EFFECTS OF STRIDE LENGTH ON LOWER LIMB STIFFNESS WHEN RUNNING WITH BODY BORNE LOAD

by

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ABSTRACT

Introduction: During military activities, soldiers are often required to run at a fixed cadence with body borne load, but these loads purportedly increase leg stiffness, leading to increased risk of musculoskeletal injury. Yet, to date, it is unknown how altering stride length when running with body borne load affects lower limb stiffness for males and females. **Purpose:** To quantify leg stiffness, and lower limb joint (hip, knee and ankle) stiffness for males and females using different stride lengths to run with body borne loads of 20 kg, 25 kg, 30 kg, and 35 kg. **Methods:** Twenty-seven (17 males and 10 females) participants (age: 21.2 ± 2.3 years, height: 1.7 ± 0.1 m, and weight: 75.5 ± 11.3 kg) had leg and joint stiffness quantified while running at 4 m/s with four load conditions (20, 25, 30, and 35 kg). With each load condition, participants performed three run trials using either: their preferred stride length (PSL) and strides that are 15% longer (LSL) and shorter (SSL) than their PSL. **Statistical Analysis:** Leg and hip, knee, and ankle stiffness were submitted to a RM ANOVA to test the main effect and interaction of load (*20, 25, 30, and 35 kg*), stride length (*PSL, PSL+15%, and PSL-15%*), and sex (*male vs female*). **Results:** Body borne load increased leg stiffness (P=0.006). Male participants decreased leg stiffness as stride lengthened from SSL to PSL and PSL to LSL (P=0.026; P<0.001), while females did not change leg stiffness with longer strides (P>0.05). Body borne load increased peak vGRF (P<0.001). Males increase peak vGRF with each increase in stride length (P=0.010; P=0.011), while females only increased peak vGRF between PSL and LSL (P<0.001). Knee (P<0.001) and ankle (P=0.013) stiffness increased with the addition of body borne load, but load had no significant effect on hip stiffness (*P*=0.723). Increasing stride length significant decreased ankle stiffness (*P*=0.003), but had no effect on hip (*P*=0.661) or knee (*P*=0.170) stiffness. Sex had no significant effect on hip (*P*=0.880), knee (*P*=0.234), or ankle (*P*=0.081) stiffness. **Conclusion:** Running with body borne load increased leg stiffness and potential risk of musculoskeletal injury. But, only male participants decreased leg stiffness and injury risk with longer strides. Both the knee and ankle increased joint stiffness, and risk of musculoskeletal injury with the addition of body borne load. The ankle, however, decreased joint stiffness with longer strides.

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LIST OF ABBREVIATIONS

MSI	Musculoskeletal Injury
GRF	Ground Reaction Force
PSL	Preferred Stride Length
SSL	Short Stride Length
LSL	Long Stride Length
CoP	Center of Pressure
PS	Peak of Stance

CHAPTER ONE: INTRODUCTION

Musculoskeletal injuries (MSIs) are an increasing problem among military personnel. The incidence of MSIs for military personnel has increased seven-fold in the last 25 years[1], with basic trainees reportedly exhibiting an injury rate of 952 per 1000 person-years between 2012 and 2014[2], and females exhibiting injury rates twice as high as their male counterparts[3]. These MSIs have a substantial financial and physical cost for the military, limiting their ability to produce combat-ready personnel. The military spends $700 million treating MSIs annually[4]; yet, a basic trainee who sustains MSIs is still three times as likely to be discharged from service[2]. The most common MSIs are overuse injuries, including sprains, strains, and damage to soft-tissue, and account for about 80% of all injuries military personnel suffer during training[5,6]. Most training-related MSIs occur in the lower limb, with approximately 25% of these injuries occurring at the knee joint[5,7,8]. Overuse injuries occur from small, repetitive loads placed on the musculoskeletal system[5] and commonly occur during the load-bearing activities required during military training, such as running or walking[5]. These body borne loads typically consist of personal fighting and protective equipment, that routinely weigh between 20 and 40 kg[9] and reportedly alter lower limb biomechanical patterns[10–13] leading to the high incidence of MSI[14–16], particularly at the knee.

During locomotion, the lower limb biomechanical adaptions that occur from the addition of body borne load may increase the risk of MSI. Body borne load reportedly increases peak vertical ground reaction force (vGRF) during walking[11,17–19] and running[9,13], increasing strain on the lower limb musculoskeletal system[19–21] and risk of MSI. In order to attenuate the increased GRF and prevent collapse of the lower limb, load carriers reportedly increase leg stiffness[13,22]. Silder et al.[13] reported individuals significantly increase leg stiffness (i.e., resistance to deformation[23]) due to both an increase in peak vGRF as well a decrease in change in leg length when running with small body borne loads (~ 20 kg or less). But to date, it is unknown if load carriers exhibit a similar increase in leg stiffness when running with the heavy body borne loads commonly worn during military activities (between 20 kg and 35 kg). While adequate leg stiffness is necessary to successfully perform many dynamic tasks[23], such as running, large magnitudes of leg stiffness increase the transmission (i.e., loading rate) of the GRF to the lower limb[13,24–26], further straining the musculoskeletal system[19,27]. The loading rate of GRF is reported to significantly increase when walking with body borne loads of 32 kg[19] and 40 kg[17], but it is currently unknown if individuals exhibit similar increases in loading rate when running with military relevant body borne loads (i.e., greater than 20 kg).

Increases in leg stiffness during locomotor activities may be attributed to greater torsional stiffness of the lower limb (hip, knee, and ankle) joints. Torsional joint stiffness is described as the change in angle that occurs when a given moment is applied[28] and may provide insight into the risk of MSI at each joint. When running with military relevant loads, participants exhibit a significant increase in peak hip and knee flexion moments, but not a similar increase in hip and knee flexion posture[9,29]. To compensate for these elevated moments, load carriers may increase the stiffness of each lower limb joint, subjecting the musculoskeletal system to greater stress and contributing to the risk of sustaining a MSI by requiring more rapid transmission of GRF through the limb[30]. In fact, a significant increase in knee joint stiffness is reported when walking with body borne load[31], but it is unknown if similar increases in joint stiffness are exhibited when running with load. Determining how the stiffness of each lower limb joint is altered during load carriage can provide the knowledge necessary to improve training methodologies and reduce risk of MSI during military activities.

During military training, personnel are often required to walk or run at a fixed-cadence. Running or walking with a fixed cadence requires all personnel to use the same stride length, potentially leading to the high incidence of MSI[32], by causing shorter individuals to over-stride and taller individuals to under-stride. Seay at al.[32] reported that changing stride length when walking with a 20 kg load altered peak GRF, and lower limb joint angles and moments. Specifically, when adopting a 15% slower cadence than preferred (i.e., over-striding), participants significantly increased peak knee flexion moment[32]; however, they may have mitigated the elevated injury risk through the use of greater peak hip and knee flexion angles, potentially reducing leg stiffness[33,34]. Conversely, when adopting a 15% faster cadence (i.e., under-striding), participants reduced peak GRF and ankle dorsiflexion moment[32], but may increase leg stiffness by reducing peak knee flexion and ankle dorsiflexion angle[34]. Considering participants increase leg stiffness to prevent collapse of the limb and maintain locomotion when running with body borne load, they may not be able to modulate lower limb stiffness when changing stride length to run with load as they do without load[35]. Yet, the effect of changing stride length on lower limb biomechanics, and subsequent risk of MSI, when running with body borne load is largely unknown.

As of 2013, females could once again serve in direct combat positions in the military[36]. When serving in these infantry positions, military personnel can be required to run with very large loads which may contribute to the high rate of injury seen by females in the military[3]. To date, sex comparisons during loaded locomotion have been limited to analyzing sagittal plane biomechanics while walking with load, where no differences between males and females have been observed[37,38]. When running without load, differences in the frontal and transverse planes, but not the sagittal plane, have been observed between males and females[39,40], yet little to no research exists directly comparing sex differences when running with military relevant loads. Considering differences in strength between males and females have been reported[41,42], it is unclear if males and females adopt similar biomechanical patterns with the addition of load or changes in stride length. This work sought to fill that critical void.

Specific Aims

Specific Aim 1

To compare lower limb stiffness during over-ground running with small changes (5, 10 and 15 kg) in body borne load. Specifically, this study quantified leg and joint (hip, knee and ankle) stiffness while participants ran over-ground at 4.0 m/s with four different body borne loads (20, 25, 30 and 35 kg).

