

Biomechanics of the Hip and Pelvis and Predictors of Injury During Military Load Carriage

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DECLARATION OF ORIGINALITY

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ABSTRACT

Musculoskeletal injuries are among the most common and costly in the British military. It is well reported that female soldiers have significantly higher rates of injury than males, particularly in the hip and pelvis. In the UK, female military trainees have been found to be up to 48 times more likely to obtain a pelvic musculoskeletal injury than male recruits. In 2016 the ban on females serving in Ground Close Combat (GCC) roles was lifted in the UK. These roles are characterised by their extremely high physical requirements and hence higher rates of injury. Male GCC trainees have a seven-fold increased risk of hip and pelvic fractures compared with standard male entrants. It is expected that as more females begin training for GCC roles, their already significantly increased risk of injury will rise further. Marching with body borne loads is an integral aspect of military training and has been suggested as a leading cause of musculoskeletal injury. The effects of carrying military loads on spatiotemporal and kinematic gait parameters are well reported in male soldiers. However, the few studies directly comparing males and females present conflicting findings and, as a result, the biomechanical factors involved in hip and pelvic injuries are still unclear.

The gait of 38 subjects walking and 19 subjects running both unloaded and with body borne military-relevant loads of 20kg and 40kg was recorded. Volunteers were either current or ex-military personnel, or physically fit civilians with load carriage experience. Gait data of the trunk and lower limb were captured using a 10 camera optical motion tracking system, force plate and surface EMG electrodes. Data was processed through an inverse dynamics based musculoskeletal model, Freebody, to obtain joint angles, torques, muscle forces and contact forces. Differences according to sex and carried load were examined using repeated measures ANOVA. Data obtained from these studies was used to develop static finite element models of the hemi-pelvis for an average male and female subject, and to simulate loading during walking with and without carried loads at three points in the gait cycle; weight acceptance, midstance, and push off.

It was found that males and females walked with similar gait parameters when carrying no load and 20kg. However when 40kg was carried, females displayed significantly shorter stride length and spent longer in double support than males, possibly due to fatigue or in an effort to reduce internal forces in the hip. When running, females were observed to have significantly shorter stride length when carrying load. Females were also found to generate a significantly greater proportion of propulsive torque at the hip compared with males, suggesting that

they may be inherently more hip dominant. Factors associated with increased hip dominance included faster speed and female sex. During both walking and running, females' hip joint contact force was angled significantly more anteriorly and superiorly than males. Results of the finite element analysis revealed that peak von Mises stresses in the pubic rami occur during toe push off. Stress in the female pelvis model was found to be higher across all load conditions. Increased adductor muscle force, pelvic obliquity and medial hip joint contact force were found to be significant predictors of greater von Mises stresses in the rami. Analysis of EMG signals from four lower limb muscles revealed that females were operating at a greater subjective workload and energetic cost than males when carrying load. Regression analyses confirmed that female sex was a significant predictor of increased hip joint contact force over the gait cycle.

The work in this thesis supports the theory that males and females adapt their gait differently in response to body born load carriage, and that females are more likely to be injured because they are working at closer to the maximum threshold where they can no longer maintain normal gait function. This, combined with inherent biomechanical differences and other intrinsic and extrinsic factors, may explain female soldiers' significantly higher rates of hip and pelvic injuries. Increased emphasis on strength training of the quadriceps and core musculature together with a more progressive approach to load carriage may improve clinical outcomes.

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List of Abbreviations

ANOVA	Analysis of Variance
ASIS	Anterior superior iliac spine
BW	Body weight
CoM	Centre of Mass
DOF	Degrees of freedom
EMG	Electromyography
FEA	Finite Element Analysis
GCC	Ground Close Combat
GRF	Ground reaction force
HSD	Honestly significant difference
JCF	Joint contact force
MRI	Magnetic Resonance Imaging
MSk	Musculoskeletal
RAF	Royal Air Force
ROM	Range of motion
SD	Standard deviation
SI	Sacro-iliac
VIF	Variance inflation factor

Chapter 1

Introduction

This chapter describes the motivation for the thesis as well as the aims, objectives and thesis structure.

1.1 Motivation

Musculoskeletal (MSk) injuries are among the most common and costly in the British military, leading to approximately 60% of all medical discharges (Ministry of Defence 2018). The majority of these injuries occur due to overuse during training. Incidence rates for MSk injuries in British Army recruits has been found to be as high as 48%, with the lower limbs most commonly affected (Sharma et al. 2015). Of these injuries, stress fractures have the longest rehabilitation times, up to 16 weeks, causing severe delays in training and lost man hours (Sharma et al. 2015; Springer and Ross 2013).

It has been well reported that injury rates in female soldiers are significantly higher than in males, particularly for the hip and pelvis (Sharma et al. 2015; Blacker et al. 2008; Mattila et al. 2007; Rauh et al. 2006). In the UK, female RAF recruits were found to be 48 times more likely to obtain a pelvic stress injury than male recruits, and over 14 times more likely to suffer from a hip injury. Standard female Army entrants were found to be over ten times more likely to obtain a stress fracture at the hip. In 2016 the ban on females serving in Ground Close Combat (GCC) roles was lifted in the UK. These roles are characterised by their extremely high physical requirements and hence higher rates of injury. Male GCC trainees have a seven-fold increased risk of hip and pelvic fractures compared with standard male entrants (Ministry of Defence 2016). It is expected that as more females begin training for GCC roles, their already significantly increased risk of injury will rise further.

Marching with body borne loads is an integral aspect of military training and has been suggested as a leading cause of MSk injuries as it requires both strength and aerobic endurance (Nindl et al. 2016; Orr et al. 2015; Knapik et al. 1996). Orr et al. (2015) found that 21% of all injuries reported in the Australian Army were directly attributable to load carriage. However, the biological mechanisms involved in these injuries are still not fully understood. While many studies have investigated the effects of load carriage on the biomechanics of males' gait, few have linked their results to the clinical injuries affecting soldiers. To date, there is limited evidence informing how females respond to carrying heavy loads and how it may influence the development of hip and pelvic injuries. Previous studies have found that males and females adapt their gait in response to load carriage, and conclude that these changes contribute to injuries by altering loading in the musculoskeletal tissues. Studies of soldiers and athletes have found correlations

between altered gait kinematics and musculoskeletal injuries. However, few studies have directly compared males' and females' gait when carrying load, and those that have report conflicting findings. Therefore, this thesis will investigate sex-specific gait adaptations that may occur during load carriage. A biomechanical approach will be used in this thesis as it allows for assessment of the motion, forces and deformation of musculoskeletal tissues which directly influence the mechanisms involved in overuse injuries.

1.2 Aims and Scope

The primary aim of this thesis is to further understanding of how carrying body borne military-relevant loads affects the gait of males and females. This thesis also aims to use a biomechanical approach to investigate if and how these gait adaptations contribute to the mechanisms involved in hip and pelvic musculoskeletal injuries commonly affecting female soldiers. This research is applicable to military populations as well as load carrying civilians, and is limited to investigating the effects of carried load, anthropometry and sex on the lower limb musculoskeletal system, in particular the hip and pelvis.

1.3 Objectives

The main objectives of this thesis are to:

- i. combine optical motion tracking methods and musculoskeletal modelling to quantify gait adaptations made during walking and running with body borne loads;
- ii. investigate how these adaptations affect the transfer of musculoskeletal forces through the hip and pelvis using finite element analysis; and
- iii. identify potential risk factors and parameters contributing to pelvic musculoskeletal injuries.

This thesis will therefore investigate the hypothesis that sex-specific gait adaptation strategies may affect musculoskeletal loading and hence injury risk. Few previous studies of load carriage have included females, and of those many reported conflicting results. As such, a broad approach will be taken in this thesis, investigating spatiotemporal, kinematic and kinetic parameters in order to identify as many relevant variables as possible.

1.4 Thesis structure

This thesis consists of seven further chapters, as follows:

Chapter 2 provides an overview of the lower limb musculoskeletal system as well as the injuries commonly obtained by soldiers. It also provides a review of the literature investigating the effects of load carriage on gait and the methods used.

Chapter 3 is a comprehensive biomechanical analysis of walking with body borne loads using optical motion tracking and musculoskeletal modelling. Spatiotemporal parameters, joint angles, torques and contact forces are presented and significant results discussed here.

Chapter 4 utilises the results found in Chapter 3 to develop static finite element models of the hemi-pelvis for an average male and female subject, and to simulate loading during walking with and without carried loads at three points in the gait cycle; weight acceptance, midstance, and push off.

Chapter 5 provides a biomechanical analysis of the lower limb during running with body borne loads using optical motion tracking and musculoskeletal modelling. Spatiotemporal parameters, joint angles, torques and contact forces are presented and discussed.

Chapter 6 investigates changes in muscle activation patterns in the lower limb while walking and running both with and without carried loads. Integrated muscle activity over the gait cycle as well as muscle burst duration and peak activation timings are presented for the soleus, vastus lateralis, biceps femoris and gluteus medius.

Chapter 7 uses multiple regression analyses to identify potential risk factors for hip and pelvic injuries and correlating factors with the aim to further the understanding of how these injuries develop.

Chapter 8 presents a summary and discussion of the thesis findings and potential applications to military and civilian populations, as well as proposing future work.

Chapter 2

Literature Review

This chapter describes the functional anatomy of the hip and pelvis, as well as biomechanical differences in the musculoskeletal systems of males and females. The epidemiology and aetiology of musculoskeletal injuries in soldiers is discussed, and computational methods for investigation of these injuries are introduced. This leads to the proposed direction of the thesis.

2.1 Functional Anatomy of the Hip and Pelvis

The complexity of the human pelvis reflects its multi-functional role in the musculoskeletal system. The pelvis both supports the weight of the upper body and aids locomotion through its attachment to the lower limbs (DeSilva and Rosenberg 2017). The bony pelvis includes the sacrum, coccyx and two os coxae, as shown in Figure 2.1. Each os coxae is made up of the ilium, ischium and pubis (Lewis et al. 2017) and attaches at the sacro-iliac (SI) and pubic symphysis joints to form the full pelvic ring. The SI joint allows little to no movement in healthy adults (Brooke 1924), while the pubic symphysis contains a fibrocartilaginous disc which allows for small translations and rotations between the two pubic bones (Hagen 1974). Additionally, the pelvis can itself rotate relative to the trunk and lower limb with three degrees of freedom (DOF). Musculoskeletal loads are transmitted between the upper body and lower limbs through the synovial articulation between the acetabulum of the os coxae bone and the femoral head. This multi-axial ball and socket joint allows for rotations with three DOFs between the pelvis and femur (Neumann 2010).

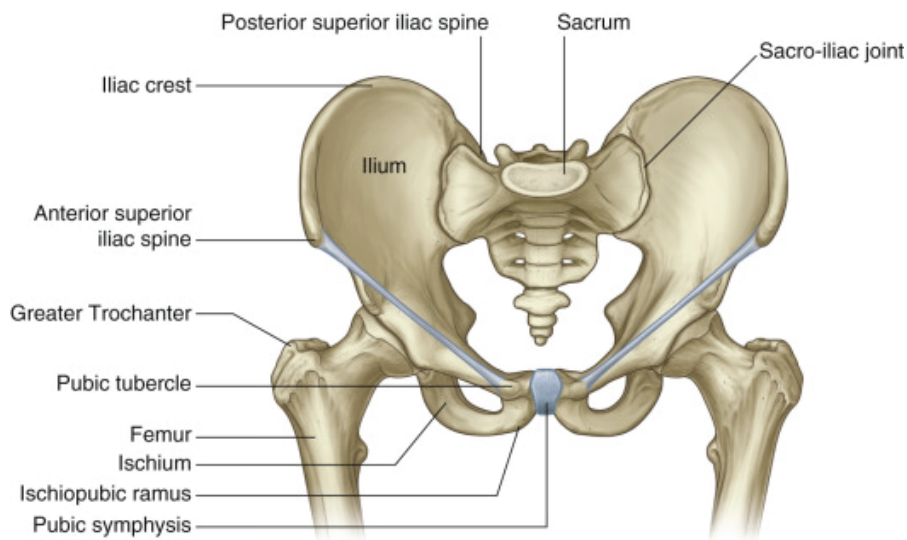


Figure 2.1: Bony anatomy of the pelvis and proximal femur. Reproduced with permission from Drake et al. (2009)

The hip joint is stabilised by the iliofemoral, pubofemoral and ischiofemoral ligaments (Drake et al. 2009) as well as the surrounding musculature (Figure 2.2(a)). These small surrounding muscles also work to externally rotate the hip together with the gluteal muscles, which also function as hip abductors. Motion of the thigh relative to the pelvis is primarily driven by the action of the quadricep, adductor, gluteal and hamstring muscle groups, many of which attach on the

pelvic bone.

The adductor and hamstring muscles in particular insert on to the narrow superior and inferior rami (Figure 2.2(b)-(c)). The quadricep muscles, including the rectus femoris, vastii and sartorius work to flex the hip (together with the large iliopsoas muscles) and extend the knee. The biceps femoris, semitendinosis and semimembranosis comprise the hamstring muscle group whose main function is hip extension and knee flexion. The adductor magnus, brevis and longus together with the gracilis and some of the small stabilising muscles work to adduct the hip (Chang et al. 2020).

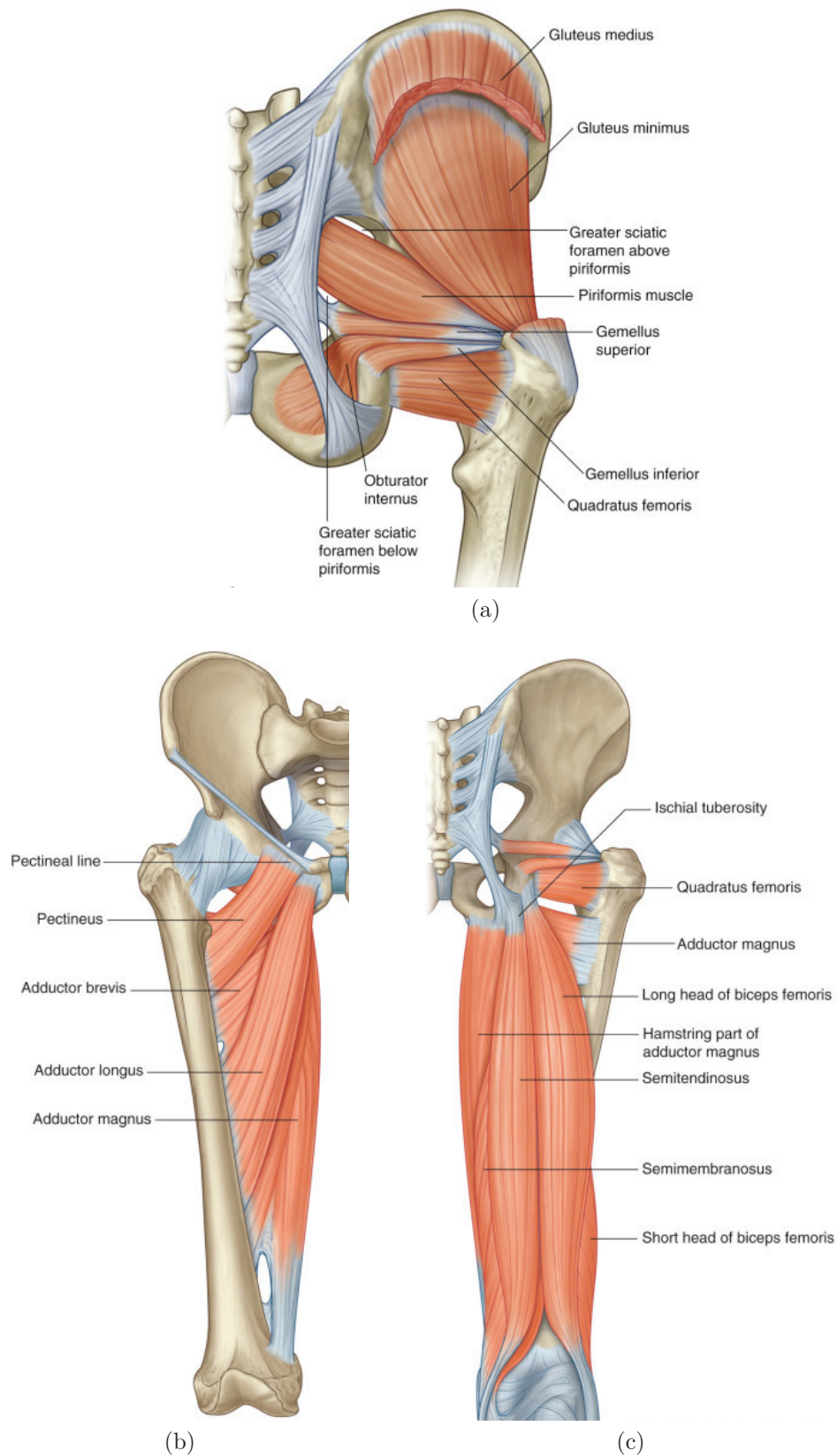


Figure 2.2: Musculature of the hip. Reproduced with permission from Drake et al. (2009)

2.1.1 Sexual Dimorphism in the Bony Pelvis

Differences in the male and female pelvis are thought to have arisen due to sex-specific selection pressures throughout evolution (Fischer and Mitteroecker 2017). The female pelvis is generally said to have less prominent ischial spines (Meindl et al. 1985; Tague 1992) and wider ischial bispinous width (Tague 1992), as shown in Figure 2.3. Though the female pelvis is often stated to be broader and flatter (Fischer and Mitteroecker 2017), significant variation in the pelvis of females has been observed (Ridgeway et al. 2008). The male pelvis is typically considered to have a longer, more curved sacrum and narrower sub-pubic arch angle (Caldwell and Moloy 1933; Tague 1992), as well as thicker bones (Abitbol 1996).

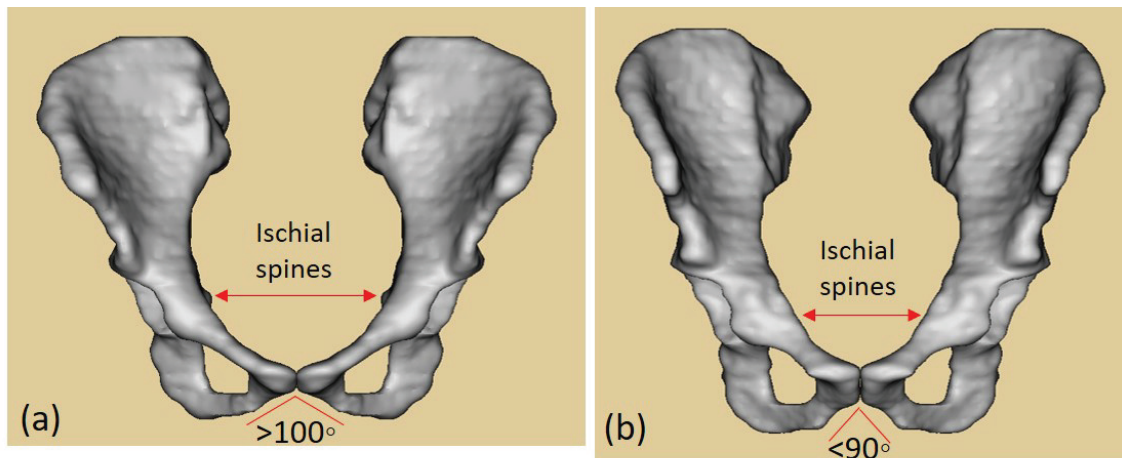


Figure 2.3: Differences in the (a) female and (b) male bony pelvis.

Whether these differences are due to allometric or obstetric demands has been debated (Rosenberg 1992; Abitbol 1996; Abouheif and Fairbairn 1997; Kurki 2011). It has been hypothesized that pelvic shape is influenced by hormonal factors (Badyaev 2002; Huseynov et al. 2016). A recent study by Fischer and Mitteroecker (2017) concluded that dimorphism in the height-width ratio and orientation of the iliac blades was largely due to allometry, while obstetric-relevant differences in pelvic shape, such as sub-pubic arch angle and distance between the acetabula, were not. Ridgeway et al. (2008) suggest that height moderately correlates with pelvic inlet and outlet and obturator foramen size in females. Limitations of their study included using a collection of disarticulated pelvises with small amounts of degeneration and no medical or ancestral history. No correlation was found between pelvic height and width.

Kuliukas et al. (2015) found that, in females, pelvic inlet shape varied from circular to oval to heart shaped (typically considered the male shape). Indeed Delprete (2017) found that there was no distinct differences in pelvic inlet shape between males and females. In addition, Albrich et al. (2015) found that the female sub-pubic arch angle can vary from 68.6-131.6°. Huseynov et al. (2016) present evidence that the bony pelvis continues to change shape throughout life, in both males and females. They suggest that between the ages of 15 and 25 years, the base of the sacrum moves posteriorly, the sizes of the pelvic inlet and outlet and the bi-acetabular distance increase in females. Factors affecting pelvis type include being exposed to strenuous physical exercise in adolescence (Abitbol 1996), and race (Rizk et al. 2004).

2.2 Gait Cycle and Mechanics

The gait cycle can be defined as the “time integral” between two successive occurrences of a gait event on the same leg, usually heel strike to heel strike (Whittle et al. 2012). This cycle is generally normalised from 0% (first occurrence of gait event) to 100% (second occurrence of gait event). The full gait cycle, including the motion of the pelvis and gait phases for the right leg, is shown in Figure 2.4.

The mechanics of the hip and pelvis vary significantly throughout the gait cycle. Following heel strike, weight is transferred to the ipsilateral leg, increasing vertical and braking (posterior) ground reaction forces. The hip joint contact force reaches a peak value of almost 250%BW (Bergmann et al. 2001) at approximately 15-20% of the gait cycle (Mathai and Gupta 2019). Eccentric contraction of the tibialis anterior promotes plantar flexion of the foot to meet the ground. The hip is generally near its most flexed position, before beginning to extend due to concentric contraction of the hamstrings and gluteal muscles (Whittle et al. 2012). The quadriceps muscles act to slow knee flexion and maintain leg stiffness in response to the increasing forces acting on the leg (Rose and Gamble 2015). The knee then begins to extend together with the hip and the contralateral foot leaves the floor, bringing the ipsilateral leg into the midstance position (Murray 1967). The hip abductors, notably the gluteus medius and tensor fascia latae, work to stabilise the pelvis while only one foot is in contact with the ground (Whittle et al. 2012).

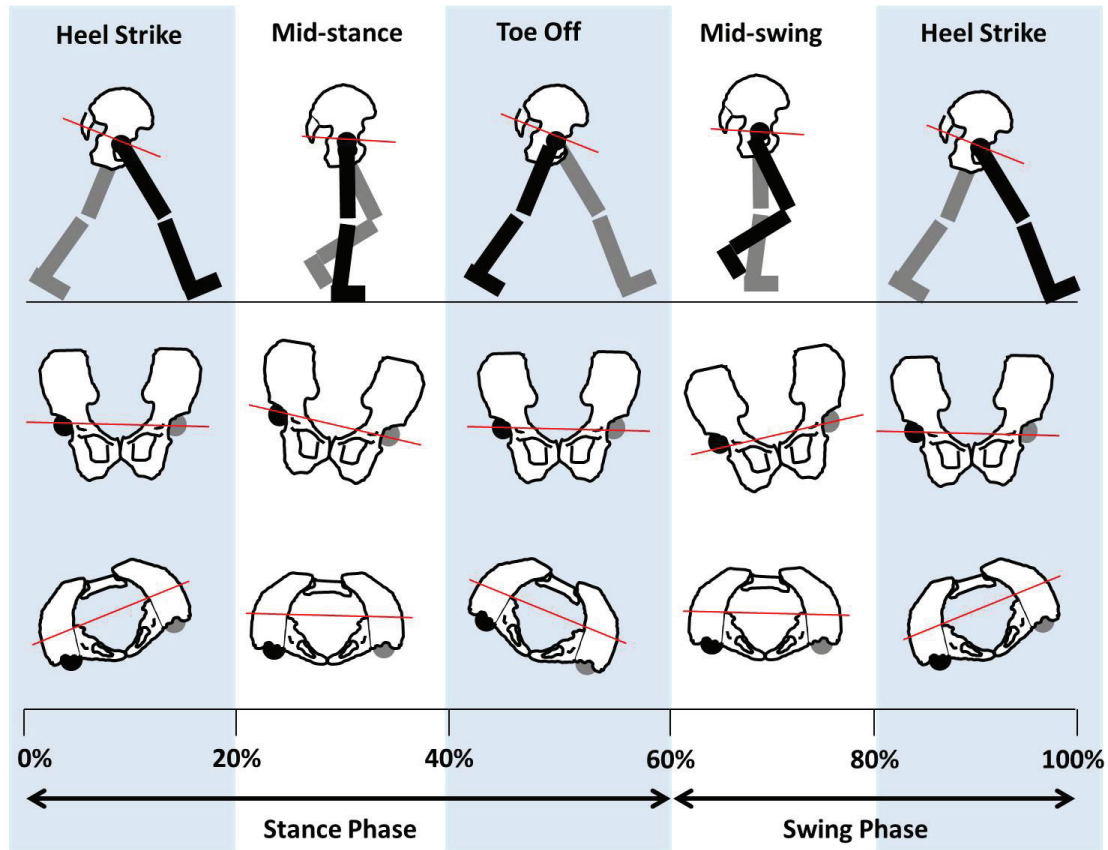


Figure 2.4: Gait cycle and motion of the pelvic girdle

Once past the vertical position, the contralateral heel strikes the floor and the moment about the hip joint switches from extension to flexion, which continues through the remaining gait cycle. Ankle plantar-flexor muscles contract to allow the foot to push off from the ground, generating another peak in the vertical and propulsive (anterior) ground reaction forces (Perry 2010). This toe push-off also corresponds with the second peak of the hip joint contact force, reaching 225%BW (Bergmann et al. 2001) at approximately 45-50% through the gait cycle (Mathai and Gupta 2019). Once the ipsilateral toe leaves the ground (~60% of the gait cycle), the leg swings forward with the knee and hip flexed due to contraction of the quadriceps and hamstrings muscles, before the knee is extended and the next heel strike of the same foot completes the gait cycle. The hip joint experiences a large change in position over the gait cycle and any study of the hip and pelvis must therefore consider the changing loading conditions at different points in the gait cycle in order to identify potential mechanisms for injury.

2.2.1 Sex-specific Differences in Normal Gait Mechanics

Males and females are generally perceived to walk differently (Frimenko and Whitehead 2014), although studies aiming to identify sex-specific differences in gait parameters have returned conflicting results. Many studies have found that females preferentially take shorter strides than males (Chao et al. 1983; Hageman and Blanke 1986; Richard et al. 1995; Smith et al. 2002; Cho et al. 2004; Chung et al. 2010; Bruening et al. 2015) and walk with higher cadence (Richard et al. 1995; Chung et al. 2010; Forczek and Staszkievicz 2012). They attribute this to females' shorter height. However some have reported no sex-specific differences in stride length and cadence (Alton et al. 1998), or differences that only emerge at fast walking speeds (Oberg et al. 1993). Analysis of stance and double support phases revealed no differences between young males and females (Kadaba et al. 1990; Laufer 2003; Cho et al. 2004; Chockalingam et al. 2012), nor did comparisons of stance and swing time (Alton et al. 1998).

Mazzà et al. (2009) found similar trunk accelerations between males and females, while other studies have observed females to walk with greater medial-lateral sway and forward flexion of the trunk at very low speeds (Goutier et al. 2010), and less trunk forward flexion at faster speeds (Goutier et al. 2010; Chung et al. 2010). Conversely a study by Bruening et al. (2015) found that females exhibited smaller medial-lateral trunk range of motion, and larger range of motion in the transverse plane compared to males when walking at self-selected speeds.

Sex-specific differences in pelvic girdle motion are widely reported but conflicting. It is generally accepted that females walk with greater pelvic obliquity (motion in the frontal plane) than males (Smith et al. 2002; Cho et al. 2004; Chockalingam et al. 2012; Bruening et al. 2015). Cho et al. (2004) found that females walked with greater anterior pelvic tilt, however this result was not supported by Chung et al. (2010). Females have been reported to display significantly greater pelvic rotation in the transverse plane (Bruening et al. 2015; Crosbie et al. 1997). However some have observed no sex-specific differences in pelvic rotation (Smith et al. 2002; Chockalingam et al. 2012), or differences that only become significant at fast walking speeds (Whitcome et al. 2017).

In the hip, females have displayed significantly greater hip flexion (Kerrigan et al. 1998; Oberg et al. 1994; Cho et al. 2004), but not range of motion (ROM) in the sagittal plane (Alton et al. 1998; Kadaba et al. 1990; Cho et al. 2004).

Females' hip ROM in the frontal plane has been found, however, to be significantly greater than males during walking (Kadaba et al. 1990; Bruening et al. 2015). Moreover, Cho et al. (2004) found that the hip was more adducted over the gait cycle in females than males, supporting research demonstrating a larger Q-angle, the angle formed from the line connecting the mid-point of the patella to the ASIS and the vertical, in females (Horton and Hall 1989; Nguyen and Shultz 2007). No sex-specific differences in transverse plane hip ROM have been observed (Kadaba et al. 1990; Cho et al. 2004). In the knee, some studies have reported greater sagittal plane knee ROM in females (Kadaba et al. 1990; Kerrigan et al. 1998) however others have not (Alton et al. 1998; Cho et al. 2004; Hurd et al. 2004). Greater ankle ROM in the sagittal plane has also been observed in females (Kerrigan et al. 1998). No sex specific differences in hip (Yamazaki and Tanino 2017) or knee (Kerrigan et al. 2000) joint torques have been found during walking. Mathematical models have been used to investigate hip joint loading during midstance. Peak hip joint contact stress was found to be larger in females than males (Iglič et al. 2001), as was hip joint contact force and force inclination angle (Stankovski et al. 1996). The authors argued that these differences were due to females' smaller femoral head, and thus smaller joint contact area.

In summary, there is a consensus that females display significantly more hip flexion, adduction and abduction during walking compared with males. There is however some ambiguity around whether these differences are due to allometry, and if they affect how internal forces are transmitted through the musculoskeletal system. Any future studies should therefore include both males and females under the same walking conditions to allow for direct comparison between them.

2.3 Musculoskeletal Injuries in the Military

Soldiers' fighting effectiveness and survivability depend on their physical readiness (Harman et al. 2008), and all military personnel must therefore maintain a high state of physical fitness. When on deployment, soldiers can be expected to undertake low intensity endurance activities such as carrying heavy loads over long distances and difficult terrain, as well as high intensity tasks involving sprinting and obstacle navigation (Harman et al. 2008). Physical training is an integral requirement in the preparation of soldiers for the demands of military duty. Basic training in the British Armed Forces varies depending on age and role. The most physically intensive training courses are designed to prepare recruits for Ground

Close Combat roles and include the Combat Infantryman's Course and the Royal Marines Commando basic training course, lasting 26 (Phase 1) and 32 weeks respectively. For many new recruits this represents a sharp and substantial increase in volume of physical activity and hence injuries are common. However, the physically demanding nature of military roles in general means that there is a high risk of injury even among trained soldiers. This costs the military in terms of medical resources, lost man hours and morale (Zambraski and Yancosek 2012).

Musculoskeletal injuries are the most common cause of medical discharge and downgrading (Ministry of Defence 2018; Schwartz et al. 2014). From 2017 to 2018 musculoskeletal disorders accounted for between 46-57% of all medical discharges in the British Armed Forces (Ministry of Defence 2018). In addition, females in the Royal Navy and Royal Air Force (RAF) were found to be significantly more likely to be medically discharged than males. In new recruits, the risk of medical discharge due to MSk injury has been reported as between two and four times higher in female entrants than males (Ministry of Defence 2016). Musculoskeletal injuries do not only present a significant issue during training. A review of US military personnel revealed the most common cause of medical evacuation from conflict zones between 2004 and 2007 was non-combat related MSk injuries, which accounted for 24% of all medical evacuations (Cohen et al. 2010). Females were found to be over twice as likely as males to be medically evacuated. In addition, 27% of American female soldiers developed a MSk injury from training when on deployment that resulted in lost duty days (Roy et al. 2015).

MSk injuries in military personnel commonly fall into two categories: trauma and overuse. Overuse injuries make up approximately 65% of all training injuries (i.e. injuries sustained during organised fitness sessions, road marches and runs, basic training etc.) with 81% of all pathologies occurring in the lower limb (Robinson et al. 2016). Overuse injuries are usually further defined in the literature as either stress fractures or non-stress fractures, and it is generally reported that females are more than twice as likely as males to obtain an overuse injury (Knapik et al. 2001; Bell et al. 2000). In addition, females have been found to be 2.5 times more likely to require time off (Bell et al. 2000) and 7.5 times more likely to be medically discharged due to an overuse injury than males (Gemmell 2002).

The most common overuse injury sites in US male soldiers are reportedly the back, ankle and knee (Knapik et al. 1993). Analysis of injuries sustained by

Australian soldiers carrying body-borne loads confirmed this result, while the primary injury sites for females were found to be the back, foot and shoulder (Orr et al. 2015). Females were also over twice as likely as males to obtain a load carriage injury in the hip. In the UK, Gemmell (2002) reported significantly higher rates of overuse injury in the Achilles tendon and back in female recruits compared to males. Female standard entrants have also been reported as significantly more likely to obtain an overuse MSk injury across all anatomical locations, but particularly in the hip, with between 5.77 and 14.38 times increased risk compared with males (Ministry of Defence 2016). Once MSk pathology at the hip occurred, females trainees were over 9 times more likely to be medically discharged as a result. This thesis will therefore focus on the biomechanics of hip and pelvic injuries.

2.3.1 Stress Fracture Injury Incidence and Treatment

Stress fractures are defined as partial or complete bone fractures resulting from the repeated application of forces, which would not otherwise be large enough to fracture the bone in a single loading situation (Martin and McCulloch 1987). During physical activity the skeleton is loaded by muscle and body forces which cause mechanical stress and deformation of the bone (Bennell et al. 1999). When these cyclic, submaximal loads are applied, osteoclastic reabsorption occurs faster than osteoblastic formation, resulting in weakened bone tissue and eventually fracture (Scully and Besterman 1982).

Female soldiers are frequently cited to be at increased risk of stress fractures, with reported fracture rates higher for females training in mixed-sex groups compared with single-sex ones (Pope 1999; Hill et al. 1996; Bergman and Miller 2001). Stress fractures have been reported to affect between 4.5 and 23.9% of female soldiers (Knapik et al. 2012; Gam et al. 2005; Beck et al. 2000; Rauh et al. 2006; Finestone et al. 2008; Lappe et al. 2001; Shaffer et al. 2006; Gemmell 2002; Wentz et al. 2011) and between 0 and 11.2% of male soldiers (Knapik et al. 2012; Beck et al. 2000; Finestone et al. 2008; Gemmell 2002; Gam et al. 2005; Sharma et al. 2015). In addition, many females recruits suffer from multiple stress fractures during training (Niva et al. 2009).

The most common stress fracture sites in females are reportedly the tibia, metatarsals, and pelvis (Kelly et al. 2000; Shaffer et al. 2006). Pelvic fractures were found to represent between 14 and 27% of all stress fractures in females

(Kelly et al. 2000; Shaffer et al. 2006; Gam et al. 2005; Pope 1999; Beck et al. 2000; Mattila et al. 2007), but less than 0.1% of those occurring in males (Gam et al. 2005; Pope 1999; Beck et al. 2000; Mattila et al. 2007). The majority of pelvic stress fractures occur in the pubic rami (Beck et al. 2000; Pope 1999). In the British Army, female standard entrants and officer trainees were found to be 10.39 and 18.75 times more likely to obtain a hip or pelvic stress fracture than their male counterparts respectively. In the Royal Air Force, female standard entrants reportedly suffered 48.4 times more pelvic stress injuries than male entrants. Finally, male infantry trainees were 7.56 times more likely to obtain a hip or pelvic stress fracture than standard male entrants (Ministry of Defence 2016), suggesting that more physically demanding training courses may increase the risk of these injuries. Since females have begun training for GCC roles, no injury data has been made available, however it is expected that their rates of injury will increase even further.

Treatment of stress fractures generally involves limiting weight bearing for between one and six weeks, and further avoidance of high impact activity for up to 12 weeks (Springer and Ross 2013). Hip and pelvic fractures in particular can take up to four months to heal, compared with around two months for tibial and metatarsal fractures (Bennell and Brukner 1997). Because of the long rehabilitation times, stress fractures account for the most lost man hours compared with other MSk disorders (Knapik et al. 1993) and are most likely to result in a medical discharge (Gam et al. 2005).

2.3.2 Risk Factors for Musculoskeletal Injury

The aetiology of musculoskeletal overuse injuries is complex and involves many intrinsic and extrinsic factors (Bennell et al. 1999). The most commonly identified and significant intrinsic risk factor in both males and females was low aerobic fitness (Robinson et al. 2016; Beck et al. 2000; Rauh et al. 2006; Jones et al. 1993; Shaffer et al. 2006; Mattila et al. 2007; Knapik et al. 2001; Knapik et al. 2012). Lesser muscle strength was also found to increase the risk of injury (Mattila et al. 2007; Knapik et al. 2001). Krauss et al. (2016) found that unfit female recruits were 1.62 times more likely to obtain a stress fracture, while physically fit female recruits with high body fat levels were 1.27 times more likely to obtain a non-stress fracture MSk injury compared with healthy recruits. Jones and Knapik (1999) speculate that low fitness increases risk of injury because soldiers are working at higher levels of physiological stress relative to their maximum capacity for the

same physical output.

Other modifiable intrinsic risk factors include low BMI (Moran et al. 2008; Robinson et al. 2016; Barrack et al. 2014; Finestone et al. 2008; Lappe et al. 2001; Mattila et al. 2007; Knapik et al. 2012), smoking (Lappe et al. 2001; Bergman and Miller 2001), poor flexibility (Leetun et al. 2004; Knapik et al. 2001) and nutritional deficiencies (Friedl et al. 2008). Extrinsic risk factors for bone stress injuries in military populations include poor fitting kit (Ling et al. 2004) and carrying heavy loads (Orr et al. 2015; Birrell et al. 2007; Park et al. 2013; Polcyn et al. 2002; Roy et al. 2016). Roy et al. (2016) found that wearing a load of more than 15%BW increased the risk of MSk injury by 5.83 times in deployed female soldiers. The effects of carrying body borne loads on soldiers' gait are discussed fully in Section 2.4.

Non-modifiable factors, such as female sex (Jones et al. 1993; Mattila et al. 2007; Knapik et al. 2012; Cohen et al. 2010; Ministry of Defence 2018; Kunte et al. 2017), age (Lappe et al. 2001; Wilkinson et al. 2011; Mattila et al. 2007; Knapik et al. 2012; Cohen et al. 2010; Ministry of Defence 2018), height (Kelly et al. 2000; Moran et al. 2008; Bergman and Miller 2001) and race (Lappe et al. 2001; Knapik et al. 2012) have also been identified as risk factors. Bell et al. (2000), found that female sex was not a significant predictor of increased injury risk once physical and demographic factors were accounted for. They concluded that being female encompasses many of the anatomical and physiological injury risk factors, such as shorter height, smaller bones and lower $\dot{V}O_2$ max. However Bell et al. (2000) considered all training related injuries, while other studies maintain that female sex is still a significant risk factor for stress fractures even when adjusted for other variables (Mattila et al. 2007). Shorter height and leg length have been attributed directly to the development of pelvic stress fractures in females, as over-striding is said to increase the pull of the adductor muscles on the narrow pubic rami (Pope 1999; Hill et al. 1996; Kelly et al. 2000).

The female athlete triad has been reported to increase bone stress injury incidence in female civilian athletes (Barrack et al. 2014; Nattiv et al. 2007), but studies of female military populations have not observed evidence of all three indicators in either recruits or active duty soldiers (Armstrong et al. 2004; Lauder et al. 1999; Orr et al. 2011). Low bone mineral density (Beck et al. 2000; Friedl et al. 2008), history of menstrual dysfunction (Rauh et al. 2006; Shaffer et al. 2006; Bergman and Miller 2001) and history of dietary restrictions (Bergman and Miller

2001; Lauder et al. 1999; Armstrong et al. 2004) have all been individually observed and attributed to increased stress fracture risk in female soldiers, as has low blood iron levels (Moran et al. 2008).

Gait changes occurring as a result of fatigue have been associated with musculoskeletal overuse injuries. Rice et al. (2017) found evidence of plantar muscle fatigue in male recruits who had completed a 12.8km overground march carrying 35.5kg. Females walking overground with 9kg were found to increase spinal curvature in as little as 18 minutes (Orloff and Rapp 2004). Other studies have attempted to fatigue subjects in a laboratory setting. Lidstone et al. (2017) studied females carrying 55%BW while walking for 60 minutes on a treadmill. They found that vertical ground reaction force and musculoskeletal loading increased with muscular fatigue and reduced ability to attenuate shock. Similar results were found by Wang et al. (2012) who used plyometric exercises to induce fatigue in males. However these studies highlight the difficulty of accurately fatiguing subjects in a laboratory setting where subjects may not fatigue in the same way as when marching overground. The effects of fatigue during a loaded endurance march will therefore not be investigated in this thesis.

Biomechanical risk factors for MSk injuries have previously been identified in male soldiers, however studies of female soldiers are limited. Springer et al. (2016) measured the gait of male soldiers walking with 10kg and found that an initial loading response time of less than 12.5% of the total gait cycle was correlated with increased injuries in the knee, ankle and foot. Studies of Royal Marine recruits have shown that greater peak heel pressure and lower range of tibial rotation increased the risk of tibial stress fractures (Nunns et al. 2016), while delayed peak toe pressure and lower foot abduction increased risk of metatarsal fractures (Dixon et al. 2019). Studies including both sexes have revealed differences in foot centre of pressure in male and female recruits with patellofemoral pain (Thijs et al. 2007), and greater knee valgus and knee internal rotation angle in military cadets with stress fractures (Cameron et al. 2013). In addition, Giladi et al. (1987) found that male and female recruits with hip external rotation angles of over 65° had higher rates of lower limb stress fractures.

2.4 Biomechanical Effects of Load Carriage

2.4.1 Spatiotemporal

Many studies, particularly of male subjects, have investigated the effects of body borne load carriage on spatiotemporal aspects of gait. Increasing load has been found to result in an increase in double support time in males (Ghori and Luckwill 1985; Harman et al. 1992; Harman et al. 2000; Park et al. 2013; Kinoshita 1985), females (Ling et al. 2004; Simpson et al. 2012b; Harman et al. 1999a), and studies of both sexes (Huang and Kuo 2014). Polcyn et al. (2002) suggest that time spent in double support is increased to limit the time when the entire load is supported by a single leg. No significant changes in double support phase were found by Majumdar et al. (2010) for males carrying between 4-18kg, or Filzah Pg Damit et al. (2017) in males carrying 15kg. They concluded that the mass was not heavy enough to induce gait changes in well trained soldiers. Direct comparison of males and females carrying between 0.5-37kg revealed the effect of load and the interaction effect of load and sex to be significant for time spent in double support, indicating that increased carried load affected the double support phase of males and females differently. However, no significant differences between males and females were found for any load condition (Martin and Nelson 1986).

Reports of load effects on other phases of gait are conflicted. Some previous studies of males observed no significant changes in stance time (Filzah Pg Damit et al. 2017; Birrell and Haslam 2010; Ghori and Luckwill 1985) or swing time (Majumdar et al. 2010; Filzah Pg Damit et al. 2017), while others report stance time to increase with carried load (Park et al. 2013; Attwells et al. 2006). Birrell and Haslam (2010) only observed changes in stance time under heavier load conditions, however they did note that stance time was significantly longer when males carried 32kg in a rucksack compared with sharing the load between a rucksack and webbing. They suggest that distributing the load may improve soldiers' ability to adapt to heavy load carriage. The results of these conflicting studies suggest that males do not increase stance time until heavier loads of around 27kg are carried, which highlights the importance of studying heavy load carriage typical in infantry roles. Furthermore, studies which found no change in stance time had relatively few subjects (<12). Stance time has also been shown to increase with load in female civilians (Simpson et al. 2012b). Martin and Nelson (1986) again observed increasing load to affect the gait of male and female soldiers differently, with significant differences in stance phase between sexes occurring when loads over 17kg were carried. Conversely, Silder et al. (2013) found no

significant differences between males and females, although they did observe stance phase to increase with carried load. It is probable that Silder et al. (2013) found no differences between males and females because loads were normalised to subjects' body mass and participants walked at a self-selected pace. When running with loads of up to 30%BW stance time was also found to increase in males and females (Silder et al. 2015; Xu et al. 2017; Liew et al. 2016; Willy et al. 2016).

The majority of previous studies have found no significant changes in stride length in either males or females with increasing carried load (Harman et al. 2000; Majumdar et al. 2010; Filzah Pg Damit et al. 2017; Ling et al. 2004; Simpson et al. 2012b; Huang and Kuo 2014; Silder et al. 2013; Lidstone et al. 2017). Harman et al. (2000) in particular found that males did not alter their stride length even when loads of up to 47kg were carried. With the exception of Harman et al. (1999a), who found stride length changes in females, the majority of studies of females used loads of less than 40%BW (~ 25 kg), which may not have been heavy enough to cause significant changes in stride length. However, Park et al. (2013) found a significant negative linear trend between stride length and carried loads between 0.06kg and 27kg in males. Attwells et al. (2006) observed a non-linear trend in males, with stride length initially increasing when lighter loads were carried, and then decreasing as load further increased to 50kg, resulting in no significant net change from walking unloaded. Martin and Nelson (1986) observed a similar non-linear trend in males' stride length, and a negative linear trend in females' stride length as load increased, resulting in females walking with significantly shorter stride length than males when carrying the heavier loads of 30kg and 36kg.

As with stride length, many studies found no changes in self selected walking speed (Majumdar et al. 2010; Park et al. 2013) or cadence (Polcyn et al. 2002; Ling et al. 2004; Harman et al. 1999a). Simpson et al. (2012b) found that females reduced their cadence instead of their stride length in order to walk at slower speeds when carrying 30% and 40%BW. Conversely, Caron et al. (2015) reported that males and females significantly increased their cadence when carrying 40%BW compared to walking unloaded. Although they do not report stride length in their results, walking speed was held constant in their study, which implies the increased cadence observed when carrying load was due to subjects' shortened stride length, as was found by LaFiandra et al. (2003). The conflicting results between studies may be explained by differing experimental protocols as Caron et al. (2015) used a treadmill while Simpson et al. (2012b) did not. This again highlights the potential

differences between treadmill and overground walking and suggests that it may not be appropriate to use a treadmill to simulate marching. Finally, Martin and Nelson (1986) observed that females walked with higher cadence than males across all load conditions, and suggest that this was due to leg length differences.

Therefore, as previous studies have found significant sex-specific differences in gait, and non-linear changes in spatiotemporal parameters as carried load increased, this thesis will directly compare males and females carrying multiple loads.

2.4.2 Kinematic, Kinetic and Energetic

Trunk

Changes in gait kinematics in response to load carriage may serve to limit musculoskeletal forces in the legs (Polcyn et al. 2002). Harman et al. (2000) suggest that increased carried load on the back requires an increase in forward trunk flexion in order to maintain centre of mass over the feet, while Kinoshita (1985) propose that increased trunk flexion facilitates better forward propulsion of the body. Many studies have reported increasing trunk forward flexion in males (Kinoshita 1985; Harman et al. 2000; Majumdar et al. 2010; Attwells et al. 2006; Al-Khabbaz et al. 2008), females (Simpson et al. 2011b; Lidstone et al. 2017; Harman et al. 1999a) and studies of both sexes (Martin and Nelson 1986; Polcyn et al. 2002; Krupenevich et al. 2015) when walking with load. Increased trunk flexion has also been observed in males running with load (Brown et al. 2014). Increasing carried load has also been reported to lead to a decrease in thoracic rotation (LaFiandra et al. 2003) and an increase in anterior-posterior postural sway (Heller et al. 2009). Electromyography of the paraspinal muscles revealed that muscle activity was similar or lower when carrying small loads compared to walking unloaded (Bobet and Norman 1984; Muslim and Nussbaum 2016; Harman et al. 1992), however this was not true for heavier loads ($\sim 50\%BW$) (Muslim and Nussbaum 2016; Harman et al. 1992). Bobet and Norman (1984) propose that a small mass carried on the back acts to counterbalance the natural forward moment on the trunk produced by the head and arms, thus reducing the paraspinal muscles work to counterbalance it. Rectus abdominus muscle activity has also been shown to increase with load (Muslim and Nussbaum 2016; Al-Khabbaz et al. 2008).

Pelvis

Studies reporting kinematic changes in pelvic girdle motion during load carriage are generally in agreement. Walking with body-borne loads leads to increased anterior pelvic tilt, decreased pelvic rotation and decreased pelvic obliquity (Birrell and Haslam 2009; Hyung et al. 2016; Smith et al. 2006; Park et al. 2014; Oberhofer et al. 2018). Only Strube et al. (2017) found no significant changes in pelvic girdle kinematics during load carriage. Park et al. (2014) argue that increased pelvic tilt is expected as is it closely linked with forward trunk flexion. Excessive anterior pelvic tilt caused by trunk flexion has been suggested as a cause of chronic lumbar pain (Smith et al. 2006). LaFiandra et al. (2002) assert that reducing pelvic rotation during load carriage acts to reduce transverse plane torque in the torso, minimising risk of low back pain. However, reduced pelvic rotation can limit stride length (LaFiandra et al. 2003), necessitating an increase in cadence to maintain speed, thus increasing risk of lower limb injury and fatigue (Park et al. 2011). In addition, increased pelvic obliquity has been linked to higher bone stresses in the cortex of the pubic rami (Volinski et al. 2018), and it is therefore possible that obliquity is reduced during load carriage to minimise internal forces in this region.

