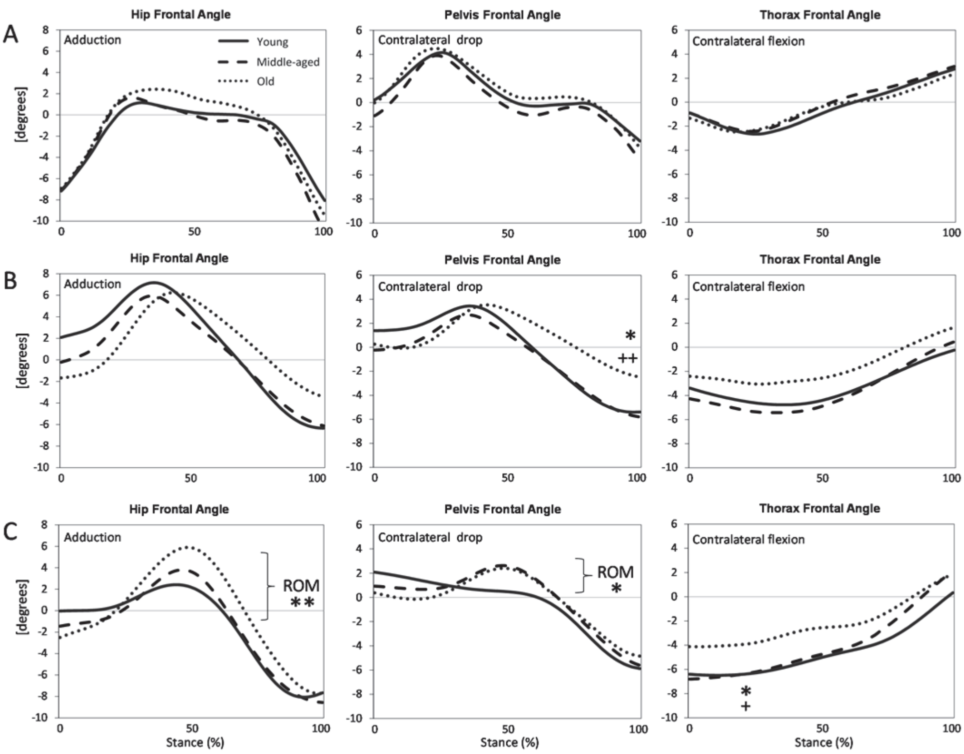


FIGURE 17. Mean frontal plane angle curves for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively (*#+P<0.05, ** ## ++P<0.01, *** ### +++ P<0.001).

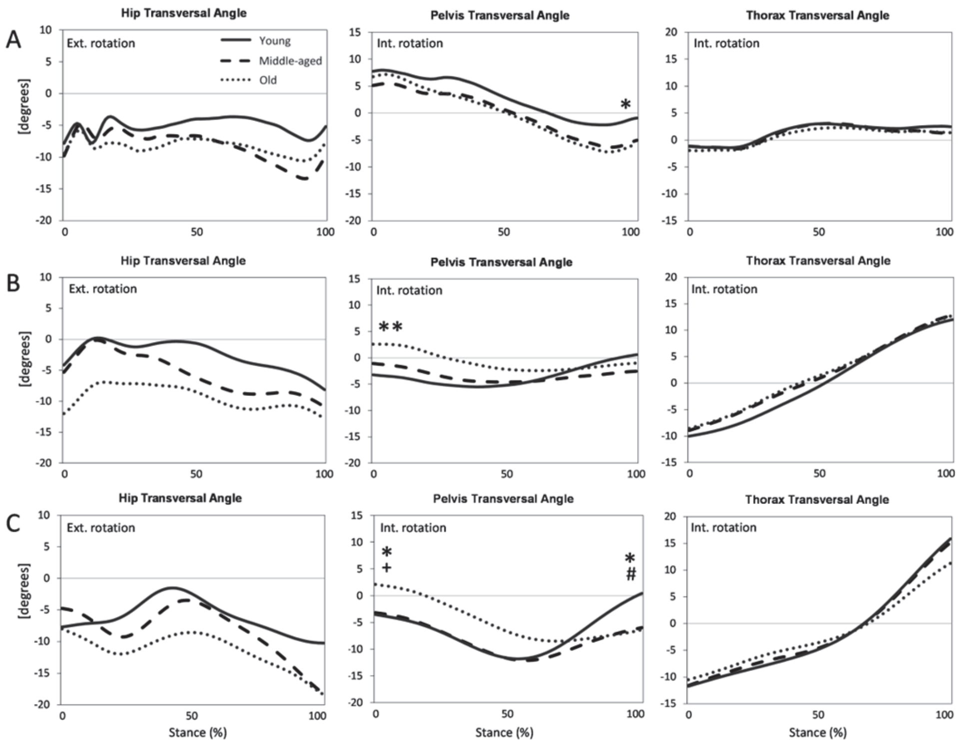


FIGURE 18. Mean transversal plane angle curves for the young, middle-aged and old groups during A) walking, B) running and C) sprinting. Symbols * # + describe statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively (*#+P<0.05, ** ## ++P<0.01, *** ### +++ P<0.001).