Hypothesis 1.1

Participants will exhibit a significant increase in leg stiffness with each incremental addition (5, 10, and 15 kg) of body borne load.

Hypothesis 1.2

With each incremental addition (5, 10, and 15 kg) of body borne load, participants will significantly increase hip, knee, and ankle joint stiffness, with the knee exhibiting the largest increase in joint stiffness.

Significance

Understanding how body borne load affects the stiffness of the stance leg during running may lead to a better understanding of the etiology of military MSIs. This information can help shape military training protocols and provide knowledge regarding the specific lower limb musculature that needs strengthening for a reduction in risk of MSI by optimizing leg or joint stiffness, and increasing the safe attenuation of GRF during running with load.

Specific Aim 2

To determine how altering stride length affects lower limb stiffness when running over-ground with small changes in body borne load (5, 10, and 15 kg). Specifically, this study determined how using preferred (PSL), long (PSL + 15%), and short (PSL - 15%) stride lengths impacts leg and joint (hip, knee and ankle) stiffness while participants run over-ground at 4.0 m/s with four different body borne loads (20, 25, 30 and 35 kg).

Hypothesis 2.1

Leg stiffness, and hip, knee, and ankle joint stiffness will significantly increase when using both the preferred stride length (PSL) and long stride length (LSL) when compared to the short stride length (SSL), and when using the LSL compared to the PSL.

Hypothesis 2.2

With the 5, 10 and 15 kg increase in body borne load, participants will exhibit a significant increase in leg stiffness, and hip, knee, and ankle stiffness when running with the PSL and LSL, but not SSL.

Significance

Determining if altering stride length leads to significant changes in stance leg stiffness when running with body borne load can help the military reduce their personnel's risk of MSI during basic and/or advanced training. The military can use this information to select exercises for military training, or prevent their personnel from using "hazardous" stride lengths during training.

Specific Aim 3

To determine if the leg and joint stiffness adaptations observed when running over-ground with small changes (5 kg, 10 kg, 15 kg) in body borne load and 15% changes in stride length are similar between males and females. Specifically, this study compared how males and females adapt leg and joint stiffness to run over-ground at 4 m/s with SSL, PSL, and LSL when carrying 20 kg, 25 kg, 30 kg, and 35 kg.

Hypothesis 3.1

With each incremental increase of 5 kg, 10 kg, and 15 kg in body borne load, males and females will not exhibit significantly different adaptations from each other in leg and hip, knee, and ankle stiffness.

Hypothesis 3.2

As participants increase stride length from SSL to PSL and from PSL to LSL, males and females will not exhibit significantly different adaptations from each other in leg and hip, knee, and ankle stiffness.

Significance

Determining if males and females exhibit different leg and joint stiffness adaptations when running with increasing body borne load and stride length can help further describe what factors may be responsible for the greater rate of injury seen among female military personnel. The military can use this information to adapt a training program that focuses on reducing injury among female personnel.

CHAPTER TWO: LITERATURE REVIEW

History of Load Carriage

The loads carried by military personnel have changed dramatically over time. Until the 18th century, loads carried by infantry rarely exceeded 15 kg as heavier loads were typically carried by auxiliary transport such as horses or carts[14]. After the 18th century, the use of these forms of transport declined and soldiers were required to carry greater loads on foot. Since the 1950s, the average load carried by a solider has increased 50% from 30 kg to 45 kg[15] with the majority of this increase occurring in the time period following the Vietnam war[43]. It is possible that the increase in load seen in recent years can be attributed increased protection and firepower carried by military personnel[14]. Methods of load carriage have increased in efficiency as the ability to measure energy expenditure has improved. Studies have found that loads placed closest to the body's center of mass are the most energy efficient[44]. This discovery has led to the load being carried primarily about the torso and has helped reduce the physiological stress of carrying load.

Injuries

An increase in the load carried by military personnel is coupled with the rising rate of injury in the military. Since the 1980's the number of injuries among military personnel has increased seven-fold[15]. This increase is mostly due to the rise in MSIs, as they are currently the second most common injury associated with load carriage following foot blisters[15]. Between the years of 2012 and 2014 the injury rate of basic trainees was reported to be 952 injuries per 1000 person-years[2], costing the military upwards of $700 million annually[4]. Kaufman et al.[7] gathered injury data from military epidemiologic studies and found that injury rates during military training were as high as 12% of male recruits per month during basic training and 30% per month during special forces training. Furthermore, females in the military are twice as likely as their male counterparts to suffer an MSI, and up to ten times as likely to suffer stress related bone injuries[7]. The two most common sites for injury in all personnel are the vertebral column and the lower extremities, consisting of 40% and 39% of all injuries, respectively[5]. While the numbers vary from 25%[5,7,8] to 50%[15], the knee is consistently the most frequent lower extremity overuse injury site. Other common lower extremity injuries were stress fractures and joint derangements, defined as "meniscal tears of the knee, loose bodies in the knee, [and] articular cartilage disorders"[5]. Overuse injuries are characterized as pain and inflammation at the injury site, typically caused by excessive stress on the musculoskeletal system due to repetitive loading cycles[5]. Biomechanical adaptations seen when performing dynamic activities such as walking and running with military relevant loads have been found to increase the stress placed on the musculoskeletal system and likely contribute to the risk of sustaining a MSI[5].

Walking with Body Borne Load

The majority of research regarding load carriage has found that the addition of load leads to increased anterior trunk and head lean[45–47]. This adjustment keeps the body's center of mass from moving backward, thus maintaining postural stability. Increased trunk and head lean are associated with higher muscular tensions which can lead to muscle strain if excessive load is carried. The use of a front pack leads to a more upright trunk posture[46] which may help to prevent such muscle strains. However, there is not enough data to support use of a front pack as a safer alternative.

Differences in GRF have commonly been reported in load carriage research is GRF. During walking, vertical and anteroposterior GRF increase proportionally to the load added[11,18,48,49]. In addition to peak GRF, the rate at which force is applied to the lower limb can have injury implications as well[19,30]. Specifically, high GRF loading rates have been linked to stress fracture risk, particularly in female runners[20]. Loading rate reportedly increases significantly when walking with loads of 32 kg[19] and 40 kg[17], however, no significant increases in loading rate have been observed when walking with a 20 kg load[17]. These findings suggest that load carriers do not exhibit similar biomechanical adaptations at all weight loads. It is thought that one contributor to increased loading rate is reduced energy absorbed by eccentric muscle contraction upon ground impact[19]. This creates a larger stress on the lower limb skeleton, potentially leading to increased stress fracture risk. Alterations in loading rate and energy absorption may be caused by alterations in leg stiffness seen during loaded walking.

During walking, leg stiffness can determine the amount of vertical excursion experienced by the body's center of mass[22,31]. Vertical leg stiffness reportedly increases linearly to the amount of load added during walking[22,31]. It is hypothesized that these increases are an adaptation made in order to keep center of mass vertical excursion constant, as alterations in center of mass excursion have been linked to increased metabolic cost[22,50]. To increase stiffness and prevent lower limb collapse, it is thought that isometric muscle activity must increase[31], which may increase delayed-onset muscle soreness[51] and potentially lead to overuse MSI risk. However, previous studies that have analyzed leg stiffness during loaded walking have not collected metabolic and neuromuscular data to definitively support this hypothesis.

As body borne load is increased, hip range of motion increases during walking[10,18,45,52]. However, changes in sagittal plane hip range of motion are not consistent. Increases in sagittal hip range of motion are typically seen when increasing load from 0 kg to 10-15 kg[18,45,52] and from 16 to 40 kg[45], however these changes are not apparent when increasing load from 40 to 50 kg[45]. Conversely, Birrell et al.[10] found no increase in hip flexion/extension range of motion for any of the load conditions studied (8, 16, 24, and 32 kg). This difference is difficult to explain as the protocols were similar for all studies. The increased range of motion seen in other studies seems to be more prevalent when walking with lighter load conditions, while heavier load conditions do not elicit the same response. One consistent finding is that body borne load increases frontal plane hip range of motion. Increases in hip abduction angle between 0 and 32 kg[10] and increased step width between 0 and 15 kg[52] have been observed during walking. This adjustment is likely made to increase the base of support and to help improve gait stability as body borne loads are added.