Hip

Increasing load carriage has been shown to increase hip sagittal plane ROM in males (Kinoshita 1985; Attwells et al. 2006; Polcyn et al. 2002; Park et al. 2014; Loverro et al. 2019; Harman et al. 2000) and females (Ling et al. 2004; Polcyn et al. 2002). This has been attributed mainly to an increase in peak hip flexion, which has also been observed to increase with carried load (Harman et al. 2000; Silder et al. 2013). Increased hip flexion due to load carriage was reported to occur during running by Silder et al. (2015) but not Brown et al. (2014). Few studies have reported hip joint angles in the frontal and transverse planes. Birrell and Haslam (2009) and Park et al. (2014) found that males increased hip abduction-adduction and internal-external rotation as carried load increased during walking. Greater hip abduction has been attributed to wider step-width, which is proposed by Birrell and Haslam (2009) to improve stability. Loverro et al. (2019) observed similar changes and also found that females exhibited greater hip adduction than males. This has implications for injury risk as larger hip adduction angle has been linked to increased prevalence of patellofemoral pain syndrome (PFPS) (Powers 2010).

Changes in hip joint angles are not necessarily reflected in the joint kinetics. Neither Krupenevich et al. (2015) nor Oberhofer et al. (2018) found load carriage to affect hip sagittal plane torque. In contrast, Loverro et al. (2019) found hip flexor, extensor and abductor torques increased with carried load, while Huang and Kuo (2014) observed increases in positive and negative hip joint work. These conflicting results may be explained by differences in study design. The studies which found no differences in sagittal plane torque used carried loads of less than 25kg and asked subjects to walk overground, while those that observed changes in hip torque used a treadmill and heavier carried loads. Harman et al. (2000) found that hip extensor and flexor torques increased in males, but only when loads of over 33kg were carried. When running, Brown et al. (2014) noted that hip flexor torques in males increased with carried load, while Liew et al. (2016) found that hip positive work decreased but hip negative work increased with carried load. Hip joint contact forces were found to increase in males by as much as 30% when walking with 49kg, and in females by as much as 37% when carrying 40kg (Polcyn et al. 2002). Xu et al. (2019) and Xu et al. (2017) calculated the hip joint contact forces for three females walking with 23kg and four females running with 30%BW. They found that peak hip JCF increased by around 39% during walking, and on average 49% during running, increasing from 11BW during unloaded running to 16BW.

Knee

Increased knee flexion is generally found in the literature when walking (Rice et al. 2017; Harman et al. 2000; Majumdar et al. 2010; Simpson et al. 2012b; Silder et al. 2013) and running (Silder et al. 2015) with load. This increase in knee flexion, particularly at weight acceptance, may facilitate better shock absorption (Kinoshita 1985; Harman et al. 2000) as high impact forces are a major risk factor for stress fractures (Knapik et al. 2001; Polcyn et al. 2002). Harman et al. (2000) argue that greater knee flexion helps to lower the body's centre of mass, which is higher due to the additional carried load on the back, thus improving stability. Corresponding increases in quadriceps muscle activity with load have been observed (Harman et al. 1992) which increase the metabolic cost of loaded walking (Hamill et al. 1995). An increase in knee ROM is reported in only some studies (Kinoshita 1985; Attwells et al. 2006), while others report decreasing knee ROM (Polcyn et al. 2002; Ling et al. 2004; Birrell and Haslam 2009; Harman et al. 1999b; Oberhofer et al. 2018). Loverro et al. (2019) found that females altered knee joint kinematics while males altered hip kinematics when walking with load

and suggest that these differences in mechanical adaptations to load may be the cause of females' significantly higher rates of injury. This thesis will therefore investigate the hypothesis that sex-specific adaptation strategies may affect musculoskeletal loading and hence injury risk.

Studies of knee kinetics report increased knee flexor torque when walking (Harman et al. 1999b; Loverro et al. 2019; Xu et al. 2016) and running (Brown et al. 2014), and knee extensor torque during walking (Loverro et al. 2019; Krupenevich et al. 2015). Knee positive joint work also increased when running with load (Liew et al. 2016). Peak tibiofemoral joint contact force was found to increase by 5.6% (Willy et al. 2016) and 10.1% (Id et al. 2018) in males running with 15kg, and by 19.9% in males running with 30kg (Id et al. 2018). Polcyn et al. (2002) reported that knee JCF increased by 35% in females walking with 40kg and 28% when males walked with 49kg compared with unloaded. Although studies of solely females were limited and involved 4 subjects or fewer, knee joint contact force was also found to increase with load when walking (Xu et al. 2019) and running (Xu et al. 2017).

Ankle

With the addition of body-borne loads, the ankle becomes more dorsiflexed (Kinoshita 1985; Majumdar et al. 2010; Attwells et al. 2006; Polcyn et al. 2002; Silder et al. 2013) resulting in an overall increase in ankle ROM (Rice et al. 2017; Polcyn et al. 2002). Kinoshita (1985) assert that increases in ankle dorsiflexion are necessary to facilitate greater flexion at the knee in response to load carriage. Increased muscle activity of the ankle plantar-flexors has been observed in response to the greater propulsion required at toe-off when carrying loads (Simpson et al. 2011a; Silder et al. 2013; Lindner et al. 2012; Harman et al. 1992). Similar kinetic changes were seen in the ankle as in the knee and hip. An increase in peak plantarflexor torque was observed in males (Krupenevich et al. 2015; Rice et al. 2017; Silder et al. 2013) and females (Krupenevich et al. 2015; Silder et al. 2013). Liew et al. (2016) observed increases in ankle positive and negative work in males and females running with load, while Huang and Kuo (2014) reported it during walking with load. They also found that the ankle showed a greater increase in propulsive work when load was added compared with the knee and hip. Polcyn et al. (2002) reported a 34% increase in ankle JCF in females and 27% increase in males when walking with 40kg and 49kg respectively. In their small studies of females, Xu et al. (2019) found between 36-48% increase in ankle JCF

during walking with 23kg, while Xu et al. (2017) found only a 4.8% increase (from 12.6BW to 13.2BW) in females' ankle JCF when running with 30%BW compared to unloaded.

Ground Reaction Forces

Ground reaction forces during load carriage are perhaps the most well investigated kinetic gait parameter. Increases in vertical ground reaction force with load have been observed by the literature (Polcyn et al. 2002; Tilbury-Davis and Hooper 1999; Birrell et al. 2007; Simpson et al. 2012b; Lidstone et al. 2017; Harman et al. 1999a; Silder et al. 2013; Wang et al. 2012; Majumdar et al. 2013; Oberhofer et al. 2018). Some studies found that vertical GRF did not increase linearly with load and suggest that protective gait adaptations were made to reduce musculoskeletal forces in the leg (Harman et al. 2000; Tilbury-Davis and Hooper 1999; Silder et al. 2013; Simpson et al. 2012b). Polcyn et al. (2002) found that mass of the subject and carried load was highly correlated with vertical GRF, accounting for between 88-90% of the variance. They also found that mass accounted for 50% and 58% of the variance in peak braking (posterior) and propulsive (anterior) GRF respectively.

Significant increases in braking and propulsive GRFs were observed in much of the literature, although the rate of increase with load was not as high as that seen in the vertical GRF (Polcyn et al. 2002; Harman et al. 2000; Tilbury-Davis and Hooper 1999; Birrell et al. 2007; Simpson et al. 2012b; Harman et al. 1999a; Oberhofer et al. 2018). Increases in medial-lateral GRF have also been measured as carried load was increased (Harman et al. 2000; Tilbury-Davis and Hooper 1999; Birrell et al. 2007; Simpson et al. 2012b; Oberhofer et al. 2018). Birrell et al. (2007) assert that this increase is due to decreased stability of the body's centre of mass caused by the carried load. During running, peak vertical GRF increased as load increased (Teunissen et al. 2007; Silder et al. 2015; Lobb et al. 2019), but was not significantly different between males and females (Lobb et al. 2019).

2.5 Computational Modelling of the Lower Limb

Computational models can be used to analyse the mechanics of physiological systems and estimate parameters that are difficult to measure *in-vivo*, such as joint contact forces and bone stresses. Traditional assessment of forces in the hip involved loading of cadaveric specimens (Ries et al. 1989; Dakin et al. 2001; Deuel et al. 2009; Silvestri and Ray 2009), or the use of instrumented hip joint implants (Rydell 1966; Davy et al. 1988; Bergmann et al. 2001; Bergmann et al. 2016). Although instrumented implants allow for the measurement of hip joint forces during activities such as walking, the subjects involved are typically not young, healthy or representative of a military population.

Computational modelling provides a non-invasive, cost-effective way to further understanding of the underlying mechanisms involved in hip pathologies, as well as the potential to diagnose conditions at an early stage (Stops et al. 2011). Two computational methods, musculoskeletal (MSk) modelling and finite element analysis (FEA) will be used in this thesis and are discussed here.

2.5.1 Musculoskeletal Modelling

Musculoskeletal models of the lower limb generally employ one of two approaches, forward or inverse dynamics. Forward dynamics models predict lower limb motion based on the inputs of joint torques or muscle activity (Erdemir et al. 2007; Geyer and Herr 2010; Silva et al. 2019; Rajagopal et al. 2016). This muscle activity must be transformed into musculotendon forces using complex muscle contraction dynamics such as those in a Hill-type muscle model (Hill 1938; Zajac and Gordon 1989), or the more complex biophysical model by Huxley (1957). Muscle fibre and tendon properties must also be defined (Zajac 1989). Forward dynamics models can be very complex unless they are highly constrained (Otten 2003). In addition, the high variability in muscle activity signals, especially during dynamic movements, makes it difficult to estimate muscle activation and contraction (Buchanan et al. 2004).

Inverse dynamics approaches have the advantage of using kinematic and external force data, such as ground reaction force, which are easily measured during standard gait analysis (Erdemir et al. 2007). Kinematic gait data is often obtained through optical motion tracking systems (Guerra-filho 2005; Duffell et al. 2014; Dorn et al. 2012; Valencia et al. 2018; Kulmala et al. 2016; Fukui et al.

2016), although tracking of body segment motion with accelerometers has also been proposed (Van den Bogert et al. 1996). Joint torques are derived from the Newtonian equations of motion for each body segment, usually solving recursively from distal to proximal (Winter 2009), although all segments may also be solved simultaneously (Cleather and Bull 2015). Muscular load sharing is then optimised for each time step of the motion by minimising an objective function such that the muscle moments are equal to the derived joint torques about each body segment end (Erdemir et al. 2007). Once muscle moment arms are derived from musculoskeletal anatomy, muscle forces, and the resulting articular forces can then be calculated (Erdemir et al. 2007). Muscle lines of action are typically defined via wrapping points that simulate interactions between muscles and bones (Klein Horsman et al. 2007; De Pieri et al. 2018; Arnold et al. 2000; Garner and Pandy 2000).

One of the difficulties of musculoskeletal modelling is the need to estimate the muscle insertion sites, as well as the mass and inertial properties of each body segment (Buchanan et al. 2004). Anatomical atlases can provide bony geometry and muscle architecture to allow for scaling of generic MSk models (Klein Horsman et al. 2007; Spitzer and Whitlock 1998). Study participants are generally height and weight matched to the nearest subject in the anatomical dataset (Ding et al. 2019). Body segment parameters can also be scaled based on regression equations using anthropometric data of the study participant and dataset subject, and has been shown to vary between males and females (Leva 1996).

Early work by Brand et al. (1982) investigated the sensitivity of hip joint forces to muscle attachment sites. They found that insertion sites were so variable between subjects that scaled geometry could not accurately predict hip joint forces in their study. Similar findings have been more recently reported by Lenaerts et al. (2009). Other studies have investigated the effects of hip joint angle on muscle moment arms (Lengsfeld et al. 1997) and the influence of changing muscle insertion sites on hip joint torques (Pressel and Lengsfeld 1998). Similarly, Röhrle et al. (1984) found that changes to muscle attachment sites greatly influenced the muscle forces calculated by their model. Muscle force predictions have also been found to vary (by as much as eightfold) with muscle cross-sectional area (Brand et al. 1986). Investigations of bony geometry scaling have shown that longer femoral neck lengths and larger femoral neck angles produce higher peak hip joint contact forces (Lenaerts et al. 2008). In addition, Lengsfeld et al. (2000) showed that peak hip JCF significantly increased when the hip joint centre of rotation was

moved 10mm posteriorly, superiorly and laterally.

Crowninshield et al. (1978) implemented a minimisation function relating muscle forces to muscle cross-sectional area, and found hip joint contact forces of between 3.6-5.2BW over the gait cycle. More recent studies have approached the problem of muscle load sharing using optimisation functions which minimise: metabolic energy consumption (Anderson and Pandy 2001; Arnold et al. 2005; Koelewijn et al. 2019), the sum of muscle and joint forces (Röhrle et al. 1984; Seireg and Arvikar 1975), muscle force to cross-sectional area (Brand et al. 1986; Modenese et al. 2011), ligament forces (Collins 1995), and fatigue (Koopman et al. 1995). Although optimisation cost functions have improved over time, accurately simulating muscle co-contraction still presents an inherent difficulty (Stansfield et al. 2003). This is because antagonist muscle activity does not occur to induce motion of the body segment with minimal force or energy etc. but rather acts to stabilise the joint (Stops et al. 2011). A study by Heller et al. (2001) accurately predicted hip joint contact forces while in stance phase of walking, but overestimated the contact force during swing phase by 33% due to inaccurate simulation of muscle co-contraction. Heller et al. (2001) also found that, because their optimisation function was solved iteratively and distal to proximal, errors propagated such that the most proximal joint (the hip) accumulated the largest error.

The musculoskeletal model used in this thesis, Freebody, is described below. Freebody has been used to simulate jumping and weightlifting (Cleather et al. 2013), investigate the effects of functional electrical stimulation of the gluteus medius (Rane and Bull 2016) and biceps femoris (Azmi et al. 2018) on knee joint forces, and to analyse the effects of strength training on lower limb forces in females during drop landings (Czasche et al. 2018).

Freebody

Freebody (Cleather and Bull 2015) is an open source inverse dynamics based musculoskeletal model of the lower limb and pelvis. Freebody includes 57 muscles (made up of a total of 163 muscle elements), including 22 muscles which cross the hip joint. Visualisation of toe-off during a gait trial can be seen in Figure 2.5, where red arrows represent joint contact forces, blue arrows represent the ground reaction force and black lines show muscle lines of action. Freebody uses a segment based approach allowing for 6 degree of freedom (DoF) motion at each joint. This has the advantage of allowing for small translations that can occur in some joints.

However the hip joint was constrained to purely rotational motion (3 DoF) to reduce its sensitivity to skin motion artefact and, because the subjects in this thesis were all healthy and without pathology, pathological translation at the hip was not diagnosed.



Figure 2.5: Freebody visualisation of toe-off with muscle lines of action, joint contact and ground reaction forces

The musculoskeletal geometry was scaled from one of ten (5 male, 5 female) anatomical data sets (Ding et al. 2016). As this thesis was most concerned with the hip, the dataset which could most accurately scale the hip joint was chosen using Equation 2.1. This equation, found by Ding et al. (2019), predicts the root mean square error between scaled and actual subject geometries where ΔLL , ΔM and ΔG are the differences in leg length, mass and gender between the subject and the anatomical dataset respectively.

$$RMSE = 9.6 + 0.38 \times \Delta LL + 0.10 \times \Delta M + 2.64 \times \Delta G \quad (2.1)$$

In Freebody, the problem of load sharing, determining which muscles should be activated and to what extent, is solved by finding the solution which minimises the cost function described in Equation 2.2:

$$J(F_i) = \sum_{i=1}^{163} \left(\frac{F_i}{F_{i,max}} \right)^3 \quad (2.2)$$

Where F is the magnitude of the i^{th} muscle force and $F_{i,max}$ is the maximum force that the i^{th} muscle can exert. Here, $F_{i,max}$ is calculated from the muscle cross sectional area multiplied by the maximum muscle stress, assumed to be $3.139 \times 10^5 Nm^{-2}$.

2.5.2 Finite Element Modelling

Finite Element Modelling (FEM) provides a non-invasive way to simulate bone stresses and strains during dynamic movements *in-silico*, and thus allows for quantification of pelvic bone stress. The analysis involves discretization of anatomical geometry into a finite number of elements where partial differential equations describe the mechanics of each element. Complex geometries and loading conditions, such as those found in the musculoskeletal system, can be solved iteratively (Stops et al. 2011). Typically anatomical geometry is obtained through CT or MRI imaging before being segmented and discretized into a finite element mesh. Material properties are often extrapolated from cadaver studies (Prendergast 1997; Miles and McNamee 1989; Finlay et al. 1986; Ries et al. 1989).

Early Finite Element studies of the hip used two dimensional models to quantify contact pressures in the acetabulum (Brown and Ferguson 1978; Brown and Ferguson 1980; Brown and DiGioia 1984; Rapperport et al. 1985). However these models failed to capture the complex geometry of the bony pelvis and acetabular cartilage (Stops et al. 2011). Much of the resulting literature focused on modelling of femoroacetabular contact and cartilage stresses, with improvements in medical imaging leading to highly detailed subject-specific cartilage analysis (Liu et al. 2016; Anderson et al. 2008; Bachtar et al. 2006; Yoshida et al. 2006; Russell et al. 2006; Speirs et al. 2007; Wang et al. 2007; Wei et al. 2005; Anderson et al. 2010). Wesseling et al. (2019) found that subject specific cartilage geometry affected acetabular contact stress more than subject-specific loading forces, while Liu et al. (2016) found that acetabular stresses were similar in models of patient-specific and constant thickness cartilage.

Finite element models of the pelvis have been used to investigate the effects of, amongst others, hip replacement (Komeili et al. 2017; Maslov et al. 2019; Dong et al. 2018; Borovkov et al. 2018; Cilingir et al. 2007), bone remodelling (Huiskes et al. 2000; Pearson and Lieberman 2004; Tsubota et al. 2009; Zaharie and Phillips 2018), bone density (Leung et al. 2009), high impact loads (Arkusz et al. 2018; Li et al. 2007) and gait (Majumder et al. 2004; Ricci et al. 2018; Volinski et al. 2018;

Dalstra et al. 1995). Dalstra et al. (1995) developed a three dimensional model of a male pelvis (without sacrum), and validated it with cadaver tests. Using that model, muscle and hip joint contact force loading at eight points in the gait cycle was simulated (Dalstra and Huiskes 1995). Muscle forces were applied as surface loads allowing for distribution of the force over the entire muscle attachment site. In their study, the sacro-iliac (SI) boundary was fully constrained while the two os coxae bones met at the pubic symphysis with no soft tissue included. Their analysis showed that stress in the cortical shell was around 50 times higher than in the trabecular bone, with highest stresses occurring in the superior acetabular rim.

A later study by Volinski et al. (2018) modelled the full pelvic ring during gait and included ligaments at the pubic symphysis. Muscle forces calculated by an inverse dynamics based musculoskeletal model were applied over the cortical shell as distributed point loads. They also found that peak von Mises stress occurred in the superior acetabular rim, measuring stresses of 25MPa in a the toe-off position. Ricci et al. (2018) again computationally measured bone stresses in the full pelvis during gait. They found that peak stresses of 31MPa occurred in the inferior pubic ramus, again during toe-off where ground reaction forces were at a peak.

In order to reduce computational expense, many studies have modelled the hemi-pelvis, necessitating the approximation of the sacro-iliac and pubic symphysis boundaries. Both boundaries are often fully constrained in six degrees of freedom (Lamvohee et al. 2014; Kaku et al. 2004; Ravera et al. 2015; Ghosh et al. 2015; Anderson et al. 2005; Kuraria and Kumar 2010). Analysis of stress distributions across full and hemi-pelvis models have shown comparable results (Watson et al. 2017). However work by Watson et al. (2017) and Phillips et al. (2007) showed that stress distribution in the pelvic cortex was more sensitive to the pubic symphysis boundary conditions than the sacro-iliac boundary. They concluded that while it is acceptable to fully constrain the si-joint, the pubic symphysis should be modelled with more physiological accuracy. Ghosh et al. (2015) fully constrained both the sacro-iliac and pubic symphysis boundaries but found that accuracy of stress distributions could still be improved by including acetabular cartilage instead of modelling the acetabulum and femoral head as a perfectly spherical joint.

Finite element analysis of the femur has shown peak stresses occur in the femoral diaphysis (Seo et al. 2014; Diffio Kaze et al. 2017), and the femoral neck (Lotz et al. 1995; Vulovic et al. 2017) during initial and terminal stance. Belaid

et al. (2019) modelled the femur during two-legged stance and observed that compressive and tensile stresses occurred on the inferior and superior femoral neck respectively. They also found that greater knee valgus angle reduced peak stress in the femoral neck while greater varus angle increased it. More recently, work has been done to integrate musculoskeletal and finite element models of the lower limb together (Xu et al. 2016; Ravera et al. 2018; Mo et al. 2018). However there is still limited analysis of pelvic bone stresses during gait in healthy female subjects, and little research into the effect of load carriage on stress distribution in the pelvis. The majority of finite element studies modelled the male pelvis, with literature of female models mainly focused on elderly females with hip pathology (Yang et al. 2018; Keyak et al. 2011; Nishiyama et al. 2014; Noda et al. 2018). This thesis will therefore model pelvic stresses at multiple points in the gait cycle for both a healthy male and female carrying body borne loads.

2.6 Conclusions

The physically demanding nature of military roles means there is an inherently increased risk of injury. Musculoskeletal pathologies are the most common cause of medical discharge and downgrading in the British Armed Forces. Females in the military are reported by much of the literature to be significantly more likely to obtain a musculoskeletal overuse injury, and particularly a stress fracture. Female soldiers and trainees display higher rates of injury than males across all musculoskeletal anatomical sites, and have been found to be as much as 48 times more likely to obtain a hip or pelvic stress fracture. The aetiology of stress fractures is complex and may involve many different factors, some of which are non-modifiable. These factors, combined with inherent differences in the male and female musculoskeletal anatomy and preferred gait patterns may explain sex-specific differences in pathology, however the biomechanical injury mechanisms for hip and pelvic injuries are still not fully understood.

Ground Close Combat roles are characterised by their requirement for carrying heavy, body borne loads, which has been identified as a major risk factor for hip and pelvic MSk injuries. Carrying loads has been shown to alter the gait patterns of healthy soldiers. It is expected that as more females begin training for Ground Close Combat roles, their already increased risk of hip and pelvic pathology will increased even further. There is still, however, limited evidence informing how females adapt their gait in response to heavy body borne loads, and what effect

this has on the internal musculoskeletal forces in the hip. This thesis will therefore investigate changes in spatiotemporal gait parameters under multiple load conditions. Differences in hip and pelvic kinematics between males and females will also be analysed, and musculoskeletal modelling will be used to quantify internal joint loading and muscle forces. Comparison of males and females under the same conditions will allow for analysis of sex-specific adaptation strategies and joint mechanics. Bone stresses in both the male and female pelvis will be modelled at multiple points of the gait cycle during load carriage using Finite Element Analysis. The results of the proposed studies will enable identification of biomechanical risk factors for hip and pelvic injuries. The following chapter uses musculoskeletal modelling to directly compare the kinematic and kinetic aspects of gait in males and females walking with heavy military-relevant loads.

Chapter 3

Biomechanical analysis of gait during load carriage: Effects of sex and load

The review of the literature highlighted that male and female soldiers are likely to obtain different injuries and that there is relatively little research on female soldiers carrying load. Therefore, this chapter presents a biomechanical analysis of gait during military load carriage. Motion capture and musculoskeletal modelling techniques were used to investigate the main and interaction effects of sex and load on gait. Both kinematic and kinetic parameters of gait were analysed and are discussed here.

3.1 Introduction

Today, soldiers in Ground Close Combat (GCC) roles could be expected to carry in excess of 60kg during training and deployment. This load can include equipment, supplies, weapons and body armour. Currently, recruits are expected to undertake training carrying body borne loads multiple times per week (Orr et al. 2015). The majority of previous load carriage studies have only considered male subjects. However with the recent recruitment of females into GCC roles, it is necessary to investigate whether females' bodies respond in the same way to heavy load carriage ($>20\text{kg}$) as their male counterparts. As discussed in Chapter 2, females are more likely to suffer from an overuse injury, possibly due to gait changes occurring during load carriage.

The effects of carrying military loads on spatiotemporal and kinematic gait parameters are well reported. Research has shown an increase in double support and stance time as load increases (Birrell and Haslam 2009, Attwells et al. 2006). It is widely theorised that these changes are an attempt to increase stability and control of the body while also reducing the internal load transferred through the standing leg. Other changes that have been observed with increased load include increased pelvic tilt and trunk flexion, and reduced pelvic rotation (Strube et al. 2017). LaFiandra et al. (2003) suggest that a reduction in pelvic rotation may reduce angular momentum and hence reduce injury risk, as described in Chapter 2.

Similar results have been observed in the few studies of females available in the literature. A study of female hikers found that females increased knee flexion and time in double support with increasing loads of 20, 30 and 40% body weight (Simpson et al. 2012b). Ling et al. (2004) also found that females' hip and knee joint flexion ranges increased. Changes in pelvic girdle motion, namely increased pelvic tilt and decreased rotation, were observed to arise from loads as low as 15% body weight (BW) (Smith et al. 2006).

Few studies have compared the effect of sex in gait analysis of military personnel. Martin and Nelson (1986) compared males' and females' gait carrying between 9kg and 36kg in webbing, a helmet and rucksack. They found that females showed greater spatiotemporal changes than males with loads over 20kg, particularly with respect to stride length and cadence. This has important implications as increased cadence has been suggested to increase over-use injury risk (Nordin and Frankel 2001).

Krupenevich et al. (2015) considered kinematic aspects of gait and found no difference between the hip torques of males and females when carrying 22kg. However further research is necessary to determine if this is also true for heavier loads. Investigation of internal forces in the lower limb is integral to understanding the formation of stress fractures. However few studies have reported hip Joint Contact Forces (JCFs) during load carriage. Bergman and Miller (2001) found that hip JCFs can reach over 250% of body weight during level walking with no load. It is expected that both joint and muscle forces will rise significantly with increasing load and may alter the way force is transferred through the pelvis.

The aim of this study was therefore to compare kinematic and kinetic parameters between males and females during loaded walking. The load conditions of 20kg and 40kg were selected as they are representative of typical military training loads and allow for comparison with much of the literature. The loads were standardised among subjects as is realistic in military training. There is some evidence that the positioning of the external load's centre of mass (COM) can affect gait (Simpson et al. 2012a), where high load positions resulted in less trunk flexion, decreased gastrocnemius activation and higher peak ground reaction forces (GRFs) compared with medium and low load positions. Therefore a rucksack with an evenly distributed load (a central COM) was used mitigate these effects. Although typically soldiers would wear a combination of webbing, rucksack, helmet and weapon on training exercises, this study only included a rucksack to avoid marker occlusion during optical motion tracking measurement.

3.2 Methods

3.2.1 Subjects

Thirty eight healthy volunteers (Table 3.1) with no history of lower limb or back injury took part in this study. Subjects were either current or ex-military personnel, or could complete the minimum military training entry fitness requirements of (a) 2.4km run in less than 14 minutes 58 seconds and (b) 40kg lift and carry for 60m. This study was granted ethical approval by the Joint Research Compliance Office and Imperial College Research Ethics Committee (Appendix A). All subjects gave written informed consent.

An a priori power analysis was conducted using a two-tailed test, medium effect size of 0.5 and alpha of 0.05 for spatiotemporal variables reported by Martin and Nelson (1986). A sample size of at least 16 participants, two groups of size $n=8$, was found to be sufficient to achieve a power of 0.8.

	Males	Females
n	19	19
Age (years)	27.1 (2.0)	28.1 (3.6)
Height (cm)*	178.1 (9.3)	168.9 (7.2)
Mass (kg)*	80.0 (13.8)	65.9 (7.8)
Leg Length (cm)*	92.3 (7.8)	87.1 (7.0)
Bi-iliac width (cm)	25.5 (2.1)	25.3 (1.7)
20kg load as %BW*	25.5 (4.2)	30.9 (4.2)
40kg load as %BW*	51.1 (8.4)	61.7 (8.5)

Table 3.1: Participant Data, mean (SD), * indicates significance between sexes for $p \leq 0.05$ tested using the TukeyHSD test.

3.2.2 Experimental Protocol

Subjects' height (measured with a stadiometer- Seca 213, Seca, Germany) and mass (measured using electronic scales- 9048BK3A, Salter, UK) were taken. Participants were then asked to put on military style boots (Outdoor Pursuits UK LLP, Glasgow, UK). Retroreflective markers were placed on the subjects' boots, legs, pelvis and trunk (Table 3.2). Marker data were captured using a 10 camera optical motion tracking system (Vicon Motion Systems, Oxford, UK) and sampled at 120Hz in line with the literature (Boyer et al. 2008, Wang et al. 2012, Whitcome et al. 2017). Ground reaction force (GRF) data was collected for the right foot

using a force plate (Kistler Instrument Corp., Winterthur, Switzerland) sampled at 1200Hz.

Segment	Marker	Anatomical Location
Trunk	1	Cluster of three markers placed on the thorax
Pelvis	2	Right anterior superior iliac spine
	3	Left anterior superior iliac spine
	4	Right posterior superior iliac spine
	5	Left posterior superior iliac spine
	6	Cluster of three markers placed on the right iliac crest
Thigh	7	Cluster of three markers placed on the left iliac crest
	8	Cluster of three markers placed on the thigh
	9	Lateral femoral epicondyle
Knee	10	Medial femoral epicondyle
	11	Cluster of three markers placed on the shank
Shank	12	Apex of lateral malleolus
Foot (Boot)	13	Apex of medial malleolus
	14	Tuberosity of 5 th metatarsal
	15	Head of 2 nd metatarsal
	16	Calcaneus
	17	Additional marker placed on the foot

Table 3.2: Anatomical locations of reflective markers

First, a static standing trial was captured to obtain the locations of markers with the body in a neutral position (Figure 3.1). Leg length and bi-iliac width were obtained from these marker positions. Subjects then walked over a 5 metre walkway at a self-selected pace carrying:

- (a) No load
- (b) 20kg rucksack
- (c) 40kg rucksack

The rucksack (Sabre 35 Litre, Karrimor, UK) was selected as it is similar to those used during military training (shown in Figure 3.2). It was filled with a mixture of sandbags (RDX Inc., Manchester, UK) and disc weights (Marcy,

Pomona, CA, U.S). The weights were evenly distributed to maintain a consistent CoM in the centre of the rucksack. Participants were instructed to tighten the shoulder and waist straps of the rucksack to a comfortable fit. The order of the load conditions was randomly assigned. A minimum of three trials per load condition were captured and the trial with least marker occlusion was selected.

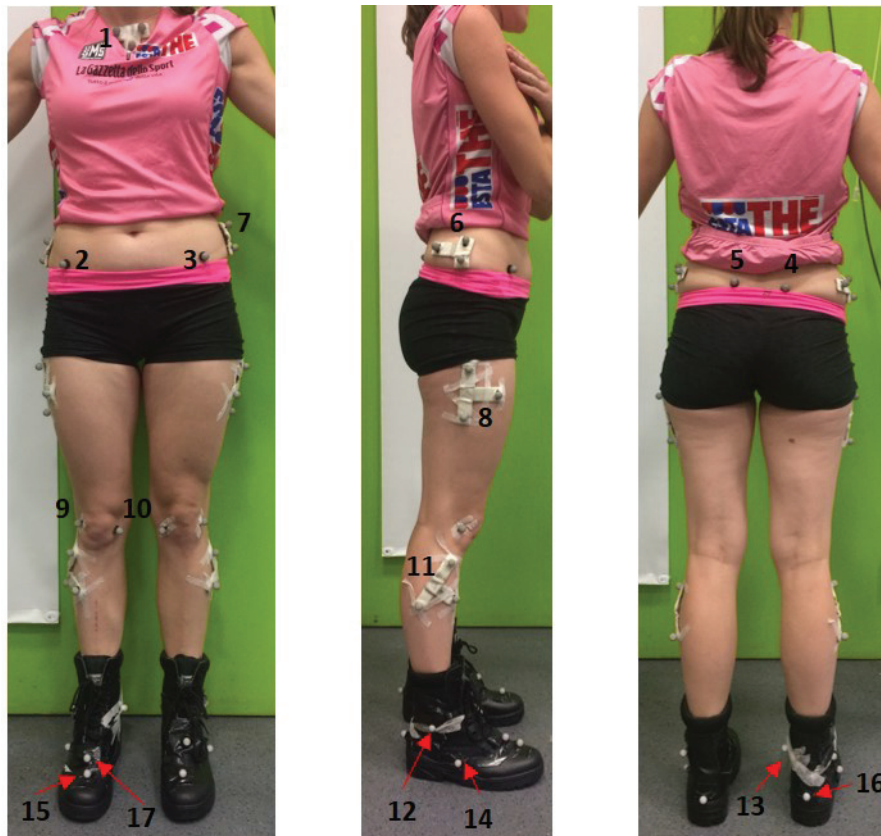


Figure 3.1: Marker positions and neutral pose



Figure 3.2: Rucksack and boots

3.2.3 Data Processing

Data was first processed using Vicon Nexus (v2.6.1, Vicon Motion Systems, Oxford, UK). Marker and GRF data were filtered using a 4Hz low pass 4th order Butterworth filter in Matlab (MathWorks, Natick, USA) as suggested by Winter (2009). GRF data was also down-sampled to 120Hz. All data was then input to the Freebody musculoskeletal model (v2.1, Ding et al. 2016), described in Section 2.5.1, to obtain lower limb joint angles, torques and contact forces. Trunk and pelvic girdle angles were calculated in Matlab from the neutral joint positions measured in the static trial. Pelvic angles were determined in the *rotation-obliquity-tilt* order as suggested by Baker (2001).

3.2.4 Statistics

All data was first tested for normality using the Shapiro-Wilk test ($p > 0.05$). Data was then evaluated using a two-way repeated measures ANOVA to test for the main and interaction effects of load and sex. A one-way repeated measures ANOVA was then used to determine the effects of increasing load on the male and female groups separately. Finally, the Tukey HSD test was performed post-hoc to determine statistical significance of results between each load condition and sex. Results were considered significant where $p \leq 0.05$. Data are presented, unless otherwise specified, as the mean $\pm$ one standard deviation.

3.3 Results

3.3.1 Spatiotemporal

Data was normalised over one gait cycle, from heel strike to heel strike on the right leg. The spatiotemporal parameters measured in this study are presented in Figure 3.3. Significant results for each sex and associated p-values are shown in Table 3.3. No significant differences were found for self-selected walking speed between sexes or load conditions, even when normalised to leg length. Females walked at 1.33m/s when carrying 40kg which is the minimum average speed required to pass the Role Fitness Test (Soldier) for regular GCC personnel (British Army 2019).

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Stride length	-	↓ 0.009	-	-	-	-
Stance phase	↑ 0.03	↑ ≤0.0001	↑ 0.004	-	↑ 0.0002	-
Double support	-	↑ ≤0.0001	↑ 0.04	↑ 0.03	-	↓ 0.02

Table 3.3: Significant changes in spatiotemporal variables with p values. ↑ indicates increase in value between load conditions, ↓ indicates decrease in value between load conditions

Females took significantly shorter strides than males when carrying 40kg ($p=0.01$). Females' stride length was also more affected by the addition of weight than males. Females showed significant reduction in stride length between 0kg and 40kg (9% decrease, $p=0.009$) while males showed no significant changes (as shown in Table 3.3). The results of the ANOVA showed significant effects of load ($p \leq 0.0001$) and the interaction effect of load and sex ($p=0.03$) for stride length.

Results of the Tukey HSD test also revealed that females saw a significant increase in double support between the 20-40kg (9% increase, $p=0.04$) and 0-40kg (19% increase, $p \leq 0.0001$) load conditions (Table 3.3). Males showed an increase then decrease (both by 7%), resulting in no significant change between the 0kg and 40kg load conditions. Females spent significantly longer in double stance than males when carrying 40kg ($p \leq 0.0001$). The effects of load ($p \leq 0.0001$), sex ($p=0.02$) and the interaction ($p \leq 0.0001$) were again significant for double support phase. Females showed significant changes in stance phase between all load conditions (3% increase, $p=0.03$ for 0-20kg, 7% increase, $p \leq 0.0001$ for 0-40kg and

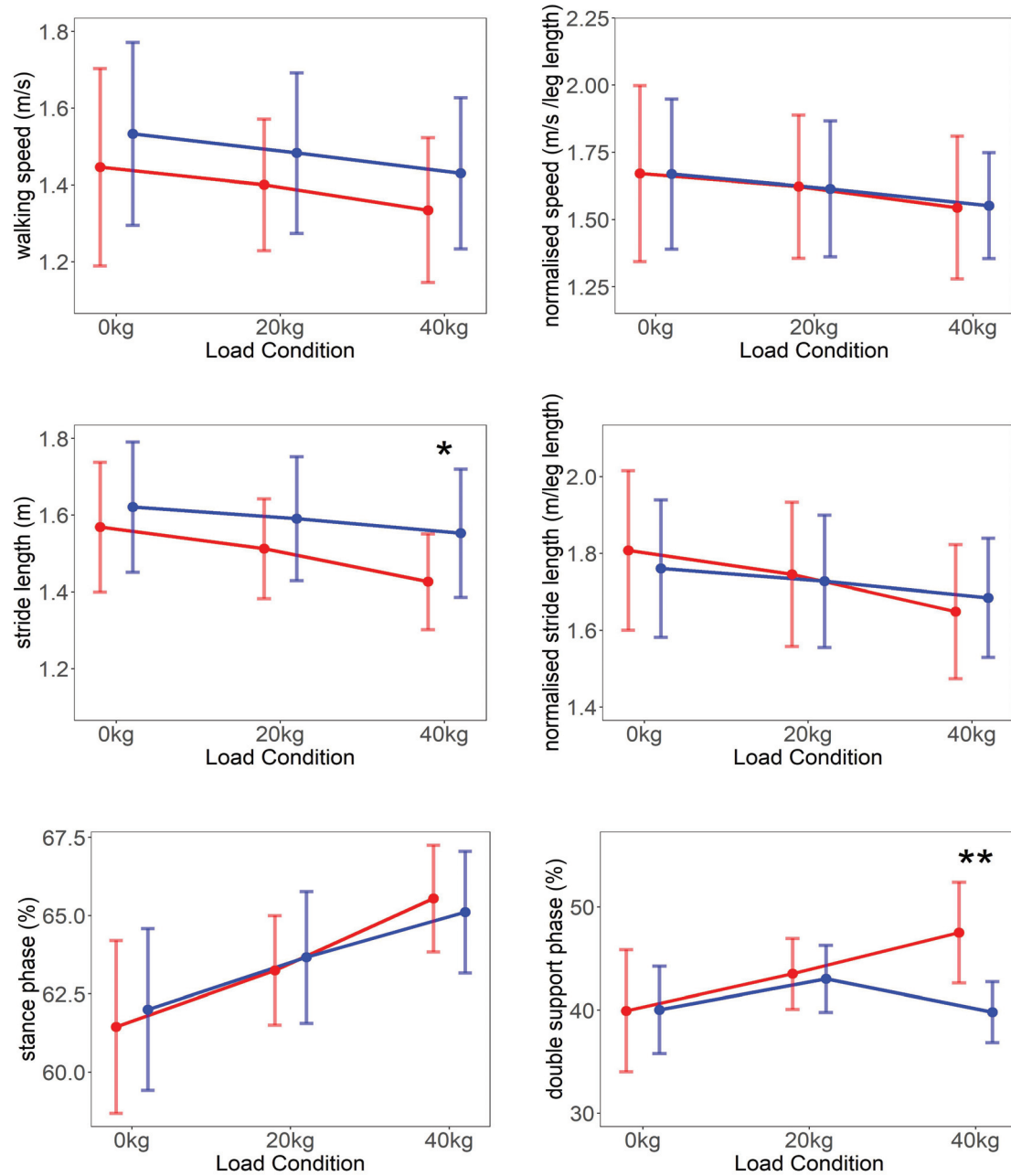


Figure 3.3: Spatiotemporal parameters, * indicates significance between sexes for $p \leq 0.05$, ** indicates significance between sexes for $p \leq 0.005$ red line- female, blue line-male

4% increase, $p=0.004$ for 20-40kg), however, males only exhibited significant changes between 0kg-40kg (5% increase, $p=0.0002$). Only the main effect of load was significant for stance phase ($p \leq 0.0001$).

3.3.2 Joint Angles

Joint angles over one gait cycle are presented in Figure 3.4. Toe off is indicated by a vertical dashed line. A significant sex by load interaction was found for peak trunk flexion angle ($p=0.01$). Both sexes displayed a significant increase in trunk flexion between the 0kg and 20kg load conditions ($p\leq 0.0001$ for both sexes). However females exhibited a greater increase in anterior trunk position from 0-20kg (187% compared with 130% for males) and were more flexed anteriorly when carrying 20kg ($p=0.03$) and 40kg ($p=0.05$) compared with males. Both sexes also displayed a further significant increase in trunk flexion between 20kg and 40kg as shown in Table 3.4 (29% increase, $p=0.008$ for females and 26% increase, $p=0.0002$ for males).

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Trunk Flexion	↑ ≤ 0.0001	↑ ≤ 0.0001	↑ 0.008	↑ ≤ 0.0001	↑ ≤ 0.0001	↑ 0.0002
Rotation	↓ 0.04	-	-	-	-	-
Tilt	↑ ≤ 0.0001	↑ ≤ 0.0001	-	↑ ≤ 0.0001	↑ ≤ 0.0001	↑ 0.02
Hip Flexion	↑ 0.0006	↑ ≤ 0.0001	-	↑ 0.005	↑ ≤ 0.0001	↑ 0.002
Hip Extension	-	↓ 0.003	-	↓ 0.04	↓ 0.0002	-

Table 3.4: Significant changes in peak joint angles with p-values. ↑ indicates increase in peak value between load conditions, ↓ indicates decrease in peak value between load conditions.

Both males and females saw a significant increase in anterior pelvic tilt when carrying the 20kg load compared with the unloaded condition (345% increase for females, 237% increase for males, $p\leq 0.0001$ for both). Males displayed a further increase between 0kg and 40kg (39% increase, $p=0.02$). Finally, a significant decrease in pelvic rotation was also found for females carrying 20kg compared with carrying no load (33% decrease, $p=0.04$). Both sexes demonstrated a significant decrease in rotation range of motion with the addition of load. The ranges of motion for pelvic obliquity showed a trend for decreasing with the addition of the rucksack. The ANOVA revealed that the main effect of load was significant for peak pelvic obliquity ($p=0.0008$) and range of motion ($p=0.04$), the interaction effect was also significant for range of motion ($p=0.0007$). Finally, the only differences between males and females were seen in peak pelvic obliquity ($p=0.02$)

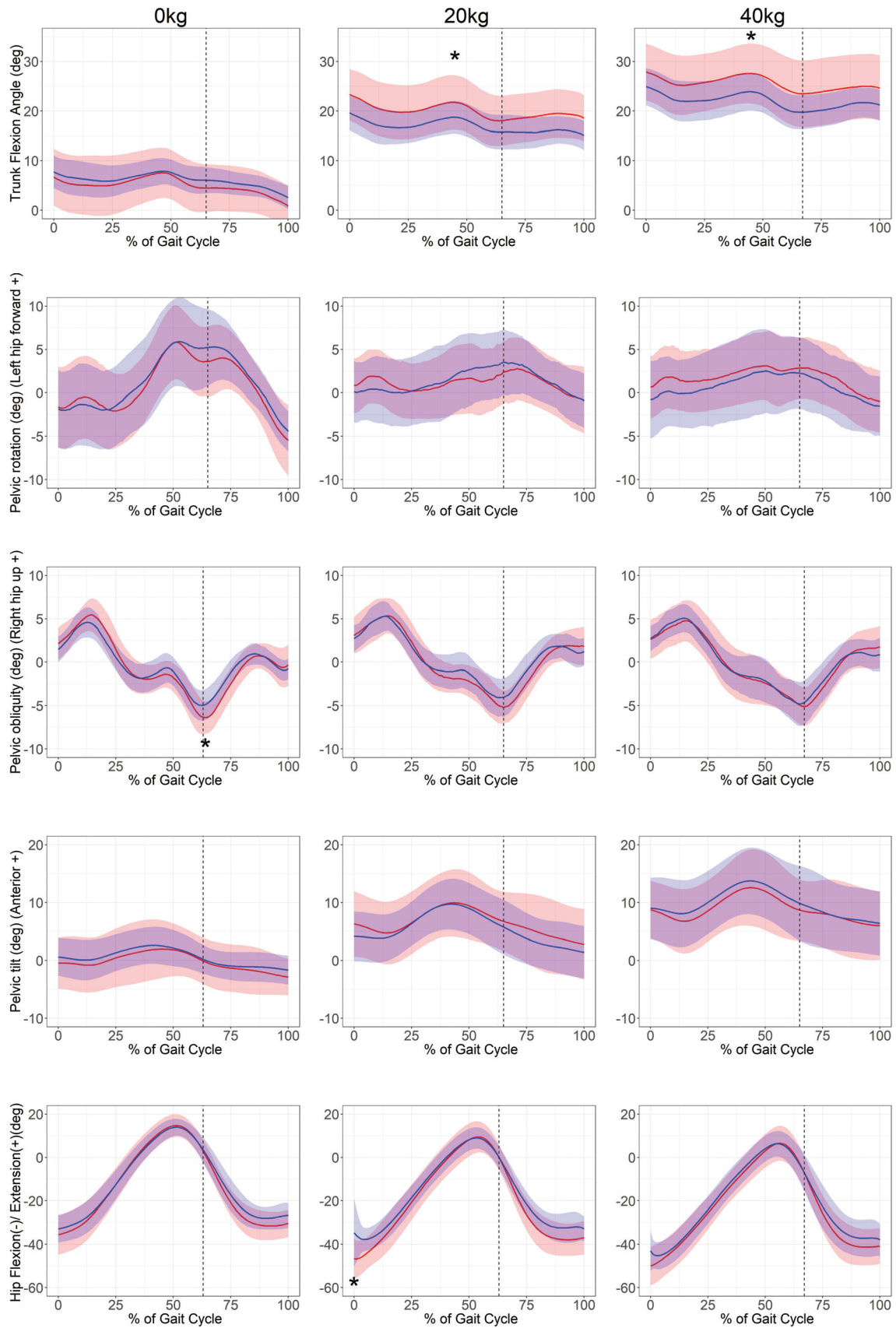


Figure 3.4: Joint angles of the trunk, hip and pelvis by load and sex (red line-females, blue line-males), * indicates significance between sexes for $p \leq 0.05$.

and obliquity range of motion ($p=0.01$) at 0kg, where females showed greater angles and range of motion.

The main effects of load ($p \leq 0.0001$) and sex ($p=0.04$) were significant for peak hip flexion angle. The sex by load interaction term was also significant ($p=0.01$). Females saw a very significant increase between the 0-20kg load conditions (32% increase, $p=0.0006$) but no further increase between 20-40kg (Table 3.4). Males meanwhile saw significant increases between all load conditions. Females were found to have significantly larger peak hip flexion angle when carrying 20kg compared with males ($p=0.001$). There were no differences between sexes for peak hip extension angle, however, both males and females saw a significant decrease in extension between 0-40kg (54% decrease for both males and females). As females did not reduce hip extension when carrying 20kg but increased peak flexion this led to a significant increase in effective range of motion of the joint at 20kg ($p=0.05$) and meant females had a larger range of motion than males ($p=0.006$).

3.3.3 Joint Torques

Hip torques normalised to body mass are presented in Figure 3.5. Significant results are included in Table 3.5. The results of the ANOVA showed the main effect of sex ($p=0.03$) and load ($p \leq 0.0001$) to be significant for the peak hip extensor (anterior) torque. Both males and females saw a significant increase in peak hip extensor torque between 0-20kg, although the increase for females was larger (29%, $p=0.006$ for females compared with 22%, $p=0.01$ for males). While males showed a further significant increase from 20-40kg (17% increase, $p=0.01$), females did not (Table 3.5). Females did, however, have a larger peak extensor torque for the 0kg ($p=0.04$) and 20kg ($p=0.01$) load conditions (Figure 3.5).

In the transverse plane, both sexes demonstrated significant increases in peak internal torque and peak external torque. As shown in Table 3.5, both males and females exhibited a significant increase in peak hip abductor torque, however in males this was not until 40kg was carried. When compared with males, females had significantly lower peak external torque when carrying load ($p=0.05$ for 20kg, $p=0.02$ for 40kg). The effects of load ($p \leq 0.0001$), sex ($p=0.03$) and the interaction ($p=0.04$) were revealed to be significant from the ANOVA for the peak external torque. Load ($p \leq 0.0001$) and the interaction ($p=0.04$) were significant for peak internal torque.