TABLE 4. Ground reaction forces and secondary plane kinematics and kinetics for young, middle-aged and old groups during walking, running and sprinting.

Parameters	Walking			Running			Sprinting		
	Young	Middle-aged	Old	Young	Middle-aged	Old	Young	Middle-aged	Old
Ground reaction forces									
Braking (% BW)	-24 ±3	-25 ±3	-24 ±4	-38 ±4	-33 ±6	-33 ±5	-58 ±11***	-51 ±5	-42 ±7+
Propulsion (% BW)	25 ±3	23 ±3	23 ±4	44 ±4***	39 ±7	33 ±5	70 ±5***	56 ±6###	45 ±6+++
Medial (% BW)	7 ±1	8 ±2	9 ±2	11 ±5	8 ±4	10 ±5	18 ±6**	14 ±6	11 ±4
Lateral (% BW)	5 ±2	5 ±2	4 ±3	6 ±4	6 ±3	6 ±6	16 ±10	14 ±8	13 ±9
Vertical (% BW)	124 ±7	127 ±10	126 ±9	319 ±33***	295 ±31	266 ±19+	337 ±33***	313 ±40	274 ±15++
Knee									
Abductor moment (Nm/kg)	0.64 ±0.22	0.69 ±0.36	0.68 ±0.19	1.75 ±0.73	1.72 ±0.75	1.22 ±0.52	1.49 ±0.58	1.80 ±0.75	1.13 ±0.36+
Hip									
Adducion at initial contact (°)	-7.2 ±4.3	-7.1 ±5.4	-6.9 ±4.7	2.1 ±3.4	-0.2 ±6.0	-1.7 ±4.0	0.0 ±2.3	-1.4 ±6.9	-2.5 ±4.7
Adduction ROM (°)	8.6 ±3.0	9.8 ±3.8	10.1 ±2.9	5.6 ±3.5	6.5 ±3.4	8.2 ±4.0	3.3 ±3.5**	5.8 ±4.6	8.6 ±3.9
Adduction at toe off (°)	-8.1 ±3.8	-10.9 ±4.0	-9.4 ±4.5	-6.3 ±4.1	-6.1 ±4.1	-3.4 ±5.1	-7.7 ±4.6	-8.6 ±3.9	-7.7 ±6.4
Abductor moment (Nm/kg)	0.85 ±0.23	0.90 ±0.28	0.95 ±0.19	2.05 ±0.53	1.96 ±0.56	1.72 ±0.52	1.76 ±0.72	2.15 ±0.52	1.75 ±0.57
Rotation at initial contact (°)	-7.8 ±6.4	-8.8 ±8.6	-9.4 ±4.0	-4.1 ±6.4	-5.2 ±7.8	-12.0 ±10.2	-7.7 ±7.3	-4.8 ±10.4	-8.0 ±6.8
Rotation at toe off (°)	-5.2 ±7.1	-8.9 ±11.4	-7.7 ±10.6	-8.1 ±5.9	-10.5 ±12.1	-12.8 ±11.4	-10.4 ±6.6	-19.2 ±11.3	-18.4 ±11.8
Pelvis									
Angle at initial contact (°)	0.2 ±2.2	-1.1 ±2.0	-0.0 ±2.4	1.4 ±1.6	-0.2 ±2.4	0.3 ±2.6	2.1 ±2.4	0.9 ±2.4	0.4 ±2.1
Pelvis drop ROM (°)	4.0 ±1.3	5.1 ±2.0	4.7 ±1.4	2.3 ±1.8	3.2 ±1.8	3.7 ±2.0	0.5 ±1.6*	2.0 ±1.9	2.5 ±2.0
Angle at toe off (°)	-3.2 ±2.6	-4.7 ±1.5	-3.7±2.5	-5.4 ±2.0*	-5.8 ±2.7	-2.5 ±3.1++	-3.36 ±1.03	-2.99 ±0.82	-2.21 ±0.70
Rotation at initial contact (°)	7.8 ±3.7	5.1 ±5.1	6.8 ±4.7	-3.2 ±3.3**	-1.1 ±5.2	2.6 ±4.8	-3.5 ±2.1*	-3.2 ±7.1	2.1 ±5.2+
Rotation at toe off (°)	-0.9 ±4.1*	-5.1 ±5.5	-5.6 ±4.4	0.6 ±4.9	-2.5 ±4.4	-1.0 ±5.6	0.4 ±5.6*	-6.0 ±5.3#	-6.4 ±6.1
Trunk									
Angle at intial contact (°)	-0.9 ±1.5	0.9 ±2.4	-1.4 ±1.9	-3.4 ±2.2	-4.3 ±3.1	-2.4 ±3.3	-6.4 ±2.4	-6.8 ±3.1	-4.1 ±2.5+
Lateral flexion max. (°)	-2.7 ±1.7	-2.5 ±2.7	-2.7 ±2.3	-5.1 ±2.0	-5.7 ±2.5	-3.4 ±3.1	-7.1 ±2.2*	-6.9 ±3.0	-4.3 ±2.6+
Rotation at initial contact (°)	-1.1 ±3.4	-1.1 ±3.8	-2.0 ±5.1	-10.9 ±3.0	-8.9 ±3.8	-8.6 ±2.6	-11.7 ±4.7	-11.6 ±8.0	-10.5 ±4.0
Rotation at toe off (°)	2.5 ±2.8	1.2 ±3.4	1.4 ±3.3	12.0 ±3.2	12.9 ±4.3	12.8 ±3.8	14.0 ±7.7	15.1 ±6.6	11.3 ±4.1