In addition to increases in range of motion, adding load when walking alters hip joint kinetics as well. Significant increases in internal hip extension moment we observed with the addition of a 32 kg load[53]. This increase in extension moment indicates a greater reaction from the hip musculature to resist being pushed into flexion. It is unclear to what degree the hip resists this action, as alterations in hip flexion-extension range of motion in response to load are inconsistent. Hip torsional stiffness has not been extensively studied when walking with borne load, however it may help explain how the hip responds to increased load. Increases in hip joint moment during stance phase suggest the possibility of alterations in hip stiffness, however, it is difficult to draw any conclusions from current literature as changes in hip flexion angle are inconsistent.

Similar to the hip, sagittal plane range of motion findings for the knee are not consistent during loaded walking. Increases in knee flexion range of motion were seen as load increased from 8 to 40 kg[45] and from 0 to 18 kg[49], while no change was seen when increasing from 40 to 50 kg[45]. However, while increases in knee flexion angle at initial contact[18,53] and greater peak knee flexion[53] were commonly seen, many of the same studies did not observe an increase in knee flexion range of motion[18,31,52–54]. Birrell et al.[10] did not find any changes in knee flexion range of motion when increasing load from 0 to 8 kg and 0 to 16 kg, however at heavier loads of 24 and 32 kg, knee range of motion was actually found to decrease. Birrell et al.[10] notes that the knee range of motion is dependent on both the knee angle at initial contact and the peak knee flexion throughout stance phase. While some studies above reported increased peak knee flexion, most did not, suggesting that load carriers may adopt a less flexed posture at mid-stance to prevent collapse of the lower limb. Additionally, the discrepancy in knee flexion range of motion may be explained by the difference in weight loads used.

Kinetic changes are also seen in the knee joint during loaded walking. The addition of body borne load has been found to increase internal knee extension moment[18,53,55]. This increase was observed when both 15%[18,55] and 30%[55] of the participant's body weight was added, as well when an absolute load of 32 kg[53] was added. All three studies found that the knee exhibited the greatest increase in moment of the lower limb joints (hip, knee, and ankle) when load was added. Similar to the hip, this increase in joint moment is likely due to the knee preventing collapse of the lower limb when load is added. While the kinematic adaptations made by the knee joint during loaded walking are inconsistent, the increase in joint moment observed is indicative of greater torsional stiffness of the knee. Previous literature has observed such an alteration in knee stiffness due to increased knee joint moment without a corresponding change in flexion angle[31] during loaded walking. While this increase in stiffness may contribute to efficient locomotion and prevent lower limb collapse, the greater moments associated with it may increase MSI risk.

While kinematic adaptations at the hip and knee have been somewhat controversial, the ankle is much more consistent while walking with load. All reviewed literature observed no change in ankle sagittal plane range of motion as load increased during walking[10,18,45,49,53,55]. Similar to some of the findings for the hip and knee, the ankle may maintain its posture as load is added in order to prevent collapse of the lower limb. While the range of motion may remain unchanged, the ankle exhibits an increase in internal plantarflexion moment with load[18,55]. Increased joint moment in combination with no change in range of motion is indicative of increased stiffness at the ankle, however there is dearth of research regarding ankle stiffness during loaded walking. During a single-leg hopping task, the ankle is the primary contributor to overall leg stiffness[56] and may explain a large amount of the increase seen in leg stiffness during loaded walking.

The alteration in lower limb biomechanics observed as load increases does not differ between male and female participants. In studies that analyzed the effects of increasing both relative[38] and absolute[37] loads during walking, males and females exhibited very similar hip, knee, and ankle biomechanics. Both males and females increased PS hip and knee flexion angle as load increased[38]. Furthermore, similar increases were seen in external knee flexion and ankle dorsiflexion moments with the addition of load[37,38]. This research is limited as only sagittal plane biomechanics were reported, however it does indicate at least some level of similarity between males and females with the addition of load during walking.

Running with Body Borne Load

Running with body borne load has not been as heavily researched as walking, yet is equally important, particularly in the military field. Similar to walking, GRFs have been found to increase during running[13,29]. One major difference however, is that the increases in GRF seen in running are less than proportional to the load added. This is thought to be a result of increased stance phase time found to be associated with increased load[13,57]. If stance phase is increased, the total GRF can be applied over a longer period of time, requiring a lower peak. Also associated with increased load carriage was a decrease in stride length[57]. The combination of longer stance time and decrease in stride length is likely adapted to increase gait stability. While increases in GRF loading rate have been observed during loaded walking[17,19], very little research exists regarding how loading rate changes when running with load. Increases in peak vertical GRF seen when running suggest the possibility of a larger loading rate, however this effect may be mitigated by increased stance phase time, allowing more time for the GRF to be applied to the body. Investigating how loading rate is affected by the addition of load when running may be very beneficial to understanding injuries in the military.

One factor that contributes to GRF loading rate is leg stiffness[13,24–26]. Increases in leg stiffness are required to perform many dynamic tasks such as hopping[28] and running[35]. Silder et al.[13] observed increases in leg stiffness when running with loads of 10, 20, and 30% of body weight. These changes were attributed to both an increase in vertical GRF and a decrease in the change in leg length, defined as the distance from the GRF center of pressure and the center of the pelvis[13]. Interestingly, between these two variables, the change in leg length contributed more to stiffness than the increase in GRF. Changes in leg length are determined by alterations in the hip, knee, and ankle joint angles and can be further examined by investigating biomechanical alterations at each joint.

When running with body borne load, load carriers exhibit greater hip flexion angle at initial contact[13,29], however no increases in sagittal plane hip range of motion have been observed. External hip flexion moment increased with the addition of load from 6 to 20 kg[9] and from 0 to 18 kg[29], however a similar increase was not seen between 20 and 40 kg[9]. Similar to walking, the increase in hip flexion angle and moment was likely a result of the additional weight acceptance required by the added load. The increased joint moment is a result of the muscles about the hip responding to the additional weight, however this response appears to be limited at higher weight loads. While hip torsional stiffness has not been directly measured during loaded running, the increased joint moment and corresponding maintenance of range of motion suggests an increase in hip stiffness at lighter weight loads. The increased muscle activity associated with this stiffness may lead to overuse injury without proper training.

Alterations in knee flexion angle have not been consistent across the literature. Silder et al.[13] found increases in knee flexion angle at initial contact with the addition of load. However, one study found an increase in peak knee flexion only when running at 5 m/s[57], while two others found no difference in knee flexion as load increased[9,29]. These differences may be attributed to different running speeds and weight loads used across studies. Knee flexion moment was found to increase with the addition of load from 20 to 40 kg, but not from 6 to 20 kg[9]. This is an interesting finding as the hip exhibited an increase during the addition of the lighter weight load but not the heavier load, while the knee exhibited the opposite. This finding supports the idea that biomechanical adaptations made at lighter weight loads may not be consistent with the adaptations made with heavier loads. It is difficult to determine any alterations in knee joint stiffness from these findings as knee flexion angle was inconsistent across studies.

While ankle joint moment has not been as extensively studied as the hip and knee during loaded running, peak dorsiflexion angle and negative power contribution increase with the addition of load[57]. An increase in negative power contribution indicates a greater involvement in weight acceptance during the first half of stance phase and may be indicative of increased eccentric muscle activity of the ankle extensors. However, these changes suggest that ankle may not contribute greatly to the increase in leg stiffness, as it does during a single leg hopping task[56].

Unlike walking with load, there is little to no research regarding a direct sex comparison between males and females running with load. However, Xu et al.[29] analyzed the effects of running with loads up to 20 kg with only female participants. The female participants exhibited an increase in hip flexion moment with no increase in knee flexion moment or angle with the addition of load[29]. While no direct comparison exists, these are similar to the findings of Brown et al.[9] when load was increased from 6 kg to 20 kg. This indicates that females and males may exhibit similar sagittal plane adaptations when running with lighter loads, however no research exists directly comparing males and females when running with heavier, military relevant body borne loads.

Changing Walking Stride Length with Load

When marching at a fixed-cadence, individuals of different heights must march with the same stride length to maintain the same speed. Because of this, shorter individuals may end up over-striding, and taller individuals may end up under-striding. These stride length alterations may have potential injury implications. Anteroposterior GRF increases as walking stride length increases from short to preferred and from preferred to long when carrying a 20 kg load[32,58]. Interestingly, Gutekunst et al.[58] found increases in vertical GRF during both shorter and longer stride length conditions when compared to preferred stride length, while Seay et al.[32] found no significant differences. This may be due to a difference in sample sizes as Gutekunst et al.[58] only had 5 subjects in their study. Loading rate was not analyzed in either study. However, shorter stance phase times must be associated with faster cadences in order to maintain the same velocity. This concept, coupled with the lack of change in vertical GRF suggests the possibility of an increased GRF loading rate when a shorter stride length is adopted. Further research is required to investigate this possibility.