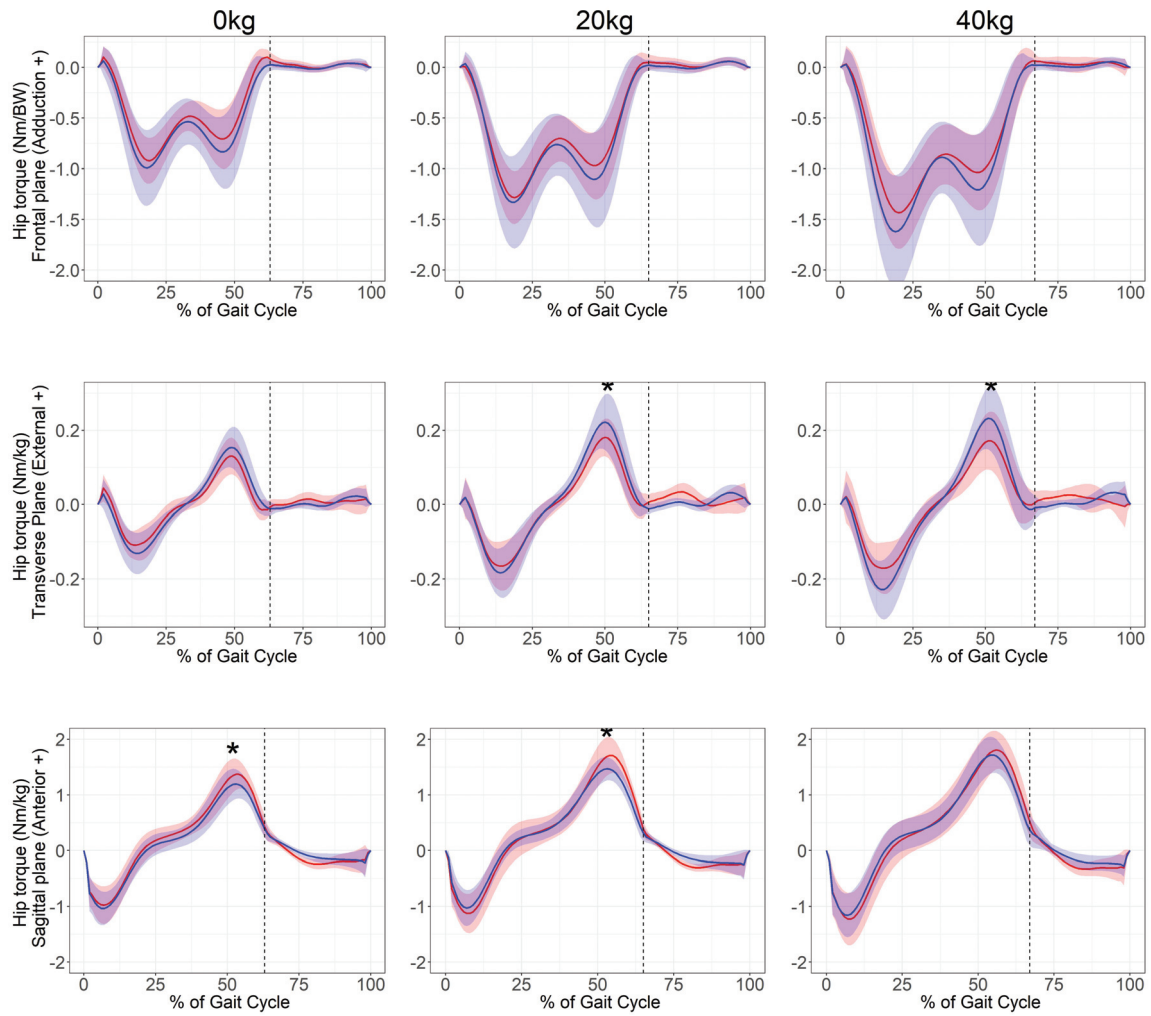


Figure 3.5: Joint torques by load and sex (red line- females, blue line-males), * indicates significance between sexes for $p \leq 0.05$. Dashed vertical line represents toe off.

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Abductor	↑ ≤0.0001	↑ ≤0.0001	-	-	↑ ≤0.0001	-
Internal	↑ 0.002	↑ 0.0002	-	↑ 0.04	↑ ≤0.0001	-
External	↑ 0.04	↑ 0.05	-	↑ 0.01	↑ 0.002	-
Extensor	↑ 0.006	↑ ≤0.0001	-	↑ 0.01	↑ ≤0.0001	↑ 0.01

Table 3.5: Significant changes in peak hip torque with p values. ↑ indicates increase in peak value between load conditions.

Total joint torques for the lower limb are shown in Figure 3.6. The percentage of ankle, knee and hip torque contributions to total lower limb torque were

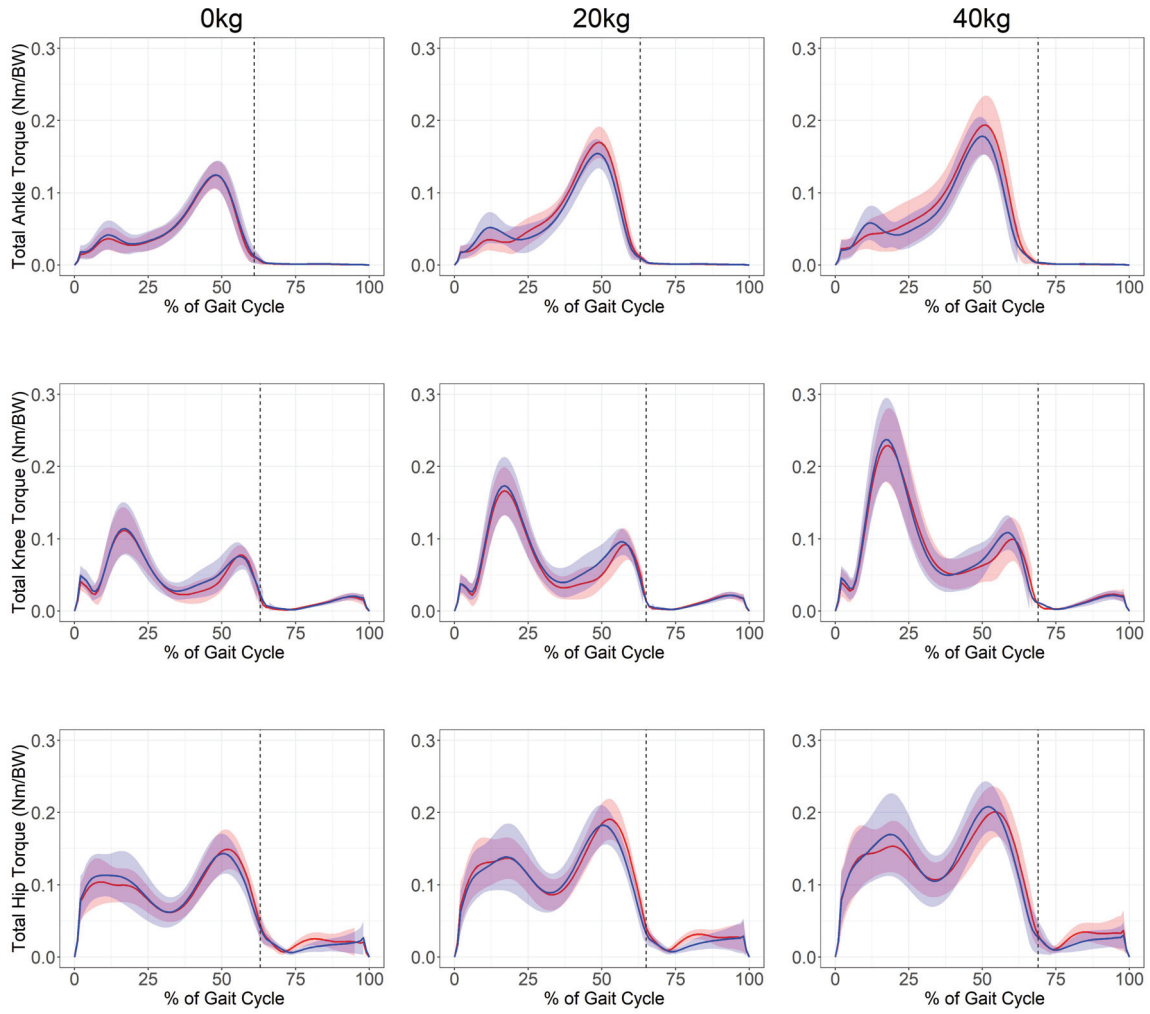


Figure 3.6: Total ankle, knee and hip joint torques by load and sex (red line- females, blue line-males). Dashed vertical line represents toe off.

compared between sexes and load conditions, as shown in Table 3.6. The hip was found to contribute significantly more torque in females than males when carrying no load ($p=0.003$) and 20kg ($p=0.002$). Males meanwhile contributed significantly more torque at the ankle for 0kg ($p=0.01$) and the knee for 20kg ($p=0.001$). Females showed a significant decrease in hip contribution between 20-40kg ($p=0.03$) and 0-40kg ($p=0.009$), and an increase in knee contribution ($p=0.02$) between 0kg and 40kg while males saw no significant changes. The main effects of sex ($p=0.005$) and load ($p=0.0002$) were significant for hip torque contribution. Main and interaction effects were significant for knee contribution ($p=0.009$ for sex, $p=0.004$ for load, $p=0.03$ for interaction).

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Ankle(%)	19(2.8) ^d	21(3.6)	21(4.5)	22(3.1) ^d	20(2.1)	21(3.4)
Knee(%)	32(2.5)	30(2.6) ^D	33(3.7) ^b	33(3.7)	35(4.6) ^D	35(4.2)
Hip(%)	49(3.2) ^D	49(2.5) ^D	46(2.8) ^{Ab}	45(4.1) ^D	45(4.6) ^D	44(4.5)

Table 3.6: Joint torque contributions as a percentage of total lower limb torque, mean (SD). *a* indicates significance from 0kg load condition, *b* indicates significance from 20kg condition, *d* indicates significance between sexes for $p \leq 0.05$. *A,B,D* for $p \leq 0.005$

3.3.4 Muscle Forces

Peak muscle forces normalised to body weight are presented in Table 3.7. Of the 22 muscles in Freebody which cross the hip, only 19 showed significant changes and have been included in Table 3.7. Muscles that showed no changes were the semitendinosus, semimembranosus and biceps femoris.

The main effect of load was found to be significant for all muscles while the main effect of sex was significant for 8 muscles. Significant differences between sexes were found in 10 muscles. For each of these muscles females showed greater normalised force than males. The most significant differences ($p < 0.005$) between sexes were seen in the psoas major and iliacus during the 20kg load condition and in the pectineus for all load conditions.

Significant increases in muscle forces in females were found in 9 muscles between the 0kg and 20kg loads and in 12 muscles between 0kg and 40kg loads. The muscles which showed the most significant increases from the unloaded condition were the rectus femoris (35% increase, $p=0.004$), gluteus maximus (51% increase, $p=0.006$) and the tensor fasciae latae (50% increase, $p=0.003$). No muscles were found to have a significant change in force between the 20kg and 40kg loads for females.

In males, 3 muscles showed increases between the 0kg and 20kg loads and 15 between the 0kg and 40kg loads. The muscles which displayed the most significant increases from the unloaded condition were the gluteus maximus (40% increase, $p=0.02$), gluteus minimus (31% increase, $p=0.04$) and the rectus femoris (25% increase, $p=0.05$). Only one muscle, the gluteus maximus, showed a further significant increase between 20kg and 40kg in males (27% increase, $p=0.02$).

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Rec. fem.	1.33 (0.3)	1.80 (0.4) ^{Ad}	1.99 (0.5) ^A	1.20 (0.4)	1.50 (0.3) ^{ad}	1.77 (0.4) ^A
Glute. max.	0.68 (0.3)	1.03 (0.3) ^A	1.27 (0.4) ^A	0.73 (0.2)	1.02 (0.3) ^a	1.30 (0.4) ^{Ab}
Glute. med.	1.32 (0.4)	1.67 (0.5)	2.01 (0.7) ^A	1.27 (0.4)	1.55 (0.5)	1.82 (0.6) ^A
Glute. min.	0.39 (0.1)	0.54 (0.2) ^a	0.61 (0.2) ^A	0.45 (0.1)	0.59 (0.2) ^a	0.71 (0.2) ^A
Psoas maj.	0.59 (0.2) ^d	0.75 (0.2) ^{aD}	0.78 (0.2) ^a	0.48 (0.2) ^d	0.55 (0.2) ^D	0.65 (0.3) ^a
Illiacus	0.58 (0.2) ^d	0.73 (0.2) ^{aD}	0.77 (0.2) ^a	0.47 (0.2) ^d	0.55 (0.2) ^D	0.66 (0.3)
Sartorius	0.38 (0.1)	0.57 (0.2) ^{ad}	0.60 (0.2) ^A	0.35 (0.2)	0.44 (0.1) ^d	0.54 (0.2) ^A
Add. mag.	0.42 (0.2)	0.48 (0.4)	0.64 (0.2)	0.40 (0.2)	0.52 (0.1)	0.59 (0.3) ^a
Add. brev.	0.07 (0.03)	0.08 (0.04) ^d	0.09 (0.02) ^d	0.05 (0.03)	0.06 (0.02) ^d	0.07 (0.03) ^d
Add. long.	0.41 (0.2)	0.48 (0.2)	0.54 (0.2)	0.32 (0.1)	0.39 (0.1)	0.48 (0.2) ^a
Gracilis	0.08 (0.04) ^d	0.08 (0.05)	0.08 (0.04)	0.06 (0.03) ^d	0.05 (0.03)	0.06 (0.04)
Pectineus	0.09 (0.03) ^D	0.10 (0.05) ^{aD}	0.11 (0.04) ^{AD}	0.05 (0.02) ^D	0.07 (0.03) ^D	0.08 (0.03) ^{aD}
Piriformus	0.07 (0.03)	0.10 (0.03)	0.11 (0.04)	0.07 (0.03)	0.09 (0.03)	0.11 (0.03) ^A
T.f.l.	0.26 (0.1)	0.39 (0.1) ^{Ad}	0.42 (0.1) ^A	0.24 (0.1)	0.31 (0.1) ^d	0.39 (0.1) ^A
Gemellus sup.	0.02 (0.01)	0.03 (0.01)	0.03 (0.01) ^A	0.02 (0.01)	0.02 (0.01)	0.3 (0.01) ^A
Gemellus inf.	0.11 (0.08)	0.12 (0.06) ^d	0.14 (0.06)	0.08 (0.04)	0.09 (0.03) ^d	0.13 (0.05) ^A
Quad. fem.	0.02 (0.01)	0.03 (0.01) ^a	0.03 (0.01) ^A	0.02 (0.01)	0.02 (0.01)	0.03 (0.01) ^a
Obturator ext.	0.3 (0.2) ^d	0.36 (0.3) ^d	0.40 (0.4) ^d	0.19 (0.1) ^d	0.16 (0.1) ^d	0.20 (0.2) ^d
Obturator int.	0.15 (0.1)	0.24 (0.1)	0.28 (0.1) ^a	0.15 (0.1)	0.21 (0.1)	0.27 (0.1) ^A

Table 3.7: Normalised peak muscle forces (N/BW), mean (SD), *a* indicates significance from 0kg load condition, *b* indicates significance from 20kg condition, *d* indicates significance between sexes for $p \leq 0.05$. *A,B,D* for $p \leq 0.005$.

3.3.5 Hip Joint Contact Forces

In both males and females, the addition of load caused significant increases in hip joint contact force (JCF) over the gait cycle as shown in Figure 3.7. Significant results and associated p values for each sex are presented in Table 3.8. Results of the ANOVA revealed the main effect of load to be significant for all components of the hip JCF ($p \leq 0.0001$ for all peak components). In addition, the main effect of sex was found to be significant in the anterior-posterior direction ($p = 0.02$).

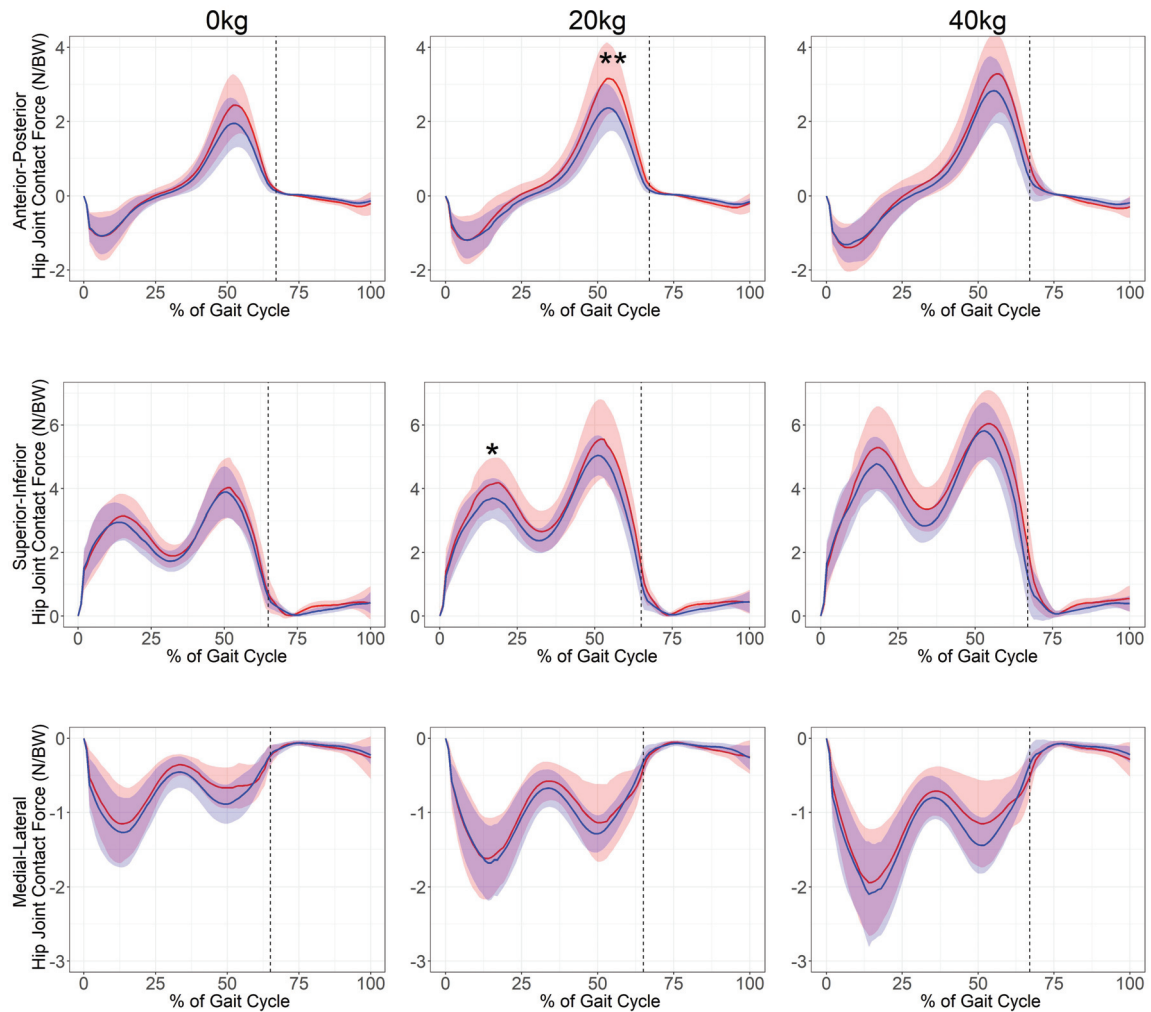


Figure 3.7: Normalised hip joint contact forces (N/BW) by load and sex, red (female) blue (male), * indicates significance between sexes for $p \leq 0.05$, ** for $p \leq 0.005$. Dashed vertical line represents toe off.

Females saw significant changes in 1st trough medial-lateral (39% decrease, $p = 0.04$ for 0-20kg, 63% decrease, $p = 0.0001$ for 0-40kg), 2nd trough medial-lateral (52% decrease, $p = 0.01$ for 0-20kg, 52% decrease, $p \leq 0.0001$ for 0-40kg), 2nd peak superior-inferior (39% increase, $p \leq 0.0001$ for 0-20kg, 49% increase, $p \leq 0.0001$ for

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Anterior-posterior (peak)	↑ 0.04	↑ 0.02	-	-	-	↑ 0.002
Superior-inferior (1st peak)	↑ 0.007	↑ ≤0.0001	↑ 0.008	↑ 0.007	↑ ≤0.0001	↑ ≤0.0001
Superior-inferior (2nd peak)	↑ ≤0.0001	↑ ≤0.0001	-	↑ ≤0.0001	↑ ≤0.0001	↑ 0.004
Medial-lateral (1st trough)	↓ 0.04	↓ 0.001	-	-	-	↓ ≤0.0001
Medial-lateral (2nd trough)	↓ 0.01	↓ ≤0.0001	-	↓ 0.001	↓ ≤0.0001	-

Table 3.8: Significant changes in hip joint contact force with p-values. ↑ indicates increase in peak value between load conditions. ↓ indicates decrease in peak value between load conditions.

0-40kg) and peak anterior-posterior joint forces (31% increase, $p=0.04$ for 0-20kg, 36% increase, $p=0.02$ for 0-40kg) between the 0-20kg and 0-40kg load conditions, but no further change between 20-40kg (Table 3.8). They also exhibited significant increases in superior-inferior (1st peak) force between all load conditions (31% increase, $p=0.0007$ for 0-20kg, 62% increase, $p\leq 0.0001$ for 0-40kg, 24% increase, $p=0.008$ for 20-40kg).

In contrast, males only displayed significant changes in force between 20-40kg for anterior-posterior (20% increase, $p=0.002$) and 1st trough medial-lateral components (24% decrease, $p\leq 0.0001$). Very significant increases in force were found between all peak loads in the superior-inferior direction (First peak: 21% increase, $p=0.0007$ for 0-20kg, 54% increase, $p\leq 0.0001$ for 0-40kg, 26% increase, $p\leq 0.0001$ for 20-40kg, Second peak: 29% increase, $p\leq 0.0001$ for 0-20kg, 51% increase, $p\leq 0.0001$ for 0-40kg, 17% increase, $p=0.004$ for 20-40kg). As with females, only significant differences between 0-20kg (41% decrease, $p=0.001$) and 0-40kg (60% decrease, $p\leq 0.0001$) were observed in the 2nd peak of the medial-lateral force component.

As shown in Figure 3.7, compared with males, females displayed significantly greater 2nd peak force in the anterior-posterior direction ($p=0.003$) and 1st peak

force in the superior-inferior direction ($p=0.04$) when carrying 20kg. Females also had a trend for higher peak force ($p=0.06$) in the anterior-posterior direction when walking with no load. No differences were found in normalised JCF magnitudes between sexes when carrying 40kg.

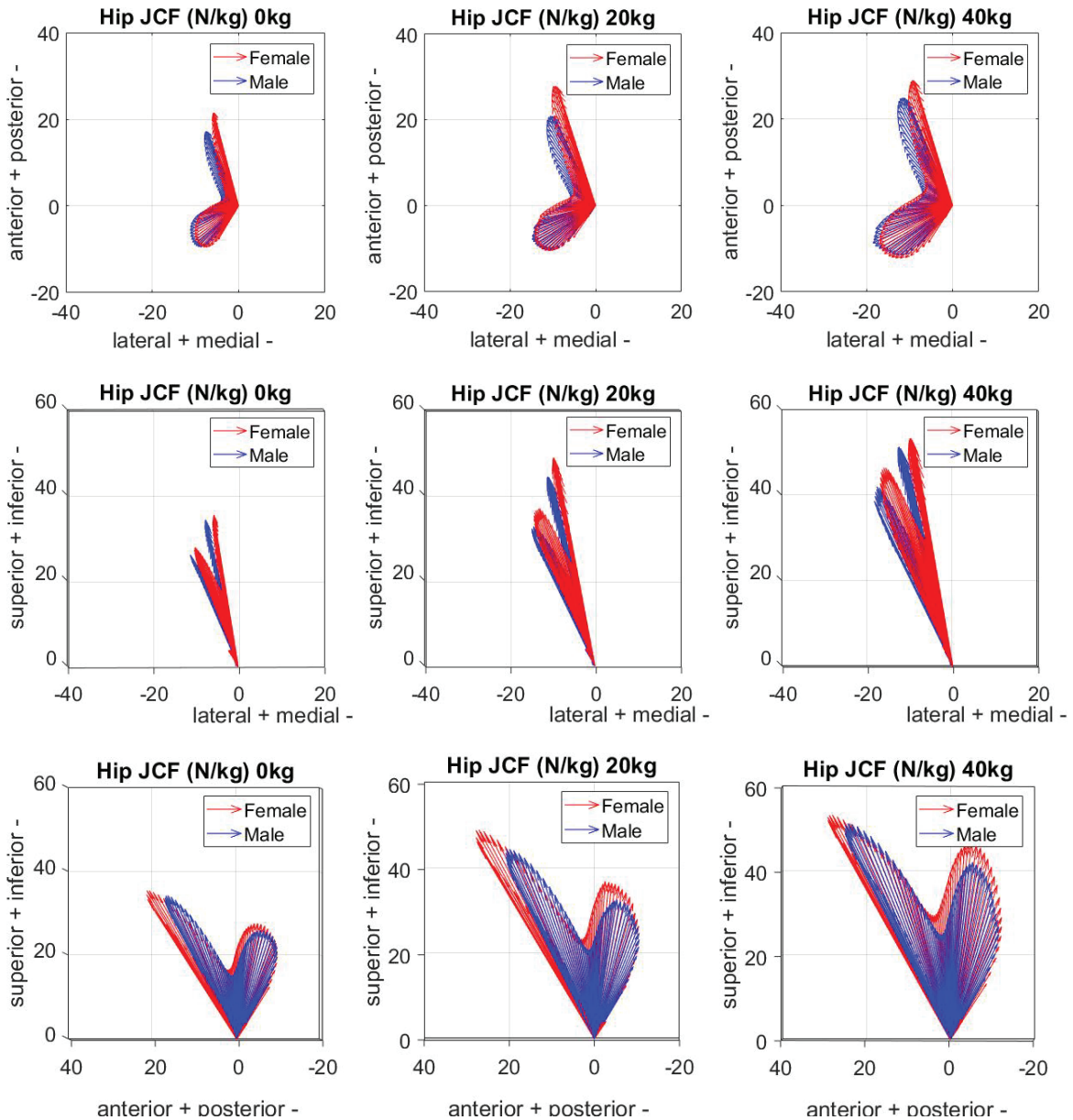


Figure 3.8: Normalised hip joint contact forces by load and sex, coordinates (0,0) represent hip joint centre

Vector components of the hip joint contact force are shown in Figure 3.8 and Table 3.9. Here, each vector arrow corresponds to the direction and magnitude of the hip JCF for each percentage of the normalised gait cycle (0-100%). A diagram defining the vector angles is shown in Figure 3.9. In the transverse plane the main

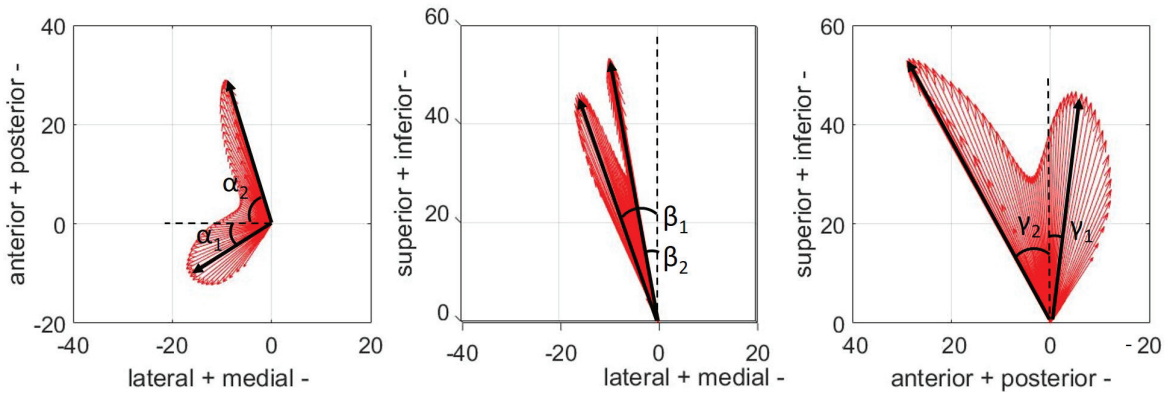


Figure 3.9: Hip JCF vector angle diagram (deg)

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
α_1	27.5 (18)	19.9 (18)	16.5 (13)	29.8 (12)	21.8 (13)	19.6 (13) ^a
α_2	72.2 (10) ^d	73.4 (9) ^d	72.6 (7) ^d	64.1 (12) ^d	65.0 (12) ^d	65.2 (11) ^d
β_1	18.9 (7)	18.9 (7)	18.8 (7)	22.2 (6)	22.2 (6)	21.9 (6)
β_2	10.1 (4) ^d	10.1 (4) ^d	10.8 (4)	13.0 (4) ^d	12.8 (5) ^d	12.9 (4)
γ_1	12.6 (11)	8.3 (9)	6.9 (7)	13.6 (7)	9.4 (7)	8.3 (7) ^a
γ_2	30.7 (8)	31.8 (7) ^d	32.1 (7) ^d	26.6 (6)	27.2 (6) ^d	27.4 (6) ^d

Table 3.9: Hip joint contact force vector angles, *a* indicates significance from 0kg load condition, *b* indicates significance from 20kg condition, *d* indicates significance between sexes for $p \leq 0.05$.

effect of load and was significant for α_1 ($p \leq 0.0001$). Sex was a significant effect for α_2 ($p = 0.05$). Females saw no change in this plane, however males saw a change in α_1 ($p = 0.04$). Females had a significantly larger α_2 for all load conditions ($p = 0.03, 0.02$ and 0.02 respectively) compared with males.

In the frontal plane the interaction effect showed a trend for β_1 ($p = 0.06$). There were also significant differences between sexes, namely in β_2 at 0kg ($p = 0.03$) and 20kg ($p = 0.04$). In the sagittal plane the main effects of load ($p \leq 0.0001$ for γ_1 , $p = 0.02$ for γ_2) and sex ($p = 0.05$ for γ_2) were significant. Males saw a significant increase in γ_1 from 0-40kg ($p = 0.05$). In addition, there were significant differences between sexes for γ_2 when carrying 20kg ($p = 0.04$) and 40kg ($p = 0.03$).

3.3.6 Key Findings

In summary, the key findings of this study were:

1. Males and females walked with similar gait patterns when carrying 0kg and 20kg. At heavier loads differences in stride length and double stance occurred.

2. Females were significantly more flexed anteriorly when carrying load, females increased hip range of motion to maintain stride length at 20kg.
3. Females generated a significantly higher proportion of torque at the hip before they altered their gait compared with males. Males generated more torque at the ankle and knee for lighter loads.
4. Females increased force in more muscles between 0-20kg than males, and then increased force in fewer muscles once their gait was altered.
5. Females consistently had a more anteriorly and superiorly angled hip JCF than males.

3.4 Discussion

Spatiotemporal effects

The results of this study show that males and females exhibit similar walking patterns when carrying up to 20kg, and that both sexes choose to walk at similar speeds regardless of leg length. However differences emerge between sexes when heavier loads are carried, namely that females take shorter strides and spend longer in double support. Increasing load has a greater effect on females' gait parameters, with females showing more significant changes than males, even when normalised to leg length, suggesting it is not just a function of anthropometric differences. This is in agreement with Martin and Nelson (1986) who also found significant differences in gait parameters of males and females when carrying 36kg. This shortening of stride lengths has implications as females will need to take more steps thereby increasing cadence and the possibility of overuse injuries in the hip due to repetitive stresses.

Similarly, increasing time spent in double support has often been suggested as a strategy to reduce the internal loads of the supporting leg by minimising the time that the load is transferred through a single leg and increasing stability and support over the gait cycle. Birrell and Haslam (2009) found that male soldiers increased double support between 0-32kg, however results of this study showed a significant decrease at 40kg. It could be that at heavier loads (over 32kg) males adapt to a different method for dealing with load carriage. Birrell and Haslam (2009) also suggested that the increase in double support is a result of the increase in stance phase, however the results of this study do not seem to corroborate this as no significant differences in stance phase between sexes were seen. In addition,

males showed significant changes in either stance phase or double support but never both.

Krupenevich et al. (2015) found no differences between males' and females' spatiotemporal parameters when carrying 22kg. They concluded that the lack of gait differences with low weights was itself a cause of injury because females were not altering their gait to account for their smaller mass and strength. By considering heavier weights however, the current study suggests that females carrying 20kg are more likely to be injured because they are working closer to their maximum threshold where they can no longer maintain natural gait patterns. At some weight between 20-40kg females will change their gait, possibly because they lack the required muscle strength or are attempting to reduce injury risk by reducing internal forces in the leg.

Kinematic effects

It has been suggested that reduction of pelvic rotation has the effect of decreasing stride length to reduce upper body torque (LaFiandra et al. 2003). Males in this study however did not show a reduction in stride length corresponding to their reduced pelvic rotation, possibly because they had been trained to walk at a set stride length. In addition there were no differences between pelvic rotation of males and females for any load condition in this study but there were differences in stride length. This could be due to differences in leg length between sexes. It seems as though males and females adopt different techniques when carrying 20kg compared with no load. Females reduce pelvic rotation but increase hip range of motion in the sagittal plane to maintain stride length while males maintain pelvic rotation and hip range of motion.

Interestingly, females did not see a significant reduction in pelvic rotation or hip range of motion corresponding to the significant reduction in stride length between 20-40kg, suggesting perhaps that they were unable to further change these parameters due to the rucksack and so used a different method to reduce stride length. Males meanwhile did demonstrate a further significant increase in hip flexion when carrying 40kg, allowing them to maintain a constant stride length across all load conditions. For 40kg the mean female pelvic rotation was positive (left hip forward) over the majority of the gait cycle, possibly due to strength imbalance between limbs or because the rucksack was affecting the motion of the pelvis.

There was no significant change in pelvic obliquity peak or range of motion as load increased, in contrast to the findings of previous studies (Smith et al. 2006). Smith et al. (2006) suggest that the addition of the backpack constrains the motion of the pelvis and although not significant, both sexes here had a trend for a small reduction in pelvic obliquity. The significant difference between males' and females' peak obliquity and range of motion at 0kg may be due to sex-specific differences in hip anatomy, namely more anteriorly angled femurs in females (Cho et al. 2004; Frimenko and Whitehead 2014).

Increased trunk flexion and pelvic tilt are necessary to stabilize the body's CoM when carrying load (Attwells et al. 2006). Pelvic tilt increased with addition of load for both sexes, in agreement with the literature (Birrell and Haslam 2009; Smith et al. 2006). Males saw a further significant increase between 20-40kg while females did not. This is possibly because they were reaching their limit of comfortable forward flexion. An increase in pelvic tilt has been suggested to increase lordosis of the spine resulting in compression of the lumbar vertebra and possibly lower back pain or injury (Birrell and Haslam 2009). Females flexed significantly more anteriorly than males when carrying load, possibly due to their lesser upper body mass and strength, which suggests that they may be more at risk of a back injury. In addition, females showed increased trunk flexion but not increased pelvic tilt when carrying 40kg, suggesting that the further forward bending is occurring in the spine and not the pelvis.

Kinetic effects

This study found that both sexes significantly increased hip extensor torque with the addition of load. This is in contrast to Harman et al. (2000), who found no change in hip extensor torque in males until over 33kg was carried. Few studies have compared hip torques between sexes when carrying loads. Krupenevich et al. (2015) found no differences between hip extensor torques of males and females carrying 22kg. They suggest it may be because the load was not heavy enough. However this study found that differences in extensor torques occurred in lower loads, before females altered their gait. This is also in contrast with Silder et al. (2013) who found no significant differences between males and females carrying relative loads of up to 30%BW, although direct comparison with their study is difficult as they did not use standardised loads. Females reduced torque contribution at the hip as load increased, possibly to reduce risk of injury or due to fatigue. It was also found that females generated a significantly higher proportion of torque at the hip, while males generated more at the knee and ankle.

This suggests that females may be naturally more hip dominant while males may be more knee dominant.

Females were found to increase the force of more muscles between 0-20kg (9 compared with 3 for males). Males increased the force in 15 muscles between 0-40kg, compared with 12 for females. In addition, significant differences between sexes were found mainly for the 20kg load condition where females had not yet altered their gait. This suggests that females were working harder to maintain normal gait function when carrying 20kg, and that by altering their gait when carrying 40kg they prevented any further increases in muscle force. Once females changed their gait (at 40kg) there were fewer differences between males and females, only significant in a few of the adductor muscles.

Correa et al. (2010) found that iliopsoas muscles contribute significantly to hip joint contact force, in particular the anterior component. This has implications as females had significantly higher iliopsoas forces for 0kg and 20kg. It is possible this was due to females' increased hip and trunk range of motion. Correa et al. (2010) also found that the gluteus medius and maximus have a significant effect on hip JCF in the superior direction at weight acceptance. Although there were no significant differences between sexes, females saw a more significant increase in gluteus maximus force between 0-20kg and 0-40kg, and gluteus medius force between 0-20kg. Finally, no differences were seen in the muscle force timings between males and females, this is in agreement with Silder et al. (2013) who found no significant differences in muscle activation patterns between sexes when carrying 30%BW.

Following their larger iliopsoas forces, females were found to have a higher normalised anterior peak hip JCF. This was significant at 20kg where females were more flexed forward, had larger hip range of motion and a larger extensor torque compared with males, which was also found in the literature (Lewis et al. 2010). Their significantly higher normalised superior JCF at weight acceptance was possibly because females were carrying a greater percentage of their body mass, and that they had not yet adjusted their stride length and gait to absorb the additional load. This suggests that females carrying loads may have a higher normalised hip JCF when walking with a similar gait to males. There was no difference in normalised hip JCF magnitude between sexes for the 40kg load conditions possibly because females were altering their spatiotemporal parameters and shifting torque production to the knee. This meant they had less increased

muscle force around the hip, resulting in no change in normalised hip JCF. However there were still sex-specific differences in direction of the JCF (in the anterior-superior direction). Females saw no change in JCF direction with the addition of load so it is possible that these sex-specific differences may be partly due to differences in anatomy and that this could affect the way force is transferred through the femur and bony pelvis. The effects of hip JCF magnitude and direction on femoral and pelvic stresses will be investigated in Chapter 4 using Finite Element Analysis.

Limitations

There are several limitations of this study. Firstly, kinematic data was collected over a flat walkway and short distance which is not representative of outdoor terrain. It is possible that altered gait patterns would emerge when carrying lighter weights if longer distances were considered. The results of this study show the gait changes occurring due to increasing carried load and not due to endurance fatigue. Loaded marches often include carrying a helmet, webbing and weapon; these were not included in this study to minimise marker occlusion. It is possible that the addition of this equipment would act to counterbalance the mass of the rucksack and therefore forward trunk flexion may be less than that measured in this study. The addition of the rucksack meant that some pelvic markers were lost and had to be reconstructed. It also meant that no markers could be placed on the spine so the trunk was modelled as a single segment preventing lumbar lordosis from being measured. Foot markers were attached to the military boots instead of bare feet which produced an error in foot kinematics and an offset of ground contact due to the thick sole. As with all motion capture studies, skin motion artefact can also affect kinematic measurements. Measurements of internal/external rotation of the thigh and shank are most susceptible to skin motion artefact (presented in Appendices B and D) and thus the measured angles will likely be larger than the actual ones. In the Freebody model subject specific geometry could not be used, resulting in a scaling error and as subjects in this study were extremely physically fit, it is probable that they had larger muscle volumes than the dataset subjects they were scaled from.

3.5 Conclusions

The results of this study found that males and females exhibited similar gait patterns when carrying up to 20kg, however females significantly altered their

stride length and double support phase when carrying heavier loads. In order to maintain normal stride length when carrying 20kg females reduced their pelvic rotation and increased hip range of motion in the sagittal plane while males did not. Females also saw an increased force in significantly more muscles suggesting that they were working harder to maintain normal gait function. Analysis of joint torque production showed that females generated a greater percentage of torque at the hip at 0kg and 20kg compared with males, suggesting that females may be naturally more hip dominant. It is thought that these factors led to females having higher normalised hip JCFs at 0kg and 20kg. When carrying 40kg females altered their gait to reduce internal forces in the leg either in an effort to reduce injury or because of fatigue. It is expected that the sex-specific differences in hip force direction are partly due to pelvic geometry and may affect the way forces are transferred through the hip joint and pelvis, which could increase risk of stress fractures although more research is needed. This study concludes that females are more likely to be injured because they are working at closer to their threshold, the maximum load where they can maintain normal gait function. It may therefore not be appropriate to have males and females walking in step when carrying heavier loads. The findings also support the suggestion that males and females may require different strength training.

In order to determine if the biomechanical differences found in this study due to load and sex could relate to increased stress fracture incidence, the next chapter will investigate the effects of joint contact and muscle forces on stress distribution through the hip and bony pelvis.

Chapter 4

Finite element analysis of the hemi-pelvis: An investigation into the effect of load carriage on stress distribution during walking

The results of Chapter 3 showed that both sex and carried load significantly affected muscle and hip joint force magnitude and direction. In order to determine the effect of these changing forces on stress distribution in the bony pelvis, two subject specific finite element models were built, one male and one female. This chapter presents the development and validation of these models, as well as the resulting stress distribution at key points in the gait cycle.

4.1 Introduction

During physical activity the pelvis is loaded by muscle and body forces which result in stress and deformation of the bone (Bennell et al. 1999). Studies have shown that, during gait, cortical bone undergoes compressive stresses at heel strike and high tensile stresses at toe-off (Carter 1978). Stress fractures develop from micro-damage caused by repetitive strain which occurs faster than the bone can remodel. The aetiology of stress fractures is complex and involves many contributing factors (Bennell et al. 1999). Intrinsic risks include low bone mineral density, smaller body size, Caucasian or Hispanic race, and female sex (Springer and Ross 2013; Mattila et al. 2007; Rauh et al. 2006; Beck et al. 2000). Other variables including lower limb flexibility, as well as hormonal and nutritional factors, may also play a part (Bennell et al. 1999). Extrinsic risks for stress fractures include lower fitness and strength, poor fitting rucksack, smoking, and gait changes (Robinson et al. 2016; Lappe et al. 2001; Ling et al. 2004; Bergman and Miller 2001).

Females in the military have been found by many studies to carry a significantly higher risk of bone injury than males (Niva et al. 2009; Mattila et al. 2007; Ministry of Defence 2016). A systematic review of 11 studies by Wentz et al. (2011) found that the incidence of stress fractures for females in the military was 9% while a review by the British MoD (Ministry of Defence 2016) found that female recruits were 3 times more likely to obtain a stress fracture compared to males, and on average 10 times more likely to obtain a stress fracture at the hip. As discussed in the review of the literature, the most common stress fracture sites in the pelvis are the superior and inferior pubic rami (Beck et al. 2000; Pope 1999).

Finite Element (FE) modelling provides an inexpensive method of estimating stress in the bony pelvis, allowing for modelling of *in-vivo* load conditions and without the limitations of cadaveric studies. The advantages of FE analysis include the ability to model stress distributions over the entire pelvic cortex, as well as to test multiple load conditions on the same bony geometry. Previous FE studies have evaluated stress in either the full pelvis and sacrum (Leung et al. 2009) or a hemipelvis (Anderson et al. 2005; Ghosh et al. 2015), with comparable results. Others have studied the effects of fixed versus ligamentous boundary conditions (Phillips et al. 2007; Watson et al. 2017). They found that fixed boundary conditions led to altered displacement and higher stresses in the cortex. More recently several studies have looked at the effect of gait on the stress in the pelvis during unloaded walking

(Volinski et al. 2018; Ravera et al. 2015; Ricci et al. 2018) and on the tibia during walking with load (Xu et al. 2019). The hypothesis of the current study was that differences in hip joint angles, muscle and joint contact forces between males and females affect stress distribution in the pubic rami and may lead to increased peak stresses. The aim was therefore to develop two subject-specific FE models of the hemi-pelvis with physiological forces applied at multiple points over the gait cycle, in order to quantify stress in the bony pelvis and identify potential fracture sites for an average male and female subject carrying 0, 20 and 40kg.

4.2 Methods

4.2.1 Geometries

Of the 38 subjects who participated in the previous study (Chapter 3), one male and one female subject, with peak hip JCFs closest to the mean of their sex, were selected (Table 4.1). Magnetic resonance imaging (MRI) scans of the full lower limbs and pelvis were collected for both subjects, from the top of the iliac crest to the ankle joint centre. The imaging was granted ethical approval by the Imperial College Joint Research Compliance Office and NRES Committee. All subjects provided written informed consent. Images were recorded using a 3T Siemens Verio MRI scanner (Siemens AG, Erlangen, Germany). A 3D T1-weighted GRE (gradient echo) sequence was used together with two body coils while the subjects were in the supine position. MRI settings were as follows: matrix= 320 x 260, slice thickness= 1mm, repetition time= 11.6ms, echo time= 4.28ms, pixel spacing= 1.4 x 1.4mm.

	Male	Female
<i>n</i>	1	1
Age (years)	24	27
Height (cm)	182	173
Mass (kg)	69	64
Leg Length (cm)	96	85
Bi-iliac width (cm)	23.8	23.3
20kg load as %BW	29	31
40kg load as %BW	58	63

Table 4.1: Participant Data

The images were manually segmented and smoothed using Mimics Research software (v19.0, Materialise, Leuven, Belgium) to obtain 3D geometry of the right

bony pelvis and proximal femur, as well as muscle attachment sites (shown in Figures 4.2 and 4.3). Muscle attachment sites were determined by tracing the 2D contour of each muscle in the transverse plane image slices until they could no longer be distinguished (Figure 4.1). These 2D contours were then reconstructed to obtain 3D volumes of the proximal end of each muscle. The number of attachment points per muscle was defined by the musculoskeletal model (Ding et al. 2019). As this study was concerned with the local stresses in the pubic rami, rather than global stress distribution throughout the pelvic ring, only the hemi-pelvis was considered in order to reduce computation time.

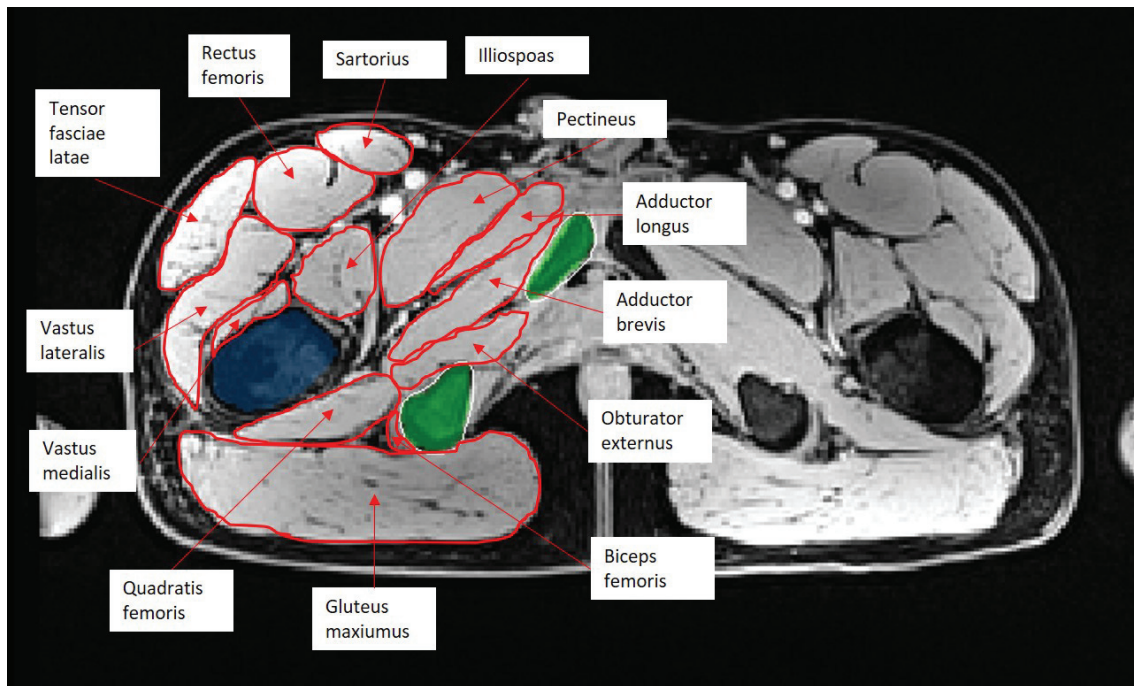


Figure 4.1: 2D contours of hip musculature defined in a transverse plane image slice. Green area- bony pelvis, blue area- proximal femur.