Data shown as mean ± SD; BW = body weight; ROM = Range of Motion
* # + statistical significance between Young and Old, Young and Middle-aged, and Middle-aged and Old, respectively.
* # + $P < 0.05$; ** ## ++ $P < 0.01$; *** ### +++ $P < 0.001$.

6 DISCUSSION

This study investigated how locomotor pattern diversity and ageing affect lower limb joint mechanics and loading across different modes and intensity of locomotion.

6.1 The effects of different walking patterns on the medial knee joint loading

Walking patterns across the normal population exhibits a large inter-individial variability especially at the knee joint level. In paper I, a cluster analysis was used to divide subjects into subgroups based on the knee joint moment patterns. Three subgroups with clearly distinct sagittal plane moment patterns were found. This finding is in accordance with the previous studies (Pedotti 1977, Simonsen et al. 1997, Winter 1984) confirming large variability in the kinetic strategies of the knee driving the gait. A novel finding was that subjects who showed a typical sagittal plane moment pattern (TYP) demonstrated clearly lower knee frontal plane moments compared with those who exhibited either knee extensor (EXT) or flexor (FLX) dominant gait patterns. Furthermore, the subjects in the TYP cluster demonstrated significantly lower knee external rotation moments than both EXT and FLX clusters.

Previous studies have shown that knee frontal plane moment plays an important role in knee joint load distribution across the medial and lateral tibio-femoral compartments (Schipplein & Andriacchi 1991, Zhao et al. 2007) and is positively correlated with OA severity and progression (Miyazaki et al. 2002, Baliunas et al. 2002, Chehab et al. 2014, Prodromos, Andriacchi & Galante 1985). Therefore, the results of the present study suggests that increased risk of OA may be present in those who demonstrate either knee extensor (EXT) or flexor (FLX) dominant gait patterns with high knee frontal plane moments. Furthermore, the subjects in the EXT and FLX clusters demonstrated significantly higher peak knee external rotation moments than TYP cluster. This finding is in line with the previous studies (Astephen et al. 2008, Gok, Ergin & Yavuzer 2002) suggesting the link between high frontal plane moment and increased external rotation moment of the knee during walking. However, while the knee frontal plane moment has been linked with the knee OA, the role of knee rotation moment in the development of the disease remains unclear.

A previous work, where an instrumented knee prosthesis was used to measure knee contact loads during walking, suggested that in addition to the knee frontal plane moment, the knee extensor moment should also be taken into account when total knee medial compartment loading is evaluated (Walter et al. 2010). Therefore, the EXT cluster, which demonstrated both the greatest knee frontal plane and knee extensor moment, most likely exhibited the highest knee medial loads during gait. This suggestion is also in accordance with a previous study, in which the knee extensor dominant gait pattern was shown to cause higher tibiofemoral joint contact force during walking than knee flexor dominant gait pattern (Simonsen & Alkjaer 2012). Importantly, recent studies of Hall et al. (2014) and Chehab et al. (2014) demonstrated that both greater knee frontal and sagittal plane moments were associated with knee medial cartilage thinning and a reduction in patients with arthroscopic partial meniscectomy and medial knee OA over the two and five year follow-up periods, respectively. In this light, the gait pattern particular to the EXT cluster may particularly increase the risk for development of knee OA.

Significant differences among clusters were observed in the sagittal plane kinematics. The greatest differences were present at the knee joint, but different profiles in the kinematic patterns were present in the hip and ankle level as well. The higher knee flexion during the entire stance phase was present in the EXT cluster, which most likely causes the greater knee extensor moment in this group. The opposite kinematic strategy showing a straighter knee during the stance phase was present in the FLX cluster, which walked with the lowest knee extensor moment and highest knee flexor moment. A similar relationship between the higher knee flexion angle and increased knee extensor moment was previously found by Simonsen and Alkjaer (2012) who also used cluster analysis to divide subjects into subgroups according to their gait patterns. They identified two distinct gait patterns, which showed the largest differences in the knee sagittal plane moment and flexion angle. Although different input parameters were used in the cluster analysis in the current study, consistent findings in the sagittal plane among clusters were present. This supports the idea that variability in gait mechanics is the largest at the knee joint level.

Conservative treatment strategies such as muscle strengthening, gait modification and laterally wedged shoes are commonly used to manage joint loading and consequent symptoms in patients with knee OA (Reeves & Bowling 2011). However, recent studies suggest that the changes in gait mechanics show inconsistent response to these strategies (Gaudreault et al. 2011, Hinman et al. 2008), and some people actually demonstrate adverse biomechanical changes leading to increased knee loading (Hinman et al. 2012b). Consequently, it would be important to better understand underlying factors behind different responses to the conservative treatment strategies. One potential possibility is that people with different gait patterns adapt differently to these modifications.