In addition to changes in GRF values, increasing from a short to long stride length increases range of motion for the hip, knee, and ankle[32]. This is likely the result of a more flexed hip and extended knee upon initial contact required in order to reach the required long stride length. Hip joint moment was unaffected by changes in stride length[32,59]. However, increased internal knee flexion moment was seen for the long stride length compared to the preferred and short stride length[32,59]. Additionally, internal ankle plantarflexion moment increased during the preferred stride length compared to the short stride length, and with the long stride length compared to the preferred and short stride lengths[32,59]. While some of the range of motion adaptations are likely just a requirement of increasing stride length, the changes in joint moments are likely used to attenuate the increased anteroposterior GRFs. It is unclear how these alterations affect the torsional stiffness at each joint, thus further analysis is required.

Changing Unloaded Running Stride Length

While there is little to no research regarding stride length alterations when running with load, the popularity of competitive and recreational running has led to research on unloaded running stride length changes. Thompson et al.[33] found that, when increasing preferred stride length by 5% and 10%, anteroposterior and vertical GRFs increased as well as knee and ankle sagittal plane moments. Additionally, reducing stride length by 10% but not 5% had the opposite effect, leading to decreased vertical GRF. This suggests that during unloaded running, using shorter strides may be beneficial for reducing injury. Edwards et al.[60] support this finding in a study where joint contact forces were used to determine injury risk when stride length is reduced. This study applied the joint contact forces calculated to a probabilistic stress fracture model and found that reduced stride length reduced stress fracture risk by approximately 3% to 6%. Additionally, Edwards et al.[60] concluded that the magnitude of impact forces are a larger contributor to stress fracture risk than the increased number of loading cycles required when running with a shorter stride length. Heiderscheit et al.[34] also found that using shorter strides when running may help reduce injury risk. In a study where 45 recreational runners where recruited, it was found that shorter strides led to less mechanical energy absorbed by the knee and hip, whereas longer strides led to an increase in energy absorption at all lower limb joints[34]. These studies seem to suggest that during unloaded running, using a shorter than preferred stride length may help reduce injury risk. It should be noted that Edwards et al.[60] did not test the probability of stress fractures when running with a longer than preferred stride length. Therefore, running with a longer stride length may not necessarily increase injury risk. Additionally, more research must be completed to determine the optimal stride length for injury prevention.

Increases in GRF are oftentimes associated with increases in leg stiffness. However, when altering running stride length, leg stiffness has actually been found to increase as stride length is reduced from long to preferred and from preferred to short[35,61]. This seems counter-intuitive as this same change in stride length is associated with decreased GRF. Therefore, this finding must be associated with a significant decrease in lower limb joint flexion, and thus, leg length[62]. The addition of load during running has been associated with just the opposite, an increase in leg stiffness[13], leading to an intriguing question of how increasing stride length while running with load will affect lower limb stiffness. While individual joint stiffness was not quantified in this literature, increases in both knee and ankle external flexion and dorsiflexion moments, respectively, were found as stride length was adjusted from preferred to +10% of preferred. This adaptation, coupled with the increase in knee moment seen with the addition of load, may be related to the large number of injuries experienced at the knee joint by military personnel.

Summary

In recent years soldiers have been required to carry historically large loads. The rise in load carried is coupled with a drastic increase in military injuries in recent years. Research has indicated that large body borne load can alter lower limb biomechanics during both walking and running with load. Furthermore, altered stride lengths associated with fixed-cadence marching and running have been shown to alter lower limb biomechanics and may pose an additional injury risk when carrying large body borne loads. Females may be especially at risk as they experience injuries at twice the rate as males in the military and, as of 2013, can serve in more demanding infantry positions. However, there is little to no research available regarding stride length alterations when running with load, nor does research exist directly comparing males and females during this task. This study will research how the leg stiffness and hip, knee, and ankle stiffness exhibited by both males and females are affected by changing stride length when running with body borne loads of 20 kg, 25 kg, 30 kg, and 35 kg.

CHAPTER THREE: MANUSCRIPT

Introduction

Musculoskeletal injuries are an increasing problem among military personnel. The incidence of MSIs for military personnel has increased seven-fold in the last 25 years[1], with female personnel exhibiting injury rates up to twice as high as their male counterparts[3]. Most training-related MSIs occur in the lower limb, with approximately 25% of these injuries occurring at the knee joint[5,7,8]. These injuries commonly occur during the repetitive load-bearing activities required during military training[5]. During training, military personnel are routinely required to run with body borne load. These body borne loads typically consist of personal fighting and protective equipment that routinely weigh between 20 and 40 kg[9] during training activities[63]. These loads are problematic because they reportedly alter lower limb biomechanical patterns[10–13], which are thought to increase risk of MSI[14–16], particularly at the knee.

During locomotion, the lower limb biomechanical adaptions that occur from the addition of body borne load may increase the risk of MSI. Body borne load reportedly increases peak vGRF during walking[11,17–19] and running[9,13], increasing the strain on the lower limb musculoskeletal system[19–21] and risk of MSI. In order to attenuate the increased GRF and prevent collapse of the lower limb, load carriers reportedly increase leg stiffness[13,22]. Silder et al.[13] reported increases in leg stiffness due to both an increase in peak vGRF as well a decrease in change in leg length when running with body borne loads of 20 kg or less[13]. However, it is unknown if load carriers exhibit a similar increase in leg stiffness with running with heavier body borne loads commonly worn during military activities (between 20 kg and 35 kg). While adequate leg stiffness may be necessary to successfully perform many dynamic tasks[23] such as running, large magnitudes of stiffness increase the transmission of the vGRF to the lower limb[13,24–26] and further strain the musculoskeletal system[19,27].

Increases in leg stiffness during locomotor activities with body borne load may be attributed to greater torsional stiffness of the lower limb (hip, knee, and ankle) joints. Torsional joint stiffness is described as the change in angle that occurs when a given moment is applied[28] and may provide insight into the risk of MSI at each joint. When running with military relevant loads, participants exhibit a significant increase in peak hip and knee flexion moments, but not a similar increase in hip and knee flexion posture[9,29], potentially increasing joint stiffness. In fact, a significant increase in knee joint stiffness is reported when walking with body borne load[31]. But it is unknown if similar increases in joint stiffness are exhibited when running with body borne load.

During military training, personnel are often required to walk or run at a fixed-cadence. Running or walking with a fixed cadence requires all personnel to use the same stride length, potentially leading to the high incidence of MSI[32], by causing shorter individuals to over-stride and taller individuals to under-stride. Participants are purported to increase peak hip and knee flexion joint angles and knee flexion moment when using longer strides (i.e., over-striding) to walk with a 20 kg load[32]. Conversely, participants exhibited a significant decrease in braking GRF and peak ankle joint angle and moment when walking with shorter strides (i.e., under-striding)[32]. Considering participants increase leg stiffness to prevent collapse of the limb and maintain locomotion when running with body borne load, they may not be able to modulate lower limb stiffness when changing stride length to run with load as they do without load[35]. Yet, the effect of changing stride length on lower limb biomechanics, and subsequent risk of MSI, when running with body borne load is largely unknown.

As of 2013, females could once again serve in direct combat positions in the military[36]. When serving in these infantry positions, military personnel can be required to run with very large loads which may contribute to the high rate of injury seen by females in the military[3]. To date, sex comparisons during loaded locomotion have been limited to analyzing sagittal plane biomechanics while walking with load, where no differences between males and females have been observed[37,38]. When running without load, differences in the frontal and transverse planes, but not the sagittal plane, have been observed between males and females[39,40], yet little to no research exists directly comparing sex differences when running with military relevant loads. Considering differences in strength between males and females have been reported[41,42], it is unclear if males and females adopt similar biomechanical patterns with the addition of load or changes in stride length.