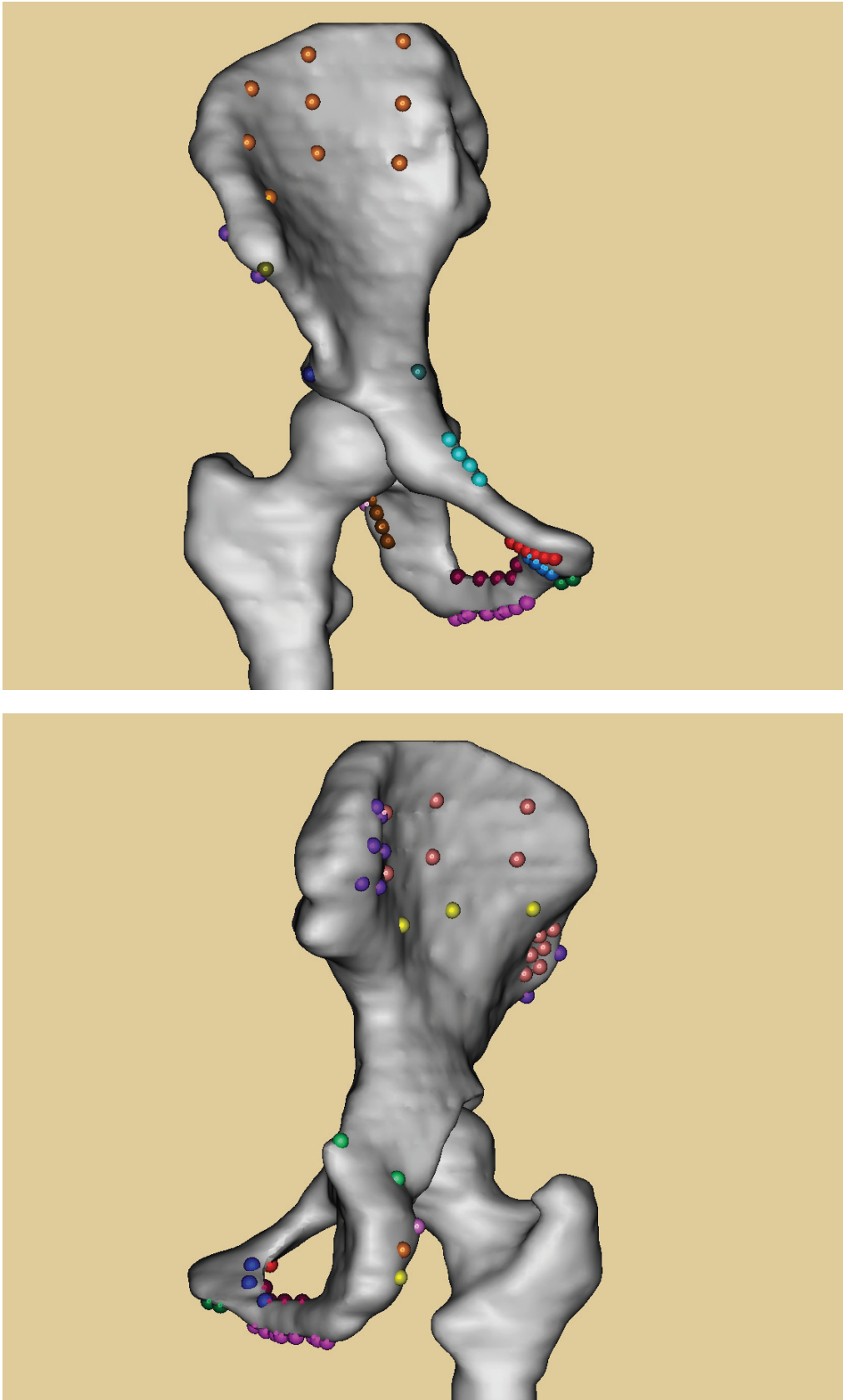


Figure 4.2: 3D geometry and muscle attachment sites for the female model. Top figure- anterior view, Bottom figure- posterior view

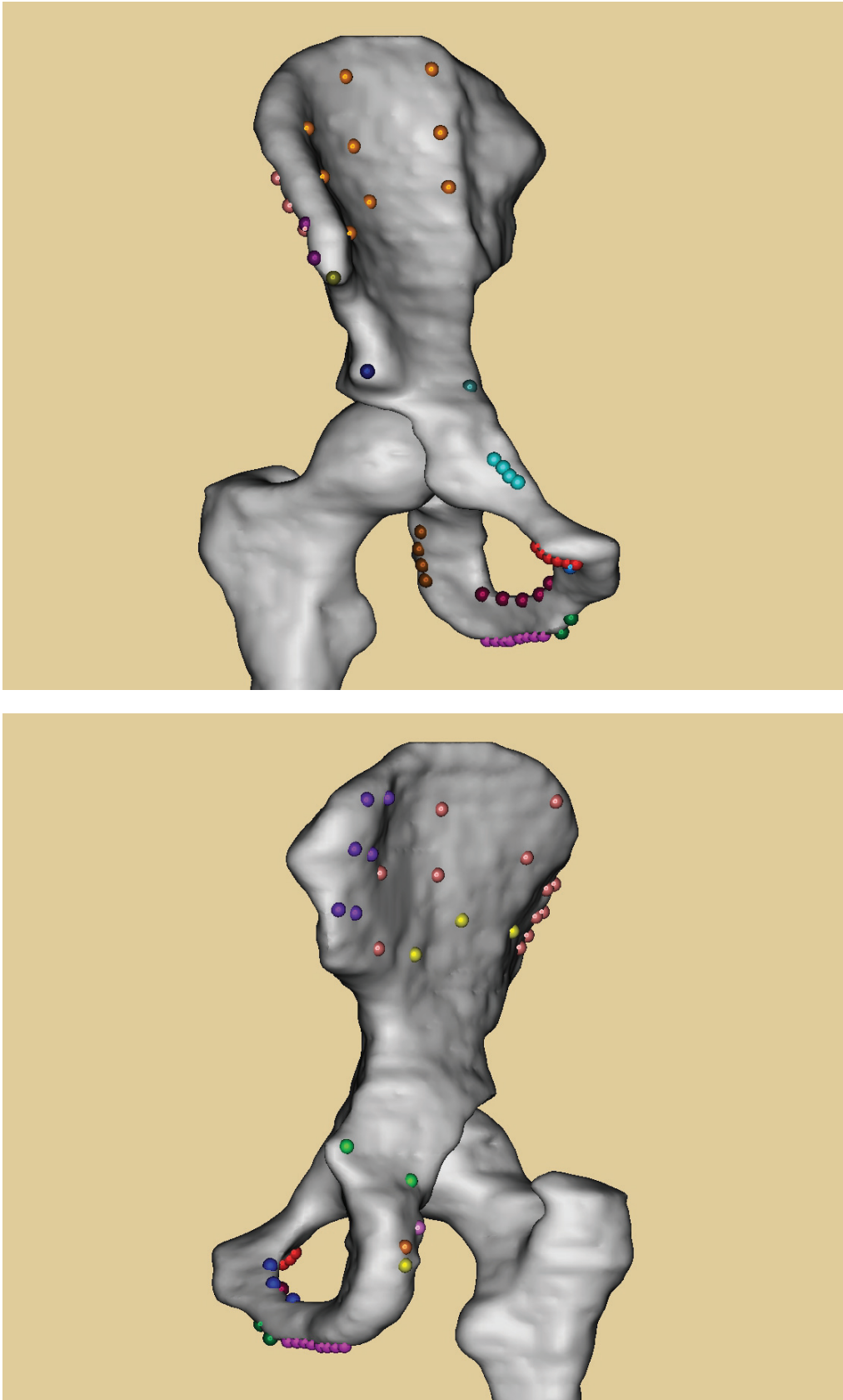


Figure 4.3: 3D geometry and muscle attachment sites for the male model. Top figure- anterior view, Bottom figure- posterior view

4.2.2 Mesh Generation

STL files exported from Mimics were converted to volumetric meshes with tetrahedral elements in HyperMesh 2017 (Altair Engineering, Troy, Michigan, USA). Due to the limited resolution of the MRI images, cortical bone thickness was assumed to be uniform at 1.4mm (one pixel width). Previous studies have used constant cortical thickness of between 1-2mm (Anderson et al. 2005; Majumder et al. 2004; Dalstra and Huiskes 1995). Computed tomography (CT) scans have found that cortical bone thickness can vary between 0.5-5mm in the pelvis. As 95% of stress is transferred through the cortical bone, an element scaling factor of 1.5 was used in the trabecular bone to reduce the number of elements in the model, with larger elements towards the interior of the trabecular bone where the stress was not of interest to this study. Because of the difficulty in identifying soft tissues in MRI images, cartilage was later added using tetrahedral elements in the FE analysis software, Marc Mentat 2018 (MSC.Software, Palo Alto, USA)(Figure 4.4). As this study was not concerned with stress in the acetabulum, no distinction was made between femoral and acetabular cartilage (Liu et al. 2016). Instead, a single layer of cartilage was created and assumed to fill the space between the femoral head and the acetabulum (Figure 4.5).

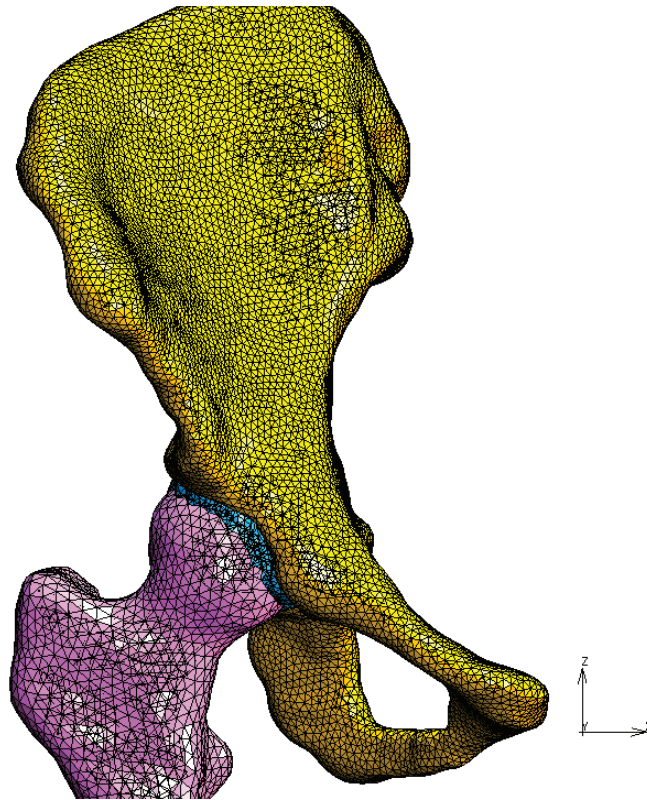


Figure 4.4: Finite element mesh of the female pelvis model, anterior view.

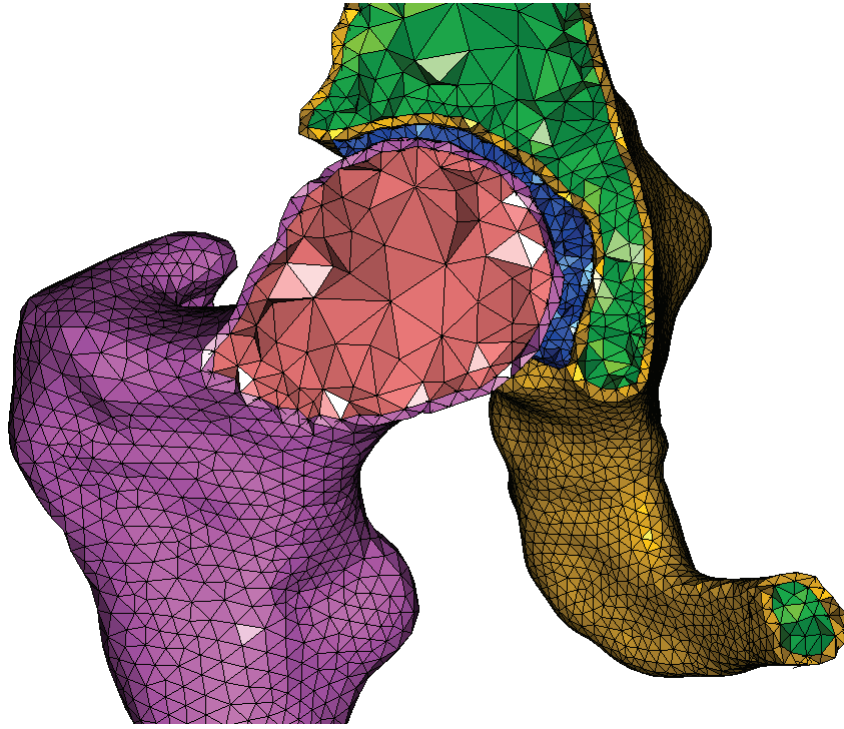


Figure 4.5: Cross section of male model showing cortical (yellow, purple) and trabecular (pink,green) bone and cartilage layer mesh (blue).

Mesh convergence analysis was performed on both the male and female model by adjusting the mesh density until the numerical solution converged. The SI-joint and pubic symphysis were fully constrained in all directions and a point load of 700N (approximately 1BW) was applied to the inferior pubic ramus of each mesh. The resulting peak stress and computation time are shown in Tables 4.2 and 4.3. Increasing the mesh density beyond 170000 elements resulted in only a small increase in von Mises Stress ($< 1.2\%$) but almost doubled the computation time. The meshes selected for both the male and female model were therefore less than 170000 elements. Element sizes in the selected meshes were 2mm for cortical bone and cartilage, and between 2mm and 7mm for the trabecular bone as shown Figure 4.5.

Number of Elements	Peak von Mises Stress	% difference	Computation Time (min)
50303	28.6	-	4
165771	101.8	255.9	19
204094	102.7	0.9	35

Table 4.2: Results of the mesh convergence analysis for the male model with a 700N load applied

Number of Elements	Peak von Mises Stress	% difference	Computation Time (min)
30989	40.8	-	3
154639	133.6	227.5	15
178563	135.2	1.2	25

Table 4.3: Results of the mesh convergence analysis for the female model with a 700N load applied

4.2.3 Material Properties

Mechanical properties for each material are shown in Table 4.4. All materials were assumed to be linear-elastic isotropic. Values of Young's modulus (E) and Poisson's ratio (ν) were taken from the literature as shown below.

Material	Young's Modulus (E)	Poisson's Ratio (ν)	Source
Cortical Bone	17GPa	0.3	Dalstra and Huiskes 1995
Trabecular Bone	70MPa	0.2	Dalstra and Huiskes 1995
Cartilage	12MPa	0.42	Hastings et al. 1998

Table 4.4: Mechanical properties of model materials

4.2.4 Boundary Conditions

As a hemi-pelvis was used in this study it was necessary to approximate the ligamentous boundaries at the sacro-iliac joint and the pubic symphysis (Figure 4.6). The sacroiliac joint generally allows very little movement in healthy adults and hence is often treated as a fixed, fully constrained boundary (Ghosh et al. 2015; Ravera et al. 2018; Anderson et al. 2005). Watson et al. (2017) found comparable results between constrained hemi-pelvis and full pelvic ring models. They noted that models were more sensitive to how the pubic symphysis boundary was represented compared with the sacro-iliac joint. In this study the sacro-iliac joint was therefore treated as a fixed boundary as it was not near the region of interest. The pubic symphysis was modelled using a series of spring/dashpot elements with high stiffness in compression (1500 N/mm) and low stiffness in tension, connected to fixed nodes as shown in Figure 4.7 (Dakin et al. 2001; Phillips et al. 2007). The hip joint contact force was applied to a node at the centre of the femoral head (Figure 4.6).

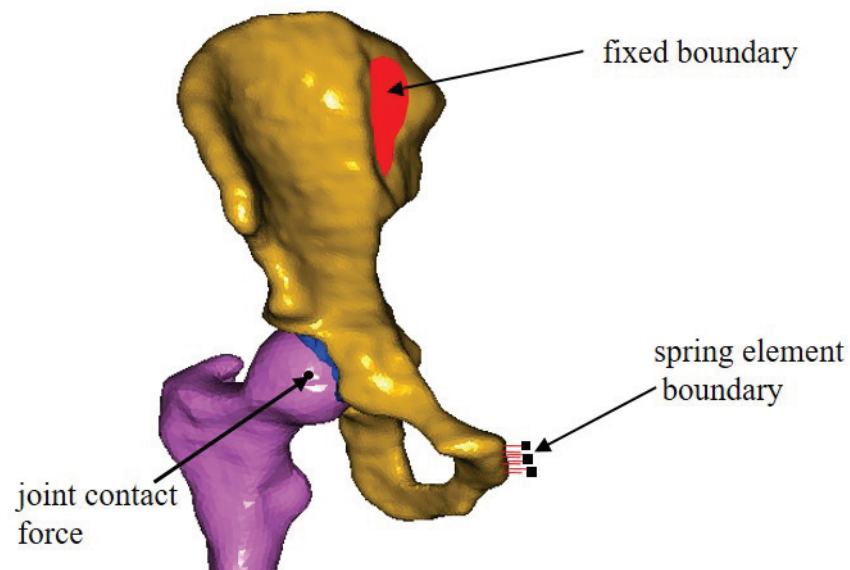


Figure 4.6: Male pelvis showing fixed boundary on the sacro-iliac joint, ligamentous boundary on the pubic symphysis joint, and load applied to the femoral head.

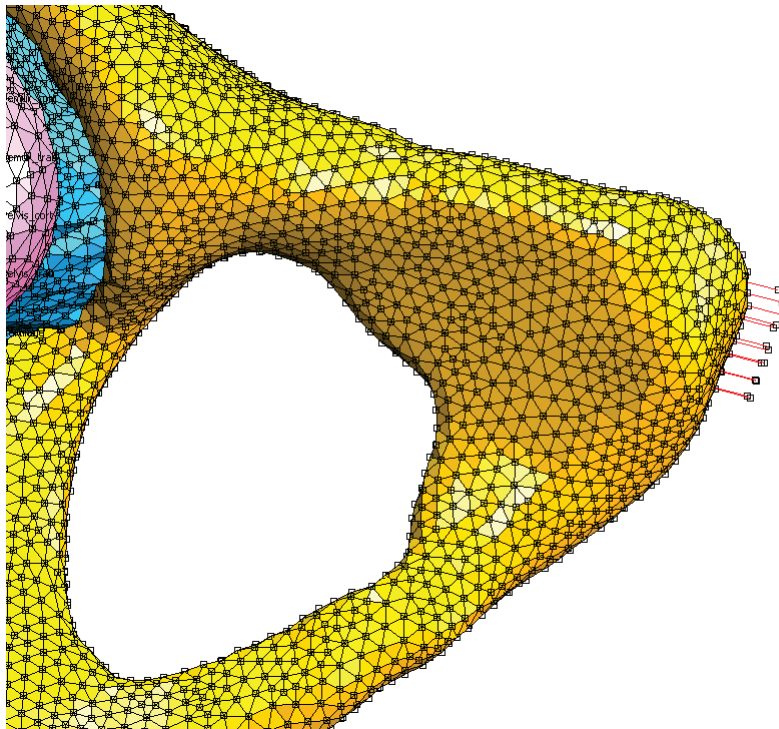


Figure 4.7: Spring/dashpot elements at the pubic symphysis connected to fixed nodes.

Subject- specific muscle and JCF's were taken from the output of the Freebody model described in Chapter 3. Three points in the gait cycle were analysed for each load condition (0,20 and 40kg), corresponding to key events and instances of peak forces for the right leg:

Position 1: First peak of hip JCF (weight acceptance)

Position 2: Midstance

Position 3: Second peak of hip JCF (push off)

Examples of how hip JCFs and key muscle forces vary for each gait position are provided in Figure 4.8. For each case, the pelvis and femur were oriented in the global frame about the pelvis centre of rotation according to the subject-specific pelvic angles measured in Chapter 3. A total of 21 muscles forces (divided into 82 muscle elements) were applied as point loads to nodes on the surface of the cortical shell at the attachment sites shown in Figures 4.2 and 4.3. Hip JCF's for each model and position are shown below in Table 4.5. Muscle forces are provided in Appendix B.

		Female			Male		
		R_x	R_y	R_z	R_x	R_y	R_z
Position 1:	0kg	814	534	1722	791	460	1839
Position 2:	0kg	287	-129	1031	218	-109	1432
Position 3:	0kg	668	-1587	2694	382	-1661	2777
Position 1:	20kg	1396	401	2491	1118	444	2393
Position 2:	20kg	392	-159	1329	329	-150	1676
Position 3:	20kg	1067	-2260	4592	690	-1968	4168
Position 1:	40kg	1154	562	2263	1197	550	2289
Position 2:	40kg	640	-234	2020	420	-171	1737
Position 3:	40kg	1092	-2197	4775	583	-2027	4666

Table 4.5: Hip JCF's (N) for each model and gait position

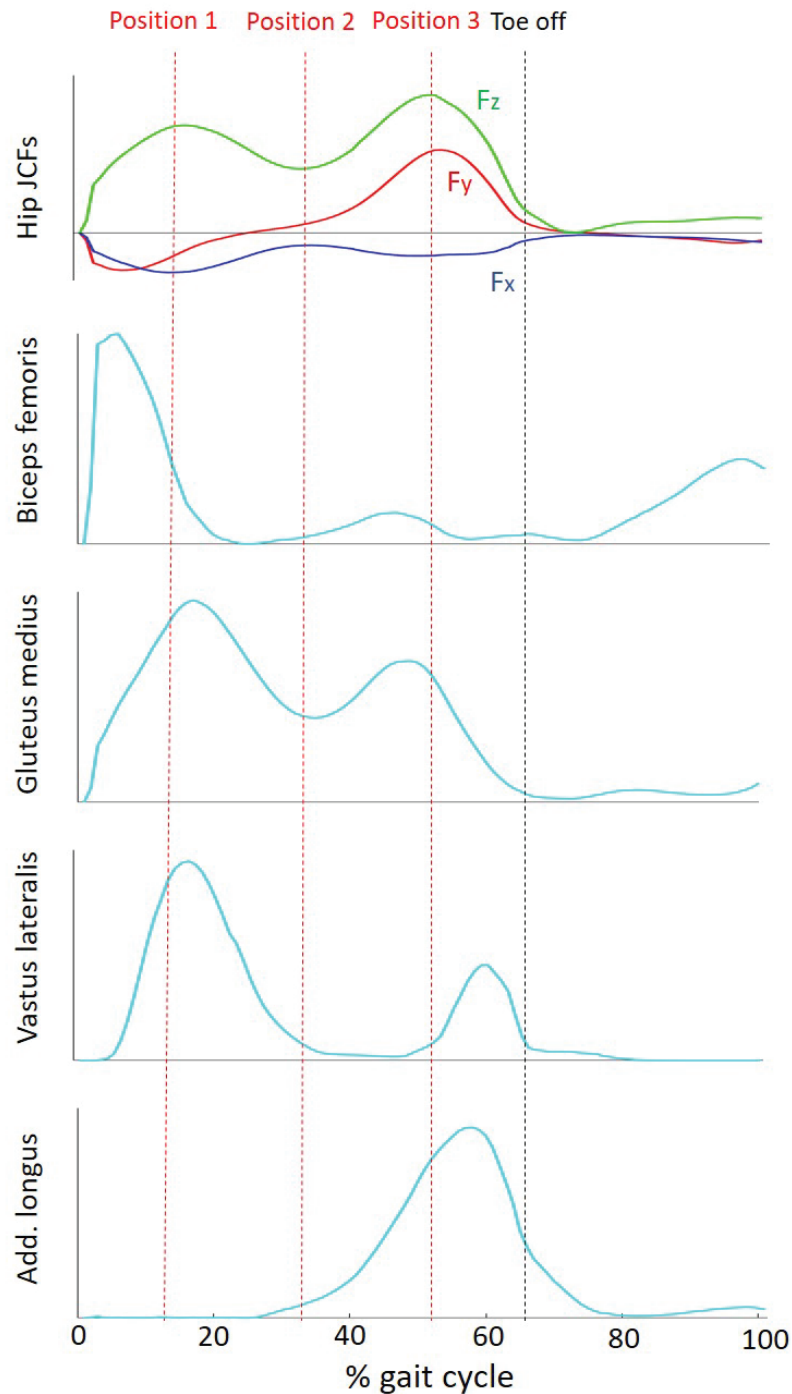


Figure 4.8: Mean hip JCFs and muscle forces at each gait position for all subjects measured in Chapter 3.

4.2.5 Validation Study

The FE models were validated using data from Dalstra and Huiskes (1995), who developed finite element models of the pelvis after performing experimental loading on cadaveric specimens. To allow for comparison with results from Dalstra and Huiskes (1995), the boundary conditions at the SI-joint and pubic symphysis of the

models were fully constrained and only the joint contact force at midstance was applied (Figures 4.9 and 4.10).

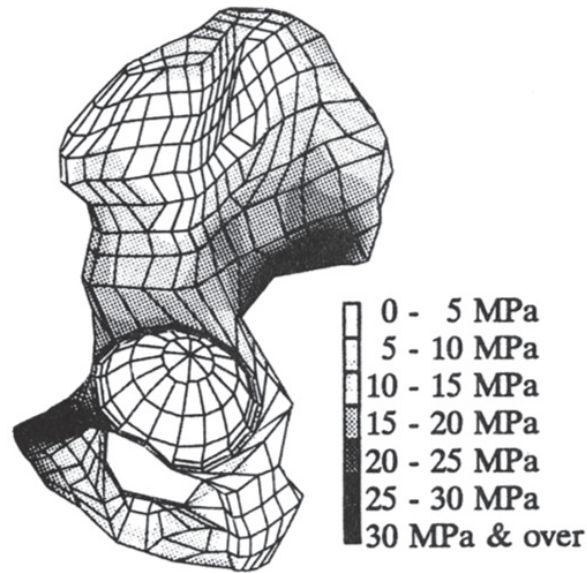


Figure 4.9: von Mises stress distribution during midstance with only the joint contact force applied (from Dalstra and Huiskes 1995).

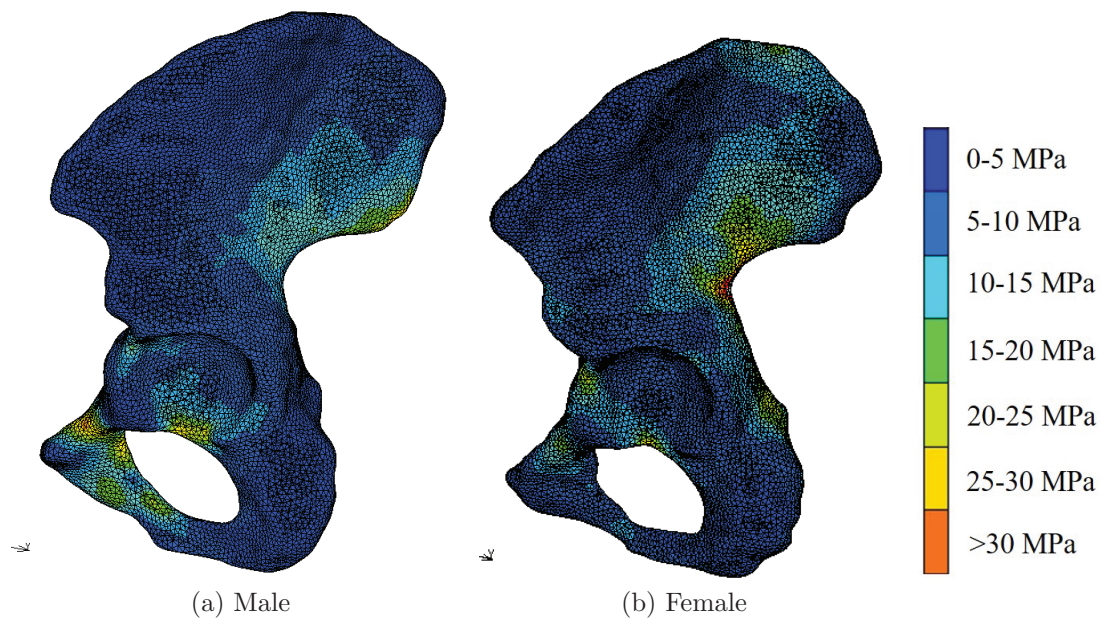


Figure 4.10: von Mises stress (N/mm^2) distribution during midstance with all boundaries constrained and only joint contact force applied.

Stress patterns in the male and female model were similar to those seen in Dalstra and Huiskes (1995). In all models stress was greatest ($>25\text{MPa}$) around the sciatic notch and in the superior ramus. As in Dalstra and Huiskes (1995) results, an area of high stress over the top of the iliac crest was observed in the

female model. However, this was not seen in the male model, possibly due to its larger size and thicker ilium. Some areas of increased stress were observed around the superior rim of the acetabulum in all models, although differences in stress distribution around the acetabulum were expected as Dalstra and Huiskes (1995) modelled the femoral head as a perfect sphere. The male and female models showed good agreement with the validation model for stress distribution in the lower pelvis, with peak stresses of 25-30MPa found in the superior rami, and very low stresses (<5 MPa) observed in the ischium.

4.3 Results

Von Mises stress distributions at weight acceptance, midstance and toe push-off are presented in Figures 4.11-4.16. Peak stress values for the superior and inferior rami are shown in Table 4.6.

4.3.1 Position 1- Weight Acceptance

Von Mises stress distribution during weight acceptance is shown for each sex and load condition in Figures 4.11 and 4.12. The initial peak hip JCF was applied to each model along with corresponding muscle forces. The boundary conditions for the SI-joint and pubic symphysis were modelling as described in Section 4.2.4. The results showed that, in the male subject, areas of increasing stress occurred in the inferior ramus on both the lateral (64% increase) and medial sides (57% increase) as carried load increased to 40kg, while the superior ramus showed no increase in stress. Higher stresses were experienced in the superior ramus when 0kg and 20kg were carried, and in the inferior ramus when 40kg was carried. The highest peak stress (22MPa) in the male subject was observed in the inferior ramus when carrying 40kg.

In the female subject, stress increased in the superior ramus on both the lateral (29% increase) and medial sides (26% increase) when carrying 20kg compared with carrying no load, and then decreased when the subject carried 40kg ($< 10\%$ decrease). In the inferior ramus, stress on the lateral side increased by 14% between 0kg and 20kg and then decreased by 8% between 20kg and 40kg. On the medial side, there was a total increase in stress of 48%. As in the male model, stress in the superior ramus was higher than in the inferior, except when 40kg was carried. The highest peak stress was seen in the superior ramus when 20kg was

carried (45MPa). In addition, peak stresses in the female model were consistently higher than in the male model, with the greatest differences seen when subjects carried 20kg (45MPa for the female model compared with 16MPa in the male model). Despite the differences in absolute stress value, stress distribution throughout the rami was similar in both models, with areas of highest stress occurring in the mid-sections of the superior and inferior rami, on both the lateral and medial sides.

4.3.2 Position 2- Midstance

Von Mises stress distribution in the male and female models during midstance on the right leg are shown in Figures 4.13 and 4.14. The male model showed very little change in stress distribution during midstance as carried load increased. An area of increased stress (10MPa) developed in the superior ramus when 20kg and 40kg were carried.

In the female model, significant increases in peak stress were seen in both the superior lateral (175% increase) and medial ramus (150% increase), as well as the inferior ramus (lateral- 186% increase, medial- 183% increase) as carried load increased. Areas of increased stress occurred in the mid-sections of both rami. The greatest peak stress was found in the lateral superior ramus when 40kg was carried (33MPa).

4.3.3 Position 3- Push-off

Von Mises stress distribution during push off is shown for each sex and load condition in Figures 4.15 and 4.16. In the male model, peak stress in the superior ramus increased as carried load increased. This was seen in both the lateral (24% increase) and medial sides (94% increase) of the pelvis. In the inferior ramus, peak stress increased between the carried load conditions of 0-20kg (14% increase), and then decreased when 40kg was carried (25% decrease). The lowest peak stress observed in the male inferior ramus occurred when 40kg was carried (12MPa). The greatest peak stress, however, was also observed when 40kg was carried, and occurred on the medial superior ramus (33MPa). There was little change in the magnitude or distribution of the stress observed on the medial side of the inferior ramus.

In contrast, very distinct areas of increasing stress were observed in the inferior ramus of the female model on both the medial and lateral sides. In the superior ramus, peak stress increased between the 0kg and 20kg load conditions (lateral-140%, medial-61%) but again showed a small decrease when 40kg was carried. The greatest peak stress observed in the female model occurred in the lateral superior ramus when 20kg was carried (60MPa). The female model was again found to exhibit greater stresses across each of the carried load conditions compared with males.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Position 1						
Superior ramus: lateral side	35	45	42	18	16	17
Superior ramus: medial side	27	34	31	20	20	20
Inferior ramus: lateral side	21	24	22	11	12	18
Inferior ramus: medial side	21	27	31	14	20	22
Position 2						
Superior ramus: lateral side	12	15	33	2	5	8
Superior ramus: medial side	10	12	25	2	10	10
Inferior ramus: lateral side	7	10	20	1	2	2
Inferior ramus: medial side	6	8	17	2	2	2
Position 3						
Superior ramus: lateral side	25	60	56	21	23	26
Superior ramus: medial side	33	53	52	17	25	33
Inferior ramus: lateral side	22	41	43	14	16	12
Inferior ramus: medial side	10	23	30	8	7	10

Table 4.6: Peak von Mises stresses (MPa) in the superior and inferior rami for each gait position.

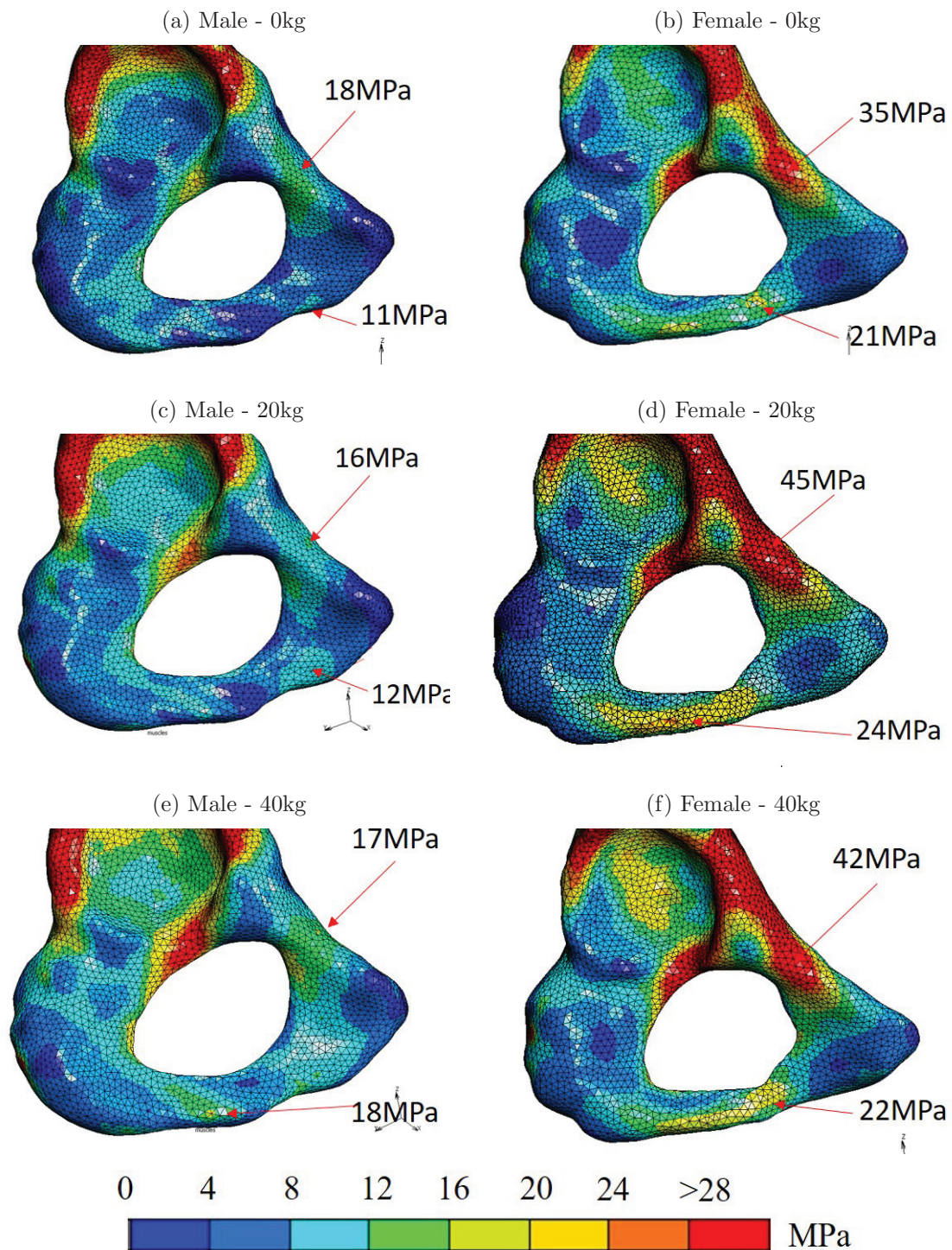


Figure 4.11: Position 1- von Mises stress (N/mm^2) distribution in the pubic rami during weight acceptance, lateral view.

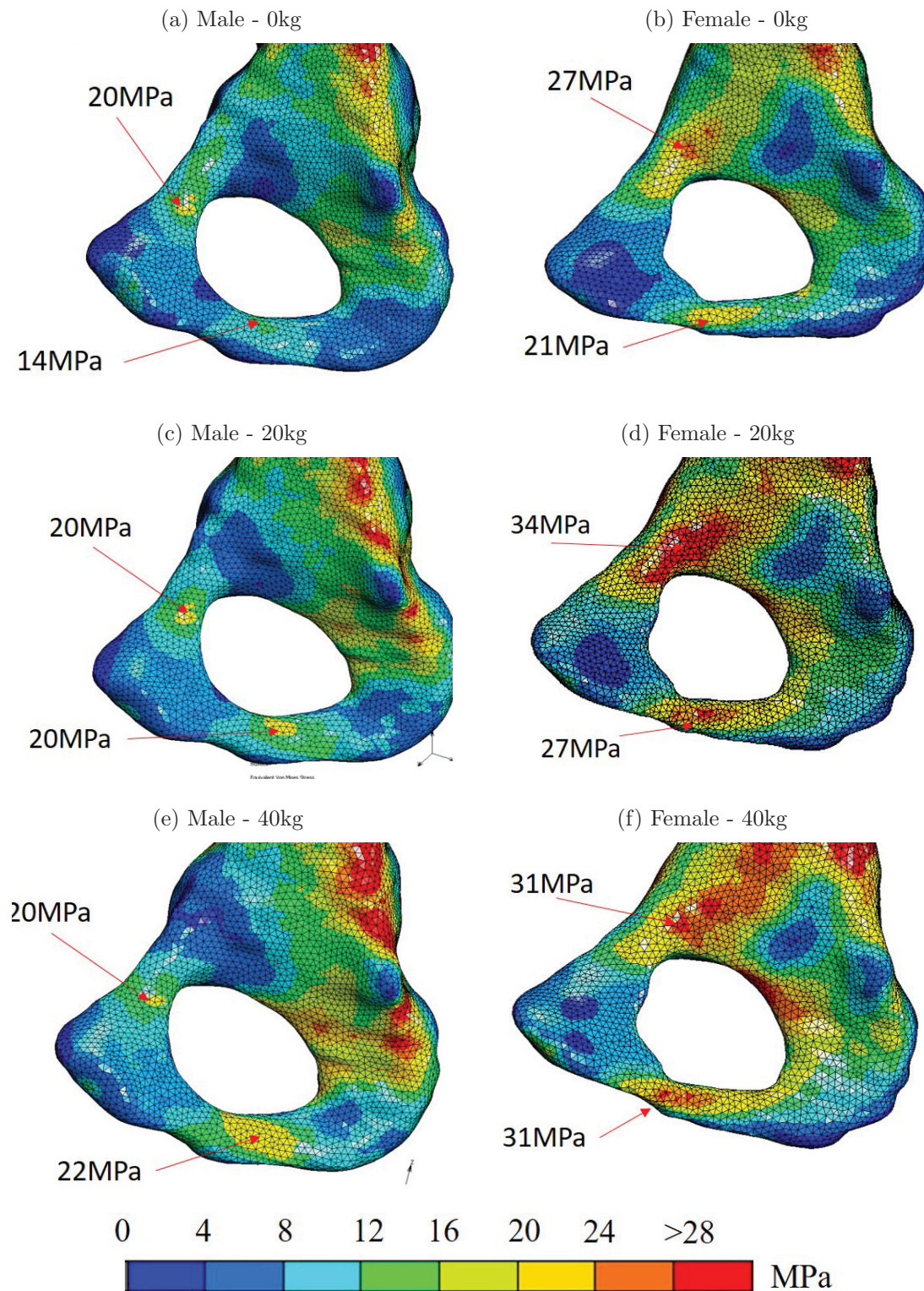


Figure 4.12: Position 1- von Mises stress (N/mm²) distribution in the pubic rami during weight acceptance, medial view.

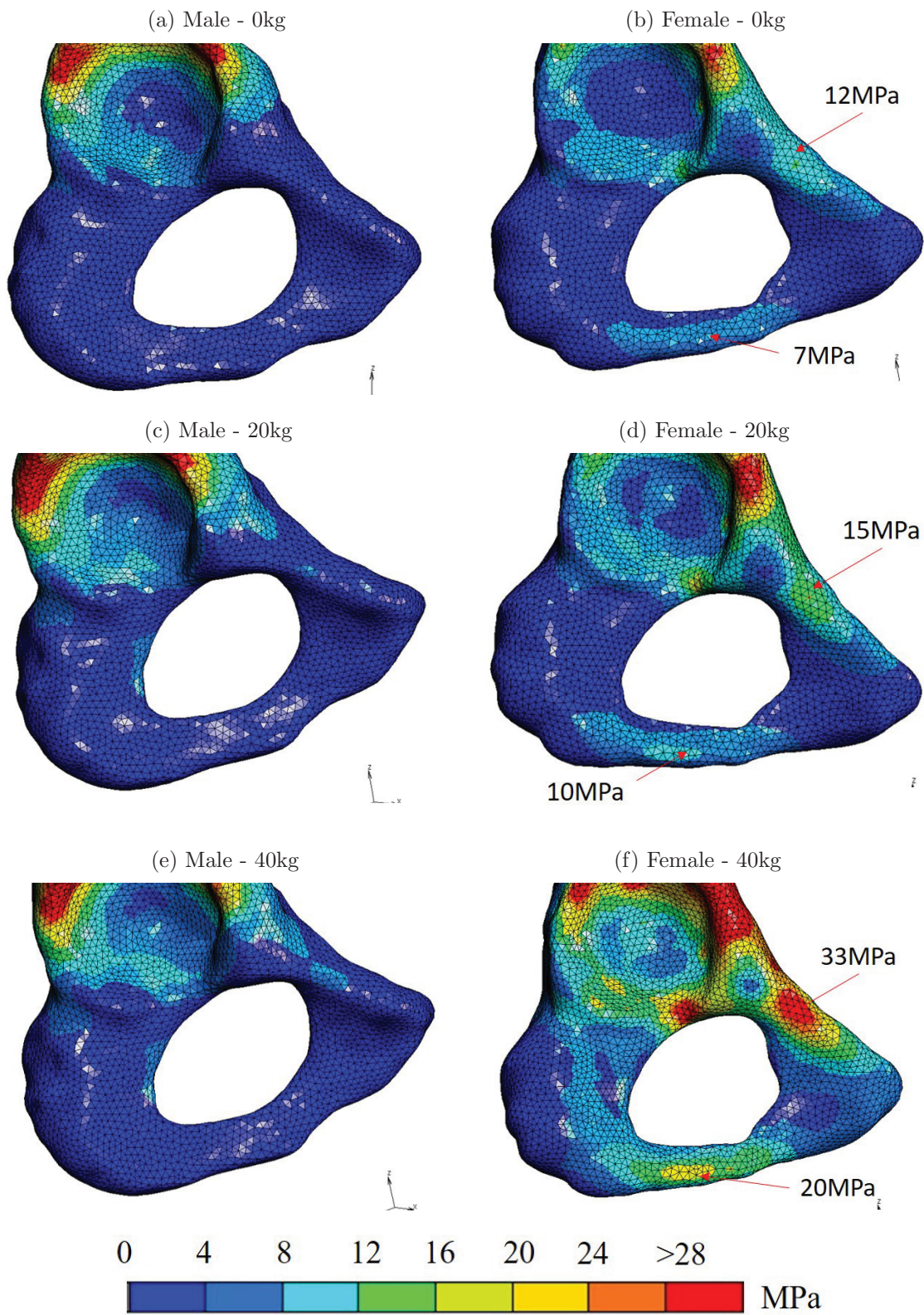


Figure 4.13: Position 2- von Mises stress (N/mm^2) distribution in the pubic rami during midstance, lateral view.

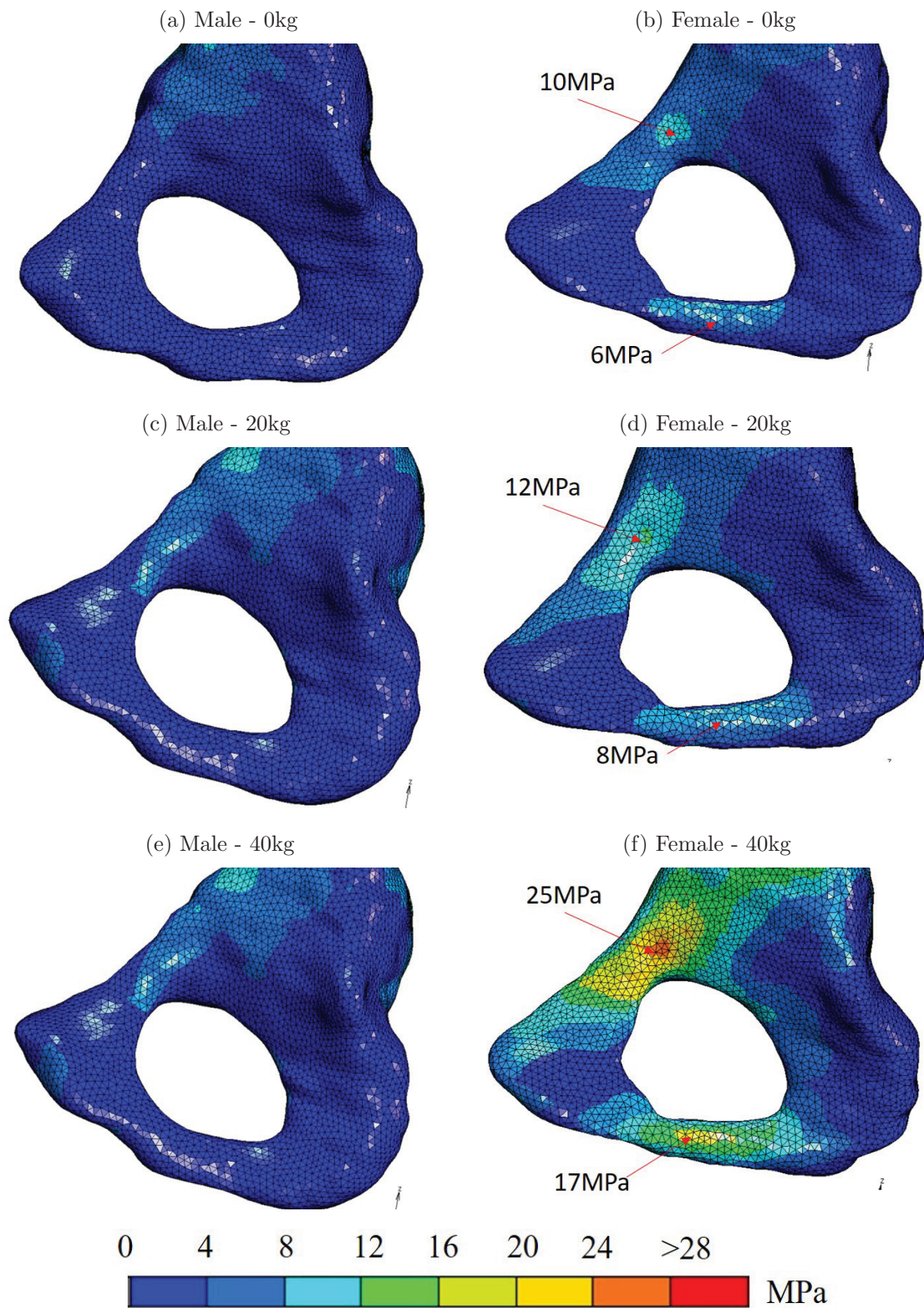


Figure 4.14: Position 2- von Mises stress (N/mm^2) distribution in the pubic rami during midstance, medial view.

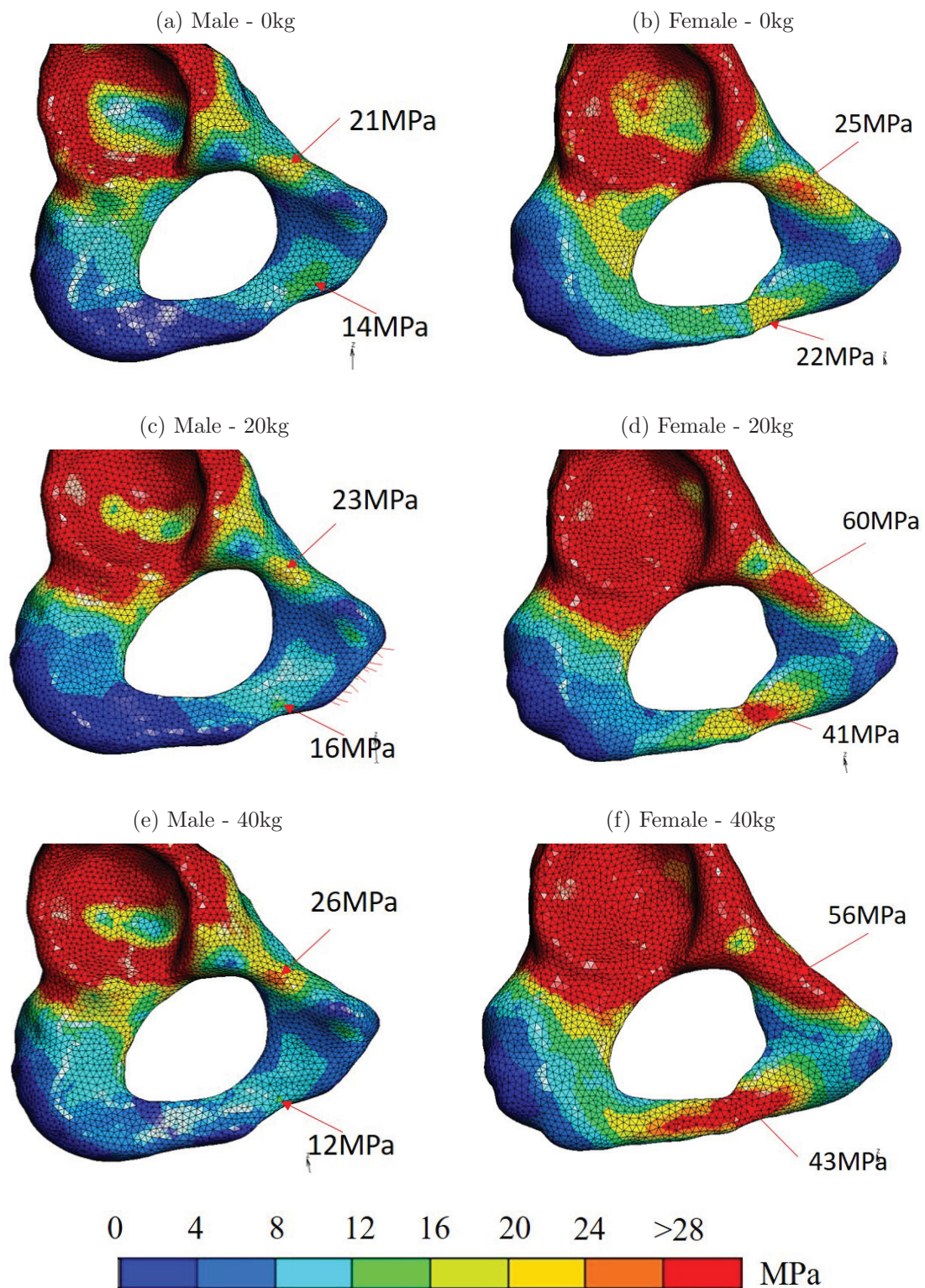


Figure 4.15: Position 3- von Mises stress (N/mm²) distribution in the pubic rami during push-off, lateral view.