6.2 Effects of foot strike pattern on the knee and ankle loading

Similar to walking, a large inter-subject variability exists in running mechanics. In paper II, the effect of a runner's foot strike pattern on knee and ankle loading were investigated. Subjects were categorized into forefoot (FFS) and rearfoot (RFS) strikers based on their landing strategy during the initial ground contact. The results revealed that those who run with a FFS pattern showed lower knee joint loading by demonstrating both lower patellofemoral stress (PFS) and knee frontal plane moment compared to runners with a RFS pattern. However, at the ankle joint level, FFS demonstrated clearly greater plantarflexor and Achilles tendon loading than RFS.

These findings may be important from a clinical point of view because one of the most widely accepted theories regarding the etiology of the overuse injuries suggest that the symptoms are resulting from repeated overloading of the musculoskeletal structures (van Mechelen, Hlobil & Kemper 1992). Therefore, it can be assumed that lower knee loading among FFS may reduce the risk of running-related knee injuries compared with RFS. However, a parallel increase in ankle plantarflexor and Achilles tendon loading may increase the risk for ankle and foot injuries among FFS runners. On the other hand, strategic modification of the striking pattern could be advisable for some runners; for example, people suffering from knee complaints may benefit from FFS technique whereas those who suffer pain at the Achilles tendon or foot areas might benefit from RFS technique.

Higher ankle contribution may be the main mechanism to explain lower knee loading among FFS because it likely reduces the role of the knee joint as an energy absorber. This idea is also supported by Arendse et al. (2004) who reported reduced eccentric quadriceps work during FFS compared to RFS (Arendse et al. 2004). Although, researchers did not report the magnitude of the knee extensor moment, PFCF or PFS in their study, they observed lower knee flexion ROM during FFS, which is in accordance with our findings. Because the quadriceps moment arm decreases as a function of increased knee flexion angle (Mason et al. 2008, van Eijden et al. 1986), greater eccentric quadriceps force is, therefore, needed to resist knee flexion during the first half of the stance when running with RFS pattern. This explains greater PFCF and PFS in RFS runners in the current study.

The spatio-temporal comparison showed lower contact time and shorter distance between COM and heel during initial ground contact for the FFS. In addition, lower hip adduction angle was present in the FFS. It has been known that the magnitude of the knee frontal plane moment is highly related to the length of the ground reaction force vector lever arm at the knee in the frontal plane (Schache et al. 2008). Because no differences were observed in the knee frontal plane angle, it can be assumed that a different limb posture with closer COM to heel distance and/or lower peak hip adduction in FFS can change the position of the GRF vector in relation to lower limb so that moment arm in the frontal plane decreases leading to a reduction of the knee frontal plane moment.

6.3 Effects of aging on the locomotion mechanics

6.3.1 Walking

The present results showed that competitive older athletes demonstrate changes in lower limb mechanics during walking but maintain similar preferred speeds as their younger counterparts. This is in line with the recent investigations of physically active older men (Cofre et al. 2011, Boyer, Andriacchi & Beaupre 2012). The alteration in gait was present in the sagittal plane mechanics where the old group demonstrated 22% lower ankle power generation compared with the young group. Furthermore, there was a trend towards lower ankle plantarflexor moment among the middle-aged and old athletes who exhibited 5% and 10% lower values compared with young athletes, respectively. As a compensatory action for reduced ankle kinetics during walking, the old groups increased hip and knee flexion and demonstrated more eccentric power at the knee joint and a trend towards higher extensor moment and power at the hip joint. These findings generally agree with the literature (DeVita & Hortobagyi 2000, Cofre et al. 2011, Boyer, Andriacchi & Beaupre 2012, Monaco et al. 2009) suggesting that during walking old adults rely less on ankle plantarflexors and more on the proximal lower limb muscles than young adults.

6.3.2 Running and sprinting

Sagittal plane mechanics.

During running and sprinting, the largest decline in sagittal plane kinetics was observed in the ankle moment and power generation. The middle-aged and old groups showed 12% and 25% lower ankle plantarflexor moment, respectively. However, during sprinting, the deficit remained roughly the same showing only slightly greater decline (15% and 27%) compared with the young athletes. A previous study (Karamanidis, Arampatzis & Bruggemann 2006) with slower running speed (2.7 m/s) found a 13% decrease in the ankle plantarflexor moment among the older adults (mean age 64 yr.), which is nearly similar with the middle-aged group (mean age 61 yr.) in the current study despite the different running speeds (4 m/s and maximum speed). Therefore, these findings suggest that age-related decline in the ankle plantarflexor moment is more related to aging than running speed. However, the same was not true in the ankle power absorption and generation, where between group differences were much larger during sprinting than during running. This most likely results from remarkably shorter ground contact times in the young group compared with the middle-aged and old groups, which require more rapid muscular force generation reflected by increased peak muscle power rather than peak muscle moments (Schache et al. 2011). Furthermore, because aging seems to have a greater effect on rapid force generation than peak force of the muscles in untrained (Candow & Chilibeck 2005, Häkkinen, Kraemer & Newton 1997, Piirainen et al. 2010) and also in speed and power trained older adults (Korhonen et al. 2009), larger age-related reduction in the ankle plantarflexor power during locomotion may, therefore, be expected.