The purpose of this study was to determine the changes in leg, and hip, knee, and ankle stiffness when using different stride lengths to run with body borne loads commonly worn during military training, and compare whether the changes in stiffness differ between male and female participants. It was hypothesized that leg, and hip, knee and ankle stiffness would significantly increase with each incremental addition of body borne load and stride length; but, significant differences in leg and joint stiffness would not be evident between male and female participants.

Methods

<u>Participants</u>

Twenty-seven participants (Table 3.1) volunteered for this study. To be included, participants were required to be healthy, physically active and self-report the ability to carry 75 pounds. Participants were excluded from the study if they had: (1) a history of previous back or lower extremity surgery; (2) pain in the back or lower extremity prior to testing; (3) a recent back or lower extremity injury (previous six months); (4) and/or a neurological disorder. Prior to testing, research approval was obtained from the local IRB and each participant provided written consent.

Table 3.1 Mean (SD) Subject Demographics by Sex

	N	Age (yrs)	Height (m)	Weight (kg)
Male	17	20.7 (1.9)	1.8 (0.1)	80.8 (9.6)
Female	10	22.0 (2.9)	1.7 (0.1)	66.6 (8.2)

<u>Biomechanical Testing</u>

Each participant completed one orientation and four test sessions. During each test session, participants donned a different body borne load configuration (20, 25, 30, or 35 kg). For each load configuration, participants were fitted with a spandex top and shorts, and carried military equipment that included: a mock weapon (M16), standard issue military helmet (ACH), and an adjustable weighted vest (Box, WeightVest.com, Inc., Rexburg, ID, USA). The military equipment weighed approximately 6.17 kg (Figure 3.1). To apply the additional load required for each condition, the weight of the vest was systematically adjusted to the necessary weight. Total weight ± 2% of the target load was accepted. Each test session was separated by at least 24 hours to limit the effects of fatigue. Prior to testing, the order in which the load configurations were tested was randomized and counter-balanced using a 4 x 4 Latin Square design.

Figure 3.1 Depiction of the military equipment (helmet, mock weapon, and adjustable weighted vest) that compose each load condition (20, 25, 30, 35 kg)

Orientation Session

Each participant performed one orientation session prior to testing. The orientation session was used to familiarize the participant with each body borne load configuration, determine their preferred stride length, and quantify hip and knee strength. During orientation, participants ran at 4 m/s (± 5%) through the motion capture area three times with each load configuration (20, 25, 30 and 35 kg). During each trial, stride length was calculated as the linear distance between two consecutive heel strikes of the dominant limb[33]. The PSL was calculated as average stride length exhibited with each load configuration, and then SSL and LSL were calculated as 85% and 115% of the PSL, respectively. Each participant had hip and knee strength quantified on an isokinetic dynamometer (System 2, Biodex Medical Systems, Inc., Shirley, NY). To quantify strength, participants performed three maximum isometric hip and knee flexion and extension contractions for three seconds. For hip flexion and extension, participants were tested while standing with their hip in a neutral position (0°)[64]. For knee flexion and extension, participants were seated with their knee at 60° for extension and 45° for flexion, respectively[65].

Biomechanical Testing Session

During each test session, participants had synchronous three-dimensional (3D) lower limb joint (hip, knee, and ankle) biomechanics data recorded during an over-ground running task. During the run task, a single force platform (AMTI OR6 Series, Advanced Mechanical Technology Inc., Watertown, MA) captured GRF data at 2400 Hz, while eight high-speed optical cameras (MXF20, Vicon Motion Systems LTD, Oxford, UK) recorded lower limb motion data at 240 Hz.

For the run task, participants ran approximately 10 m at 4 m/s (± 5 %) through the motion capture volume using either their PSL, SSL or LSL. During each run, two sets of infrared timing gates (TracTronix TF100, TracTronix Wireless Timing Systems, Lenexa, KS), placed 4 m apart in the motion capture volume, quantified running speed. Participants performed three successful trials with each stride length (PSL, SSL, and

LSL). To minimize the number of incorrect trials, each participant's required stride length was marked on the floor with tape according to Allet et al.[59]. The order each participant performed the stride lengths was randomized using a 3 x 3 Latin Square design and assigned prior to testing. A trial was considered successful if the participant ran at the correct speed, used the correct stride length, and only contacted the force platform with their dominant limb. During testing, participants were given water and provided adequate rest between each trial to minimize the effects of fatigue.

Biomechanical Analysis

During each trial, lower limb joint rotations were quantified using the 3D coordinates of thirty-four retro-reflective (15mm diameter) and four virtual (digitized) markers (Table 3.2). Reflective markers were precisely attached using double sided tape and secured using elastic tape (Cover-Roll Stretch, BSN medical GmbH, Hamburg, Germany) over specific landmarks. The virtual markers were created by digitizing their location in the global coordinate system using a Davis Digitizing Pointer (C-Motion Inc., Rockville, MD). After the markers were placed, a high-speed video recording of the participants standing still in anatomic (static) position was captured. The static recording was used to construct a kinematic model with eight segments (trunk, pelvis, and bilateral thigh, shank, and foot) and 27 degrees-of-freedom using Visual 3D v6.00 (C-Motion Inc., Rockville, MD). Each segment of the kinematic model was assigned a local coordinate system with an origin located at a virtual joint center and three orthogonal axes (x, y and z). For the orthogonal axes, the x-axis was laterally directed with positive values directed to the right, the y-axis was anteriorly directed with positive values directed forward, and z-axis was axially directed with positive values directed proximally. The pelvis was defined in relation to the global coordinate system with its origin calculated at the midpoint between the right and left iliac crests and was assigned six degrees of freedom (three rotational and three translational)[66]. The hip joint center was determined during a functional range-of-motion trial according to Schwartz and Rozumalski[67], and the local coordinate system assigned three degrees of freedom. In accordance with previous literature[68,69], knee and ankle joint centers were calculated as the midpoint between the medial and lateral femoral epicondyles and the medial and lateral malleoli, respectively, and each local coordinate system assigned three degrees of freedom. The origin of the trunk segment was calculated as the point of intersection of the line connecting the left and right acromion processes and the line connecting the C7 vertebra and the sternum jugular notch. Its local coordinate system was also assigned three degrees of freedom.

Table 3.2 Marker Placement for the Kinematic Model

	Markers
Trunk	Acromion processes, *Sternum jugular notch*, *C7 vertebrae*, *Xiphoid process*, and *Midpoint between the inferior angles of the scapulae*
Pelvis	**Anterior-superior iliac spines**, **Posterior-superior iliac spines**, and Iliac crests
Thigh	Greater trochanter, Distal thigh, **Medial** and **Lateral femoral epicondyles**
Shank	Tibial tuberosity, Lateral fibula, Distal tibia, **Medial** and **Lateral malleoli**
Foot	Posterior heel, Midpoint between first and fifth metatarsal heads, **First metatarsal head**, and **Fifth metatarsal head**

Note: Bold indicates calibration markers, *Italic* indicates virtual markers

For each trial, the synchronous 3D GRF and marker trajectory data were low pass filtered using a fourth-order Butterworth filter with a cutoff of 12 Hz[70]. The filtered marker trajectories were processed by the Visual 3D software to solve for 3D hip, knee, and ankle joint rotations. Joint rotations were expressed relative to each participant's static position using a joint coordinate system approach[68,69]. Then, the filtered kinematic and GRF data were submitted to standard inverse dynamics analyses to obtain 3D moments at each lower limb joint[71]. The inertial properties for each lower limb segment were established according to Dempster et al.[72]. The hip, knee and ankle joint moments were characterized with respect to the cardanic axes of their respective joint coordinate systems[71]. Hip and knee joint moments are expressed as flexion-extension, while the ankle joint moments are expressed as dorsiflexion-plantarflexion. Joint moments were normalized to subject mass (kg) and height (m), while GRF was normalized to subject body weight (N).