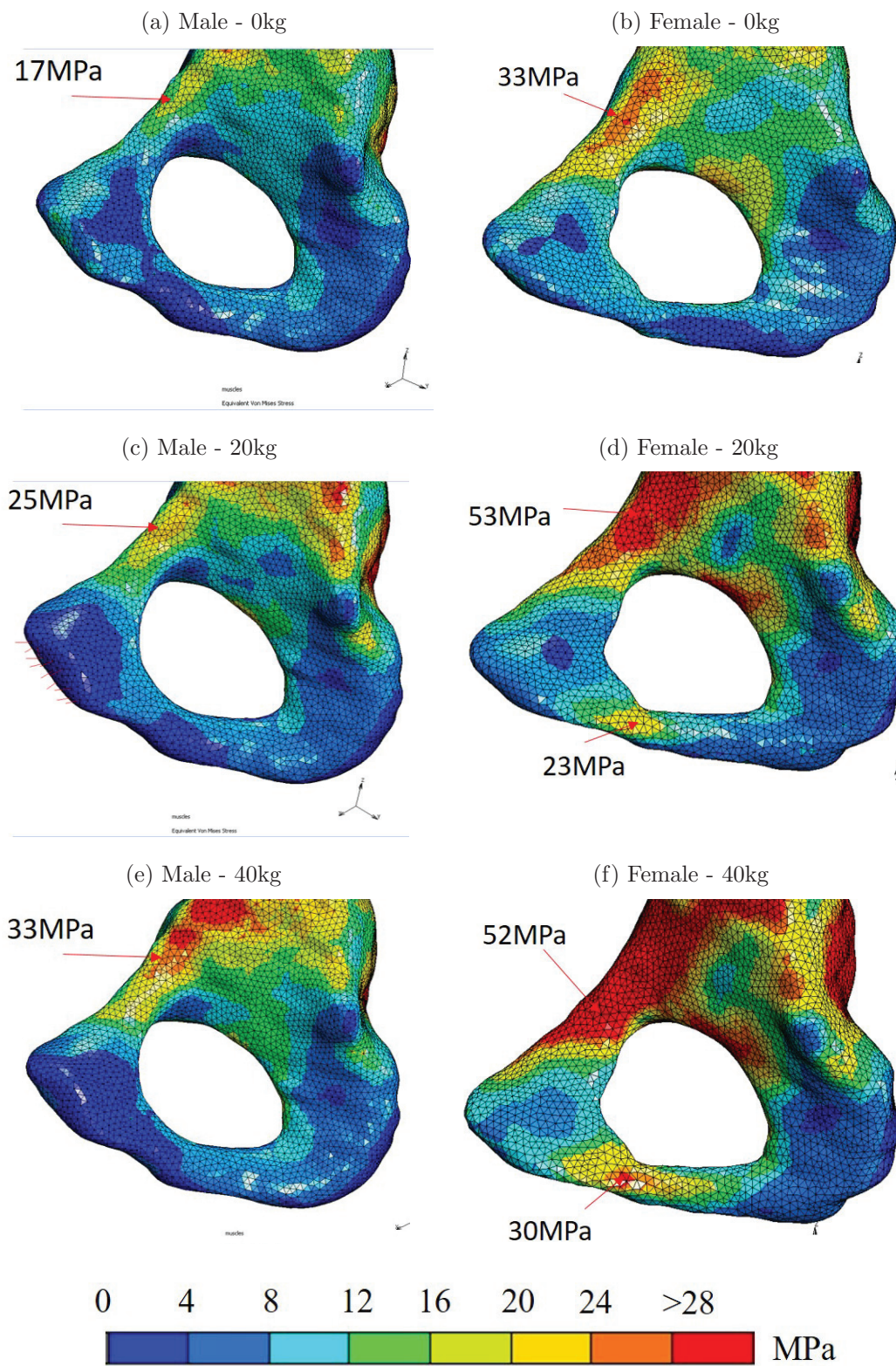


Figure 4.16: Position 3- von Mises stress (N/mm^2) distribution in the pubic rami during push-off, medial view.

4.4 Discussion

The results of this study show that von Mises peak stress and distribution in the rami are affected by both gait position and carried load. It is generally suggested in the literature that force is transmitted through the acetabulum around the pelvic ring via the SI-joint and pubic rami, resulting in areas of increased stress in these regions (Ricci et al. 2018; Ghosh et al. 2015). During weight acceptance (Position 1) the hip was flexed and the hip joint contact force was angled posteriorly. Volinski et al. (2018) considered the full pelvic ring in their model of a 50th percentile male during unloaded walking. They found peak stresses of 35MPa in the superior ramus during weight acceptance. Ricci et al. (2018) also modelled the full male pelvic ring, and found a peak stress of 25MPa in the superior ramus during weight acceptance. The results of this study are in good agreement, with peak stress in the female superior ramus also found to be 35MPa (Position 1). Peak stress in the male model was likewise located in the superior ramus but with a lower value of 20MPa in Position 1. This suggests that the hemi-pelvis models and boundary conditions used in this study were appropriate and can provide comparable results with full pelvis models.

Although finite element methods have been used previously to model the tibia (Xu et al. 2016) and shoulder (Epstein et al. 2017) during load carriage, no studies which considered the hip or pelvis could be found. It was therefore not possible to compare the loaded walking results of this study with literature. When carrying 20kg and 40kg, peak stresses in the female model increased significantly compared with the unloaded condition during weight acceptance. Perhaps surprisingly, these stresses were greater when carrying 20kg than when carrying 40kg. This could be due to the reduction in joint contact and adductor magnus muscle force shown by the female subject between 20kg and 40kg (Table 4.5). Furthermore, the female model showed an area of increasing stress in the inferior ramus corresponding to the increase in adductor magnus and obturator internus muscle forces as carried load increased.

In the male model, only the inferior ramus exhibited increased stress in Position 1 as carried load increased. Again areas of high stress formed in the mid-section of the ramus, where the bone was narrow. Across all carried load conditions greater peak stresses were found in the female model than the male model. This could be due to the smaller bony geometry, as well as the greater R_x and R_y causing increased torsion around the pubic symphysis (Volinski et al.

2018). In addition, it has been suggested that increased hip obliquity can increase stress in the superior ramus (Volinski et al. 2018).

Stress distribution in the pelvis during single leg stance has been well documented by many studies (Phillips et al. 2007; Dalstra and Huiskes 1995; Ravera et al. 2015; Ghosh et al. 2015). During midstance (Position 2) the pelvis was generally in a near neutral position and the full weight was supported by the single, standing leg. At this point in the gait cycle Phillips et al. (2007) suggest that the bony pelvis acts primarily as a column, transferring the joint contact force upwards from the acetabulum through the ilium to the sacro-lumbar region. The results of this study show that, during unloaded walking, stress in the rami are relatively low ($<12\text{MPa}$), in agreement with the literature (Phillips et al. 2007; Volinski et al. 2018; Ravera et al. 2018).

As carried load increased, little change was noted in the male model during midstance. In contrast, the female model exhibited significant increases in peak stress in both rami, particularly between the 20kg and 40kg load conditions. At this point in the gait cycle, there was a large increase in female hip JCF and gluteal muscle force required to stabilise the pelvis due to the large carried load. As in Position 1, the female model displayed larger stress values and areas compared with the male model. This could be due to the female having smaller bony geometry and a larger hip JCF relative to body mass.

During push off (Position 3) the hip was extended and the hip joint force was angled anteriorly. At this point in the gait cycle, Volinski et al. (2018) observed superior ramus peak stress of greater than 20MPa , which is in good agreement with the results of this study. Similarly, Ricci et al. (2018) found peak stresses of 25MPa and 18MPa in the superior and inferior rami respectively. They found that, during push off, the rami undergo bending with the lateral side of the pelvis in compression and the medial side in tension. When carried load was increased to 20kg, a significant increase in peak stress was observed in the female model, with areas of high stress forming at the narrowest points of the rami. Conversely the male model displayed very little change in stress distribution. As in Position 1, when load was increased to 40kg, the female model saw a reduction in superior ramus peak stress but an increase in inferior ramus stress. Interestingly, the opposite was found to occur in the male model. The female subject was observed to reduce pelvic obliquity angle between the 20kg and 40kg loads while the male subject increased obliquity. This confirms results from Volinski et al. (2018) and

suggests that changes in pelvic inclination may affect how force is transmitted through the rami, with greater obliquity causing larger stresses in the superior ramus and lower obliquity increasing stress in the inferior ramus.

This study found that maximum von Mises stresses occurred during push off (Position 3). This corresponded to the maximum JCF and peak of many of the muscle forces seen over the gait cycle. Xu et al. (2016) observed peak stresses of 58.5MPa in the tibia when females carried 30% of their body weight, which is in a similar range to the results found in this study for load carriage. Maximum stress was found to be 60MPa and 33MPa in the female and male models respectively. Ultimate tensile strength of cortical bone has been reported in the range of 80-120MPa (Ott et al. 2010; Arola et al. 2010)). Ricci et al. (2018) suggests that the endurance limit of bone may be around 40% of this range (i.e. 32-48MPa), as it is for some other materials. This implies that carrying load could significantly increase the risk of pelvic rami stress fractures, particularly in females. Further work should investigate the sensitivity of peak stresses to individual muscle forces and hip kinematics in order to identify potential biomechanical risk factors for pelvic stress fractures during load carriage.

There were a number of limitations to this study. Firstly, only a hemi-pelvis model was used for each subject. Although results comparable with full pelvic ring models have been found, a more accurate distribution of stress could be achieved if both sides of the pelvis were included in this study. In addition, the use of a hemi-pelvis model required the construction of boundary approximations at the SI-joint and pubic symphysis, potentially resulting in increased stress around these regions. Muscle forces were applied as point loads to the cortical shell which could increase localised stresses. In order to prevent this, muscle forces could be applied as surface loads in future. Finally, due to limitations with the imaging technique, cortical bone thickness and density were assumed to be constant for both subjects while in reality they have both been shown to vary. These limitations may affect the accuracy of the peak stress values measured, however they still allow for direct comparison between gait positions and carried loads, in line with the main aims of this study.

4.5 Conclusions

This study found that the von Mises stress distribution in the pubic rami changed over the course of the gait cycle and as carried load was increased. Peak stresses did not necessarily increase linearly as carried load increased, suggesting that gait alterations made by subjects in response to carrying load may affect bone stress in the pelvis. Regions of high stress developed in the mid-sections of the superior and inferior rami, corresponding to the areas reported in the literature as most likely to incur a stress fracture (Beck et al. 2000; Pope 1999). Von Mises stress in the female model was consistently higher than in the male model, possibly due to narrower rami, as well as differences in joint contact and muscle forces and pelvic obliquity. These, along with other factors such as bone density, could indicate why females are so much more likely to obtain a pelvic stress fracture.

This chapter has quantified stresses in the pelvic cortex for an average male and female walking with and without load. Changes in hip kinematics and MSk forces, due to both load carriage and gait position, affect stress distribution in the pubic rami. In order to further investigate the effects of military load carriage and gait on forces in the lower limb, the next chapter will include a biomechanical analysis of running with body borne loads.

Chapter 5

Biomechanical analysis of running during load carriage: Effects of load and sex

The results of Chapters 3 and 4 revealed that there are significant differences between males and females when walking with load. This study aimed to investigate if these differences were exacerbated when undergoing more strenuous activity. Therefore, this chapter presents a biomechanical analysis of running during military load carriage. Motion capture and musculoskeletal modelling techniques were used to investigate the main and interaction effects of sex and load on running. Both kinematic and kinetic parameters were analysed and are discussed here.

5.1 Introduction

The incidence rate of MSk injuries in British Army recruits has been found to be as high as 48% (Sharma et al. 2015). Of these injuries, femoral stress fractures were found to require the longest rehabilitation time (Sharma et al. 2015). MSk injuries typically arise in training during repetitive load-bearing exercise such as running (Hauret et al. 2010). As discussed in Chapter 2, military personnel are regularly required to train carrying heavy loads, further increasing their risk of injury (Andersen et al. 2016).

Carrying body born loads has been shown to affect lower limb biomechanics when running. Teunissen et al. (2007) reported a 13% increase in peak vertical ground reaction force, while Silder et al. (2015) found that leg stiffness increased when running with load. They defined leg stiffness as the peak vertical GRF divided by the change in leg length during stance phase. A study of 31 subjects (both male and female) carrying up to 20% BW found that increasing load and speed while running significantly increased joint work (Liew et al. 2016). In males, running with body borne loads of up to 40kg were found to increase peak trunk, hip and knee flexion moments (Brown et al. 2014). However, running with load has not been found to affect the distribution of work produced in the lower limb joints (Liew et al. 2016; Brown et al. 2014). Increases in joint contact forces while running have been reported in the knee in males carrying as little as 15kg (Willy et al. 2016) and in the hip in females carrying 30% BW (Xu et al. 2017). This would be expected due to the increased GRF's and joint moments.

When running without load, Giarmatzis et al. (2015) found that increasing speed resulted in an increase in hip JCF, and suggested that peak JCF is best predicted by hip adduction moment. Several studies have considered the effect of sex on running without load. A study of 20 males and 20 females found that, compared to males, females displayed greater hip joint work in the frontal and transverse planes (Ferber et al. 2003). Studies have also found that female runners exhibit significantly greater peak hip adduction and internal rotation (Ferber et al. 2003; Chumanov et al. 2008). Hobara et al. (2012) found that increasing stride (in males) increased peak GRF, which has implications for females who preferentially take shorter strides (Milner et al. 2006b). Finally, only Lobb et al. (2019) compared males and females when running with load. They found that, compared with females, males exhibited greater leg stiffness when running with short strides and more significant decreases in leg stiffness as stride length increased.

In summary, running with load has been found to increase leg stiffness, joint work and joint contact forces. In addition, differences between males and females have been seen in both unloaded and loaded running. The aim of this study was therefore to compare kinematic and kinetic parameters between males and females during running with and without load. The standardised loads of 20kg and 40kg were again selected (Chapter 3) as they are representative of typical military training loads (Andersen et al. 2016). As in Chapter 3, a rucksack with an evenly distributed load (a central CoM) was used. Although typically soldiers would wear a combination of webbing, rucksack, helmet and weapon on training exercises, this study only included a rucksack to avoid marker occlusion during optical motion tracking measurement.

5.2 Methods

5.2.1 Subjects

Eighteen healthy volunteers (Table 5.1) with no history of lower limb or back injury took part in this study. All volunteers had previously participated in the study described in Chapter 3. Subjects were either current or ex-military personnel, or could complete the minimum military training entry fitness requirements of (a) 2.4km run in less than 14 minutes 58 seconds and (b) 40kg lift and carry for 60m. This study was granted ethical approval by the Joint Research Compliance Office and Imperial College Research Ethics Committee (Appendix A). All subjects gave written informed consent.

	Males	Females
<i>n</i>	9	9
Age (years)	26.8 (2.6)	29.0 (3.5)
Height (cm)	178.2 (10.4)	170.8 (4.3)
Mass (kg)*	77.6 (12.4)	66.9 (5.8)
Leg Length (cm)	94.0 (9.8)	89.7 (4.8)
Bi-iliac width (cm)	24.2 (2.1)	25.6 (1.9)
20kg load as %BW*	26.4 (3.9)	30.1 (2.7)
40kg load as %BW*	52.8 (7.9)	60.2 (5.3)

Table 5.1: Participant Data, mean (SD), * indicates significance between sexes for $p \leq 0.05$

5.2.2 Experimental Protocol

The experimental protocol, apparatus and marker set used in this study were the same as those described in Chapter 3, Section 3.2.2. After the retro-reflective markers were placed on anatomical landmarks (Table 3.2, Chapter 3), a static standing trial was taken with the body in a neutral position. Subjects were then instructed to run at a comfortable pace over a 5 meter walkway carrying:

- (a) No load
- (b) 20kg rucksack
- (c) 40kg rucksack

A minimum of three trials per load condition were captured and the order of the loads was randomly assigned. The trial with least marker occlusion was selected.

5.2.3 Data Processing

Data was processed using Vicon Nexus (v2.6.1, Vicon Motion Systems, Oxford, UK), Matlab (MathWorks, Natick, USA) and Freebody (V2.1, Ding et al. 2016) to obtain spatiotemporal, kinematic and kinetic gait parameters. Full details of the data processing can be found in Chapter 3, Section 3.2.3.

5.2.4 Musculoskeletal Model

The Freebody musculoskeletal model (Cleather and Bull 2015) was used in this study (described in Chapter 2, Section 2.5.1).

5.2.5 Statistics

Data was tested for significance between each load condition and sex using two-way repeated measures ANOVAs and the Tukey HSD test post-hoc. A full description of the statistical treatment of the data can be found in Chapter 3, Section 3.2.4. Results were considered significant where $p \leq 0.05$.

5.3 Results

5.3.1 Spatiotemporal

Data was normalised over one gait cycle, from heel strike to heel strike on the right leg. The spatiotemporal parameters measured in this study are presented in Figure 5.1. Significant results for each sex and associated p-values are shown in Table 5.2.

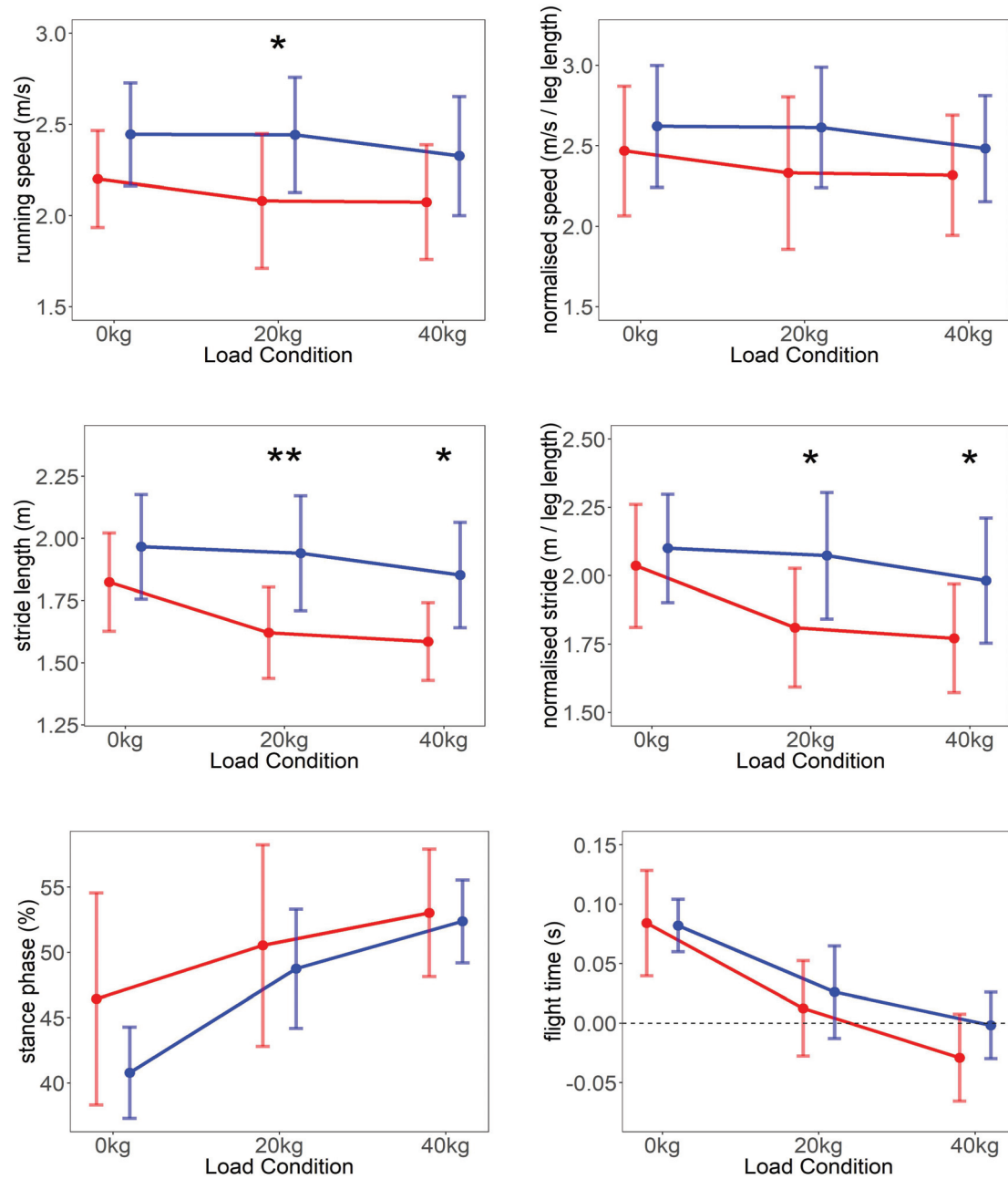


Figure 5.1: Spatiotemporal parameters, * indicates significance between sexes for $p \leq 0.05$, ** for $p \leq 0.005$ (red line- females, blue line- males)

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Stride length	↓ (0.06)	↓ 0.02	-	-	-	-
Stance phase	-	-	-	↑ 0.0004	↑ ≤0.0001	-
Flight time	↓ 0.003	↓ ≤0.0001	-	↓ 0.002	↓ ≤0.0001	-

Table 5.2: Significant changes in spatiotemporal variables with p-values. ↑ indicates increase in value between load conditions, ↓ indicates decrease in value between load conditions

No significant changes were found for self-selected running speed between load conditions, however females ran significantly slower than males when carrying 20kg ($p=0.04$). When normalised to leg length there were no significant differences in running speed. Results of the ANOVA showed sex to have a significant effect on speed ($p=0.003$).

Females took significantly shorter strides than males when carrying 20kg ($p=0.005$) and 40kg ($p=0.007$). This difference remained even when normalised to leg length ($p=0.02$ for 20kg and $p=0.05$ for 40kg). Females also exhibited a significant reduction in stride length between 0-40kg (13% decrease, $p=0.02$) while males saw no change. The reduction in females' stride length between 0-20kg approached significance (11% decrease, $p=0.06$). Results of the ANOVA again showed sex to be a very significant effect ($p<0.0001$).

There were no differences between sexes for any load condition in stance phase, however males significantly increased their time spent in this phase between 0-20kg (20% increase, $p=0.0004$) and 0-40kg (28% increase, $p=0$). Results of the ANOVA showed the effects of both load ($p=0.05$) and sex ($p=0.01$) to be significant for stance phase. Both sexes saw significant reductions in flight time between 0-20kg (85% decrease, $p=0.003$ for females, 68% decrease, $p=0.002$ for males) and 0-40kg (134% decrease, $p=0$ for females, 102% decrease, $p=0$ for males) with females showing greater percentage reductions between load conditions. Negative flight time represented the time spent in double support i.e. there was no point in the gait cycle where both feet left the ground. The ANOVA showed the effects of load ($p<0.0001$) and sex ($p=0.02$) to be significant for flight time.

5.3.2 Joint Angles

Joint angles over one gait cycle (normalised from heel strike to heel strike on the right leg) are presented in Figure 5.2. Toe off is indicated by a vertical dashed line. As there were no significant differences in stance phase between males and females, the total mean toe off of all subjects is shown. Significant changes in joint angles and the associated p-values are shown in Table 5.3. A significant load effect was found for peak trunk flexion angle ($p=0.0001$). Females showed a 76% increase in peak trunk flexion between 0-20kg ($p=0.0002$) and a 111% increase between 0-40kg ($p<0.0001$). Similarly, males showed an increase of 73% between 0-20kg ($p=0.001$) and 107% between 0-40kg ($p<0.0001$) for trunk flexion. There were no significant differences found between males and females.

The effect of sex was found to be significant for peak pelvic rotation angle ($p=0.04$), however, no other significant changes were seen. In females, the decrease of peak pelvic obliquity approached significance between 0-20kg (46% decrease, $p=0.06$). Both sexes exhibited a 68% increase in peak pelvic tilt angle between the 0kg and 40kg load conditions ($p=0.03$ for females, $p=0.05$ for males).

Results of the ANOVA showed the effect of sex to be significant for peak hip extension angle ($p=0.03$). Both males and females were found to decrease peak hip extension between 0-40kg (136% decrease, $p=0.04$ for males and 258% decrease, $p=0.02$ for females) and increase hip flexion (37% increase, $p=0.0004$ for males and 40% increase, $p=0.02$ for females). This resulted in the hip being more flexed over the gait cycle with the addition of load, but did not cause significant changes to the total hip range of motion.

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Trunk Flexion	↑ 0.0002	↑ ≤0.0001	-	↑ 0.001	↑ ≤0.0001	-
Obliquity	↓ (0.06)	-	-	-	-	-
Tilt	-	↑ 0.03	-	-	↑ 0.05	-
Hip Flexion	-	↑ 0.02	-	-	↑ 0.004	-
Hip Extension	-	↓ 0.02	-	-	↓ 0.04	-

Table 5.3: Significant changes in joint angles with p values. ↑ indicates increase in peak value between load conditions, ↓ indicates decrease in peak value between load conditions.

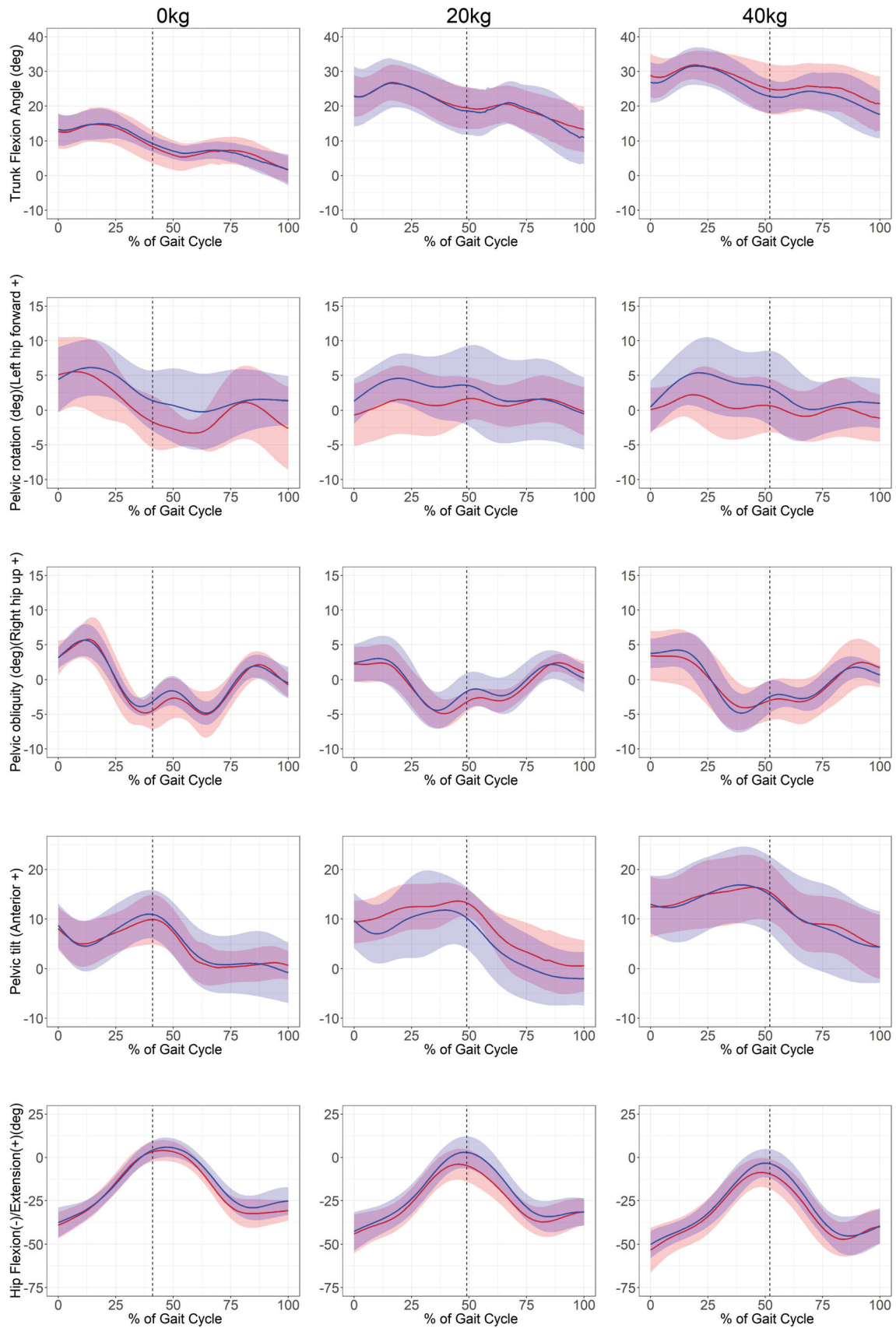


Figure 5.2: Joint angles (deg) of the trunk, hip and pelvis by load and sex (red line-females, blue line-males). No significant differences were found between sexes.

5.3.3 Joint Torques

Hip torques normalised to BW are presented in Figure 5.3. Significant changes in hip torque and associated p-values can be found in Table 5.4. In the frontal plane, the main effect of sex was significant for peak abduction torque ($p=0.05$), however no other significant differences were found between sexes or load conditions. In the transverse plane, the effects of load ($p=0.0003$) and the interaction effect of load and sex ($p=0.02$) were found to be significant for peak internal hip torque. Males also showed a very significant increase in peak internal torque between 0-40kg (128% increase, $p=0.0002$) and a smaller but still significant increase between the 20kg and 40kg load conditions (53% increase, $p=0.02$).

In the sagittal plane, the main effects of sex were significant for peak flexor and extensor torque ($p=0.0006$ and $p=0.05$ respectively). In addition, the main effect of load approached significance ($p=0.06$). As shown in Table 5.4, both females and males saw a significant increase in peak flexor torque between 0-40kg (39% increase, $p=0.01$ for females and 47% increase, $p=0.004$ for males). Males also showed an additional increase between 0-20kg (42%, $p=0.01$). The only significant difference between sexes was found for peak flexor torque when carrying 0kg ($p=0.03$), where females had a greater normalised peak torque.

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Internal	-	-	-	-	↑ 0.0002	↑ 0.02
Flexor	-	↑ 0.01	-	↑ 0.01	↑ 0.004	-

Table 5.4: Significant changes in peak hip torque with p values. ↑ indicates increase in peak value between load conditions.

Total joint torques for the lower limb are shown in Figure 5.4. The percentage of overall ankle, knee and hip torque contributions to total lower limb torque were compared between sexes and load conditions, as shown in Table 5.5. The hip was found to contribute significantly more torque in females than males when carrying 0kg ($p=0.02$), 20kg ($p=0.01$) and 40kg ($p=0.05$). Males meanwhile contributed significantly more torque at the knee while carrying 20kg ($p=0.02$). There were no sex-specific differences for ankle torque contribution for any load condition, however both males and females showed an increase between 0-40kg ($p=0.01$ for females and $p=0.05$ for males). In the hip, females exhibited a significant decrease in torque contribution between 0-20kg ($p=0.03$) and 0-40kg ($p=0.0004$) while males only showed a significant decrease when 40kg was carried ($p=0.008$). Results

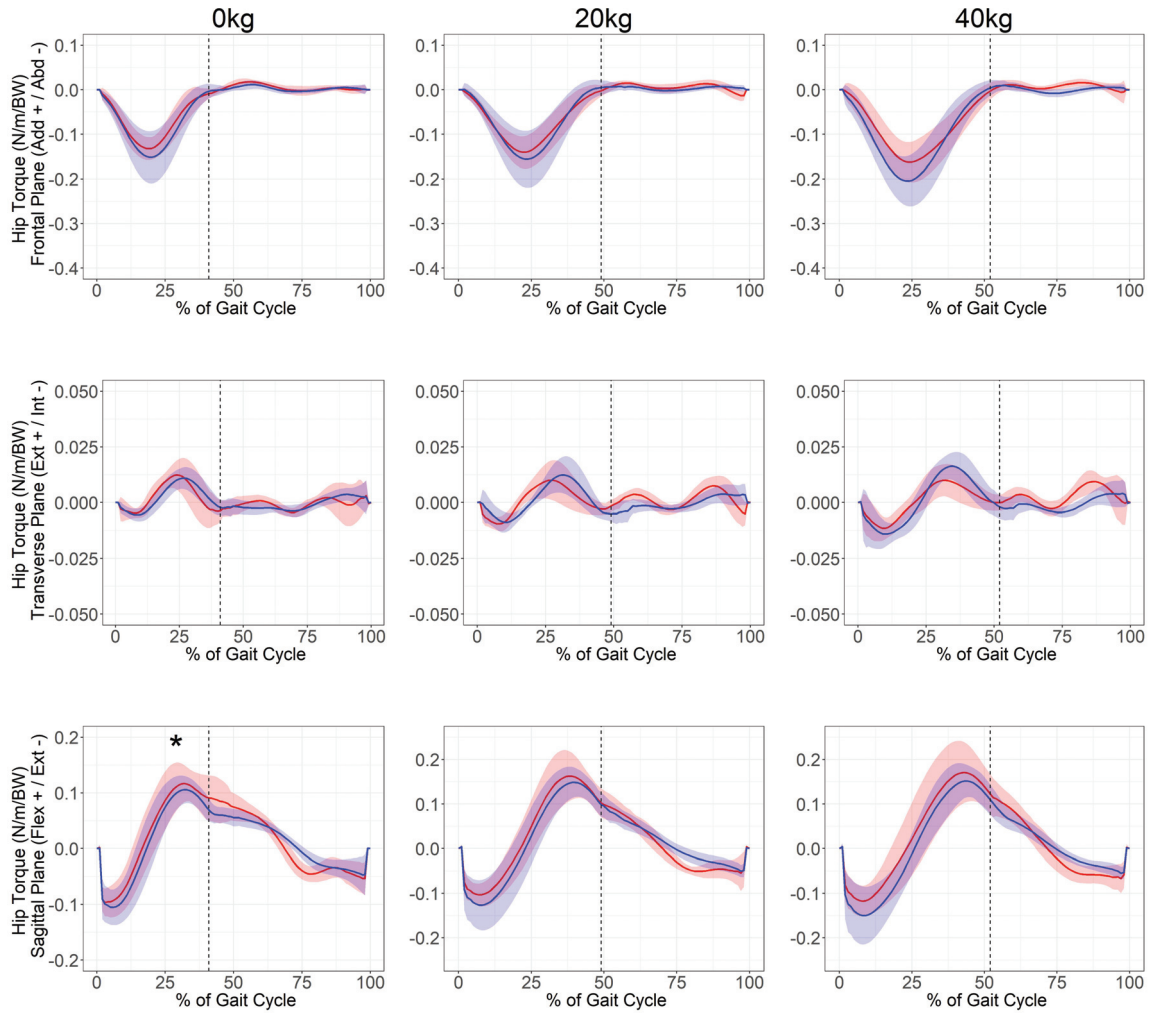


Figure 5.3: Joint torques by load and sex (red line- females, blue line-males), * indicates significance between sexes for $p \leq 0.05$. Dashed vertical line represents toe off.

of the ANOVA showed that the main effect of load was significant for ankle torque contribution ($p=0.03$). The main effect of sex was found to be significant for the knee ($p=0.003$) and highly significant for the hip contributions ($p=0.0003$).

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Ankle(%)	9(1.9)	12(2.6)	13(3.1) ^a	10(2.3)	11(3.0)	13(2.7) ^a
Knee(%)	31(2.3)	31(2.7) ^d	32(3.1)	33(3.9)	35(3.1) ^d	34(3.4)
Hip(%)	58(1.8) ^d	55(2.4) ^{ad}	54(2.1) ^{Ad}	55(2.6) ^d	52(2.5) ^d	51(3.5) ^{Ad}

Table 5.5: Joint torque contributions as a percentage of total lower limb torque, mean (SD). ^a indicates significance from 0kg load condition, ^b indicates significance from 20kg condition, ^d indicates significance between sexes for $p \leq 0.05$. ^{A,B,D} for $p < 0.005$

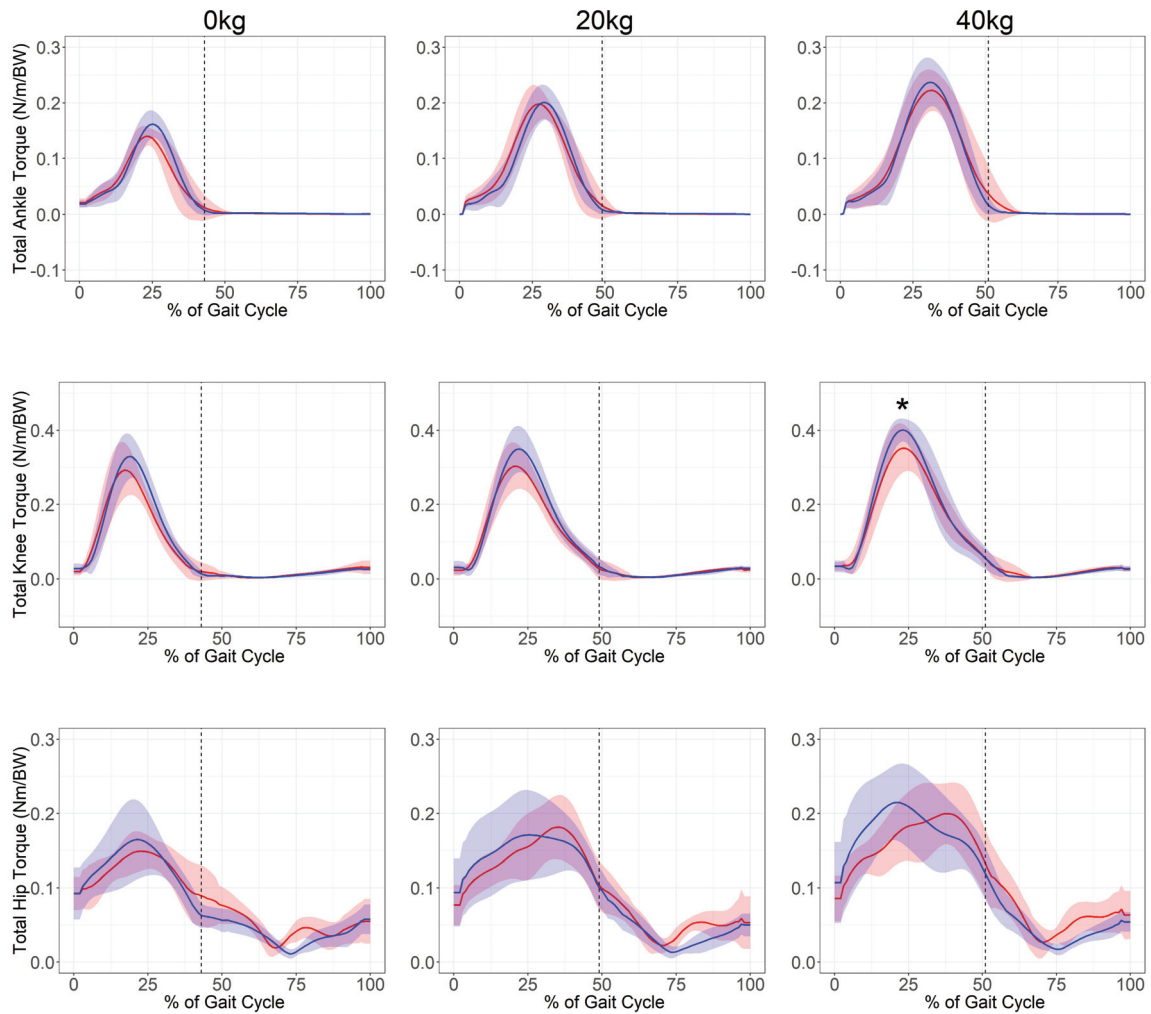


Figure 5.4: Total ankle, knee and hip joint torques by load and sex (red line- females, blue line-males), * indicates significance between sexes for $p \leq 0.05$. Dashed vertical line represents toe off.

5.3.4 Muscle Forces

Peak muscle forces normalised to body weight are presented in Table 5.6. Of the 22 muscles in Freebody which cross the hip, only 16 showed significant changes and have been included in Table 5.6. Muscles that showed no changes were the semitendinosis, semimembranosis, biceps femoris, rectus femoris, piriformus and gluteus minimus.

The main effect of sex was found to be significant for 14 muscles, most notably in the adductor magnus ($p < 0.0001$) and the gluteus maximus ($p = 0.0002$). The main effect of load was significant for 7 muscles, most notably in the obturator internus ($p = 0.003$). The interaction effect of load and sex was not found to be significant for any of the muscles crossing the hip. Compared to males, females

were found to have significantly greater normalised force in 5 muscles. Similarly, males were found to have significantly greater normalised force in 5 muscles when compared to females. The greatest absolute differences in muscle forces between sexes were found in the gluteus medius, gluteus maximus and the adductor magnus for the 20kg and 40kg load conditions.

In males, significant increases in force were found in 6 muscles between the 0kg and 20kg loads. The most significant of these were the tensor fasciae latae (55% increase, $p=0.01$) and the pectineus (71% increase, $p=0.02$). Between the 20kg and 40kg conditions, only the gluteus maximus (32% increase, $p=0.04$) and the gemellus superior (43% increase, $p=0.02$) displayed significant changes. Significant increases in force were found in 12 muscles between the 0kg and 40kg loads. The muscles which showed the most significant increases from the unloaded condition were the adductor brevis (110% increase, $p=0.0005$) and the obturator externus (115% increase, $p=0.0007$).

In females, 9 muscles displayed significant increases in force between the 0kg and 40kg loads. The most significant of these was found to be the gluteus maximus (76% increase, $p=0.0003$). No muscles showed any significant differences between the 0-20kg or 20-40kg conditions.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Glute. max.	0.86 (0.2) ^d	1.18 (0.4)	1.27 (0.3) ^{Ad}	1.18 (0.3) ^d	1.44 (0.4)	1.9 (0.4) ^{Abd}
Glute. med.	2.02 (0.6)	1.79 (0.4)	1.93 (0.7) ^d	2.28 (0.7)	2.27 (0.6)	2.79 (0.7) ^d
Psoas maj.	0.49 (0.1) ^d	0.61 (0.2)	0.64 (0.2)	0.34 (0.1) ^d	0.52 (0.2) ^a	0.52 (0.1) ^a
Iliacus	0.48 (0.1) ^d	0.53 (0.2)	0.56 (0.2)	0.33 (0.1) ^d	0.51 (0.2) ^a	0.51 (0.1) ^a
Sartorius	0.41 (0.08) ^D	0.53 (0.2)	0.56 (0.1) ^{ad}	0.28 (0.09) ^D	0.53 (0.3) ^a	0.42 (0.1) ^d
Add. mag.	0.45 (0.02) ^d	0.47 (0.06) ^D	0.56 (0.07) ^D	0.69 (0.03) ^d	0.83 (0.05) ^D	0.95 (0.05) ^D
Add. brev.	0.06 (0.03) ^d	0.11 (0.04)	0.14 (0.02) ^a	0.09 (0.03) ^d	0.14 (0.02)	0.19 (0.03) ^A
Add. long.	0.05 (0.02)	0.06 (0.02)	0.06 (0.02)	0.04(0.01)	0.07 (0.03) ^a	0.06 (0.02) ^a
Gracilis	0.24 (0.1)	0.26 (0.2)	0.30 (0.2)	0.13 (0.08)	0.26 (0.1)	0.29 (0.1) ^a
Pectineus	0.31 (0.1) ^d	0.37 (0.1)	0.37 (0.1)	0.22 (0.07) ^d	0.38 (0.2) ^a	0.37 (0.1) ^a
T.f.l.	0.07 (0.02) ^d	0.09 (0.03)	0.10 (0.03) ^a	0.05 (0.01) ^d	0.08 (0.02) ^a	0.08 (0.02) ^A
Gemellus sup.	0.08 (0.03)	0.12 (0.05)	0.15 (0.04) ^A	0.11 (0.03)	0.13 (0.05)	0.19 (0.05) ^{Ab}
Gemellus inf.	0.02 (0.008)	0.03 (0.01)	0.04 (0.01) ^A	0.03 (0.008)	0.04 (0.01)	0.06 (0.04) ^a
Quad. fem.	0.31 (0.06)	0.39 (0.1)	0.42 (0.1) ^a	0.27 (0.09)	0.36 (0.1)	0.34 (0.1)
Obturator ext.	0.13 (0.06) ^d	0.25 (0.2)	0.36 (0.2) ^a	0.20 (0.06) ^d	0.29 (0.1)	0.42 (0.1) ^A
Obturator int.	0.02 (0.008)	0.03 (0.02)	0.04 (0.02) ^A	0.02 (0.007)	0.03 (0.01)	0.06 (0.04) ^a

Table 5.6: Normalised peak muscle forces (N/BW), mean (SD), *a* indicates significance from 0kg load condition, *b* indicates significance from 20kg condition, *d* indicates significance between sexes for $p \leq 0.05$, *A,B,D* for $p < 0.005$.

5.3.5 Hip Joint Contact Forces

Hip joint contact forces normalised to body weight (BW) are presented in Figure 5.5. Significant changes and associated p-values are shown in Table 5.7. The main effect of sex was significant for all components of the normalised hip JCF with p-values of between 0.00003 and 0.04. The main effect of load was only significant for the anterior-posterior ($p=0.004$) and the superior-inferior components ($p=0.02$). The interaction effect of load and sex was not significant.

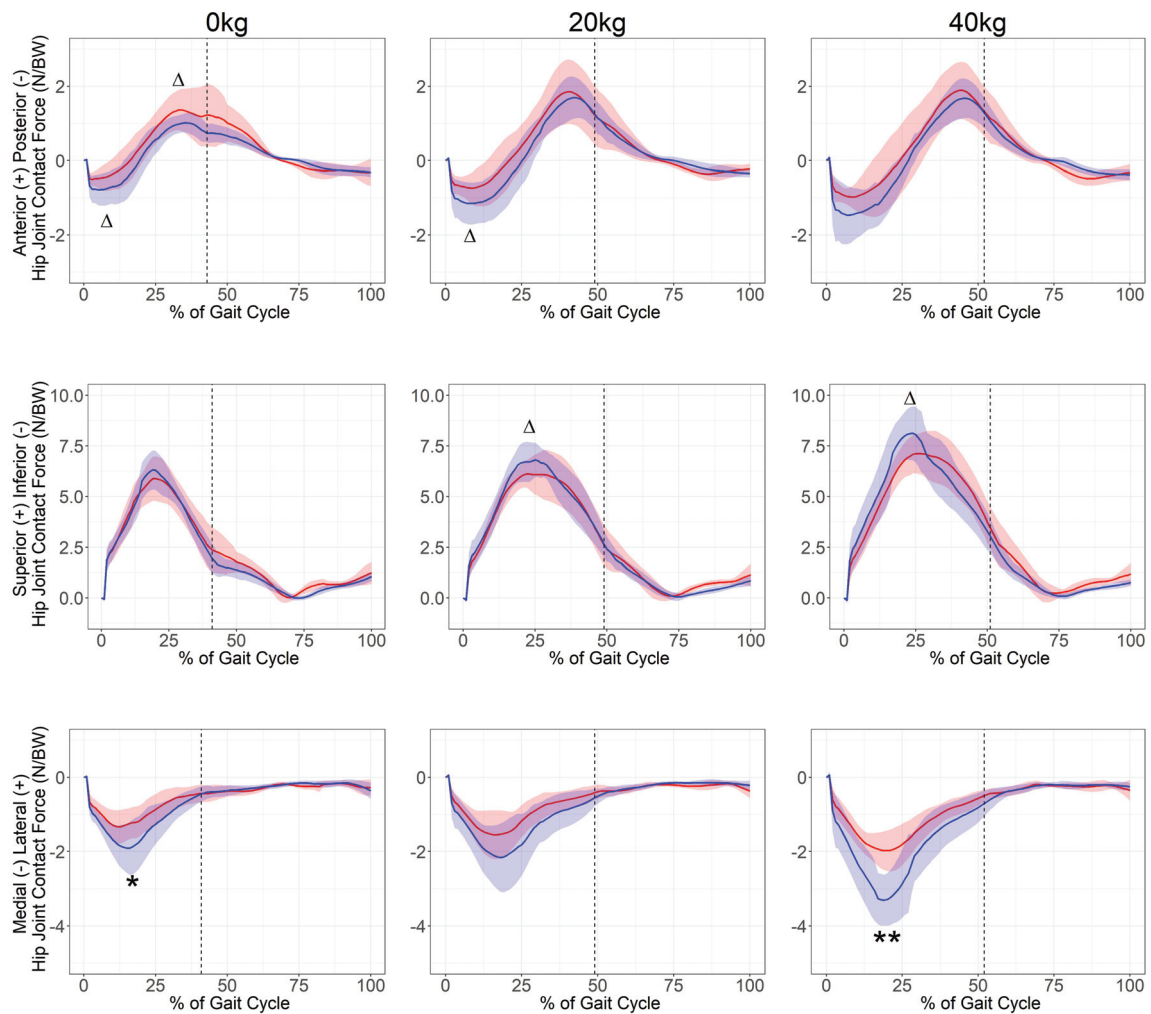


Figure 5.5: Hip joint contact forces by load and sex, red (female) blue (male), * indicates significance between sexes for $p \leq 0.05$, ** for $p \leq 0.005$. Δ indicates approaching significance ($0.05 < p \leq 0.07$). Dashed vertical line represents toe off.