In addition, the present study showed that older athletes tended to compensate for the reduced ankle contribution by increasing the effort of the hip extensors. Although older people are known to demonstrate a redistribution of power output to more proximal muscles during walking to maintain similar speed with their younger counterparts (DeVita & Hortobagyi 2000, Cofre et al. 2011, Boyer, Andriacchi & Beaupre 2012, Monaco et al. 2009) this compensation strategy has not been demonstrated before in running. A distal to proximal shift in joint kinetics was found especially among subjects in the old group who exhibited reduction in the ankle kinetics, but instead, increased hip extensor power during stance phase of running when compared with the young group. The lack of distal to proximal shift in the previous studies (Karamanidis, Arampatzis & Bruggemann 2006, Fukuchi et al. 2014) in older runners may be explained by the lower mean age of the subjects (60-64 vs. 78 yr.) and slower running speed (2.7 vs. 4 m/s) when compared with the current study.

The old group in the present study also showed a tendency towards higher foot strike angle during both running and sprinting, which suggests the presence of a rearfoot running pattern (Altman & Davis 2012). It is possible that alteration in the landing strategy in the old group is a mechanism to shift the contribution from ankle plantarflexors to the more proximal muscles such as the knee and hip extensors by changing lower limb alignment in relation to the GRF vector (reduction in the ankle but increase in the knee joint distance in relation to the GRF vector).

It was also found that the old group demonstrated a more flexed hip joint during running and sprinting. Previously an age-related shift towards hip flexion has been observed during walking (DeVita & Hortobagyi 2000, Cofre et al. 2011, Monaco et al. 2009), but the findings of the current study suggest that altered hip movement is present in running and sprinting as well. This change may also contribute to impaired performance by placing the old athletes in a suboptimal position to generate propulsion forces. It remains unclear, however, whether or not a more flexed hip joint during the whole stance phase is resulting from an age-related reduction in the hip ROM as suggest by previous studies (Kerrigan et al. 1998, Kerrigan et al. 2003), or alternatively, is an adaptation strategy that may help older people to optimize their performance and/or balance control during locomotion.

During sprinting, a reduction in the joint moments and powers was present only at the hip joint but surprisingly, not at the knee joint. This finding suggests that, besides ankle plantarflexors, an age-related decline in the muscular output during sprinting occurs in the hip extensors and flexors, while the contributions of the knee extensors remain essentially similar. Therefore, the knee extensors' capacity may not be a limiting factor of the sprinting performance in older age. Such a result may be explained by the findings that, in contrast to ankle and hip joints, the kinetics of the knee joint during the stance phase of sprinting appears to change very little as running speed increases from 5 m/s to 9 m/s (Schache et al. 2011), and thereafter with higher speeds, the extensor moment of the knee may even decrease (Kuitunen, Komi & Kyrolainen 2002, Bezodis, Kerwin & Salo 2008). In line with these findings, a recent computer simulation study suggested that the contribution of the ankle plantarflexor to GRF generation increases, whereas knee extensor contribution decreases during sprinting with greater speeds than 7.0 m/s (Dorn, Schache & Pandy 2012). Because, in this study, the young adults reached remarkably higher sprinting speeds (9.3 m/s) than the middle-aged (7.9 m/s) and old athletes (6.6 m/s), the need for a larger knee extensor moment or power generation among the young group may be diminished.

Secondary plane mechanics.

The age-groups showed similar amounts of hip abductor moment across all measured activities even though the old group generated substantially lower vertical GRF during running and sprinting. This finding is somewhat surprising, when considering the function of the hip abductors as an antigravity muscle group responsible for the frontal plane stability of the pelvis and trunk (Winter 1995). Furthermore, increased external forces have been typically shown to be related to greater internal joint moments that need to be generated by the lower limb muscles that provide support for the body over the period of ground contact (Schache et al. 2011, Arampatzis, Bruggemann & Metzler 1999). The present results of the whole body frontal plane kinematics offer, however, a potential biomechanical explanation for this finding. By demonstrating greater trunk lateral flexion and lower hip adduction during running and sprinting, the athletes in the young and middle-aged groups align their body more closely to support leg than the athletes in the old group. As a result, young and middle-aged athletes likely shift the displacement of their COM more lateral to reduce the hip frontal plane lever arm, which consequently lead to similar hip abductor moments among groups despite distinct amounts of vertical GRF.

The performance of the lower limb muscles (Candow & Chilibeck 2005), including hip abductors (Johnson et al. 2004, Ramskov et al. 2014), declines with age. Therefore, the present finding that hip abductor moment did not differ between age groups, brings into question: why did the young and middle-aged athletes align their body more laterally to reduce hip frontal plane moment, although their hip abductor performance is likely better? However, rather than keeping hip abductor loading similar to old athletes, greater trunk lateral flexion among young and middle-aged groups may be a strategy to assist hip adductors to counteract powerful force production of the antagonist muscle group (hip adductors) during running and sprinting.