Leg and joint stiffness were calculated with custom written MATLAB (Mathworks, Natick, MA) code. Leg stiffness (K_l) was calculated at the frame where leg length reached its minimum, and defined as the component of GRF directed from the center of pressure (CoP) to the hip joint center (F_e) divided by the change in leg length (L_e)[73]. Leg length was defined as the linear distance between the CoP and the hip joint center[74]. Joint stiffness (K_{tors}) at the hip, knee, and ankle was calculated and defined as the difference between the maximum and minimum joint moment divided by the corresponding change in joint angle[25]. All biomechanical data was normalized from 0 % - 100 % of stance phase, and resampled at 1 % increments (N = 101). Stance phase was defined as heel strike to toe off, which were respectively defined as the moment GRF first exceeded and fell below 10 N.

$$\text{Leg Stiffness: } K_l = \frac{F_e}{\max(\Delta L_{e})}$$

$$\text{Joint Stiffness: } K_{tors} = \frac{\Delta Moment}{\Delta Angle}$$

Statistical Analysis

Predefined biomechanical variables related to leg and joint stiffness were submitted to statistical analysis. These dependent variables consisted of leg, and hip, knee and ankle joint stiffness, peak of stance (PS, 0% - 100%) hip and knee flexion, and ankle dorsiflexion joint angle and moment, peak vertical GRF, and change in leg length. Each dependent variable was averaged across three successful trials to create a participant-based mean. Then, the participant-based means were submitted to a repeated measures ANOVA to test the main effects of and interaction between load (*20 kg, 25 kg, 30 kg, and 35 kg*), stride length (*PSL, SSL, and LSL*), and sex (*male vs female*). Significant interactions were submitted to a simple effects analysis, and a Bonferroni correction was used for multiple comparisons[75]. Independent *t-tests* were used to compare hip and knee flexion and extension strength between sexes. For significant pairwise comparisons, effect size was calculated using Cohen's *d*[76]. All statistical analysis was performed using SPSS v23 software (IBM, Armonk, NY) with alpha level 0.05.

Results

Leg Stiffness

The ANOVA revealed a significant stride length versus sex interaction for leg stiffness (P=0.011) (Figure 3.2). Males decreased leg stiffness using LSL compared to PSL (P<0.001, d=0.89) and SSL (P<0.001, d=1.23), and PSL compared to SSL (P=0.026, d=0.39), while females exhibited no significant differences in leg stiffness between strides. Further, males exhibited greater leg stiffness compared to females with SSL (P=0.047, d=0.87), but not with PSL (P=0.299, d=0.44) or LSL (P=0.867, d=0.07) strides.

Load (P=0.006) and stride length (P<0.001), but not sex (P=0.218, d=0.52) had a significant effect on leg stiffness (Table C.1). Leg stiffness increased with the 35 kg load compared to the 20 kg load (P=0.002, d=0.63), but no significant differences were evident between any other loads. Further, participants decreased leg stiffness with LSL compared to PSL (P<0.001, d=0.75) and SSL (P<0.001, d=0.91), but no significant difference was observed between the PSL and SSL (P=0.520, d=0.22) strides.

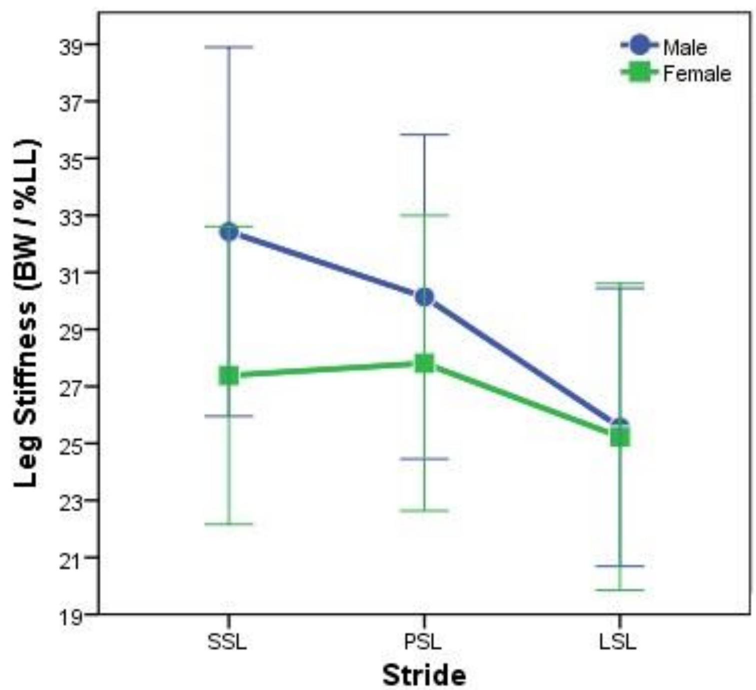

Figure 3.2 Peak leg stiffness exhibited by male and female participants running with SSL, PSL, and LSL.

Peak Vertical GRF

There was a significant stride length versus sex interaction effect for peak vGRF (P=0.039) (Figure 3.3). Males exhibited a significantly larger peak vGRF compared to females with SSL (P=0.043, d=0.88) and PSL (P=0.004, d=1.31), but not with LSL (P=0.086, d=0.72). Both males and females exhibited significantly greater peak vGRF with LSL compared to PSL (P=0.011, d=0.25; P<0.001, d=1.26) and SSL (P<0.001, *d*=0.69; *P*=0.004, *d*=1.25), but only males exhibited greater vGRF with PSL compared to SSL (*P*=0.010, *d*=0.46).

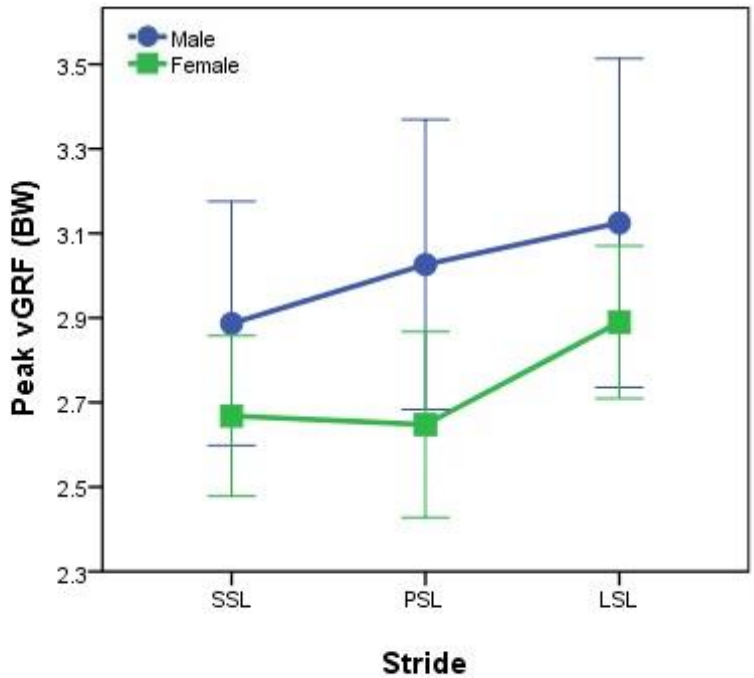

Figure 3.3 Peak vGRF exhibited by male and female participants running with SSL, PSL, and LSL.

Load (*P*<0.001) (Figure 3.4), stride length (*P*<0.001), and sex (*P*=0.021, *d*=1.02) had a significant effect on peak vGRF (Table C.1), with males exhibiting greater vGRF compared to females. Specifically, peak vGRF was larger with 35 kg compared to 30 kg (*P*=0.003, *d*=0.35), 25 kg (*P*<0.001, *d*=0.48), and 20 kg (*P*<0.001, *d*=0.84) loads, and with the 30 kg (*P*<0.001, *d*=0.49) and 25 kg (*P*=0.004, *d*=0.32) compared to the 20 kg load. Peak vGRF was also greater with LSL compared to PSL (*P*<0.001, *d*=0.44) and SSL (*P*<0.001, *d*=0.76), but differences were not observed between PSL and SSL (*P*=0.311, *d*=0.26).

<u>Leg Length</u>

Stride length (*P*<0.001), but not sex (*P*=0.490, *d*=0.31) or load (*P*=0.725) had a significant effect on the change in leg length (Table C.1). The change in leg length was larger for LSL compared to both PSL (*P*<0.001, *d*=0.86) and SSL (*P*<0.001, *d*=0.98), but there was no significant difference between PSL and SSL (*P*=1.000, *d*=0.14).