In males, significant changes were seen in the anterior-posterior (peak) forces between 0-20kg (69% increase, $p=0.008$) and 0-40kg (70% increase, $p=0.007$), and in the anterior-posterior minimum force between 0-40kg (77% decrease, $p=0.03$). Males also showed very significant changes between the 0kg and 40kg load

conditions in the superior-inferior (32% increase, $p=0.0009$) and medial-lateral components (67% decrease, $p=0.003$). A further significant decrease was seen in the medial-lateral force between 20-40kg (47% decrease, $p=0.02$).

Females displayed significant increases in JCF force in the posterior (84% increase, $p=0.004$) and medial directions (50% increase, $p=0.02$). They also showed a significant increase in peak JCF in the superior-inferior direction (22% increase, $p=0.04$). Compared with males, females had significantly lesser medial-lateral (trough) force when carrying 0kg ($p=0.04$) and 40kg ($p=0.0007$). As shown in Figure 5.5, 5 of the other peak forces measured showed a trend for significance between sexes with p -values of between 0.05 and 0.07.

	Female			Male		
	0-20kg	0-40kg	20-40kg	0-20kg	0-40kg	20-40kg
Anterior-posterior (peak)	-	-	-	↑ 0.008	↑ 0.007	-
Anterior-posterior (trough)	-	↓ 0.004	-	-	↓ 0.03	-
Superior-inferior (peak)	-	↑ 0.04	-	-	↑ 0.0009	-
Medial-lateral (trough)	-	↓ 0.02	-	-	↓ 0.003	↓ 0.02

Table 5.7: Significant changes in peak hip joint contact force with p values. ↑ indicates increase in peak value between load conditions. ↓ indicates decrease in peak value between load conditions.

As described in Chapter 3, the resultant mean hip contact forces over the gait cycle are shown in Figure 5.6 and Table 5.8. Each arrow represents the magnitude and direction of the hip JCF at each percentage of the gait cycle from 0-100% (normalised from heel strike to heel strike). A diagram describing the angles at which the peak JCF occurs is shown in Figure 5.7. In the transverse plane females displayed a significantly more anteriorly oriented peak JCF, α , when carrying 40kg compared with males ($p=0.05$). The ANOVA also showed the effect of sex to be significant for α ($p=0.007$).

In the frontal plane, significant differences in β were seen between sexes when

carrying 0kg ($p=0.04$) and 40kg ($p=0.009$), with females having a less medially angled peak force. Males also showed a significant decrease in β between the 20kg and 40kg load conditions. The main effect of sex was also found to be significant for β ($p=0.002$). In the sagittal plane, γ at 0kg approached significance between sexes ($p=0.07$) and the main effect of sex from the ANOVA was again significant ($p=0.01$).

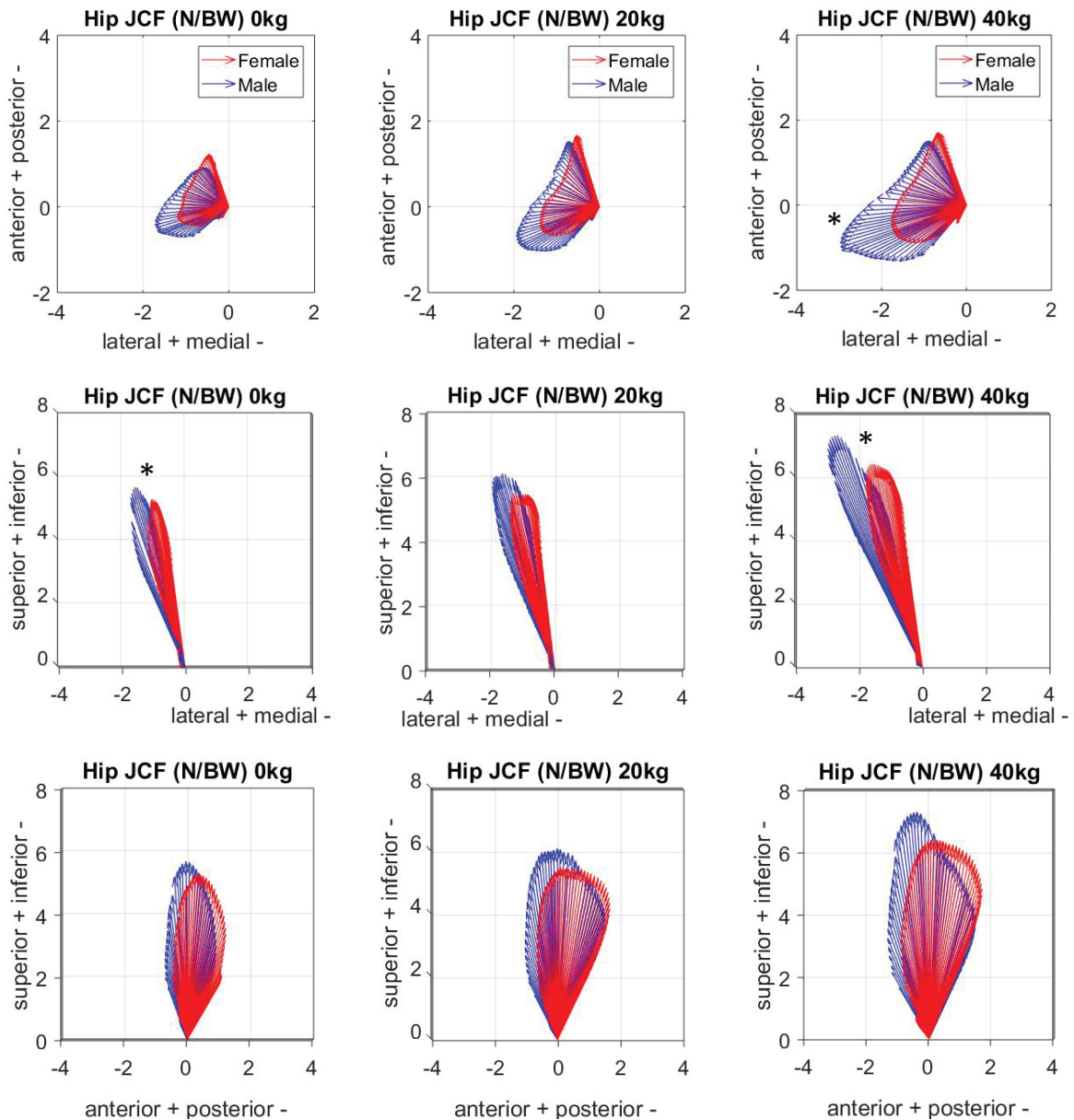


Figure 5.6: Hip joint contact forces by load and sex, coordinates (0,0) represent hip joint centre. * indicates significance between sexes for $p \leq 0.05$.

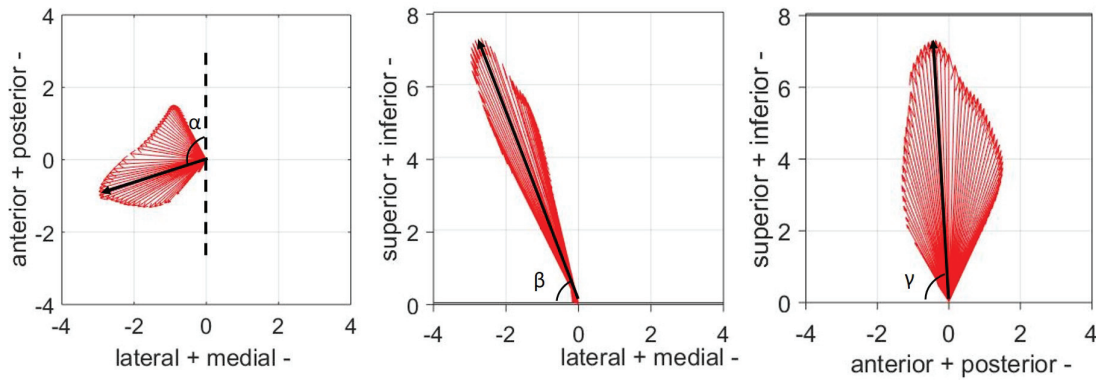


Figure 5.7: Hip JCF vector angle diagram (deg)

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
α	65.3 (32)	61.2 (41)	75.0 (34) ^d	87.4 (29)	81.7 (32)	100.9 (12) ^d
β	79.4 (4) ^d	78.1 (7)	76.1 (6) ^d	74.1 (6) ^d	74.8 (6)	69.4 (3) ^{bd}
γ	94.7 (7)	95.3 (12)	91.7 (10)	88.7 (6)	91.0 (10)	85.8 (5)

Table 5.8: Hip joint contact force vector angles (deg), a indicates significance from 0kg load condition, b indicates significance from 20kg condition, d indicates significance between sexes for $p \leq 0.05$.

5.3.6 Key Findings

In summary, the key findings of this study were:

1. Both sexes ran with similar gait patterns when unloaded. When carrying load, females had significantly lesser stride length compared with males.
2. Both males and females significantly increased trunk flexion, pelvic tilt and hip flexion when carrying load.
3. Females generated a significantly greater proportion of torque at the hip compared with males. Males generated more torque at the knee when carrying 20kg.
4. Females showed no increase in muscle force between 0-20kg after reducing their stride length. Males increased stance duration but still showed muscle force increases between 0-20kg.
5. The greatest absolute differences in normalised muscle forces between sexes were found in the glutes and adductor magnus when carrying load.
6. Females had a significantly more anteriorly and superiorly angled peak hip JCF.

5.4 Discussion

Spatiotemporal effects

The results presented in Section 5.3.1 show that males and females exhibit significantly different running patterns when carrying load. Although neither sex significantly altered their running speed, there was a significant difference between males' and females' speed when carrying 20kg, with females running slower. Silder et al. (2015) found that subjects' preferred unloaded running speed was approximately 3.34 m/s. The subjects in this study ran at a slower pace, possibly due to the limited walkway length and thus short running time. Teunissen et al. (2007) did not see significant changes in spatiotemporal parameters until 30% BW was carried, while Carretero-Navarro et al. (2019) observed changes in males carrying 10% and 20% BW. Although difficult to directly compare, this study found gait changes in males occurred between 0-20kg but not between 20-40kg. As in this study, both Teunissen et al. (2007) and Carretero-Navarro et al. (2019) observed an increase in stance phase in males. Liew et al. (2016) and Willy et al. (2016) again found that increasing load increased stance phase duration in males, however they observed that it also affected stride length, in contrast with the results found here.

Results of this study show that males adapted to carrying load by increasing stance time while females reduced stride length. This suggests that males and females adopt different methods in response to load carriage, possibly to reduce internal loads in the leg or as a result of sex-specific muscle weaknesses. In addition, females ran with significantly shorter stride length when carrying load compared with males. This was true even when normalised to leg length, suggesting that it was not solely caused by anthropometric differences between sexes. This also supports the theory that it may not be appropriate to have males and females running with the same fixed cadence and stride. A similar result was reported by Lobb et al. (2019), who found that males and females alter leg stiffness differently in response to changing stride length.

Both males and females showed a significant reduction in flight time with the addition of load, in agreement with the literature (Teunissen et al. 2007; Carretero-Navarro et al. 2019). When carrying 40kg females did not produce the propulsive force necessary to leave the ground and instead stayed in double support, possibly to reduce injury risk or due to muscle weakness. Males' mean flight time was 0 ± 0.03 seconds when carrying 40kg, indicating that they were able

to maintain single stance but also did not produce the required force to leave the ground. Finally, there were no significant changes in spatiotemporal parameters between the 20kg and 40kg load conditions for either sex. This suggests that gait adaptations are made at lower loads and subjects then either cannot or will not alter them further.

Kinematic effects

As with the spatiotemporal results, significant changes in joint angles were only observed between 0-20kg and no further changes were seen between 20-40kg. It is interesting that males and females showed significant differences in spatiotemporal parameters when carrying load, but not in joint angles. As expected, this study found that trunk flexion angle significantly increased with the addition of load for both males and females. This is in agreement with Brown et al. (2014), who found that trunk flexion increased in males carrying loads of 20kg and 40kg. It has been suggested that this increase in trunk flexion is to bring the center of mass back over the feet and reduce the propulsive force required for forward motion (Brown et al. 2014). It is surprising that females did not adopt a more forward flexed stance considering their lesser upper body mass but implies that females could experience greater demand on their core musculature in order to obtain the same trunk flexion as males when carrying load.

Xu et al. (2017) found that hip flexion increased as loads increased from 10% to 30% BW in females. The results of this study however suggest that significant increases in hip flexion, extension and pelvic girdle motion do not occur until heavier loads are carried. Although the clinical significance of altered hip kinematics due to load carriage is not clear, differences in hip joint angles have been reported between healthy and injured runners (Esculier et al. 2015). No significant differences were seen in hip flexion or pelvic obliquity between males and females, in agreement with the literature describing sex-specific differences in unloaded running (Ferber et al. 2003; Chumanov et al. 2008). Interestingly, there was no change in pelvic rotation or hip range of motion corresponding to females' decrease in stride length. This could be due to the rucksack constraining the motion of the pelvis, causing females to adopt a different method to reduce stride length. Additionally, males and females increased trunk flexion but not pelvic tilt between 0-20kg suggesting that the forward bending was occurring in the spine which has implications for increased back injury risk.

Kinetic effects

This study found that only males increased hip flexor torque between 0-20kg, possibly because females reduced their stride length resulting in reduced forces at the hip. Both sexes significantly increased hip flexor torque between 0-40kg, although in females this was less significant as they again had a reduced stride length compared with the unloaded condition. These findings are supported by the literature. Xu et al. (2017) found peak flexion moment increased by 10% when subjects carried 30% BW and Brown et al. (2014) found that, in males, hip flexion moment increased between 6-20kg but, as in this study, did not show a further increase between 20-40kg. When running with load this study found no significant differences between males and females, despite females' significantly lesser stride length and speed. In addition, when running unloaded (where there were no sex-specific differences in gait parameters) females displayed a significantly higher peak flexion torque compared with males. This suggests that females may have inherently higher torque at the hip. However these results are in contrast with Ferber et al. (2003) who found no differences in hip torque between sexes in unloaded running.

As in Chapter 3, females were found to generate a significantly higher proportion of torque at the hip, suggesting that they could be more hip dominant than males. Although previous studies did not consider joint dominance when reporting sex-specific differences in running, they did find that females displayed greater hip joint moments and work in the frontal and transverse planes (Ferber et al. 2003; Gehring et al. 2014). In this study both sexes displayed significant redistribution of torque production with the addition of load (proximal to distal). This is in contrast with the literature, where no redistribution of joint powers during steady-state running was found when changing load or speed (Brown et al. 2014; Farris and Sawicki 2012). In addition, Farris and Sawicki (2012) found that the ankle contributed most to total positive joint power when running compared with the knee and hip. Although direct comparison is not possible, this study found that the ankle contributed least to total lower limb torque and it is hypothesised that this was due to the military boots constraining flexion/extension of the ankle, reducing its propulsive ability and requiring the knee and hip to contribute more torque (Kulmala et al. 2016).

Significant differences in force were observed in 9 muscles between sexes for the unloaded condition. However when carrying 20kg, where females ran with significantly shorter strides and speed, there were only differences in the adductor

magnus, with females displaying significantly lesser normalised force. This suggests that females must run slower and with shorter strides to achieve the same normalised muscle forces as males. In addition, females did not see an increase in any muscle forces between 0-20kg while males did, suggesting that reducing stride length is more effective than increasing stance time at reducing internal forces in the leg. There was no sex-specific difference in trunk forward flexion for any load condition, however at 0kg females had significantly greater iliopsoas muscle forces, corresponding to their greater peak hip flexor torque. Males were found to have significantly greater peak gluteus maximus force when running unloaded, in agreement with Vannatta and Kernozek (2018) but not with Willson et al. (2012). Neither sex showed significant changes in rectus femoris or hamstring muscle force, possibly due to reducing flight time and hence reducing the propulsive force required by these muscles.

Xu et al. (2017) reported peak hip contact forces in females of between approximately 11 BW (running unloaded) and 16 BW (running with 30%BW load), much greater than those found in this study. This may be due to their subjects running faster ($3.34 \pm 0.22\text{m/s}$ compared to $2.20 \pm 0.27\text{m/s}$ in this study), and also wearing a weight distributed vest rather than a rucksack, as load position has been shown to affect gait (Simpson et al. 2012a). This difference in results highlights the potential importance of speed and load distribution on hip joint contact forces. Giarmatzis et al. (2015) found peak hip JCFs of between approximately 5-10 BW for unloaded running at similar speeds which is within the range of this study (4-8 BW).

As discussed in Chapter 3, Correa et al. (2010) found that the iliopsoas muscles contribute to hip contact force during walking, in particular the anterior component. They also found that the gluteal muscles contributed significantly to the medial JCF component. This study found that during running, when females displayed greater iliopsoas muscle forces, they also showed significantly greater anterior hip JCF. Similarly, where males showed greater gluteus maximus force, they also showed significantly greater peak medial JCF compared with females. This suggests that similar muscles contribute to JCF components when walking and running. Compared to males, females generally showed less significant increases in JCF with increasing load, and were able to prevent anterior peak force from increasing even with the addition of load, again highlighting the importance of altering gait parameters to limit internal forces in the leg when running with load and supporting the argument that it may be inappropriate to have all soldiers

run with the same gait.

In summary, this study found that running with load caused spatiotemporal, kinematic and kinetic changes in both males and females, with implications for increased injury risk. A more forward flexed posture could result in increased pressure in the intervertebral discs, leading to chronic back pain (Smith et al. 2006). Although the clinical significance of altering hip and pelvic girdle angles is not clear, it has been reported that increased hip rotation and adduction could be predictors of future injury (Noehren et al. 2007; Khamis and Yizhar 2007). In addition, increased muscle forces due to load carriage have been associated with greater compressive forces on the spine (Cook et al. 1987). The addition of load also resulted in sex-specific differences in spatiotemporal and kinetic parameters. Females preferentially shortened their stride length when carrying load, resulting in increased cadence, which Xu et al. (2017) suggest may lead to an increase in the cumulative stress on the MSk system. Furthermore, research has found that significant weakness of the hip abductors and external rotator muscles are related to injury in female collegiate athletes (Cichanowski et al. 2007). Strength training programmes designed to strengthen hip flexor and pelvis stabilizing muscles (i.e. the gluteus medius) have been shown to be effective in preventing injury in basic training (Henning et al. 2011; Harman et al. 1997). Results of this study suggest that there may also be inherent differences in the direction of the hip JCF between males and females. It is not yet known if there is any clinical significance of JCF direction while running, however it has been found to vary significantly both within a group of subjects with hip replacements (Bergmann et al. 2016), and between healthy and pathological subjects while walking (Ng et al. 2018).

Limitations

The full limitations of this study are detailed in Chapter 3 Section 3.4. The short length of the walkway used was a particularly significant limitation here as subjects were only able to run over 5 metres.

5.5 Conclusions

The results of this study found that males and females displayed similar spatiotemporal parameters when running with no weight, however they did display significant differences in hip joint kinetics. This suggests that males' and females' bodies manage internal loads differently and may require different strength training regimes. When carrying load, females significantly reduced stride length,

either to reduce the risk of injury of because of fatigue. They also ran with shorter stride length and lower speed compared with males, supporting the argument that soldiers should not be forced to run with a predetermined gait. The addition of load also caused significant changes in trunk and pelvic motion of all subjects and further research should be done to investigate the long term effects of this.

Across all load conditions females generated a significantly greater proportion of torque at the hip, suggesting that females may be inherently more hip dominant. Finally, by significantly reducing their stride and the proportion of torque produced at the hip, females were able to prevent significant increases of hip torque and JCF between the 0kg and 20kg load conditions. However, the results of this study suggest that for loads over 20kg this is no longer effective. This study concludes that there may be inherent differences between males and females both when running unloaded and in how they adapt their gait to run with load. The next chapter will consider lower limb electromyographic data when walking and running in order to investigate the timings of muscle activations and the subject workload between sexes and load conditions.

Chapter 6

Electromyography of lower limb muscles when walking and running with body borne military-relevant loads

Results of the previous chapters have highlighted that females are more hip dominant than males, and reduce their stride length in response to increasing carried load, leading to significant differences in estimated muscle forces. Investigation of measured muscle activity can provide further insight into these differences, as well as the subjective workload of the lower limb, which has been linked to MSk injury. This chapter therefore presents an analysis of the effects of load and sex on the activity patterns of four lower limb muscles when walking and running.

6.1 Introduction

Analysis of lower limb muscle activity has the potential improve our understanding of mechanisms that contribute to the development of musculoskeletal injuries during load carriage. Increased muscle strain in the lower limbs has been linked to overuse injuries in soldiers (Kaufman et al. 2000). Wearing military issue combat boots has been shown to increase tibialis anterior and rectus femoris muscle activity in soldiers walking on a treadmill (Schulze 2011). In civilians, modified activity of the tibialis anterior has been linked to an increased risk of shin splints (Mizrahi et al. 2000). Runners with patellofemoral pain syndrome (PFPS) have presented altered rectus femoris, soleus and gluteus medius activity (Esculier et al. 2015; Ribeiro et al. 2010). Furthermore, modified gastrocnemius and rectus femoris activity have been found to promote the development of Achilles tendinopathy in runners (Azevedo et al. 2008).

Chumanov et al. (2008) compared muscle activity in males and females when walking and running unloaded. They found that females displayed greater gluteus maximus muscle activity compared with males, and that females' gluteus medius activity significantly increased with running speed, while males' did not. Similar differences in gluteal muscle activity between males and females were found by Willson et al. (2012). Another study also found distinct sex-specific differences in EMG signals when running with no load, namely that females had greater gastrocnemius and tibialis anterior activity just after heel strike, and that males and females showed different activation timings of the hamstrings and quadriceps muscles later in the gait cycle (Von Tscharnner and Goepfert 2003). Similar results were reported by Wall-Scheffler and Myers (2017), who observed that males increased hamstring activation earlier in the gait than females during walking, and that females showed significantly higher gluteus medius activation at the beginning of swing phase.

Despite these differences between males and females, studies investigating muscle activity during load carriage have largely considered only male subjects. Harman et al. (1992) measured EMG activity in the tibialis anterior, gastrocnemius, rectus femoris and biceps femoris of male soldiers carrying between 6kg and 47kg. They found that activity increased in the gastrocnemius and rectus femoris as carried load increased. A more recent study reported increases in gastrocnemius activity in males carrying loads up to 33kg, but no further change when heavier loads were carried (Harman et al. 2000). Carrying loads higher up on

the back has been shown to increase erector spinae and trapezius EMG amplitude in males carrying 20kg (Bobet and Norman 1984). Cook et al. (1987) and Ghori and Luckwill (1985) included both males and females in their studies, however they did not consider the effect of sex in their results. As in the studies involving only male subjects, Cook et al. (1987) found that lumbar paraspinal EMG activity decreased when carrying low loads of up to 20% BW. Ghori and Luckwill (1985) found that quadriceps muscle burst duration increased but hamstring muscle burst duration did not when loads of 50% BW were carried compared to walking unloaded. Simulations of the lower limb MSk system during walking with 25% BW load suggest that the quadriceps and gluteal muscles are the primary activators during weight acceptance (McGowan et al. 2010).

A study of female hikers found that carrying a load of 30% BW high on the back was associated with lower gastrocnemius activation levels than medium and low load positions (Simpson et al. 2012a). Previously, females had been shown to increase vastus lateralis and gastrocnemius activation, as well as biceps femoris muscle burst duration when carrying 20% BW to 40% BW (Simpson et al. 2011a). Very few studies have directly compared males' and females' muscle activity while carrying load. Silder et al. (2013) found no differences in the muscle activity patterns of males and females carrying between 10% BW and 30% BW. However, body weight normalised loads are not representative of realistic military load carriage and there is therefore limited knowledge of EMG muscle activity in female soldiers carrying load.

The results of Chapters 3 and 5 revealed that females significantly reduced their stride length and hip torque contribution in response to increasing carried loads. Analysis of EMG data may reveal if these adaptations result in changes to muscle activity patterns that could reduce energetic cost. The purpose of this study was therefore to investigate the effects of standardised, military-relevant load carriage and sex on lower limb muscle activation level and timing during walking and running.

6.2 Methods

6.2.1 Subjects

Nineteen healthy volunteers (Table 6.1) with no history of lower limb or back injury took part in this study. All subjects participated in the motion capture studies described in Chapters 3 and 5, with EMG data recorded simultaneously. Subjects were either current or ex-military personnel, or could complete the minimum military training entry fitness requirements of (a) 2.4km run in less than 14 minutes 58 seconds and (b) 40kg lift and carry for 60m. This study was granted ethical approval by the Joint Research Compliance Office and Imperial College Research Ethics Committee (Appendix A). All subjects gave written informed consent.

	Male	Female
<i>n</i>	8	11
Age (years)	27.3 (2.6)	29.2 (4.2)
Height (cm)*	180.6 (10.0)	168.8 (6.9)
Mass (kg)*	81.4 (13.4)	65.7 (8.0)
Leg Length (cm)	93.0 (7.9)	86.9 (7.3)
Bi-iliac width (cm)	24.7 (2.3)	25.6 (1.8)
20kg load as %BW*	25.3 (4.3)	31.0 (4.3)
40kg load as %BW*	50.5 (8.5)	62.0 (8.7)

Table 6.1: Participant Data, * indicates significance between sexes for $p \leq 0.05$

6.2.2 Data Collection and Processing

Muscle activity data were recorded at 2160Hz using four wireless surface EMG sensors (Delsys Incorporated, Natick, MA, USA) during the motion capture studies described in Chapters 3 and 5. Briefly, participants walked and ran over a 5m walkway carrying (a)no load, (b)20kg rucksack and (c)40kg rucksack. Sensors were placed on the muscle bellies of the soleus, vastus lateralis, biceps femoris and gluteus medius according to SENAIM recommendations (Hermens et al. 2000) on the subject's right leg. These muscles were selected to represent hip flexor, extensor and abductor muscle groups, as well as the foot plantarflexors. It was not possible to record hip adductor muscle activity as the sensors were repeatedly dislodged by the left leg when walking and running. Similarly it was not possible to record abdominal muscle activity due to the location of the waist belt of the

rucksack.

The raw EMG data were full-wave rectified and filtered using a low-pass fourth order zero-phase-shift Butterworth filter with 4Hz cut-off frequency to obtain linear envelopes (Buchanan et al. 2004). Muscle activity over the entire gait cycle was calculated by integrating each signal and normalising the value relative to the subjects' unloaded trial to account for variability between subjects. A threshold of 10% of the value of the peak activation was used to determine when muscles were considered active and allow for muscle burst duration to be calculated from the linear envelope (Simpson et al. 2011a). Muscle burst duration was normalised to the subjects' unloaded trial. The percentage of the gait cycle at which the peak muscle activity occurred was also recorded for each of the four muscles.

6.2.3 Statistics

Normality of the data was confirmed using the Shapiro-Wilk test ($p > 0.05$). Results were tested for significance between each load condition and sex using two-way repeated measures ANOVA and the Tukey HSD test post-hoc. Results were considered significant where $p \leq 0.05$. Data are presented, unless otherwise stated, as the mean value $\pm$ one standard deviation.

6.3 Results

6.3.1 Walking

Integrated muscle activities normalised to the unloaded walking condition are presented in Figure 6.1. Results of the ANOVA revealed the main effect of load to be significant for the soleus ($p = 0.01$), vastus lateralis ($p \leq 0.0001$) and gluteus medius ($p = 0.02$). The main effect of sex and the interaction effect of load and sex were not found to be significant for any muscle. When carrying 20kg, females were found to significantly increase integrated muscle activity in the soleus ($p = 0.05$). When carrying 40kg, females displayed significant increases in integrated muscle activity for the soleus ($p = 0.05$), vastus lateralis ($p = 0.004$) and gluteus medius ($p = 0.04$) while males only showed an increase in the vastus lateralis ($p = 0.008$). The increase in vastus lateralis integrated muscle activity between 20kg and 40kg approached significance in males ($p = 0.07$). No significant differences were found

between males and females.

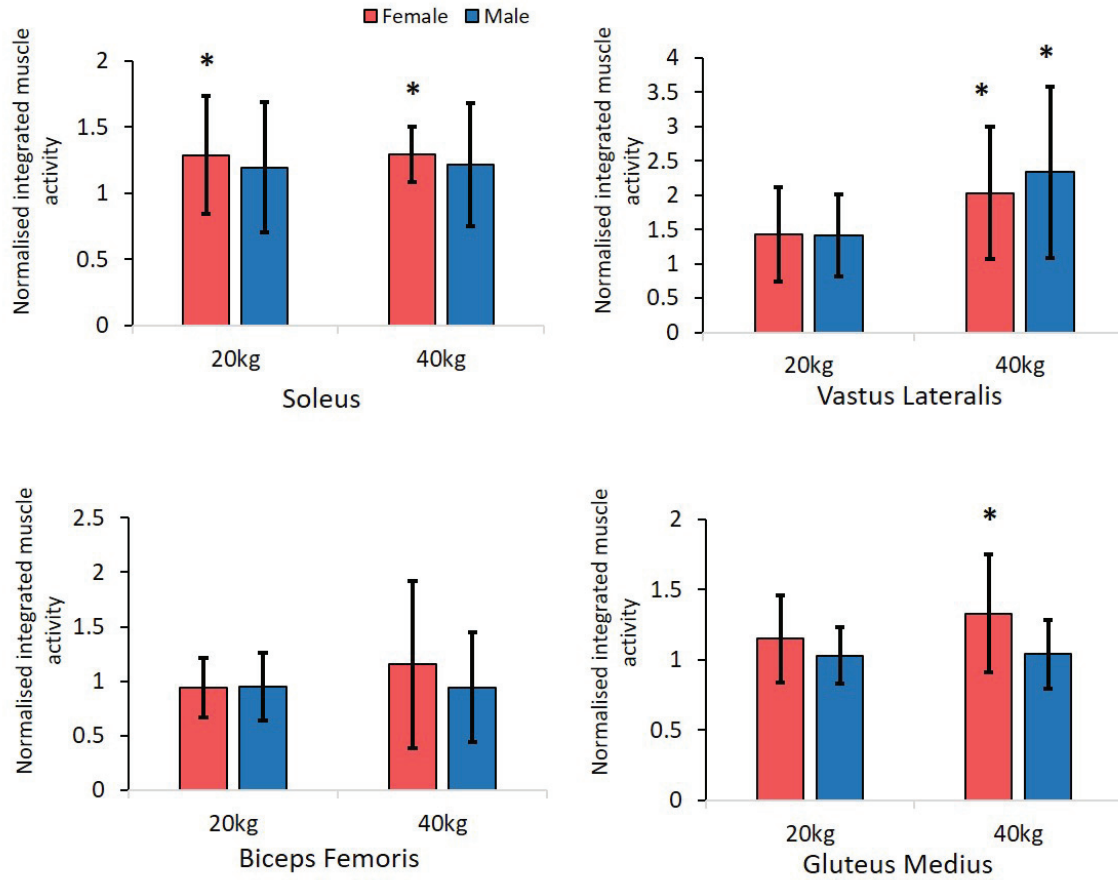


Figure 6.1: Integrated muscle activity over one gait cycle normalised to the unloaded walking condition. * indicates significance from the unloaded walking condition for $p \leq 0.05$.

Muscle burst durations normalised to the unloaded walking values are shown in Figure 6.2. The main effect of load was found to be significant for the soleus ($p=0.0004$), vastus lateralis ($p=0.0005$) and gluteus medius muscles ($p=0.02$). The effect of load tended towards significance for the biceps femoris ($p=0.06$). Similarly, the main effect of sex approached significance for the soleus ($p=0.06$). When carrying 20kg, females displayed significantly longer muscle burst duration than males in the soleus ($p=0.01$). In females, soleus muscle burst duration was significantly longer when carrying 40kg compared to the unloaded condition ($p=0.004$). Muscle burst duration in the gluteus medius also increased significantly between 20kg and 40kg in females ($p=0.02$). Males showed no significant changes in muscle burst duration across any of the load conditions.

The mean percentage of the gait cycle at which peak muscle activation

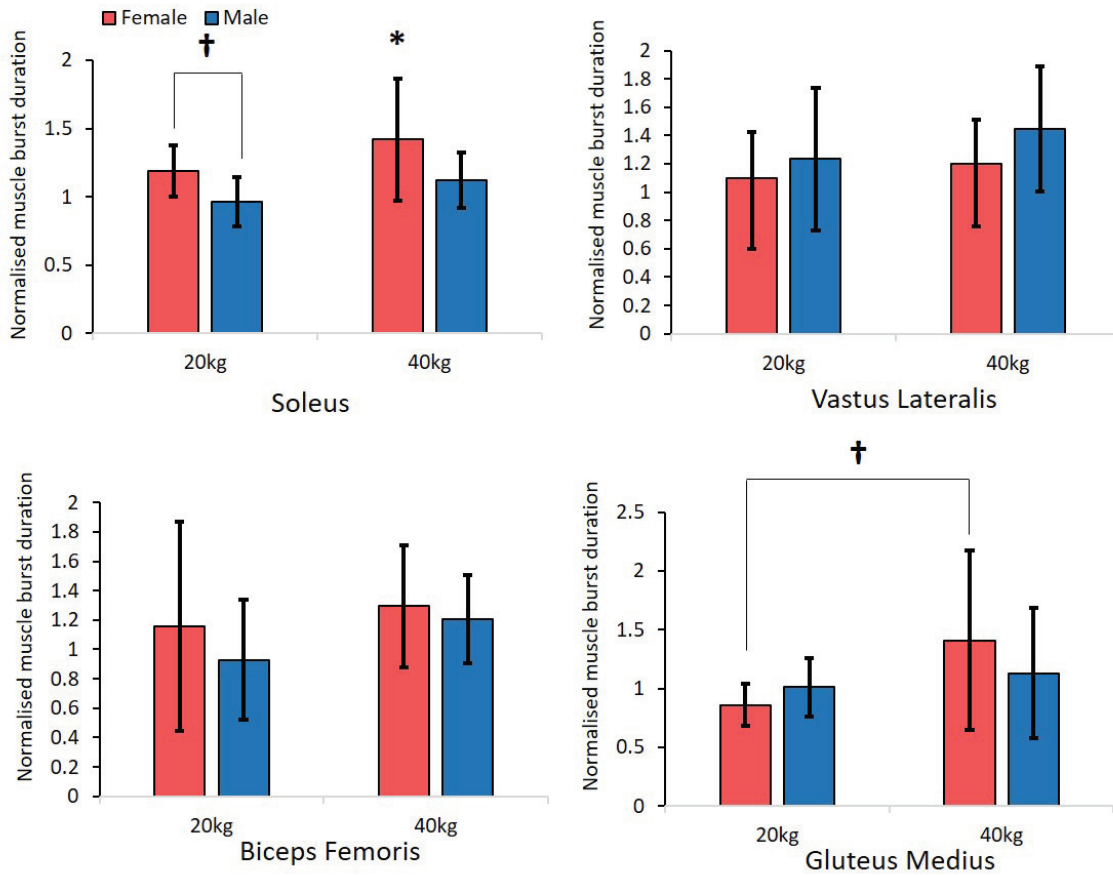


Figure 6.2: Normalised muscle burst duration. * indicates significance from the unloaded condition for $p \leq 0.05$. † indicates significance between load conditions or sexes for $p \leq 0.05$.

occurred is included in Table 6.2. Significant changes in peak activation timing were found only in the biceps femoris. The main effects of load ($p=0.05$) and sex ($p=0.04$) were both significant for the biceps femoris peak activation timing. In addition, there were significant differences in the timing of the biceps femoris peak activation between males and females when carrying no load ($p=0.05$) and 40kg ($p=0.03$), with females' peak activation occurring sooner in the gait cycle.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Soleus	63 (9)	60 (7)	60 (6)	63 (7)	61 (9)	63 (7)
Vastus Lateralis	23 (8)	25 (7)	26 (7)	28 (7)	29 (7)	30 (7)
Biceps Femoris	10 (6) ^d	15 (8)	13 (4) ^d	18 (11) ^d	22 (14)	21 (10) ^d
Gluteus Medius	18 (5)	19 (3)	21 (6)	23 (8)	23 (7)	21 (6)

Table 6.2: Mean (SD) percentage of the gait cycle at which peak muscle activation occurred during walking. ^d indicates significance between sexes ($p \leq 0.05$).

6.3.2 Running

Integrated muscle activities normalised to the unloaded running condition are shown in Figure 6.3. Results of the ANOVA revealed the main effect of load to be significant for the gluteus medius activation ($p=0.01$) and approaching significance for the biceps femoris activation ($p=0.06$). Females displayed a significant decrease in integrated muscle activity in the biceps femoris between the 20kg and 40kg load conditions. Males showed no changes for any of the four muscles and, in addition, there were no significant differences observed between sexes.

Normalised muscle burst durations during running are presented in Figure 6.4. The main effect of load was significant for the soleus ($p=0.008$) and the gluteus medius ($p=0.006$) from the ANOVA. The interaction effect of load and sex was significant for muscle burst duration in the biceps femoris ($p=0.04$), and approached significance for the soleus ($p=0.06$). Females significantly increased muscle burst duration for the soleus when running with 40kg compared to the unloaded condition. Compared with males, females also displayed a significantly shorter muscle burst duration in the biceps femoris when carrying 40kg. Males again showed no significant changes between load conditions for any of the four muscles.

Table 6.3 shows the mean percentage of the gait cycle at which peak muscle activation occurred during running. The main effect of load was significant for the soleus peak activation timing ($p=0.003$), and approached significance for the biceps femoris ($p=0.08$). In addition the main effect of sex approached significance for the gluteus maximus peak activation timing ($p=0.07$). The interaction effect of load and sex was significant for the soleus ($p=0.01$) and gluteus medius ($p=0.002$). In males, peak activation in the gluteus medius occurred significantly earlier in the

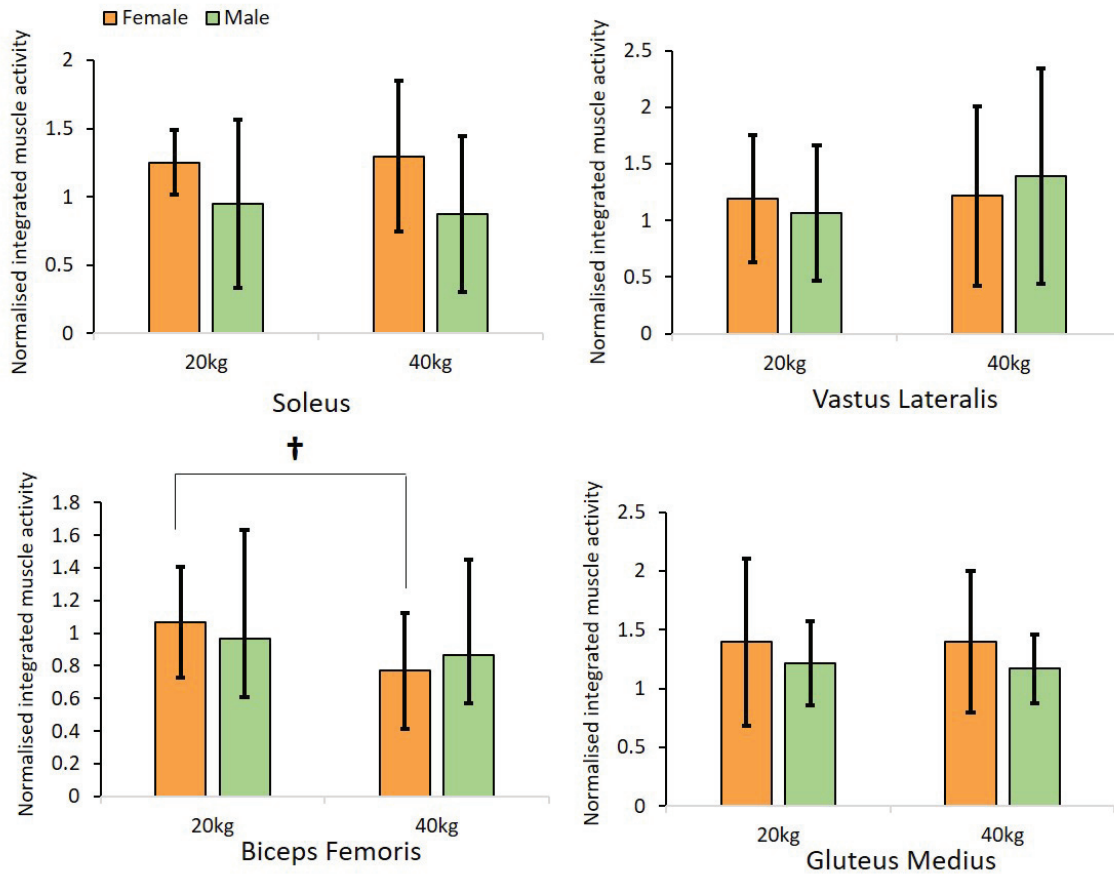


Figure 6.3: Integrated muscle activity over one gait cycle normalised to the unloaded running condition. † indicates significance between load conditions for $p \leq 0.05$.

gait cycle when carrying 40kg compared with running unloaded ($p=0.03$) and with 20kg ($p=0.03$). Males also shifted peak soleus activation to later in the gait cycle when carrying 40kg compared with no load ($p=0.03$). Females showed no significant changes in peak activation timings for any muscle. However, when running with 40kg, females' peak soleus activation occurred significantly sooner in the gait cycle compared with males ($p=0.02$), and peak gluteus medius activation occurred significantly later ($p=0.0007$).

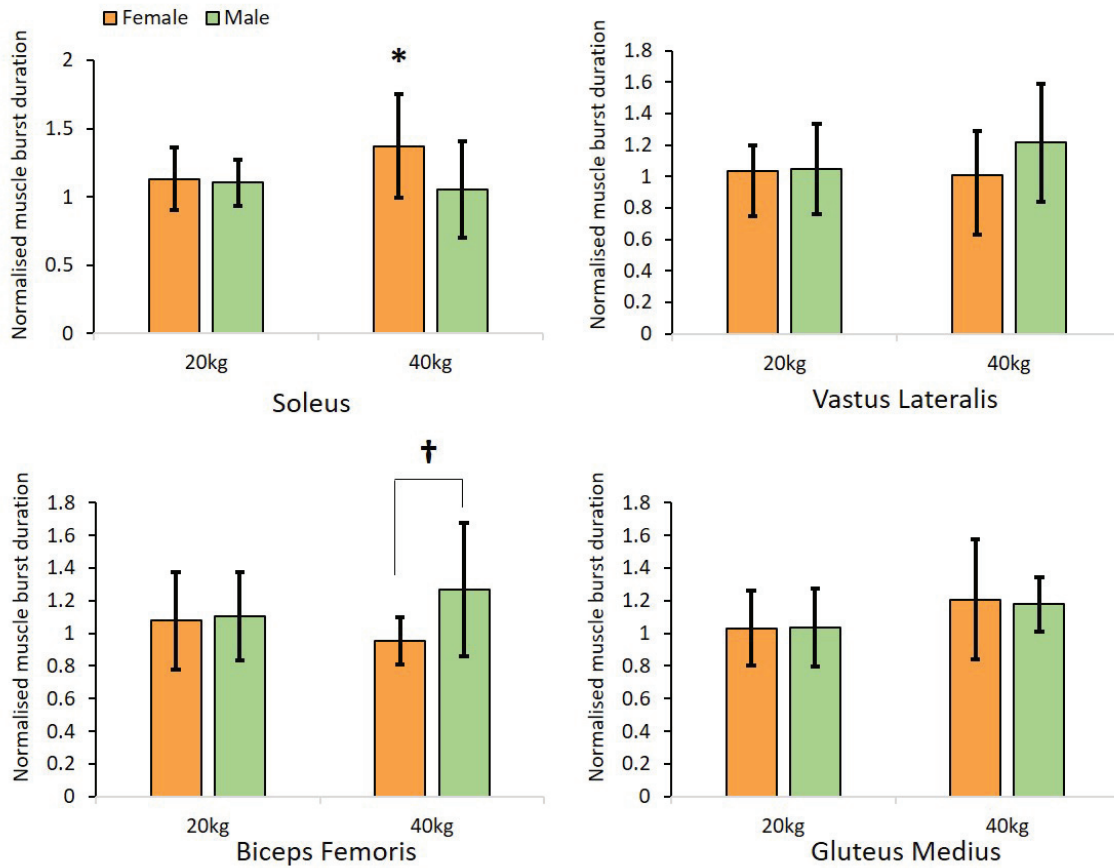


Figure 6.4: Normalised muscle burst duration. * indicates significance from the unloaded condition for $p \leq 0.05$. † indicates significance between load conditions or sexes for $p \leq 0.05$.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Soleus	41 (15)	44 (14)	44 (14) ^d	41 (16)	48 (16)	59 (7) ^{ad}
Vastus Lateralis	41 (14)	45 (12)	43 (14)	33 (11)	41 (14)	36 (14)
Biceps Femoris	28 (11)	32 (10)	28 (12)	29 (12)	32 (15)	22 (8)
Gluteus Medius	32 (9)	36 (7)	37 (9) ^D	33 (11)	33 (8)	21 (6) ^{abD}

Table 6.3: Mean (SD) percentage of the gait cycle at which peak muscle activation occurred during running. ^a indicates significance from the unloaded running condition, ^b indicates significance from the 20kg load, ^d indicates significance between sexes for $p \leq 0.05$. ^{A,B,D} for $p < 0.005$

6.4 Discussion

Walking

When walking, the main effect of load was significant for integrated soleus activity. The increased integrated activity of the ankle plantar flexors is thought to be due to the increased propulsion required at toe push off and the greater muscular effort necessary to overcome the inertia associated with increased backpack load (Attwells et al. 2006). Only females displayed significant increases in integrated soleus activity, which would come at a greater energetic cost (Hamill et al. 1995), and suggests that females were under a greater subjective workload than males (Simpson et al. 2012b).

Females' significantly greater soleus muscle burst duration when carrying load, and significant muscle burst duration increase when carrying 40kg from the unloaded condition are reflected in their significant increase in peak ankle torque (Chapter 3, Figure 3.5). When walking unloaded, females generated a significantly lesser proportion of overall torque at the ankle compared with males. However when carrying load, when soleus muscle activity was increased in females, there was no longer a significant difference in ankle joint torque proportion between sexes (Chapter 3, Table 3.6). This increase in soleus activity may reflect females' requirement for increased ankle joint and overall centre of mass stability over of the gait cycle due to carrying a greater percentage of their body weight. It is also possible that these changes in muscle activity may be an indicator of females attempting to shift torque production away from the hip. In addition, simulations of walking with load have suggested that the soleus is one of the main muscles responsible for generating the second peak of the ground reaction force (Lui et al. 2008; McGowan et al. 2010), where stresses in the pelvis are highest. It is also therefore possible that females must significantly increase their soleus activity and muscle burst duration during load carriage to compensate for their relatively smaller muscle mass and strength when generating the propulsive force required at toe push-off.

The main effect of load was significant for vastus lateralis integrated muscle activity consistent with the literature for males (Harman et al. 1992) and females (Simpson et al. 2011a; Silder et al. 2013). This effect of modifying vastus lateralis activity may be required to maintain leg stiffness and knee stability, particularly during weight acceptance, in response to increasing load. As in the previous studies, significant increases in integrated vastus lateralis muscle activity were only

displayed when heavier loads were carried. Harman et al. (2000) suggest this is due to muscles around the knee being more heavily recruited during heavy load carriage, while at lighter loads, other muscles may play a greater role. Analysis of joint dominance in Chapter 3 confirmed this finding in females, who did not significantly increase knee torque contribution until 40kg was carried, but not in males (Table 3.6). Rice et al. (2017) found that males increased peak vastus lateralis activity but not integrated activity when carrying 35.5kg. In their study of female hikers, Simpson et al. (2011a) found that females showed significantly delayed peak vastus lateralis activation when carrying 40% BW compared with no load. They suggest that this later activation is a result of increased workload on the vastus lateralis to slow knee flexion during weight acceptance. Although no significant changes were observed in the absolute percentage of the gait cycle at which peak vastus lateralis activity occurred in this study, females did delay their peak activity by an average of 3% of the gait cycle when carrying 20kg and 40kg compared to walking unloaded. Similarly, males were found to delay peak vastus lateralis timing by an average of 2% of the gait cycle when walking with 20kg and 3% when walking with 40kg. No significant differences in vastus lateralis muscle activity between males and females were observed, in agreement with the literature (Chumanov et al. 2008; Silder et al. 2013).