Unfortunately, an assessment of the function of the hip adductor muscle group is difficult based on the hip joint moment data because the inverse dynamics approach provides information only about the net joint moments, thus, ignoring antagonist forces acting on the opposite direction on the joint (Winter 2009, 107-136). Nevertheless, there is evidence to show that hip adductors play an important role in human locomotion. First, the fact that this muscle group makes up one third of the mass of the thigh musculature (Ito et al. 2003) suggest that, besides causing the mere function of hip adduction, it likely plays a role in other hip movements as well. Moreover, electromyographic data shows that hip adductors are activated during both swing and stance phases of sprinting suggesting a contribution to hip sagittal plane movements (Wienmann & Gunter 1995). Finally, an analysis of abductor magnus muscle architecture suggests that especially the structure and function of the distal part of the abductor magnus with long muscle fiber length appears to be designed to move the thigh through a large ROM whereas the proximal part of this muscle is oriented more towards stabilizing the hip joint in the frontal plane (Takizawa et al. 2014). Based on these findings, it can be assumed that the hip adductor muscle group contributes to the thigh sagittal plane movement during running and sprinting, and at the same time, generates force toward hip adduction. As a result, hip abductors must counteract both gravity and force generated by a large adductor muscle group. Consequently, the greater lateral trunk alignment seen in the young and middle-aged groups likely reduced hip frontal plane moment, which in turn, assisted hip abductors to maintain lateral stability for the pelvis and trunk over the period of ground contact.

Previously, aging has been associated with reduced hip adduction ROM during walking and running (Ko et al. 2009, Ko et al. 2012, Fukuchi et al. 2014). However, the present study found increased hip adduction ROM with age especially during running and sprinting. One explanation for this discrepancy is that walking and running speeds in the present study were substantially higher compared with the previous studies, which likely pose greater demands on the locomotor system. Therefore, it is possible that larger hip adduction ROM in the old group may result from insufficient force production of the hip abductors to prevent lowering of the pelvis to the contralateral side during single leg support. Thus, aside from declines in ankle and hip sagittal plane kinetics, increased hip adduction ROM, perhaps owing to reduced hip abductor force production, may also contribute to the age-related locomotor decline.

Previous studies have suggested that altered frontal and transversal plane mechanics such as increased maximal hip adduction and internal rotation angles and/or greater knee abductor moment may increase the risk knee overuse injuries (Noehren, Hamill & Davis 2012, Stefanyshyn et al. 2006, Noehren, Davis & Hamill 2007). Although age-related increases in the hip adduction ROM were observed during sprinting, differences in the peak hip adduction and internal rotation angles between age-groups were not significant. In addition, knee abductor moment showed no age-related effect during walking and running, whereas during maximal sprinting, the middle-aged group exhibited higher abductor moment when compared with the old group. However, the present study found no relationship between age-related secondary plane changes and the gait characteristics related to knee overuse injuries. This suggest that alterations in the secondary plane mechanics with age may not explain higher prevalence of the lower limb disorders among older people (Woolf & Pfleger 2003, McKean, Manson & Stanish 2006).

6.3.3 Mechanisms of age-related locomotor decline

The majority of the research to date examining age-related locomotion ability and muscular force capacities have focused on the proximal lower limb muscles such as knee extensors rather than ankle plantarflexors (Liu & Latham 2009, Hairi et al. 2010, Geirsdottir et al. 2012). Findings of these experiments generally suggest a strong link between walking ability and knee extensor strength. However, in the present study we found age-related deficit in the ankle moment and power generation during locomotion, whereas no deterioration was present at the knee moment or power. This suggests that the force generation capacity of the ankle plantarflexors rather than knee extensors may play a critical role in maintaining locomotor performance in older age. The fact that the age-related declines in force generation capacity occur similarly in different lower limb muscle groups (Candow & Chilibeck 2005, Karamanidis & Arampatzis 2006), may be an explanation for strong association between walking ability and knee extensor capacity in the previous studies (Liu & Latham 2009, Hairi et al. 2010, Geirsdottir et al. 2012), even though it may not be the limiting factor of the walking performance.

Because an age-related loss of muscle force generation capacity occurs similarly in different lower limb muscle groups (Candow & Chilibeck 2005, Karamanidis & Arampatzis 2006), one might ask why the reduction in the lower limb joint moments and powers during locomotion take place mainly at the level of ankle plantarflexors instead of knee and hip muscles. One possible explanation for this question is a recent notion that a much greater percentage of the maximal force generation capacity is required from the ankle plantarflexors than knee and hip muscles to drive the motion during walking gait (Beijersbergen et al. 2013). By expressing joint moments during walking in relation to the maximal available effort of each joint, the researchers (Beijersbergen et al. 2013) pointed out that old adults walk near the maximal capacity of the ankle plantarflexors while the knee and hip muscles operate at much lower relative effort. Furthermore, the results of a previous modelling study, where researchers progressively weakened all lower limb muscle groups in gait simulation suggest, that walking gait is remarkably robust to weakness of the knee and hip extensors but very sensitive to weakness of the ankle plantarflexors (van der Krogt, Delp & Schwartz 2012). These findings together with the current results show that age-related decline in muscular capacity first challenges the normal function of ankle plantarflexors during locomotion, thus leading to compensatory actions such as shorter steps and shift of the joint kinetics from ankle toward more proximal joints.