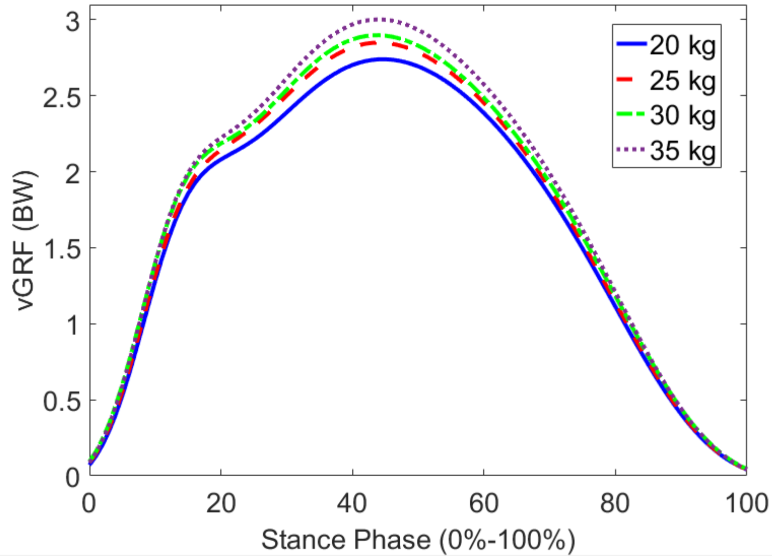

Figure 3.4 Stance phase (0% - 100%) vGRF exhibited when running with each body borne load (20 kg, 25 kg, 30 kg, and 35 kg).

Joint Stiffness

There was a significant effect of body borne load on knee (*P*<0.001) (Table C.3) and ankle (*P*=0.013) (Table C.4), but not hip stiffness (*P*=0.723) (Table C.2). Participants exhibited greater knee stiffness with the 35 kg compared to 25 kg (*P*=0.001, *d*=0.67), and with 35 kg (*P*<0.001, *d*=1.11), 30 kg (*P*=0.002, *d*=0.76) and 25 kg (*P*=0.006, *d*=0.48) compared to 20 kg. But, no significant difference was observed between 30 kg and 25 kg (*P*=0.189, *d*=0.38), or between 35 kg and 30 kg (*P*=1.000, *d*=0.19). After correcting for Type I error, the post-hoc analysis revealed no significant difference in ankle stiffness between body borne loads.

Stride length had a significant effect on ankle (*P*=0.003) (Table C.4), but not hip (*P*=0.661) (Table C.2) or knee stiffness (*P*=0.170) (Table C.3). Specifically, a decrease in ankle stiffness was observed with LSL compared to PSL (*P*=0.024, *d*=0.44) and SSL

(*P*=0.030, *d*=0.50), but no difference was observed between PSL and SSL (*P*=1.000, *d*=0.09).

Sex had no significant effect on hip (*P*=0.880), knee (*P*=0.234) or ankle stiffness (*P*=0.081).

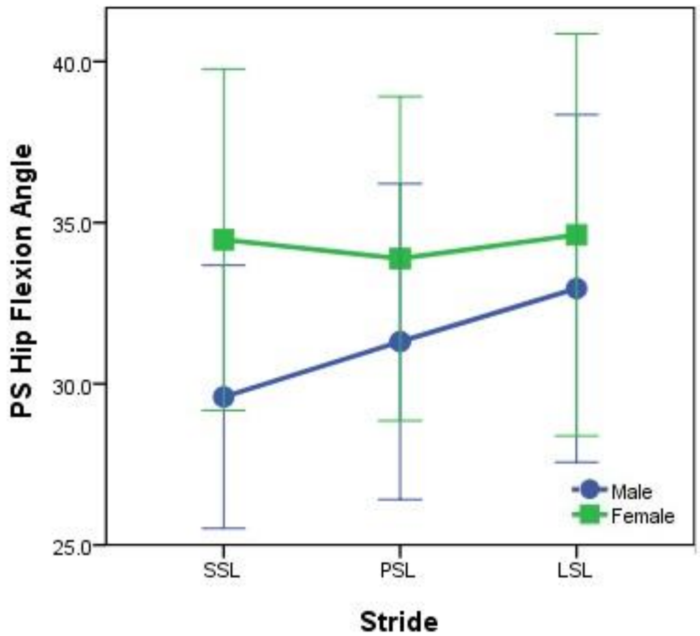

Figure 3.5 Peak stance (0% - 100%) hip flexion angle exhibited by male and female participants running with SSL, PSL, and LSL.

Joint Flexion Angles

The ANOVA revealed a significant stride length versus sex interaction for PS hip flexion angle (*P*=0.010) (Figure 3.5). Females exhibited significantly greater PS hip flexion compared to males with SSL (*P*=0.013, *d*=1.11), but not with PSL (*P*=0.203) or LSL (*P*=0.472). Furthermore, males increased PS hip flexion angle with LSL (*P*<0.001, *d*=0.73) and PSL (*P*=0.008, *d*=0.39) compared to SSL, and with LSL compared to PSL (*P*=0.041, *d*=0.33), while female participants exhibited no significant change in PS hip flexion angle between strides.

There was a significant load versus stride length (P=0.028) interaction for PS knee flexion angle (Figure 3.7). Participants exhibited greater PS knee flexion angle with LSL compared to both PSL and SSL when running with the 20 kg (P=0.001, d=0.50; P<0.001, d=0.76), 25 kg (P<0.001, d=0.54; P<0.001, d=0.68), and 30 kg (P=0.004, d=0.46; P<0.001, d=0.85), but not the 35 kg load (P=0.760, d=0.27; P=0.614, d=0.39). Furthermore, participants exhibited a significant reduction in PS knee flexion angle with 35 kg compared to 20 kg when using the LSL (P=0.021, d=0.40), but no differences were evident between loads for PSL (P=0.063, d=0.22) or SSL (P=1.000, d=0.05).

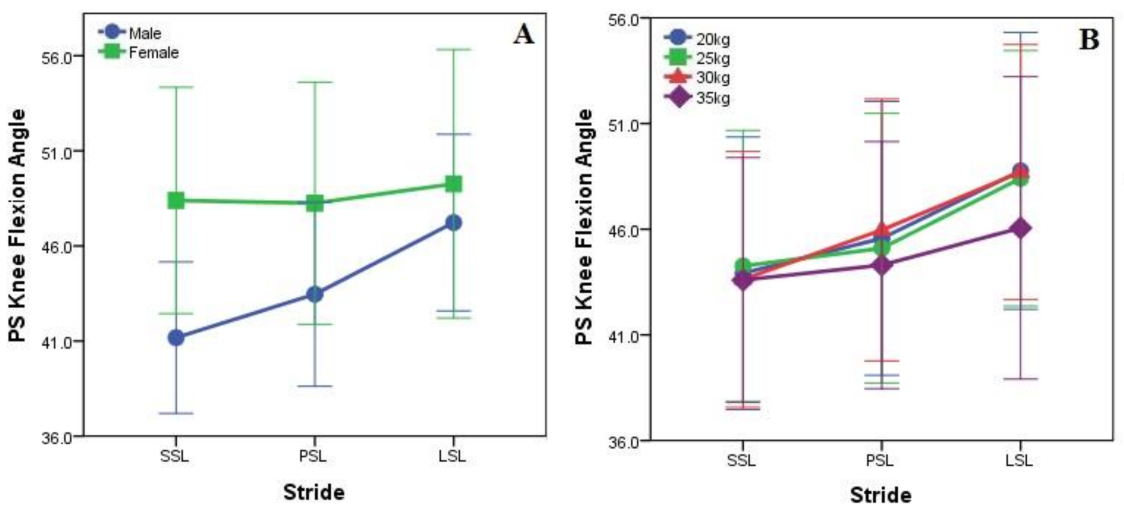

Figure 3.6 Peak stance (0% - 100%) knee flexion angle exhibited when running with SSL, PSL, and LSL for both sexes (A) and with each body borne load (B).

A significant stride length versus sex (P<0.001) interaction was observed for PS knee flexion angle (Figure 3.7). Post-hoc analysis revealed significantly greater PS knee flexion angle for females compared to males with SSL (P=0.001, d=1.57) and PSL (P=0.037, d=0.92), but not LSL (P=0.373). Males, however, exhibited a significant increase in PS knee flexion angle with LSL (P<0.001, d=1.44) and PSL (P=0.001, d=0.53) compared to SSL, and with LSL compared to PSL (P<0.001, d=0.82), while females did not change knee flexion angle between stride lengths.