Although the main effect of load approached significance for the biceps femoris muscle burst duration, no significant changes in integrated activity or muscle burst duration of the biceps femoris were found for either sex, in agreement with Ghori and Luckwill (1985), but in contrast with much of the more recent literature (Simpson et al. 2011a; Silder et al. 2013; Lindner et al. 2012). It is interesting that no changes in biceps femoris muscle activity were found because the pull of the hamstrings has been suggested to contribute to stress fractures in the pelvis in female US naval recruits (Kelly et al. 2000). Significant sex-specific differences were found in the peak activation timing of the biceps femoris for the unloaded and 40kg load conditions, namely that females' peak activation occurred sooner in the gait cycle. The biceps femoris has been linked to hip control during stance phase (Simpson et al. 2011a) and so it is possible that females activated the biceps femoris sooner to provide greater control of the hip during weight acceptance. At this point in the gait cycle, the quadriceps and hamstrings co-contract to allow for shock absorption and lower limb stability. Altered co-contraction timings, such as those observed between males and females when walking unloaded and with 40kg, have been linked to increased loading and injury risk in the knee (Palmieri-Smith et al. 2009; Wakeling et al. 2003), but their affect on the hip has not yet been

studied.

The main effect of load was significant for integrated muscle activity and burst duration of the gluteus medius, however, only females displayed significant changes in muscle activity patterns. The main function of the gluteus medius is to provide pelvic stability during early to mid stance and forward propulsion during late stance (Lui et al. 2008). When carrying 40kg females significantly increased integrated activity and muscle burst duration compared to the unloaded and 20kg conditions respectively. As with the soleus, this suggests that females were working harder to maintain pelvic stability and underwent increased energetic costs when carrying 40kg. This has implications for pelvic stress injuries as the gluteus medius has been suggested as the main contributor to medial and superior components of the hip joint contact force (Correa et al. 2010). Wall-Scheffler and Myers (2017) found that, during unloaded walking, females activated their gluteus medius significantly more at the initiation of swing. They suggest that this may be to resist pelvic drop during single leg stance. Although this study did not consider sex-specific differences in muscle activity at different points of the gait cycle, no significant differences in muscle activity between males and females were observed in the gluteus medius, in consensus with Chumanov et al. (2008).

Running

The main effect of load was significant for soleus muscle burst duration when running. There were no significant sex-specific differences in integrated soleus activity or muscle burst duration, however females significantly increased muscle burst duration between the unloaded and 40kg conditions. This led to the interaction effect of sex and load approaching significance for the soleus ($p=0.06$). Increased soleus muscle burst duration has been found in recreational runners with patellofemoral pain syndrome compared with healthy controls (Esculier et al. 2015). This increase in muscle burst duration may be a mechanism to maintain ankle stability in stance or to generate the necessary increase in propulsion (Dorn et al. 2012). Peak soleus activity was found to increase with distance over a 4km unloaded run, correlating with a shift in foot strike pattern as a result of a compensatory strategy due to fatigue (Wu et al. 2007). This suggests that females may be displaying similar soleus activity patterns when running short distances with 40kg as seen in fatigued unloaded runners. Xu et al. (2017) studied four female runners and found that peak soleus activity when running unloaded occurred at 10% of the gait cycle, in contrast to this study where it occurred around toe off. This could be due to differences in foot strike pattern, running

speed, or differences between treadmill and overground running. The only significant difference between males' and females' soleus activity was seen in the peak activity timing when running with 40kg. Males' peak timing occurred significantly later in the gait cycle, again this could be due to differences in foot strike pattern or as a result of sex-specific gait adaptations.

No significant changes were found in the integrated muscle activity or burst duration of the vastus lateralis, in contrast to Bourdin et al. (1995), who found male runners significantly increased integrated muscle activity when carrying loads as little as 10% BW. Similarly, Abe et al. (2011) found significant increases in integrated vastus lateralis activity of males carrying 2.3kg during the phase of gait where the muscle eccentrically contracted, but not where it concentrically contracted. Xu et al. (2017) found that peak vastus lateralis activity occurred at approximately 20% of the gait cycle while this study found it occurred later, nearer toe off. Again this could be the result of differing foot strike patterns, running speeds or the influence of the treadmill vs overground running techniques. No sex-specific differences were observed in any of the vastus lateralis activity patterns. Compared with males, Chumanov et al. (2008) found that females had significantly greater integrated EMG activity during the initial loading phase of gait when running unloaded, but not over the entire gait cycle.

The interaction effect of load and sex was found to be significant for muscle burst duration of the biceps femoris. Females significantly reduced integrated muscle activity over the gait cycle when carrying 40kg compared to 20kg and correspondingly, had shorter muscle burst duration than males when running with 40kg. Hamstring activity was found to increase with distance in males running unloaded (Wu et al. 2007), which suggests that the decrease in biceps femoris activity may be due to gait adaptations and not necessarily due to muscle fatigue. Consistent with results of other muscles, Xu et al. (2017) found that peak activation occurred at heel strike for the biceps femoris, around 20% of the gait cycle earlier than this study.

The main effect of load was significant for the integrated gluteus medius muscle activity and muscle burst duration, both of which were positively correlated with increasing load. Despite this, no significant changes in integrated gluteus medius activity or muscle burst duration were observed for either sex. In agreement with Chumanov et al. (2008) and Willson et al. (2012), no significant differences were seen between males and females for integrated gluteus medius activity, muscle

burst duration or peak activity timings during unloaded running. When carrying 40kg however, females displayed significantly delayed peak activity timing compared to males, which has implications for injury, as later gluteus medius activation has been seen in females with PFPS running unloaded (Willson et al. 2011) and both sexes with PFPS when undertaking stair stepping tasks (Brindle et al. 2003).

Limitations

Limitations to this study include the short length of walkway and lab setting, which are not representative of typical training terrain and thus the measured EMG signals may not accurately reflect muscle activation patterns during training. In addition, this study did not take into account the effects of fatigue, which are likely to significantly alter muscle activity. The results do, however, capture changes in muscle activity as a direct result of increasing carried load, without the influence of fatigue. Finally, surface EMG electrodes are susceptible to skin motion artefact, electrical noise and nearby muscle interference, which can also influence the recorded signals. However, the influence of this interference was reduced in this study by normalising signals to the unloaded conditions, allowing for direct comparison between load conditions and reducing variability between subjects.

6.5 Conclusions

Changes to integrated muscle activity and burst duration during walking suggest that females were working at a higher energetic cost than males in order to maintain leg stability. When walking with 20kg and 40kg females significantly increased integrated soleus activity to produce the propulsive force required by the added load. When walking with 40kg females utilised the soleus, vastus lateralis and gluteus medius to maintain leg and pelvic stability while males only showed changes in the vastus lateralis. These changes to muscle activation patterns were reflected in the analysis of joint torque distributions reported in Chapters 3 and 5. Sex-specific differences in the timing of peak activation of the biceps femoris may have been due to females requiring greater control of the hip during weight acceptance. Running with 40kg led to females altering muscle activity patterns in the biceps femoris, as well as changes in peak muscle timings of the soleus and gluteus medius in males. This supports the theory that males and females adapt their gait differently in response to heavy body borne loads. The fewer significant changes in activity patterns when running compared with walking may be due to

variability in running techniques between subjects as this has been shown to affect muscle activity. However, the differences that were observed in females compared to males, namely increased soleus activity and delayed gluteus medius activity, have both been observed in runners with PFPS. This suggests that increasing load has similar links with muscle activity in females as with lower limb pathologies when running.

This chapter has found that lower limb muscle activity is influenced by both sex and carried load. The next chapter will further investigate the influence of these parameters by identifying potential biomechanical risk factors for increased forces in the hip and stress in the pelvis using regression analysis.

Chapter 7

Biomechanical risk factors for hip and pelvic injuries related to military load carriage

Results from previous chapters have confirmed that males and females respond differently when carrying standardised loads. This chapter reports a multiple regression approach to determine if these biomechanical differences could be risk factors or correlating variables that may contribute to hip and pelvic musculoskeletal injuries.

7.1 Introduction

The biomechanical gait differences between males and females are well reported. Previous studies have observed sex-specific differences in spatiotemporal, kinematic and kinetic parameters during walking (Kerrigan et al. 1998; Cho et al. 2004; Callisaya et al. 2008), and running (Gehring et al. 2014). Studies have also attempted to describe gait variation due to spatiotemporal and anthropometric variables through non-linear (Wannop et al. 2012), linear (Lelas et al. 2003), and multiple linear (Senden et al. 2012; Lee et al. 1999; Dixon et al. 2014; Macellari et al. 1999) regression analysis.

More recently, regression analysis has been used to quantify intrinsic and extrinsic risk factors for injury. Jacobs et al. (2014) suggest it may be more useful to reclassify factors as modifiable (e.g. physical fitness, gait parameters, footwear) or unmodifiable (e.g. age, race, history of injury). Springer et al. (2016) measured the gait of male soldiers walking with 10kg and used linear regression to predict injury rates against spatiotemporal parameters. They found that an initial double stance phase of less than 12.5% of the gait cycle was correlated with increased injuries in the knee, ankle and foot. Studies of Royal Marine recruits have shown that greater peak heel pressure and lower range of tibial rotation increased the risk of tibial stress fractures (Nunns et al. 2016), while delayed peak toe pressure and lower foot abduction increased risk of metatarsal fractures (Dixon et al. 2019). Logistic regression has revealed differences in foot centre of pressure in male and female recruits with patellofemoral pain (Thijs et al. 2007).

In studies of non-biomechanical parameters, low cardiovascular fitness was found to be a predictor of injury, but sex was not, when controlled for age and ethnicity (Blacker et al. 2008; Bell et al. 2000). However these studies considered all injuries occurring during training, not specifically musculoskeletal ones. Being female has been found to significantly increase risk of bone stress injuries in military recruits (Mattila et al. 2007), as discussed in Chapter 2. Lower muscle strength and aerobic fitness have been identified as modifiable risk factors for stress fractures in female soldiers (Rauh et al. 2006).

Willems et al. (2006) used Cox regression analysis and observed that runners with patellofemoral pain (PFP) displayed altered foot mechanics before their injury occurred. Studies of runners have found that magnitude, timing and direction of peak ground reaction force predicted PFP (Thijs et al. 2008) and

Achilles tendinopathy (Van Ginckel et al. 2009) using logistic regression. Female runners with PFPS were found to display weaker hip abductor and external rotator muscles in the injured leg (Cichanowski et al. 2007). There is still, however, limited evidence of how biomechanical gait parameters may correlate with hip injuries, particularly in females.

Although cohort injury data was not available for this study, it was assumed that factors such as high internal MSk loads may be indicators of increased risk of MSk injury (Attwells et al. 2006; Simpson et al. 2011a). This study therefore aimed to identify biomechanical risk factors for high internal musculoskeletal hip loads and pelvic stresses. Another major finding of this thesis, that males and females preferentially produce propulsive torque at different joints, is also investigated in this chapter in order to determine if biomechanical factors affect joint dominance.

7.2 Methods

In order to account for the anthropometric differences between sexes and to identify variables which may contribute to injury during walking with load, multiple regression analysis was used, as shown in Equation 7.1 below (Macellari et al. 1999):

$$y_i = \beta_0 + \beta_1 \cdot x_1 + \dots + \beta_n \cdot x_n + \varepsilon \quad (7.1)$$

Where y_i is the dependent variable, β_0 is the intercept term, $\beta_1 \dots \beta_n$ are the coefficients of the independent variables, $x_1 \dots x_n$ are the independent variables and ε is the error term.

Spatiotemporal, kinematic and kinetic data collected in Chapter 3 were analysed. All statistical analysis was performed in RStudio (v1.1.463). Variance Inflation Factors (VIF) were first calculated in order to identify and remove collinear redundant variables from consideration ($VIF > 5$). All data was previously checked for normality (described in Chapter 3). Independent variable selection was performed using a stepwise forward selection process. Variables were included in the model if they improved the accuracy by increasing the adjusted R^2 value. If variables did not improve the accuracy of the model they were rejected. Variables

were considered to significantly improve model accuracy for $p \leq 0.05$. Models were calculated for each load condition for the dependent variables of hip, knee and ankle dominance, as well as both the peak hip JCF and the integrated area under the JCF curve. This area was taken to give a measure of the total force applied through the joint over one gait cycle.

For all multiple linear models, the variance explained by each independent variable was determined using ANOVA to calculate the sum of squares for each variable. The proportion of each variable's sum of squares to the total sum of squares in the model was taken as the variable's contribution to variance in the model, as shown in Equation 7.2 (Olejnik and Algina 2003):

$$V_i = \frac{SS_i - (n - 1) \times MSE}{\sum_{i=1}^n SS_i + MSE} \times 100 \quad (7.2)$$

Where V_i is the percentage of total variance in the model described by the independent variable, SS_i is the sum of squares of the variable, MSE is the mean square error and n is the number of variables in the model.

In order to determine which independent variables were associated with higher stresses in the pubic rami, pelvic joint angles, contact forces and muscle forces for 11 randomly selected subjects (6 female, 5 male, mean mass= 68.9kg) carrying 20kg (data collected in Chapter 3) were applied to the female hemi-pelvis finite element model described in Chapter 4. The 20kg load condition was selected as no differences in spatiotemporal gait measurements were observed between males and females, however significant differences in internal joint loads were (Chapter 3). Subject specific loads for gait position 3 (toe push-off) were used as this was where the highest von Mises stresses occurred over the gait cycle. Although the von Mises stresses calculated were not likely to be accurate for each subject due to differences in pelvic morphology, comparing multiple subjects' load conditions on the same model allowed for identification of potential risk factors for higher pelvic stresses while controlling for shape variations in the bony pelvis. Due to the relatively small sample size ($n=11$) multiple regression analysis was not used. Variables were instead individually assessed against the von Mises stress values for the superior and inferior rami using the Pearson correlation coefficient, r , value. Again variables were considered significant for $p \leq 0.05$.

In order to identify variables corresponding to high forces in the hip while running with load, a combination of multiple regression and Pearson correlation were used, as described above. For all dependant variables, multiple linear regression analysis was used when data from ten subjects per independent variable were available. Where there were less subjects, multiple linear regression could not be performed and Pearson correlation was used instead.

7.3 Results

7.3.1 Variables correlated with MSk loads in the hip during walking

Variables (and p-values) associated with changes in propulsive torque produced at the ankle, knee and hip are shown in Table 7.1. Sex ($p \leq 0.0001$) and walking speed (≤ 0.0001) were found to be the most statistically significant variables, with female sex and faster walking speed associated with a greater percentage of propulsive torque being produced at the hip. Male sex and slower speed were found to correlate with greater ankle and knee dominance. Figure 7.1 shows how propulsion is moved from the knee up to the hip as walking speed increases, with females more hip dominant and males more knee dominant when walking at the same speeds. Greater hip extension angle was also a significant predictor of hip dominance ($p=0.02$), while greater knee internal rotation angle neared significance for increasing knee dominance ($p=0.08$). Lower hip extension angle and ankle external rotation improved the R^2 values for the knee and ankle regression models respectively, however they were not significant. Carried load did not significantly affect the proportion of propulsive torque produced in any joint.

The regression equation for hip dominance was the most accurate, with an R^2 value of 0.59. Walking speed accounted for 30% of variance in the model, while sex and hip extension angle accounted for 14% and less than 1% respectively. An R^2 of 0.31 was found for the knee regression model. Speed accounted for less variance than in the hip model (15%), as did sex (9%). Hip extension and knee internal rotation both accounted for 2% of variance in the model. For the ankle regression, speed accounted for 9% of variance in the model, while sex and ankle external rotation described 1% of the variance each.

	Sex	Walking speed	Hip extension angle	Knee internal rotation angle	Ankle external rotation angle	R ²
Hip	F ≤0.0001	↑ ≤0.0001	↑ 0.02			0.59
Knee	M ≤0.0001	↓ ≤0.0001	↓	↑ (0.08)		0.31
Ankle	M 0.04	↓ 0.001			↓	0.13

Table 7.1: Parameters correlating with percentage of propulsive torque produced at the hip, knee and ankle joints including p-values, F indicates being female corresponds with higher joint dominance, ↑ indicates an increase in the variable corresponds to an increase in joint dominance and vice-versa. Green cells indicate which variables were included in each regression model to maximise the adjusted R² value, even if not significant.

Peak and integrated joint contact force model variables, associated p-values and R² values are shown in Table 7.2. Mass (taken here as body mass not including rucksack load) was found to be the most statistically significant factor in all joint contact force models ($p \leq 0.03$ for all models). Female sex was a significant predictor for greater integrated hip JCF over the gait cycle for the unloaded ($p=0.03$) and 20kg ($p=0.04$) load conditions and improved the accuracy of the 40kg model, though it was not significant. Sex was not found to impact peak hip JCF for any load condition. Walking speed was found to be significant in 2 of the models and improved the accuracy of 4 out of the 6 models. For peak JCF when carrying load, stride length was found to improve the models more than walking speed. They were not both included as variables due to the high collinearity between them. Pelvic tilt was found to be significant for the integrated JCF in the unloaded condition ($p=0.003$), and neared significance for the peak JCF ($p=0.07$). Trunk flexion angle was not found to be statistically significant for any of the models. Smaller peak ankle external rotation angle was significantly correlated with a greater integrated JCF when carrying 20kg ($p=0.009$) and 40kg ($p=0.05$).

For integrated joint contact force over the gait cycle, mass accounted for between 30-33% of variance in the models while sex explained between 3-8%. Walking speed accounted for 1-15% of the variance. For all variables, variance percentages were highest for the unloaded condition and lowest for the 40kg condition. Pelvic tilt explained 5% of the variance in the unloaded integrated and peak JCF models. Ankle external rotation angle accounted for 9% (20kg) and 5% (40kg) of the variance when carrying load. For the peak JCF models, mass

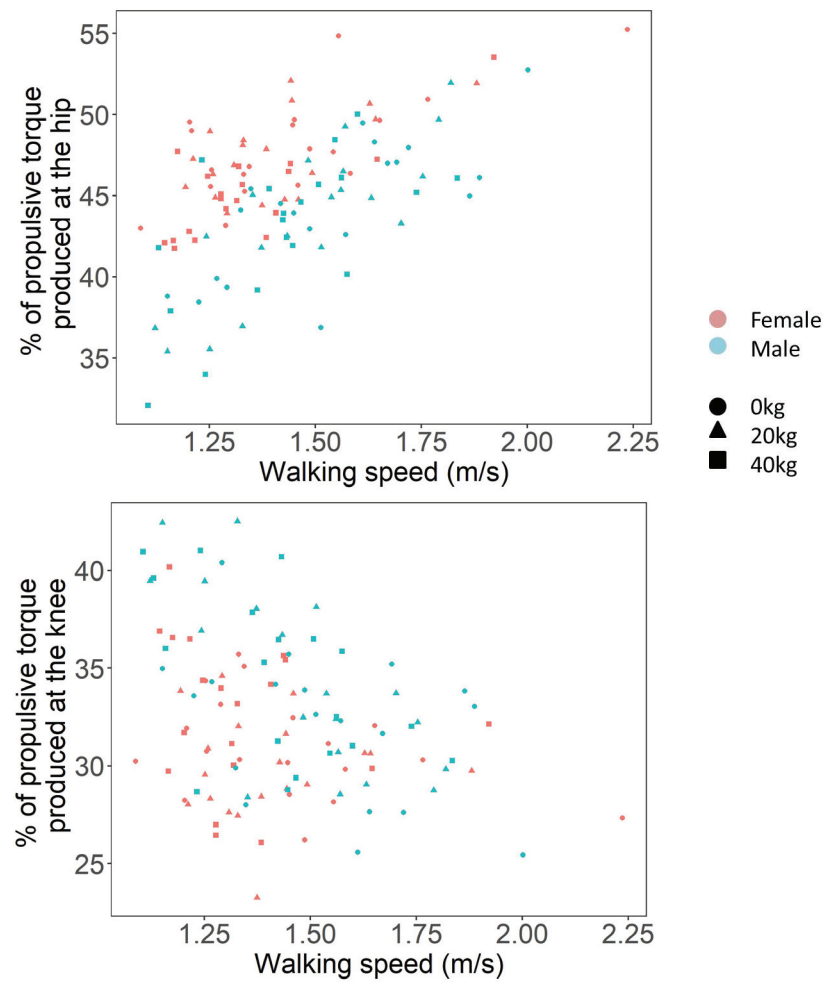


Figure 7.1: Variation in hip and knee dominance with walking speed and sex for all load conditions.

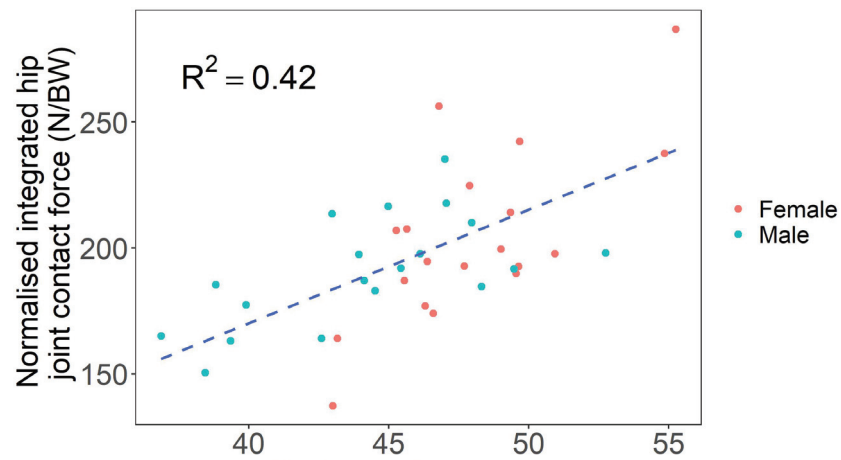
accounted for between 13-30% of variance while stride length only accounted for between 7-16% when carrying load. Walking speed was responsible for the most variation in the unloaded peak JCF model (19%).

Figure 7.2 shows the moderately positive correlation between increasing hip dominance and hip joint contact force (normalised to body weight) over the gait cycle. For all load conditions a larger percentage of propulsive torque produced at the hip corresponded significantly to a larger integrated joint contact force (Pearson coefficient: $r=0.65$, $p \leq 0.0001$ for 0kg, $r=0.54$, $p=0.0005$ for 20kg, $r=0.48$, $p=0.002$ for 40kg).

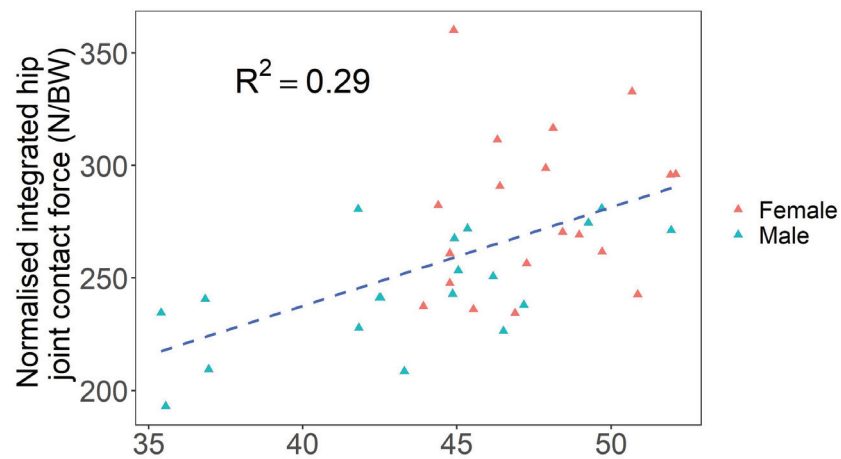
Sex	Mass	Walking speed	Stride length	Knee abduction angle	Pelvic tilt	Ankle plantar flexion angle	Knee internal rotation angle	Ankle external rotation angle	R ²
0kg integrated	F 0.03 ↑ ≤0.0001	↑ 0.01			↑ 0.003		↓		0.80
0kg peak		↑ 0.004	↑ 0.01		↑ (0.07)	↓			0.48
20kg integrated	F 0.04 ↑ ≤0.0001	↑ (0.07)						↓ 0.009	0.6
20kg peak		↑ 0.03	↑ 0.0005	↑				↑	0.57
40kg integrated	F ↑ ≤0.0001	↑						↓ 0.05	0.57
40kg peak		↑ 0.02	↑ 0.01	↑				↑	0.59

Table 7.2: Parameters correlating with Hip JCF including p-values, F indicates being female corresponds with higher JCF, ↑ indicates an increase in the parameter corresponds to an increase in JCF. Green cells indicate which parameters were included in each regression model to maximise the adjusted R² value.

(a) 0kg



(b) 20kg



(c) 40kg

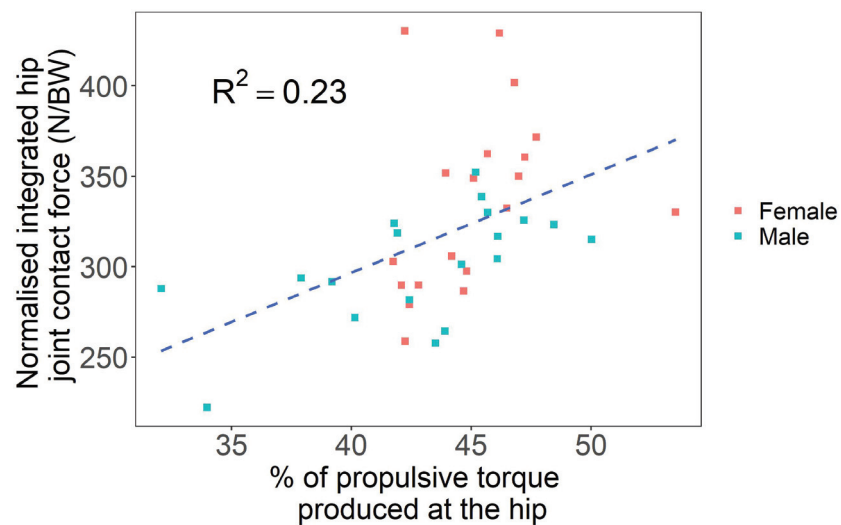


Figure 7.2: Relationship between hip dominance and integrated joint contact force over the gait cycle for (a) 0kg (b) 20kg (c) 40kg.

7.3.2 Variables correlated with von Mises stress in the pubic rami

Variables significantly correlated with peak von Mises stress in the superior and inferior rami are shown in Figures 7.3 and 7.4 respectively. Larger pectineus muscle force was found to strongly correlate with increased peak stress in the superior ($r=0.85$, $p=0.0009$) and inferior rami ($r=0.92$, $p\leq 0.0001$). In the superior ramus, the medial component of the hip JCF also showed a strong positive correlation ($r=0.83$, $p=0.001$) with peak stress. Larger adductor longus muscle force ($r=0.60$) and pelvic obliquity ($r=0.58$) were also significantly correlated with peak stress in the superior ramus ($p=0.05$ for each). Adductor brevis muscle force showed a moderate positive correlation with increased stress which tended towards significance ($r=0.55$, $p=0.07$). In the inferior rami (Figure 7.4), the adductor longus ($r=0.79$, $p=0.003$), adductor brevis ($r=0.72$, $p=0.01$) and medial hip JCF forces ($r=0.80$, $p=0.003$) showed strong significant positive correlations with peak von Mises stress. As in the superior ramus, pelvic obliquity was moderately correlated with increased peak stress in the inferior ramus ($r=0.65$, $p=0.03$). Statistical significance between the peak stresses of males and females could not be tested, due to the small sample size.

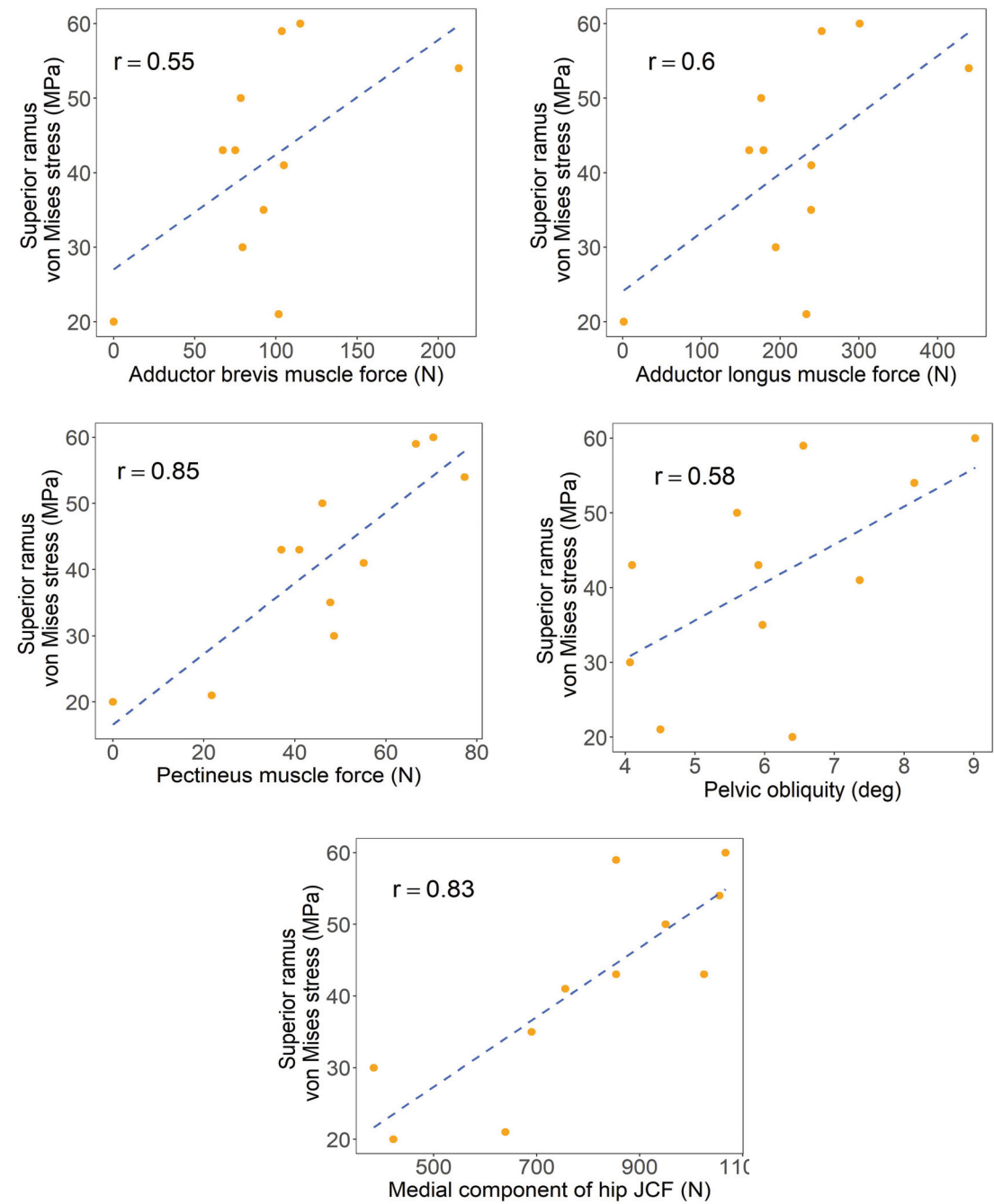


Figure 7.3: Pearson correlation, r , for significant variables against superior ramus peak von Mises stress.

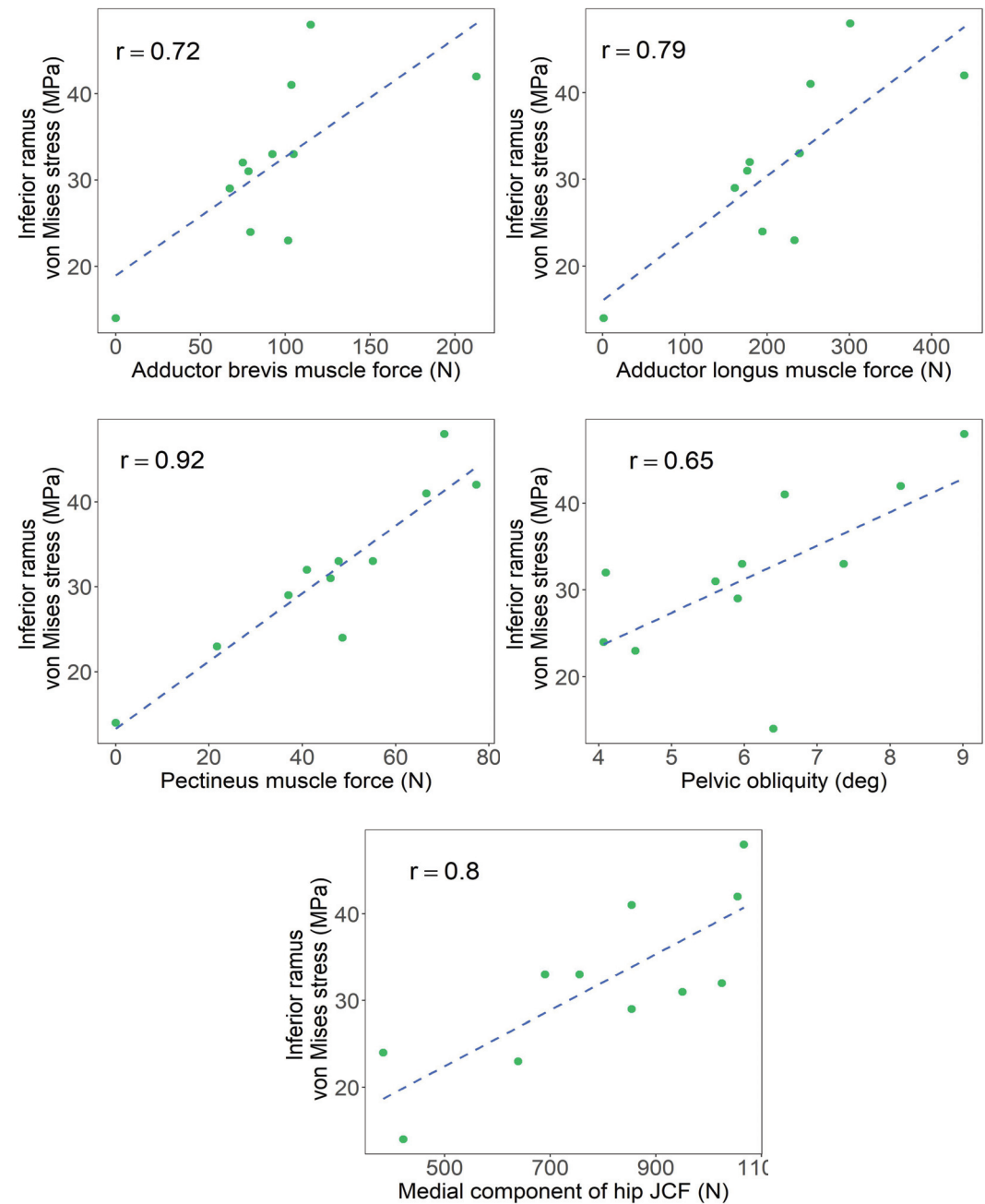


Figure 7.4: Pearson correlation, r , for significant variables against inferior ramus peak von Mises stress.

7.3.3 Variables correlated with MSk loads in the hip during running

Variables (and p-values) associated with changes in propulsive torque produced at the ankle, knee and hip when running are shown in Table 7.3. Female sex was found to correlate with increased hip dominance during running ($p \leq 0.0001$). Similarly being male was associated with increased knee dominance ($p = 0.007$). Faster running speed ($p \leq 0.0001$) correlated with increased hip dominance, while greater carried load resulted in reduced hip dominance ($p \leq 0.0001$) and increased propulsion produced at the ankle ($p = 0.0003$). Larger ankle plantarflexion also correlated with increased ankle dominance ($p = 0.003$). Carried load did not significantly affect the proportion of propulsive torque produced at the knee, however less knee flexion was significantly correlated with increased knee dominance ($p = 0.03$). The relationship between running speed and hip dominance is shown in Figure 7.5 for both males ($p = 0.0008$) and females ($p = 0.0002$), with females exhibiting greater hip dominance for across all speeds. Similarly, Figure 7.6 shows the significant positive correlation between flight time (time spent with both feet off the ground) and proportion of propulsive torque produced at the hip ($p \leq 0.0001$ for females, $p = 0.0003$ for males).

	Sex	Running speed	Stride length	Carried load	Knee flexion	Ankle plantar flexion	R ²
Hip	F ≤ 0.0001	$\uparrow$ ≤ 0.0001		$\downarrow$ ≤ 0.0001			0.75
Knee	M 0.007				$\downarrow$ 0.03		0.23
Ankle			$\downarrow$ 0.0007		$\uparrow$ 0.0003	$\uparrow$ 0.003	0.46

Table 7.3: Parameters correlating with hip, knee and ankle joint dominance including p-values, F indicates being female corresponds with higher joint dominance, $\uparrow$ indicates an increase in the variable corresponds to an increase in joint dominance and vice-versa. Green cells indicate which variables were included in each regression model to maximise the adjusted R² value.

As with walking, the regression equation for hip dominance was the most accurate, with an R² value of 0.75. Running speed accounted for 34% of variance in the model, while the effects of carried load and sex accounted for 24% and 17% respectively. An R² of 0.23 was found for the knee regression model. Sex accounted for less variance than in the hip model (15%). Knee flexion angle

accounted for 8%. For the ankle regression, stride length accounted for 22% of variance in the model, while carried load and ankle plantarflexion angle described 14% and 10% of the variance respectively.

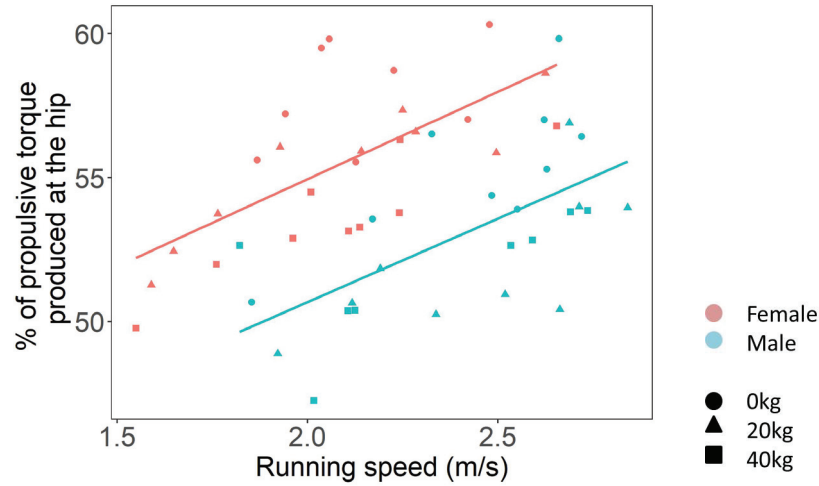


Figure 7.5: Relationship between running speed and hip dominance for all load conditions. Female, $r=0.66$. Male, $r=0.62$

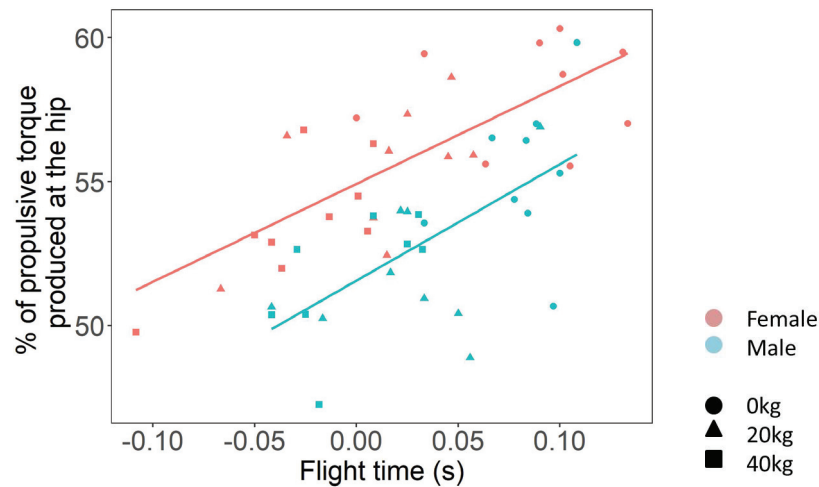


Figure 7.6: Relationship between flight time and hip dominance for all load conditions. Female, $r=0.77$. Male, $r=0.65$

The R^2 and p -values for variables which significantly correlated with integrated hip joint contact force during running are shown in Table 7.4. The hip JCF for all load conditions and sexes were evaluated against each variable separately due to the non-linear relationships between groups. Greater mass was found to correlate strongly with increased JCF in males for all load conditions ($p<0.01$), but not females. Running speed was significant for females ($p=0.03$) and males ($p=0.0006$) when carrying 40kg. A moderate positive correlation with stride length was

significant for females running unloaded ($p=0.04$), and males both unloaded ($p=0.02$) and carrying 20kg ($p=0.03$). Larger trunk forward flexion ($p=0.007$) as well as ankle ($p=0.05$) and hip external rotation ($p=0.01$) were positively correlated with increased hip JCF in females running unloaded. Ankle eversion angle was positively correlated in females carrying 40kg ($p=0.02$). No variables were found to significantly correlate with hip JCF in females carrying 20kg. Pelvic obliquity ($p=0.02$), hip internal rotation ($p=0.05$) and hip extension ($p=0.03$) positively correlated with males' hip JCF when running unloaded. Trunk flexion angle ($p=0.03$), pelvic obliquity ($p=0.05$) and hip extension angle ($p=0.03$) were significant for males carrying 40kg. No kinematic measures were found to be significant for males' hip JCF for the 20kg load condition.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Mass				↑0.8 (0.001)	↑0.74 (0.003)	↑0.63 (0.01)
Speed			↑0.51 (0.03)		↑0.39 (0.07)	↑0.68 (0.006)
Stride length	↑0.49 (0.04)	↑0.39 (0.07)		↑0.54 (0.02)	↑0.49 (0.03)	
Trunk flexion	↑0.67 (0.007)					↑0.53 (0.03)
Pelvic obliquity	↑0.41 (0.06)			↑0.56 (0.02)		↑0.44 (0.05)
Hip external rotation	↑0.62 (0.01)					
Hip internal rotation				↑0.42 (0.05)		
Hip flexion	↓0.4 (0.06)					
Hip extension				↑0.52 (0.03)		↑0.5 (0.03)
Ankle eversion			↑0.57 (0.02)			
Ankle external rotation	↑0.44 (0.05)					

Table 7.4: Variables, R^2 (p-value), correlating with integrated hip JCF when running for males and females.

The relationship between percentage of propulsive torque produced at the hip and integrated joint contact force normalised to body weight is shown in Figure 7.7. Weak positive correlations were found for males running unloaded ($r=0.21$) and with 40kg ($r=0.32$). Moderate positive correlations were observed for females running unloaded ($r=0.55$) and with 40kg ($r=0.52$). When running with 20kg, a significant strong positive correlation between hip dominance and normalised integrated hip JCF was found for both males and females ($r=0.79$, $p=0.01$ for each).

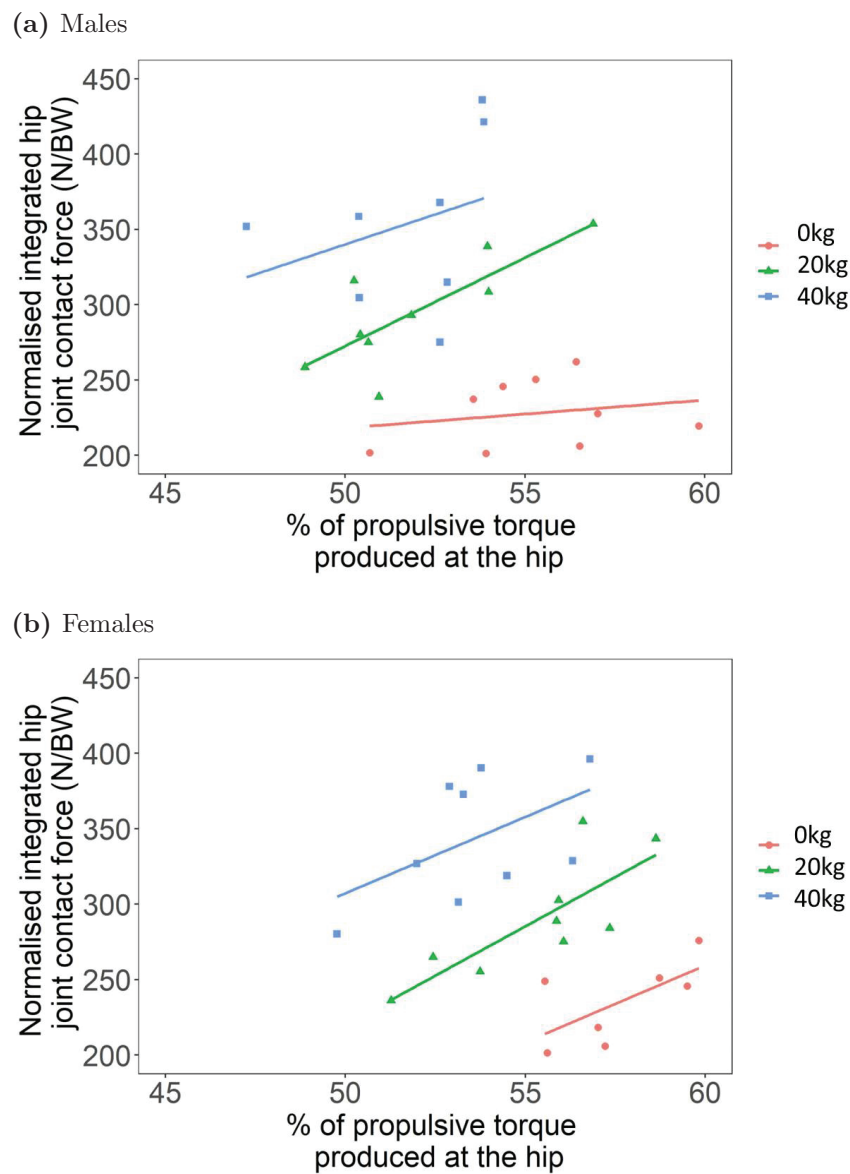


Figure 7.7: Relationship between hip dominance and integrated JCF during running for (a) males (b) females.

7.3.4 Key Findings

1. Female sex and faster walking speed were the most significant predictors of hip dominance.
2. Propulsive torque production is exchanged from the knee to the hip as walking speed increases.
3. Female sex is a significant predictor of greater integrated hip JCF over the gait cycle when walking and running.
4. Hip dominance showed a significant positive correlation with integrated hip JCF when walking.
5. Adductor muscle forces, medial hip JCF and pelvic obliquity significantly correlated with peak von Mises stress in the rami when carrying 20kg.
6. Female sex, faster speed and carrying heavier loads have the most significant effects on hip dominance during running.
7. Sex and carried load affected which variables correlated with integrated hip JCF when running.
8. Increased hip dominance significantly correlated with larger integrated hip JCF when running with 20kg.

7.4 Discussion

Walking

Results of the multiple regression analysis revealed that walking speed accounted for the most variance in the model for hip dominance. Female sex was also a significant factor, indicating that when walking at the same speed, females preferentially produced a greater percentage of propulsive torque at the hip compared with males. Slower walking speed significantly correlated with increased ankle and knee dominance, as did being male, indicating that males utilised the other lower limb joints to produce propulsive torque more than females. Additionally, plots of speed against joint dominance illustrated how propulsion was transferred from the knee to the hip as walking speed increased. This could be due to the larger muscles surrounding the hip being able to produce the higher propulsive forces required at faster walking speeds. Although not directly comparable, Lelas et al. (2003) observed that hip power generation increased quadratically with walking speed, and at a greater rate than knee power

generation. Their results further suggest that the hip is the main joint responsible for propulsion at faster walking speeds. In the current study, larger peak hip extension angle was also correlated with increased hip dominance, possibly because subjects were extending the hip further in order to maximise stride length and peak extensor torque in an effort to increase hip propulsive performance.

Although the results from Chapter 3 showed differences in males' and females' hip JCF's, it was necessary to determine if this was due to the significant mass differences between the groups and hence females carrying a greater percentage of their body mass. Accounting for anthropometric and gait parameters showed that being female was still a significant predictor for greater integrated hip JCF when walking with similar gait to males. This suggests that females may have inherently higher internal forces in the hip, possibly because females must generate more propulsion in the hip in order to compensate for their relatively weaker knee and ankle musculature. Differences in joint level kinetics may also be due to differences in pelvic geometry, as it has been suggested that female pelves may have evolved for improved efficiency during long-distance walking (Wall-Scheffler and Myers 2017). Speed was also found to influence integrated hip joint force. This is in agreement with a study by Weinhandl et al. (2017), which found that even a 10% increase in walking speed resulted in significant increases in hip joint contact force. Pelvic tilt correlated only with the 0kg condition, possibly because the rucksack was affecting pelvic girdle motion. When carrying load, a more externally rotated ankle was found to correlate with lower integrated hip JCF. Larger ankle external rotation moment has also been associated with reduced risk of tibial stress fractures in females (Milner et al. 2006a), suggesting that transverse plane ankle motion may be related to internal force transmission in the leg. Neely (1998) suggested that increasing ankle external rotation may be an adjustment used to reduce ankle dorsiflexion demands during push-off.

Mass was again found to be responsible for the most variation in peak hip JCF. Stride length correlated more strongly than walking speed with peak joint force when carrying load. This has implications for injury as a longer stride length was found to correlate with larger peak ground reaction force in this study. Results from the literature have found that greater vertical loading rates (Milner et al. 2006b) and peak heel pressure (Nunns et al. 2016) increased shock attenuation in the leg and led to increased risk of tibial stress fractures. It was expected that the regression models would be less accurate as load increased due to compounding of modelling errors, however this was only the case for integrated hip force. When

carrying load, R^2 values of approximately 0.6 were found, suggesting that there are more factors accounting for around 40% of variance in hip contact force that have not been considered in the models.