Although the current results showed no age-related effects in the hip frontal plane moment during locomotion, differences in the frontal plane kinematics during running and especially sprinting may reflect a compensatory movement strategy due to diminished hip abductor muscle force generation capacity during single leg support. Because the inverse dynamics approach provides information only about the net joint moments, thus, ignoring antagonist forces acting on the opposite direction on the joint (Winter 2009, 107-136), there is need for future studies, that account for hip adductor muscle force generation, to further understand the frontal plane mechanics and function of the hip during locomotion.

6.4 Limitations

Certain limitations of this thesis should be considered when interpreting its findings. First, a cross-sectional design was used in the studies. While the present thesis suggest implications for prevention and management of stress-related disorders and age-related locomotor decline, longitudinal studies are needed to evaluate a potential positive outcomes of intervention strategies such as gait modifications or specific training programs. Second, the participants in the studies I, III and IV were males and in the study II females. Because of certain sex-related differences in walking (Boyer, Beaupre & Andriacchi 2008) and running (Ferber, Davis & Williams 2003, Gehring et al. 2014) mechanics, caution should be used in generalizing these results to both genders. In addition, only limited functional parameters, such as muscle force production capacities or joint ROMs, were measured and compared between different subject-groups. Future work examining lower limb mechanics together with these factors will provide more detailed insights into the mechanisms behind different locomotor patterns as well as age-related locomotor decline.

Methodological concerns of this thesis are those which are commonly associated with the motion analysis and inverse dynamics methods. First, because movement data are calculated based on skin marker trajectories, it is highly dependent upon marker placement and may be also influenced by soft-tissue movement artifact (Leardini et al. 2005). Second, despite the inverse dynamics approach being used widely to quantify joint kinetics during locomotion, it is limited in its ability to account for the effects of in-series elastic structures and the energy transport via two-joint muscles from one joint to another (Cronin et al. 2013, van Ingen Schenau 1989). Also, inverse dynamics are unable to account for antagonist forces acting on the opposite direction on the joint, which may lead to underestimation of the amount of internal moments and forces acting on the lower limb joints (Brechter & Powers 2002). However, because all subject-groups are equally exposed to these methodological issues, it is very unlikely that the consistent group differences in this thesis were due to measurement errors.

7 PRIMARY FINDINGS AND CONCLUSIONS

The present study showed that locomotor pattern diversity during both walking and running largely affects lower limb mechanics and the amount of joint loading during the stance phase of gait. In addition, locomotion analysis of different age-groups revealed joint specific attenuations in muscle moment and power generation with age. The main findings and conclusions of the present study are as follows:

1) Different walking patterns largely affected the amount of knee frontal plane moment, which is known to correspond with medial knee joint loading. Individuals who demonstrated either knee extensor (EXT) or flexor (FLX) dominant gait patterns exhibited significantly higher peak knee frontal plane moment during the stance phase of gait compared with those who had a typical gait pattern. Due to established relationships between medial knee loading and the presence of OA, people in EXT and FLX subgroups may be at a higher risk for development of knee OA.

2) Rearfoot (RFS) versus forefoot (FFS) striking techniques were associated with different running-induced loading of the lower limb. FFS runners demonstrated lower patellofemoral stress and knee frontal plane moment but higher Achilles tendon force compared with the RFS runners pattern. As a result, FFS technique may reduce the risk of running-related knee injuries compared with RFS. However, higher ankle plantarflexor and Achilles tendon loading may increase risk for ankle and foot injuries among FFS.

3) Ageing resulted in joint specific alterations in lower limb mechanics during locomotion. The most prominent decline in joint kinetics in older age occurred at the ankle joint level across different modes and intensities of locomotion. As a compensatory action for impaired ankle propulsion, old adults demonstrated higher muscular efforts of the more proximal

lower limb muscles than their younger counterparts when walking and running at the same speed. During maximal sprinting, age-related reduction in joint kinetics was present at the ankle and hip, but interestingly, not at the knee joint. These findings indicate that age-related locomotor decline depends mainly on impaired ankle propulsion. As a result, exercise interventions designed to prevent age-related locomotor impairments should focus on improving especially ankle plantarflexor capacity.

4) Whole body secondary plane mechanics showed no age-related changes in walking. However, a tendency towards larger hip adduction ROM and pelvic drop motion but reduced trunk lateral flexion during running and sprinting were observed in the old group. These alterations in the frontal plane kinematics likely reflect a compensatory strategy to maintain lateral balance during single leg support but may not explain higher prevalence of the lower limb disorders among older people.

YHTEENVETO (FINNISH SUMMARY)

Yksilöllisen liikkumistyylin ja ikääntymisen vaikutukset alaraajan mekaniikkaan ja kuormitukseen kävelyssä ja juoksussa

Yksipuolisesti toistuva biomekaaninen kuormitus nähdään yhä enenevässä määrin keskeiseksi riskitekijäksi alaraajojen kuormitusperäisten vammojen syntymisessä. Valtaosa tästä kuormituksesta on peräisin ihmisen liikkumisesta eli kävelystä ja juoksusta. Vaikka kävely- ja juoksutyylien tiedetään vaihtelevan yksilöllisten erojen sekä ikääntymisen seurauksena, näiden vaikutukset alaraajan biomekaaniseen kuormitukseen tunnetaan huonosti.

Tämän väitöskirjatutkimuksen tavoitteena oli tuottaa uutta tietoa siitä, miten yksilölliset liikkumistyylit sekä ikääntyminen vaikuttavat alaraajan liikemekaniikkaan ja kuormitukseen kävelyssä ja juoksussa. Aihetta tutkittiin mittaamalla alustan reaktiovoimia voimalevyantureilla sekä kehon liikeratoja 3D–liikeanalyysimenetelmällä. Alaraajan mekaniikkaa ja nivelkuormitusta arvioitiin käänteisen dynamiikan avulla.

Aikaisempien tutkimusten mukaisesti suurimmat erot ihmisen kävelytyyleissä tunnistettiin polven ja erityisesti reisilihasten toiminnassa. Keskeisenä löydöksenä havaittiin etureisi- ja takareisipainotteisen kävelytyylin aiheuttama suurentunut polven sisemmän nivelpinnan kuormitus tavalliseen kävelytyyliin verrattuna. Tutkimuksessa juoksutyylit jaettiin kanta- ja päkiäaskellukseen jalkaterän alkukontaktin aikaisen kulman perusteella. Juoksun aikainen polven kuormitus oli selvästi matalampi päkiäaskeltajilla, mikä näkyi sekä pienempänä polvilumpion alapinnan että polven sisemmän nivelpinnan kuormituksena kanta-askeltajiin verrattuna. Päkiäaskellus vähensi myös alkukontaktin aikaisia törmäysvoimia, mutta suurensi akillesjänteen ja pohjelihasten kuormitusta.

Nuorten ja iäkkäiden liikkumisen mekaniikassa selkein ero havaittiin nilkan ojentajalihasten toiminnassa. Ikääntyneillä miehillä nilkan ojentajalihasten työntövoimassa ilmeni vajetta kävelyssä sekä erityisesti juoksussa, mutta he pystyivät kuitenkin ylläpitämään saman nopeuden nuorten miesten kanssa lisäämällä polven ja lonkan lihasten tehontuottoa. Maksimaalisessa pikajuoksussa suuremman nopeuden saavuttaneet nuoret miehet tuottivat enemmän lihastehoa nilkan ojentajalihaksilla sekä lonkan ojentaja- ja koukistajalihaksilla iäkkäisiin miehiin verrattuna. Nuoret miehet myös kallistivat pikajuoksussa ylävartaloa enemmän puolelta toiselle, mikä saattoi mahdollistaa lonkan lähentäjälihasten tehokkaamman osallistumisen lonkan ojennusliikkeen tuottamiseen. Yllättävää sen sijaan oli, että iäkkäiden miesten liikkumisessa ei havaittu vajetta polvinivelen tehontuotossa.

Johtopäätöksenä voidaan todeta, että yksilöllisillä kävely- ja juoksutyyleillä on keskeinen vaikutus alaraajan nivelten kuormitukseen ja siten mahdollisesti kuormitusperäisten vammojen syntymiseen. Toisaalta kuormituksen pienentäminen liikkumistyylin muutoksilla saattaa olla potentiaalinen keino vähentämään nivelkuormitusta ja siitä johtuvia kiputiloja. Havainnot nilkan ojentajalihasten heikentyneestä toiminnasta iän myötä puolestaan rohkaisevat tämän lihasryhmän harjoittamiseen vanhemmalla iällä.

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ORIGINAL PAPERS

I

KNEE EXTENSOR AND FLEXOR DOMINANT GAIT PATTERNS INCREASE THE KNEE FRONTAL PLANE MOMENT DURING WALKING

by

Kulmala, J.P., Äyrämö S. & Avela, J 2013

Journal of Orthopaedic Research 45(12):2306-13

II

FOREFOOT STRIKERS EXHIBIT LOWER RUNNING-INDUCED KNEE LOADING THAN REARFOOT STRIKERS

by

Kulmala, J.P., Avela, J., Pasanen, K. & Parkkari, J. 2013

Medicine and Science in Sports and Exercise 45(12):2306-13

III

WHICH MUSCLES COMPROMISE HUMAN LOCOMOTOR PERFORMANCE WITH AGE?

by

Kulmala J.P., Korhonen M.T., Kuitunen S., Suominen H.,
Heinonen A., Mikkola A. & Avela J. 2014

Journal of the Royal Society Interface. 11(100): 1-10

IV

AGE-RELATED DIFFERENCES IN FRONTAL PLANE MECHANICS ACROSS DIFFERENT LOCOMOTION MODES

by

Kulmala J.P., Korhonen M.T., Kuitunen S., Suominen H.,
Heinonen A., Mikkola A. & Avela J. 2014

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