Comparison of the percentage of propulsive torque produced at the hip and integrated joint contact force revealed significant moderate positive correlations between the two. For all load conditions, an increase in hip dominance corresponded to an increase in hip JCF over the gait cycle. In addition, the rate of increase was largest when 40kg was carried, suggesting that small changes in hip dominance cause greater changes in JCF when heavier loads are carried. As females were found to be more hip dominant, this further supports the theory that females are prone to higher internal forces in the hip.

Von Mises stress in the rami

Adductor longus and adductor brevis muscle forces were found to strongly correlate with inferior ramus peak stress and moderately correlate with the superior ramus. This is perhaps not surprising as these large muscles attach directly to the inferior ramus and have been attributed to the development of pelvic stress fractures (Kelly et al. 2000). For both the inferior and superior rami, pectineus muscle force was most highly correlated with peak stress. The attachment of the pectineus on the top of the superior ramus and direction of the muscle force may create torsion in the rami around the pubic symphysis joint. Similarly, the medial component of the hip joint contact force has been associated with bending of the rami, increasing compression on the lateral side of the pelvis (Ricci et al. 2018). Results of this study therefore suggest that stress in the rami are more sensitive for forces in the frontal plane. Additionally, greater pelvic obliquity angle significantly correlated with increased stress in both the superior and inferior rami. This has also been found in the literature to alter how force is transferred through the rami and pubic symphysis to the contralateral pelvis (Volinski et al. 2018). Gluteal and hamstring forces were not found to significantly correlate with peak stress in the rami, possibly because their force is transmitted directly down through the ilium to the femur (Phillips et al. 2007).

Running

As with walking, speed was found to correlate most significantly with hip dominance when running. However, unlike during walking, carried load also significantly affected the percentage of propulsive torque produced at the hip and ankle. Subjects may shift torque production to other joints when carrying heavier

loads to reduce internal forces at the hip due to the relatively higher musculoskeletal loading during running. Again female sex was found to be significant for increased hip dominance, with females generating a greater proportion of propulsion at the hip compared to males for all running speeds and flight times. Male sex was significant for knee dominance, suggesting they will preferentially produce more of the propulsion required to leave the ground at the knee than females. Less knee flexion was also a predictor of greater knee dominance, while more knee flexion was found to correlate with increased ankle dominance. It is possible that the subjects with larger knee flexion did not have the required quadriceps strength to maintain leg stiffness and produce enough force to extend the knee and generate propulsion, and hence used the ankle to generate propulsive force instead. In addition, greater ankle plantar-flexion at toe off correlated significantly with increased ankle dominance, in agreement with a study by Huang et al. (2015), which found a direct relationship between plantar-flexion angle and ankle push-off power and work in healthy adults.

Mass was significantly correlated with integrated hip JCF in males when running, accounting for as much as 80% of the variation in joint force. In females, however, mass explained only around 30% of the variation, and was not significant for any load condition. This suggests that running technique may have a more significant impact on the hip JCF of females than males. Longer stride length correlated significantly with greater integrated hip JCF, which has implications for injury risk as longer stride length has also been found to increase ground reaction force when running with and without load (Lobb et al. 2019). Larger pelvic obliquity was a significant predictor of greater integrated hip JCF, and has been associated with poor running form due to muscle weakness in collegiate long distance runners (Ford et al. 2013) and muscle fatigue in non-runners (Tsatalas et al. 2013). It is possible that larger trunk flexion angle correlated with greater hip JCF as it increased torque required by the hip to maintain the position of the centre of gravity. Hip external rotation angle showed a strong positive correlation with integrated hip JCF in females running unloaded. A significant correlation between large hip external rotation angles ($> 65^\circ$) and lower limb stress fracture incidence has been found in studies of military recruits (Finestone et al. 1991; Giladi et al. 1987). In addition, larger ankle eversion, which correlated significantly in females running unloaded, has been linked to greater risk of exercise related lower limb injury (Pohl et al. 2008; Willems et al. 2006).

Finally, comparison of hip dominance versus integrated hip JCF revealed a

positive correlation between the two variables. This correlation was stronger in females than males for all load conditions. Furthermore, the rate of increase of hip JCF with hip dominance was greater in females than males. This suggests that females' joint contact force will increase faster as more propulsive torque is produced at the hip, and that they may be more prone to larger internal loads in the hip, and hence more susceptible to injury.

7.5 Conclusions

Faster walking speed and female sex were the most significant predictors of hip dominance, indicating that females may be more at risk of high internal hip forces when walking and running at the same speeds as males. Hip dominance was found to positively correlate with integrated JCF, suggesting that females may be more prone to greater hip JCF's over the gait cycle. Results of this study also suggest that muscle strength may significantly affect injury risk as adductor muscle forces were found to significantly correlate with higher peak stresses in the rami. In addition, pelvic obliquity, which has been associated with muscle weakness and fatigue, correlated with both stress in the rami when walking and hip JCF when running. The evidence indicates that males and females adapt their gait differently in response to carrying heavy loads and changing speeds, which combined with inherent biomechanical differences, leads to differences in musculoskeletal loading of the lower limb.

This chapter has identified sex-specific biomechanical differences during walking and running with load, as well as kinematic factors which correlate with high MSk forces and pelvic stresses. The next chapter will present the final summary and conclusions of the thesis findings. The clinical significance of these findings will be discussed and recommendations on how the biomechanical risk factors may be mitigated will be presented.

Chapter 8

Conclusions and Future Work

This chapter presents a summary and discussion of the main thesis findings as well as their applications to military and civilian populations. The work is evaluated in the broader context of the issue, and future work is proposed.

8.1 Summary and Conclusions

Female soldiers are often reported as being at a significantly greater risk of MSk overuse injuries, particularly in the hip and pelvis (Ministry of Defence 2016). With the recent decision to open Ground Close Combat roles in the British Armed Forces to females for the first time, it is expected that reported injury rates will rise, especially in the hip and pelvis. Carrying heavy body borne loads is often required during military training, and has been identified as a primary contributor to overuse injuries, leading to gait changes and increased forces in the MSk system (Birrell et al. 2007; Tilbury-Davis and Hooper 1999). A number of studies have identified both intrinsic and extrinsic injury risks factors in male and female soldiers, including low BMI (Moran et al. 2008) and poor aerobic fitness (Robinson et al. 2016). While biomechanical risk factors for injuries in male soldiers have been studied extensively, those for female soldiers had not been fully investigated. As a result, the biomechanical parameters involved in the development of hip and pelvic overuse injuries were still not understood. The work in this thesis therefore assessed the gait parameters and internal MSk forces in males and females carrying military-relevant loads through the use of optical motion tracking and musculoskeletal modelling.

Gait analysis of walking both with and without loads revealed that females and males displayed similar speed and stride length preferences when unloaded and carrying 20kg. This suggests that mixed-sex training groups may be acceptable when marching with light loads. However, a major finding of this work was that females preferentially produced a greater percentage of lower limb torque at the hip, while males produced more at the knee and ankle. As a result, when carrying 20kg females had larger peak hip joint contact forces than males, and showed more significant increases in hip muscle forces compared with walking unloaded. This increased hip dominance in females may be due to relative weaknesses of the knee and ankle musculature. Greater emphasis on heavy resistance training of the knee flexor and extensor muscles, to target type II muscle fibres associated with power (Nindl 2015), may encourage females to shift torque production down the kinetic chain, thus reducing forces at the hip.

When walking with 40kg, females significantly reduced their stride length and increased their time spent in double support, leading to significant gait differences between sexes. It is possible that females made these adaptations in an effort to limit the increasing forces at the hip, or because they could no longer maintain

normal gait due to lacking the required muscle strength. These results suggest that it is therefore not appropriate to have male and female soldiers walking or marching in step when carrying heavy loads. Analysis of kinetic variables indicated that the gait adaptations made by females were successful in reducing forces at the hip, significantly reducing the percentage of torque produced at the joint. As a result, females no longer displayed significantly greater peak hip JCFs than males, despite carrying a significantly greater percentage of their body weight. Females also showed less significant increases in hip joint torques, muscle forces and JCFs between the 20kg and 40kg load conditions due to these gait adaptations. Although successful at limiting hip MSk loading, these findings still suggest that females will be more likely to be injured when carrying heavy loads, as they are working past the limit where they can maintain normal gait function.

Analysis of the joint kinematics presented in Chapter 3 revealed that females displayed significantly more trunk forward flexion than males when carrying loads. It was concluded that this was due to females' lesser upper body mass and strength. Upper body resistance training has been shown to improve load carriage performance at almost the same rate as lower body resistance training (Knapik et al. 2012). Therefore training regimes for female soldiers should place greater emphasis on upper body and core strength and power development. When carrying loads, males were also observed to increase hip flexion during the weight acceptance phase of gait in order to attenuate the increased forces acting on the leg. This was not seen in females however, suggesting that they may not possess the quadriceps strength required to absorb the increased shock when carrying loads.

The finite element analysis presented in Chapter 4 identified that peak bone stresses occurred in the mid-sections of the pubic rami at toe push-off when carrying 20kg. Higher stresses were observed in the female model across all gait positions. Furthermore, peak von Mises stress did not increase linearly with load in either the male or female model, suggesting that gait adaptations may affect stress distribution in the pelvic cortex. Previous literature has suggested that pelvic stress fractures develop due to the pull of the adductor and hamstring muscles which insert onto the narrow pubic rami (Pope 1999; Hill et al. 1996). The work in Chapter 7 confirmed that adductor muscle force was highly correlated with peak stress values in the rami, however hamstring muscle forces were not. This thesis also identified the previously unreported effects of the pectineus muscle and medial hip JCF on peak rami stress values when carrying body borne load.

These findings have important implications for pelvic stress injury rates in soldiers, as female subjects displayed more significant increases in medial hip JCF with the addition of load than males, and had greater pectineus muscle force across all load conditions. The work in this chapter also confirmed that increased pelvic obliquity resulted in higher rami stresses. Pelvic obliquity is often attributed to weak gluteus medius and oblique abdominals, and hence training programmes should focus on increasing strength in these muscles.

When running with load, males and females were observed to adapt their gait differently. Females reduced their stride length while males increased the time spent in stance phase. When running with 40kg, females displayed periods of double support, possibly because they did not possess the muscle capacity required to propel themselves off the ground. As with walking, females were found to take significantly shorter strides than males when running with load. This was true even when leg length differences were accounted for. This finding supports the theory that males and females should not march together in double time while carrying heavy loads, and that the practice of having shorter soldiers march at the front of the troop may not be enough to prevent injuries. The gait adaptations made by females in response to the increasing load worked to reduce hip JCFs when running. As a result, males displayed significantly greater hip JCFs than females when carrying load, and showed significant increases in more hip muscle forces and torques than females.

Evaluation of muscle activity in Chapter 6 revealed that females significantly increased activation of the soleus, vastus lateralis and gluteus medius when carrying load. This suggests that they were more affected by the addition of loads than males, and thus under a greater subjective workload and energetic cost. It may therefore be beneficial for females to follow a more progressive load carriage training regime to allow more time to build the required muscle strength. In order to limit overuse injuries in females, it may also be beneficial to prioritise low-volume, high-intensity training to build aerobic capacity in place of endurance exercises, where females will fatigue sooner than males (Nindl 2015).

Finally, regression analysis identified that biomechanical risk factors for high forces in the hip included increased pelvic tilt when walking unloaded, and a more internally rotated ankle when walking with load. During running, biomechanical risk factors included longer stride length, greater trunk forward flexion and greater pelvic obliquity. In females, a more externally rotated and less flexed hip

corresponded to increases in hip JCF when running, as did increased ankle eversion and external rotation. In males, a more internally rotated and extended hip was found to increase hip JCF when running. Furthermore, an increase in hip dominance was found to correspond to a more significant increase in hip JCF in females compared with males, suggesting that females may be inherently more susceptible to higher musculoskeletal loads in the hip. These results highlight the importance of including both males and females in studies of the MSk system.

8.2 Future Work

This thesis has identified potential biomechanical risk factors for high MSk forces in the hip and pelvis, and suggested possible mitigation strategies. Although the evidence suggests that higher mechanical loading is a predictor for overuse injuries (Attwells et al. 2006; Bennell et al. 1999; Milner et al. 2006b), monitoring injuries and gait differences in a cohort of females training for GCC roles could provide greater insight into specific biomechanical risks. Gait changes associated with injuries in male soldiers (Springer et al. 2016; Nunns et al. 2016; Dixon et al. 2019) and male and female civilian athletes (Thijs et al. 2007; Willems et al. 2006; Van Ginckel et al. 2009; Cichanowski et al. 2007) are well reported, as discussed in Chapter 2. Although many intrinsic and extrinsic risk factors have been identified, there is still a lack of research into sex-specific biomechanical risk factors for overuse injuries in female soldiers.

The data used in this thesis was collected in a laboratory setting. In order to more accurately capture natural gait patterns, data collection should be done in a realistic training environment. This would allow for analysis of more body borne equipment including webbing, helmet and weapons. Measuring gait during a loaded training march would also allow for the influence of fatigue to be investigated. Fatigue has been suggested as a cause of MSk injury (Behrens et al. 2013; Nordin and Frankel 2001). Studies have attempted to induce load carriage fatigue in a laboratory setting through plyometric exercises (Wang et al. 2012) and treadmill walking (Lidstone et al. 2017; Filzah Pg Damit et al. 2017), however whether these fatigue the body in the same way as overground endurance marching is debatable.

Overstriding as a result of marching in step has been cited as a leading cause of pelvic stress fractures (Hill et al. 1996; Kelly et al. 2000). Although the effects of

overstriding on leg stiffness and ground reaction forces have been investigated during load carriage (Lobb et al. 2019), the effects on hip MSk forces have not yet been studied. It might therefore be useful to investigate how pelvic stresses are affected when subjects over- or under-stride. The results from Chapter 3 showed that, during walking with load, mean pelvic rotation was predominantly left-side anterior over the gait cycle. This suggests that leg length discrepancies or muscle weakness in the non-dominant leg may be exacerbated when carrying loads. Future work could therefore include analysis of body asymmetries, as female endurance athletes have been shown to be more likely to obtain an injury on their weaker leg (Cichanowski et al. 2007). Muscle strength tests could also be performed so that relative exertion rates could be better derived from EMG data.

The musculoskeletal model used in this thesis, Freebody, was scaled from the closest of ten anatomical datasets, resulting in scaling errors. In addition, it is probable that soldiers would have greater muscle mass and strength than the average person, which would not be accurately represented using the muscle and inertial parameters currently in the model. Using subject-specific geometry could reduce these errors but would involve imaging each participant and defining muscle volumes and attachment sites. Instead, the number of subjects in the reference data set could be increased, or a dataset representative of the military population could be created to improve the reliability of the model predictions. The musculoskeletal model outputs could be further improved by including muscle co-contraction considerations in the load sharing algorithm.

Finite element analysis has been used to model stress in the pelvic cortex during walking in males (Dalstra and Huiskes 1995; Majumder et al. 2004; Ricci et al. 2018; Volinski et al. 2018), however modelling of the pelvis during load carriage has not been previously investigated. This thesis presented stress distribution in the pubic rami for an average male and female subject walking with and without load. Future work on this area could include modelling more participants using subject-specific geometry and MSk forces to investigate the effect of bony geometry on stress distribution. Implementation of an automatic image segmentation technique would make it quicker and more clinically useful to develop subject-specific models. Statistical shape modelling could be used for automatic segmentation or to improve the scaling of reference models, however current shape models of the pelvis are not highly accurate due to the complexity of the anatomy (Meller and Kalender 2004; Kainmueller et al. 2009; Lamecker et al. 2004).

The finite element models of the hemi-pelvis developed in this thesis could be further improved by including the anatomy of the full pelvic ring. This would allow for more accurate representation of the SI-joint and pubic symphysis, as well as allowing for load transfer between the contralateral and ipsilateral pelvis to be studied. Simulation of the boundary conditions could be further improved through the inclusion of subject-specific femoroacetabular cartilage layers and contact properties (Liu et al. 2016). The finite element models developed in this thesis, and in much of the literature on gait modelling (Dalstra and Huiskes 1995; Volinski et al. 2018; Ricci et al. 2018), used static analysis of selected points in the gait cycle, however implementation of a dynamic model would allow for analysis over one or multiple full gait cycles to more accurately model mechanical fatigue of the tissues (Noda et al. 2018; Seo et al. 2014; Wesseling et al. 2019). Finally, the musculoskeletal and finite element models could be integrated together to create a single framework capable of determining joint kinetics, muscle forces and bone stresses (Xu et al. 2016; Ravera et al. 2018; Mo et al. 2018).

8.3 Broader Context

The work presented in this thesis describes a computational modelling approach to investigate the effects of military-relevant load carriage on gait. Potential biomechanical risk factors for high musculoskeletal loads in the hip were identified. There are, however, other approaches to reduce overuse injuries in the military. Pope (1999) implemented several prevention strategies including reduced marching speed, promotion of self-selected stride length and running on softer surfaces at lower intensities. They found that the incidence of pelvic stress fractures in female recruits fell significantly. Harman et al. (1997) implemented a specially designed conditioning system focused on improving weight lifting performance in female soldiers. After 24 weeks, the number of females who could qualify for "very heavy" US Army roles increased from 24% to 78%. The work presented in this thesis has the potential to further improve training programmes for female soldiers, with the aims of increasing performance and reducing injury rates.

With advancements in defence technology, such as the development of load bearing exoskeletons (Gorgey 2018), the operational demands on soldiers are changing and physical training approaches must adapt to reflect this (Nindl 2015). The sex-specific joint level mechanics and adaptations to load carriage presented in

this thesis could inform future equipment design. The finding that females are preferentially more hip dominant when walking and running, even when not carrying loads, has implications not just for military populations but also civilian athletes when developing training programmes and technique.

8.4 Concluding Remarks

This thesis presents one of the largest studies to directly compare males and females carrying load, and one of few to include females carrying the heavy loads (>25kg) required of Ground Close Combat soldiers. This work is also the first to model pelvic stresses during load carriage over the gait cycle in females. The results have highlighted the importance of including both sexes in studies of the MSk system. By directly comparing males and females this thesis has also identified sex-specific gait adaptation techniques that are proven to reduce joint torques, muscle forces and joint contact forces at the hip.

From the perspective of the individual soldier, this thesis has identified specific biomechanical risk factors for increased MSk forces in the hip, which may be corrected by strength training or a more progressive approach to load carriage. On a military-wide level, it is clear that males and females adapt their gait differently in response to load carriage, even when anthropometrics are accounted for, and that it may therefore not be appropriate to have males and females marching in step. The work in this thesis has confirmed that females work at a greater subjective workload and energetic cost when maintaining the same gait as males. These findings, combined with the influence of hormonal and other physiological factors, indicate that males and females will benefit from different training strategies to maximise performance and reduce the risk of injury.

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Appendix A

Documentation for Ethical Approval

Imperial College London

Imperial College Research Ethics Committee
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Anthony M J Bull
Bioengineering Department
Royal School of Mines
Imperial College London
London, SW7 2AZ

22nd January 2018

Dear Professor Bull

Study Title: Analysis of gait kinematics and kinetics during load carriage walking and the walk-to-run transition.

ICREC reference: 17IC4270

The above study was approved by your Head of Department on date and by the Joint Research Compliance Office on 22nd January 2018.

Under the Imperial College Research Ethics Committee process, a study that has been reviewed by the Joint Research Compliance Office and Head of Division/Department (or Principal), where no significant ethical issues have been identified in the protocol or ethics application, can be approved without requiring it to go to full committee.

Documents

The documents reviewed were:

- ICREC Application form
- Protocol (v2 22/01/18)
- JRCO Sponsorship and Insurance Request
- Information Sheet (v2 22/01/18)
- Consent Form (v2 22/01/18)
- Email advert (v1)
- Poster advert (v1 01/05/17)

Yours sincerely,



Ruth Nicholson,
Research Governance Manager,
Imperial College London

Investigators: Prof Anthony MJ Bull
Prof Alison McGregor
Suzanne Shanel

Analysis of gait kinematics and kinetics during load carriage walking and the walk-to-run transition.

Participant Information Sheet

Version 2, 22/01/2018

We would like to invite you to join a research study. Before you decide it is important for you to understand why the research is being done and what it will involve. This form gives detailed information about the research study, and is yours to keep. Please take time to read the following information carefully and discuss it with others if you wish before you decide whether or not you wish to take part. Ask us if there is anything that is not clear or if you would like more information.

What is the purpose of the study?

The aim of this study is to measure changes in walking patterns of participants while carrying a military pack. Women in the military are more likely to obtain a musculoskeletal injury than men. This study will help to determine the reasons for this increased risk and the findings could help to reduce injuries to female military personnel.

Why have I been chosen?

We are asking you to consider participating in this study because you are a healthy, physically fit person between the ages of 18 and 36 years, you are not pregnant and you don't have any systemic inflammatory or connective tissue disorders, or an allergy to adhesives. We are recruiting a total of 40 people.

We are looking for healthy, physically fit volunteers, who can complete basic military fitness tests of (a) lift 40kg (b) run 2.4km in 14 minutes. You will NOT be asked to undertake these tests, just to be fit enough to do them.

Analysis of gait kinematics and kinetics during load carriage walking and the walk to run transition
Version 2 (22/01/2018)

Do I have to take part?

No, this is entirely up to you. If you would like to take part, you will be asked to sign a consent form. Even after you have signed this consent form and agreed to join the study, you are free to withdraw from the study at any time. If you decide not to take part, or withdraw from the study, it will not affect any future interactions that you may have with Imperial College London. If you chose to withdraw from the study at any time, the information collected up until that point will be retained and used in the study; however, no further data will be collected. Please inform any member of the research team if you no longer wish to participate in the study.

What will happen to me if I take part?

Once you have decided to take part in this research, you will be asked to come to the Motion Analysis Laboratory on the South Kensington Campus of Imperial College London, SW7 2AZ where our research team will discuss the study with you and answer any questions you may have. If you are still happy to take part, we will ask you to sign the consent form. You will be asked to attend one session lasting approximately 60 minutes. The investigator of this study will measure your height and weight, and will ask you to provide us with some information regarding your age, sex, or any history of lower limb pain or injury. This collected information will be safely stored and will only be used to process the data to allow for comparison across participants. No personal information, such as surname, will be used in this study or any publications that result.

To track your movement during walking, reflective markers will be stuck to your bare leg at the start, you will therefore be required to wear shorts. With your permission, we may record video footage of you during the study. Electromyographic sensors will be placed on your leg as to measure the activity of the muscles. You will then be asked to walk and jog along a short walkway at a comfortable speed. You will be asked to repeat this three times, the first time will not involve wearing a military pack, to measure your natural walking pattern. The second and third times will involve wearing a pack with either a 20 or 40kg weight inside.

What are the side effects, and are there any risks in taking part?

The adhesive used to attach the reflective markers to your skin may cause some redness. If you have any allergies or skin conditions (e.g. eczema, dermatitis, etc.), please inform the researcher prior to the placement of the markers, as you may be unable to take part. If you notice any irritation during or immediately following the study, please inform the researcher and a first aider will be called to assist you. The area will be cleaned with soap and water. If

the irritation persists for more than 24 hours, please contact us and you may be advised to consult your local medical clinic.

If you experience any muscle or joint pain and/or discomfort during the session please inform the researcher, and if you need to rest and/or want to stop, you can do so at any time.

What are the possible benefits of taking part?

There are no direct benefits to you as a participant, other than contributing to biomechanics and military research. However, there is the potential benefit to the clinical and research military community. This study could contribute to helping improve training and injury prevention for women in the military.

Will my taking part in this study be kept confidential?

Any information you give us will be kept confidential. If the study is published in a book or scientific journal, no individual will be identified in any way. Your personal details will be stored on a computer which belongs to Imperial College London. The study data (demographic information regarding age, sex, height, weight and data collected during trial) will be stored on a password protected computer and the paper forms (such as anthropometric data and demographic data) will be kept in secure filing cabinet at Imperial College London, accessible by only the PI and co-investigator. Personal details such as consent forms containing the personal information (such as forename and surname) will be kept in a locked filing cabinet, separate to the study data. Only the principal investigator and co-investigator named on this application will have access to the data.

What will happen to the results of the research study?

The results of the study will be analysed by the research team and presented at biomechanical and clinical conferences and published in scientific journals. No individual subject will be identified in any report or presentation arising from the research. We are unable to provide you with your individual results; however, you can be provided with a summary report of our findings at the end of the study, upon your request.

Will I be paid for taking part in the study?

You will not be paid for your participation in the study.

Who has reviewed the study?

Analysis of gait kinematics and kinetics during load carriage walking and the walk to run transition
Version 2 (22/01/2018)

This study was reviewed by the Dean of the Faculty of Engineering and the Imperial College Joint Research Compliance Office (JRCO).

Contacts for further information:

If you would like to consider this study further before you make your decision, please take your time to do so. To request further information please contact Miss Suzanne Shanel, who may be reached via email at s.shanel15@imperial.ac.uk.

Appendix B

Additional Results for Chapter 3

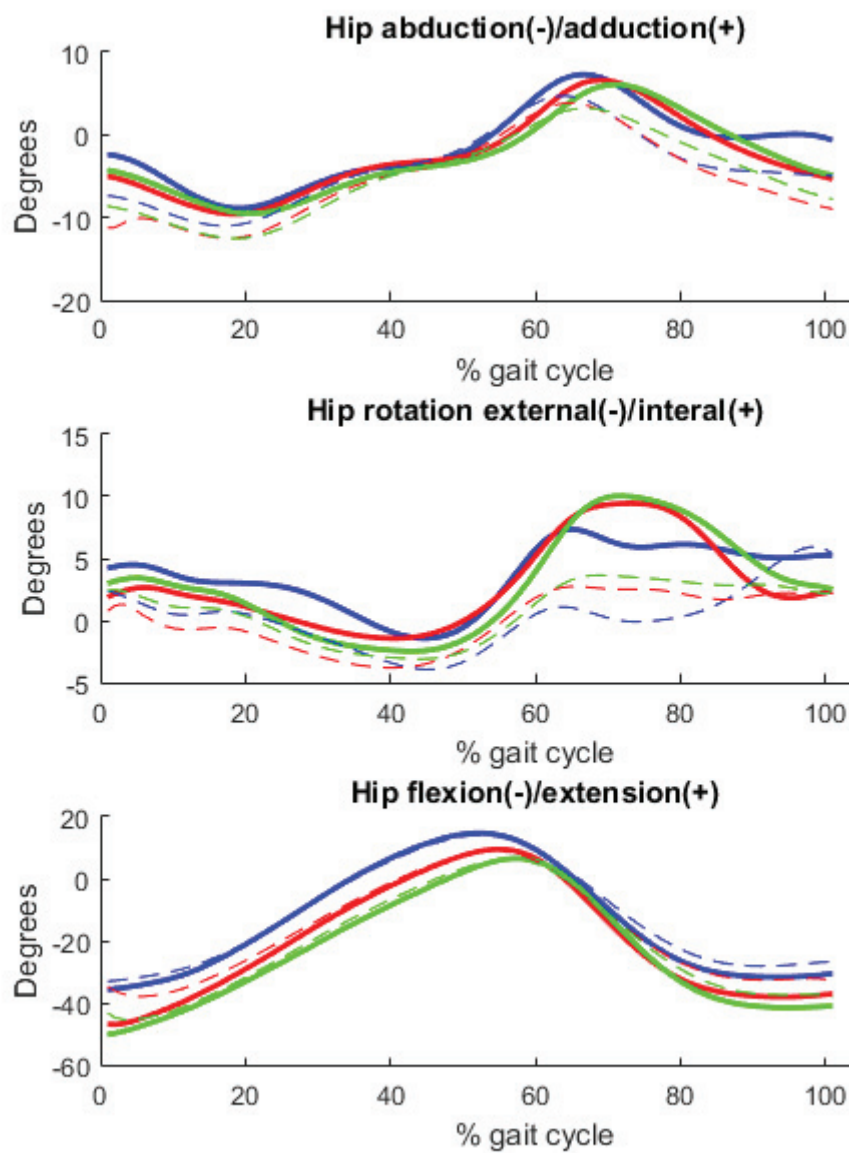


Figure B.1: Hip joint angles during walking. Blue-0kg, Red-20kg, Green-40kg, Solid line-females, dashed line-males.

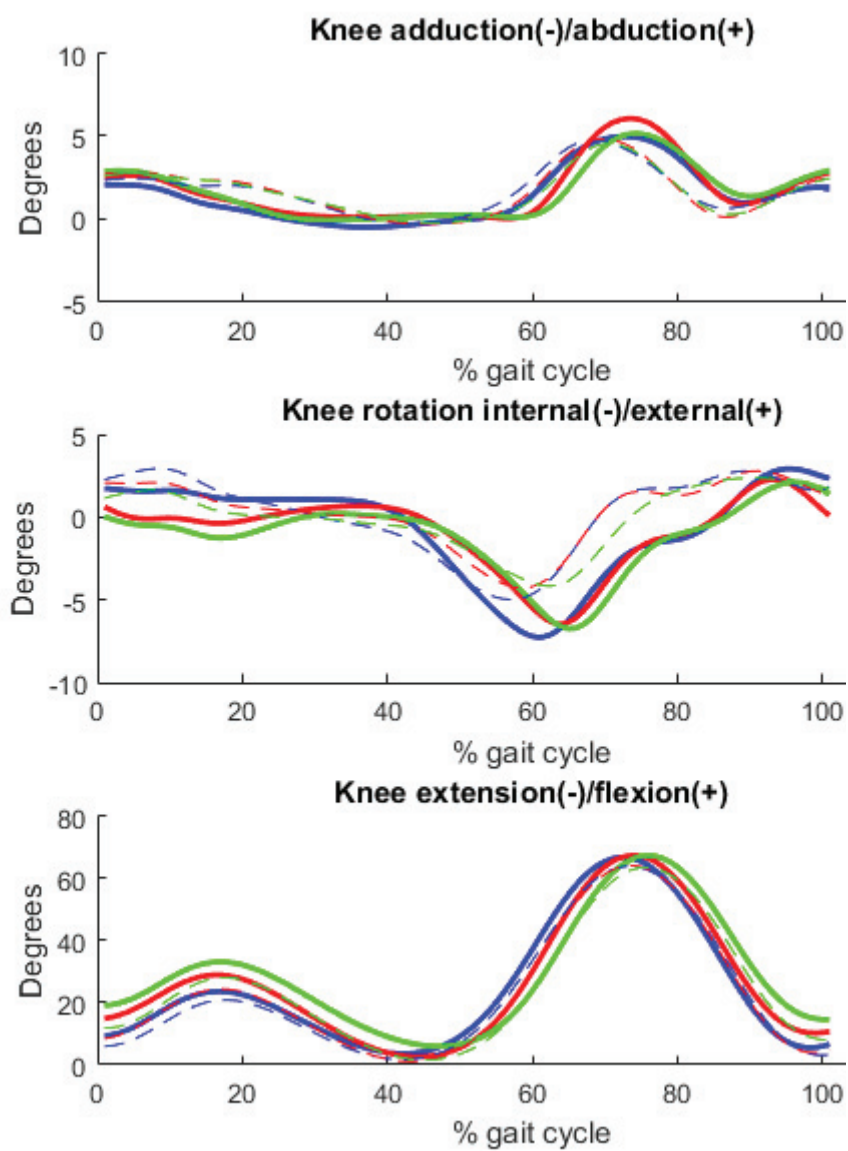


Figure B.2: Knee joint angles during walking. Blue-0kg, Red-20kg, Green-40kg, Solid line-females, dashed line-males.

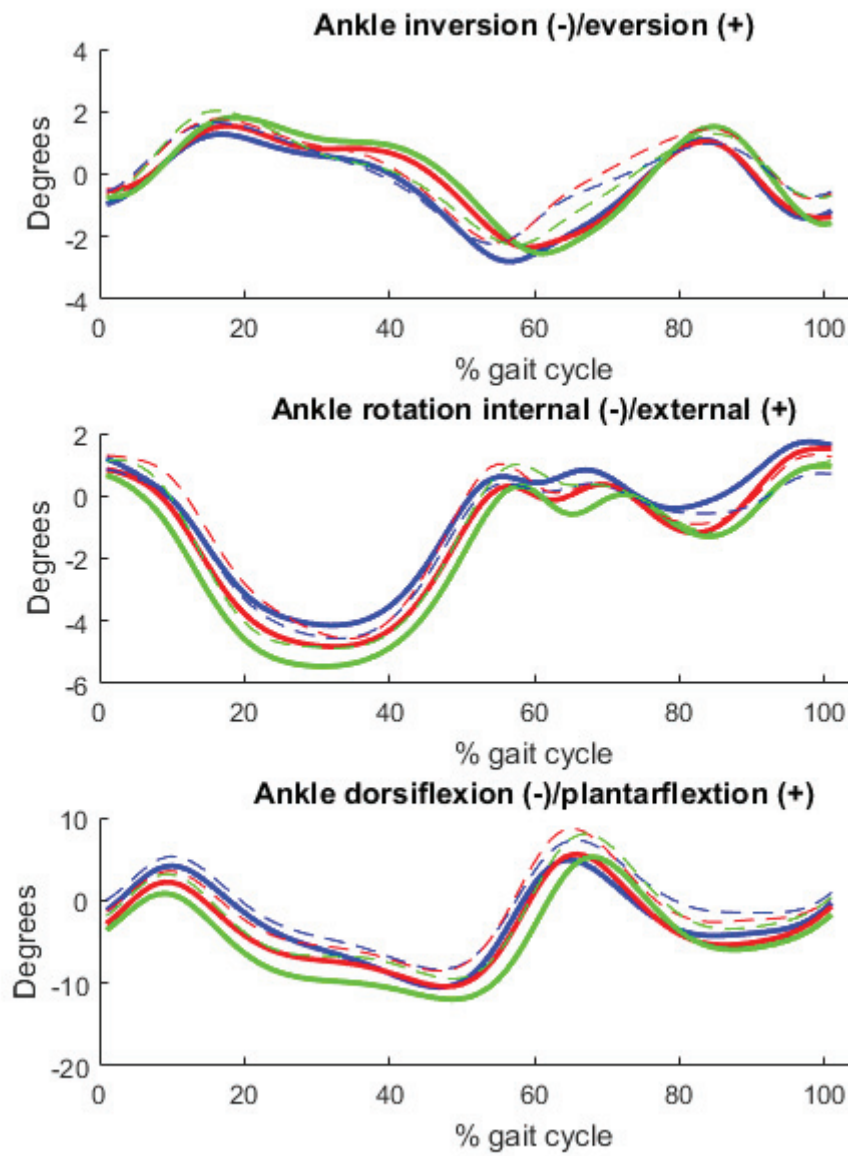


Figure B.3: Ankle joint angles during walking. Blue-0kg, Red-20kg, Green-40kg, Solid line-females, dashed line-males.

Appendix C

Additional Results for Chapter 4

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Adductor brevis	0.02	0.02	0.01	0.02	0.01	0.01
Adductor longus	0.02	0.02	0.01	0.02	0.01	0.01
Adductor magnus	96.07	96.07	82.79	231.48	213.07	299.60
Biceps femoris	262.02	262.02	272.32	190.14	66.27	93.53
Gemellus	25.28	25.28	39.83	25.03	35.66	42.14
Gluteus max.	402.04	402.04	600.09	361.31	582.92	532.30
Gluteus med.	485.47	485.47	654.49	549.53	859.12	629.25
Gluteus min.	121.34	121.34	174.84	113.55	198.09	115.36
Gracilis	0.01	0.01	0.00	0.02	0.00	0.01
Iliacus	0.05	0.05	0.03	0.04	0.02	0.02
Obturator ext.	0.43	0.43	32.11	13.20	0.06	60.65
Obturator int.	109.68	109.68	179.84	107.48	141.20	200.20
Pectineus	0.02	0.02	0.01	0.02	0.01	0.01
Piriformus	48.80	48.80	69.97	43.24	63.82	67.56
Psoas minor	0.02	0.02	0.02	0.02	0.01	1.17
Quadratus femoris	39.51	39.51	65.58	51.36	60.55	85.73
Rectus femoris	0.01	0.01	0.01	0.02	0.01	0.05
Sartorius	0.01	0.01	0.00	0.01	0.00	0.00
Semimembranosis	115.94	115.94	114.01	110.55	91.77	65.59
Semitendinosis	98.26	98.26	99.11	85.19	48.00	45.11
Tensor f. l.	0.01	0.01	0.00	0.01	0.00	0.00

Table C.1: Muscle forces (N) for gait position 1: Weight Acceptance.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Adductor brevis	0.12	0.07	0.04	0.08	0.02	0.02
Adductor longus	0.17	0.09	0.05	0.14	0.02	0.02
Adductor magnus	0.07	0.03	0.02	0.02	0.01	0.02
Biceps femoris	0.01	0.00	0.00	0.00	0.00	0.00
Gemellus	0.06	0.13	1.76	0.01	0.01	0.03
Gluteus max.	169.40	288.07	409.01	142.28	269.83	332.27
Gluteus med.	288.99	292.55	539.76	204.47	396.32	364.28
Gluteus min.	102.43	135.85	188.03	143.59	199.09	174.93
Gracilis	0.02	0.01	0.00	0.00	0.00	0.00
Iliacus	40.67	48.78	75.36	71.31	0.47	0.27
Obturator ext.	0.11	0.07	0.04	0.04	0.02	0.03
Obturator int.	0.06	0.09	0.06	0.01	0.01	0.02
Pectineus	0.82	2.70	4.21	2.45	0.03	0.03
Piriformus	0.06	8.89	27.80	0.01	0.01	0.02
Psoas minor	0.76	0.98	0.38	0.01	0.00	0.00
Quadratus femoris	0.04	0.02	0.01	0.01	0.01	0.01
Rectus femoris	168.63	234.54	350.53	385.10	335.46	297.78
Sartorius	47.78	78.57	92.33	98.41	88.13	74.60
Semimembranosis	0.01	0.00	0.00	0.00	0.00	0.00
Semitendinosis	0.01	0.00	0.00	0.00	0.00	0.00
Tensor f. l.	38.51	59.28	75.93	80.21	78.57	68.29

Table C.2: Muscle forces(N) for gait position 2: Midstance.

	Female			Male		
	0kg	20kg	40kg	0kg	20kg	40kg
Adductor brevis	84.21	115.04	121.25	100.95	92.40	105.33
Adductor longus	197.47	301.03	307.40	228.46	239.14	237.11
Adductor magnus	91.27	0.00	0.00	0.01	0.00	0.01
Biceps femoris	0.00	0.00	0.00	0.00	0.00	0.00
Gemellus	0.00	0.00	0.00	0.00	0.00	0.00
Gluteus max.	296.04	536.51	547.20	59.76	265.96	36.35
Gluteus med.	368.37	715.89	593.17	318.86	598.77	302.98
Gluteus min.	135.78	244.25	183.68	183.34	366.24	177.91
Gracilis	39.14	26.82	24.86	12.34	0.00	14.08
Iliacus	330.26	461.98	457.56	308.88	356.72	316.00
Obturator ext.	180.74	45.48	59.02	151.98	0.01	163.87
Obturator int.	0.01	0.00	0.00	0.00	0.00	0.00
Pectineus	49.21	70.45	76.89	52.86	47.82	54.90
Piriformus	0.00	0.00	0.00	0.00	0.00	0.00
Psoas minor	7.30	13.75	13.73	4.99	0.31	5.46
Quadratus femoris	20.77	0.00	0.00	0.00	0.00	0.00
Rectus femoris	885.32	1460.57	1526.73	1059.19	1358.57	1079.08
Sartorius	238.03	453.06	488.63	273.04	394.96	276.62
Semimembranosis	0.00	0.00	0.00	0.00	0.00	0.00
Semitendinosis	0.00	0.00	0.00	0.00	0.00	0.00
Tensor f. l.	169.81	318.89	346.06	196.25	281.91	198.08

Table C.3: Muscle forces (N) for gait position 3: Push-off.

Appendix D

Additional Results for Chapter 5

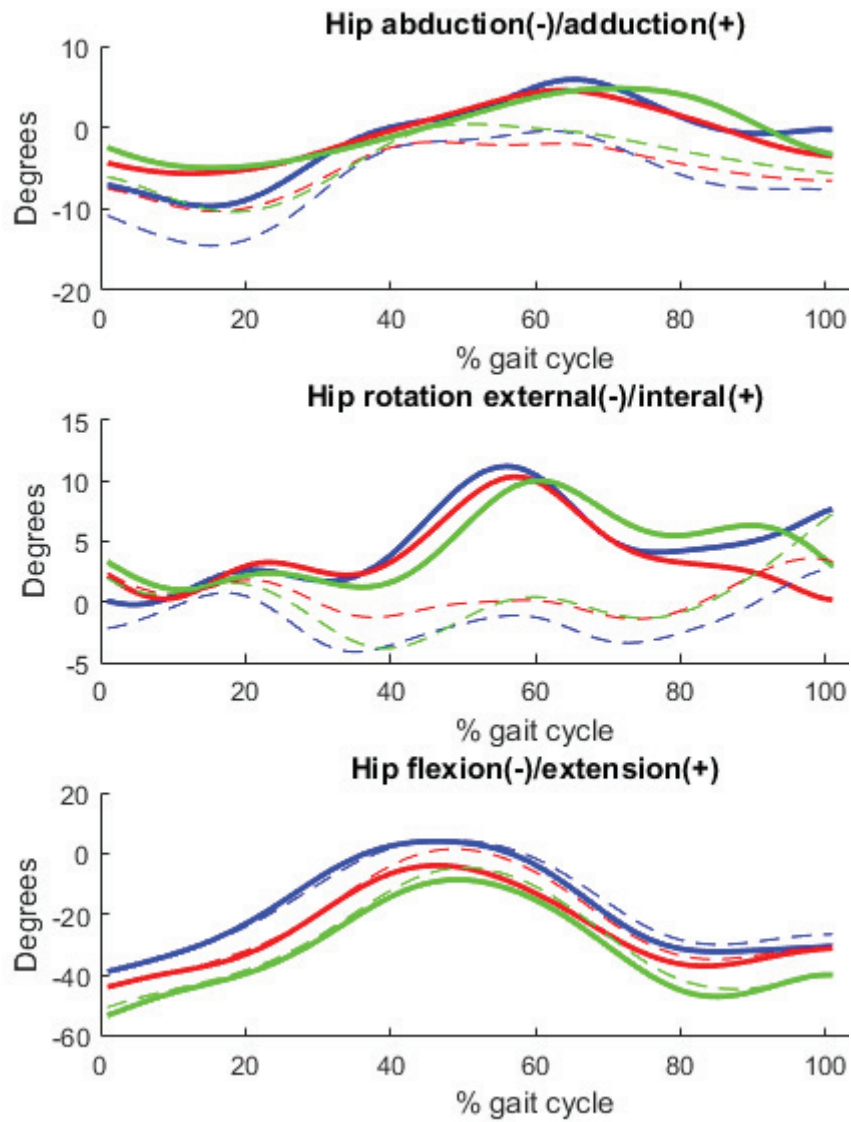


Figure D.1: Hip joint angles during running. Blue-0kg, Red-20kg, Green-40kg, Solid line-females, dashed line-males.

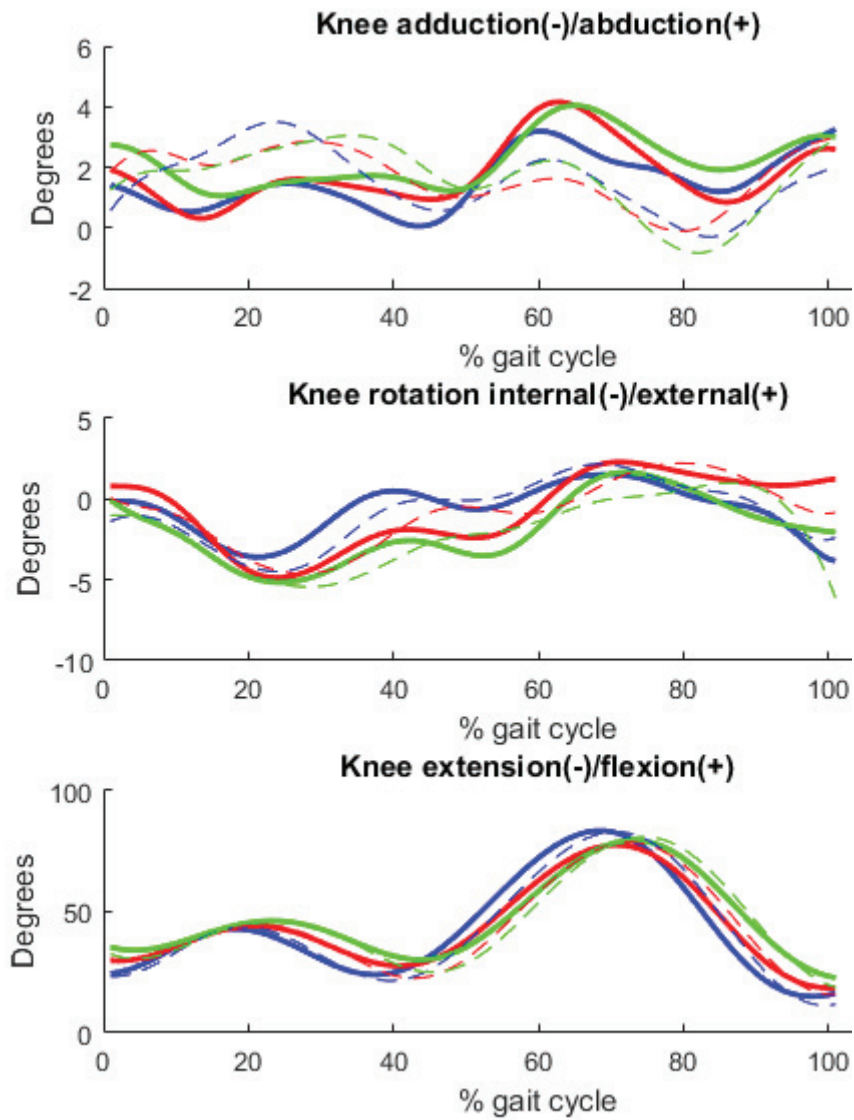


Figure D.2: Knee joint angles during running. Blue-0kg, Red-20kg, Green-40kg, Solid line-females, dashed line-males.

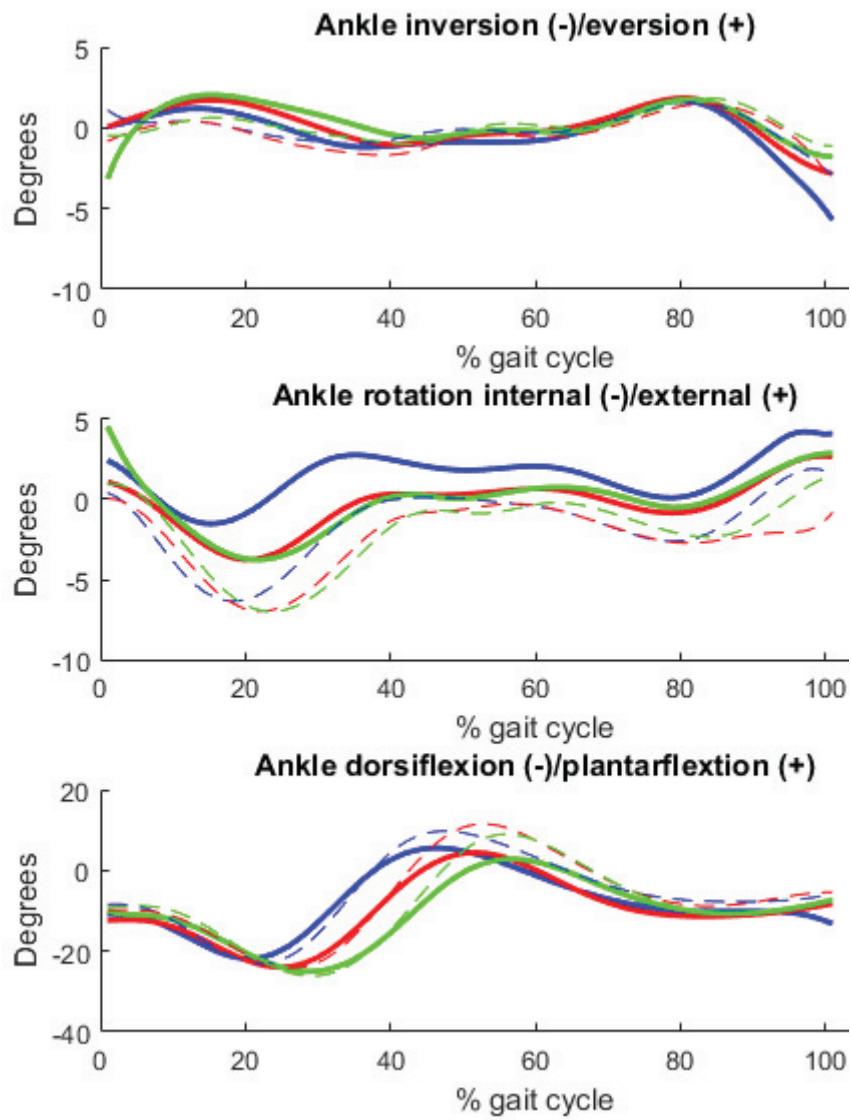


Figure D.3: Ankle joint angles during running. Blue-0kg, Red-20kg, Green-40kg, Solid line-females, dashed line-males.