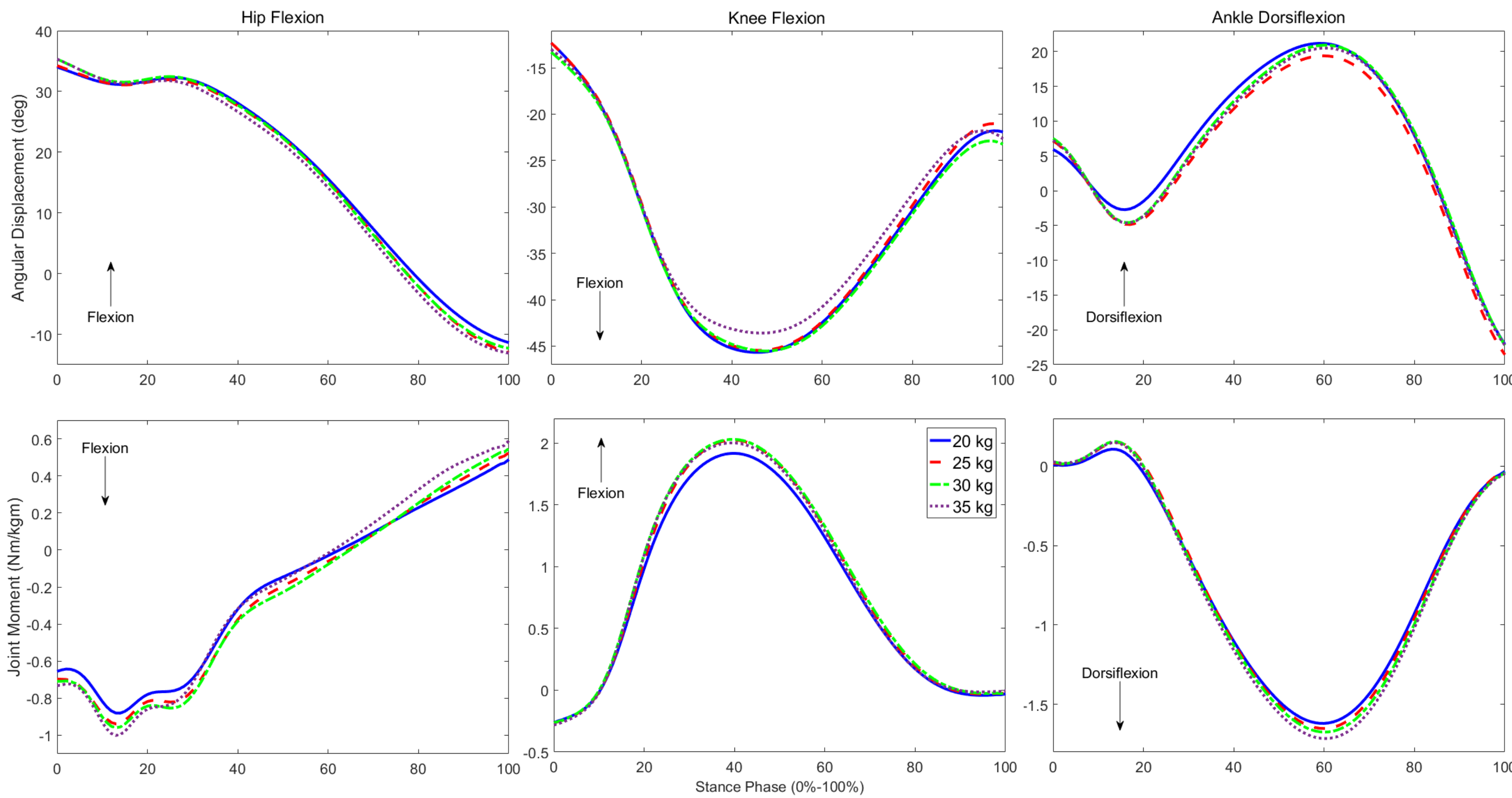

Figure 3.7 Stance phase (0% - 100%) hip and knee flexion, and ankle dorsiflexion joint angles and moments exhibited when running with each body borne load (20 kg, 25 kg, 30 kg, and 35 kg).

Body borne load had a significant effect on PS knee flexion (*P*=0.014), but not PS hip (*P*=0.704) and ankle flexion (*P*=0.130) angles (Figure 3.6). Specifically, participants decreased PS knee flexion angle with 35 kg compared to 20 kg (*P*=0.002, *d*=0.25), but significant changes were not observed between any other loads.

Stride length had a significant effect on PS hip (*P*=0.005) and knee (*P*<0.001) flexion, and ankle dorsiflexion (*P*<0.001) angles (Figure 3.9). Participants increased PS hip (*P*=0.023, *d*=0.41) and knee (*P*<0.001, *d*=0.73) flexion, and ankle (*P*<0.001, *d*=0.46) dorsiflexion angle with LSL compared to SSL, and knee flexion (*P*<0.001, *d*=0.49) and ankle dorsiflexion (*P*=0.005, *d*=0.29) with LSL compared to PSL. But, no significant differences in hip flexion were seen between LSL and PSL (*P*=0.084, *d*=0.25), or for hip (*P*=0.587, *d*=0.18) and knee (*P*=0.076, *d*=0.19) flexion, and ankle (*P*=0.266, *d*=0.17) dorsiflexion between PSL and SSL strides.

Females exhibited significantly greater PS knee flexion angles compared to males (*P*=0.030, *d*=0.95) (Table C.3), but there were no sex differences for PS hip flexion (*P*=0.130) or ankle dorsiflexion (*P*=0.219).

<u>Joint Flexion Moments</u>

The ANOVA revealed a significant load versus sex (*P*=0.016) interaction for PS knee flexion moment (Figure 3.8). Males exhibited greater PS knee flexion moment compared to females with the 30 kg (*P*=0.006, *d*=1.23), but not 20 kg (*P*=0.574, *d*=0.23), 25 kg (*P*=0.066, *d*=0.78), or 35 kg (*P*=0.173, *d*=0.56) loads. Males also increased PS knee flexion moment with 25 kg (*P*=0.009, *d*=0.86), 30 kg (*P*=0.004, *d*=1.06), and 35 kg (*P*=0.014, *d*=0.79) loads compared to the 20 kg, while females exhibited no significant difference between any loads.

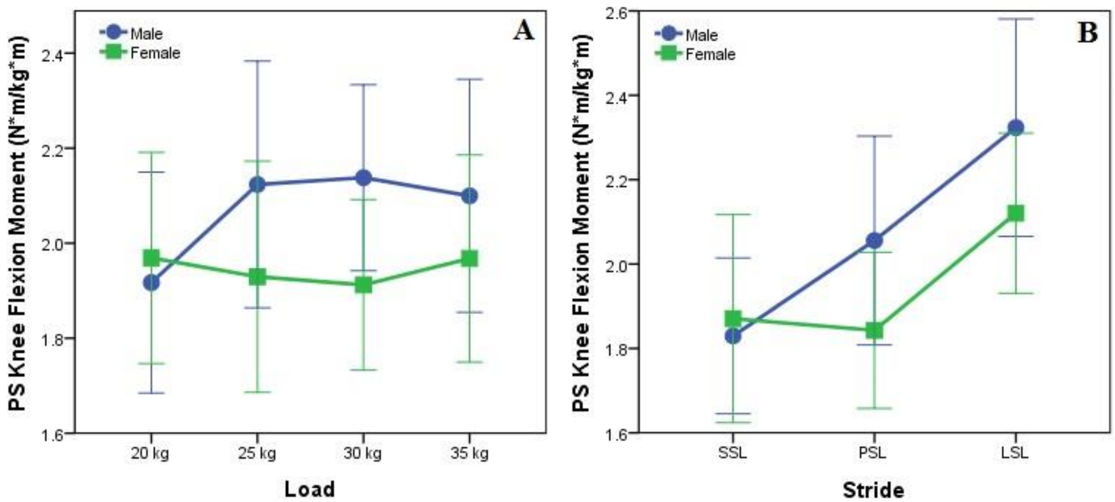

Figure 3.8 Peak stance (0% - 100%) knee flexion moment exhibited by male and female participants when running with each body borne load (20 kg, 25 kg, 30 kg) (A) and stride length (SSL, PSL, and LSL) (B).

There was also a significant stride length versus sex (*P*=0.012) interaction for PS knee flexion moment (Figure 3.8). Males exhibited greater PS knee flexion moment compared to females with PSL (*P*=0.027, *d*=0.98) and LSL (*P*=0.041, *d*=0.89), but not SSL (*P*=0.629). Furthermore, both males and females significantly increased PS knee flexion moment with LSL compared to PSL (*P*<0.001, *d*=1.09; *P*<0.001, *d*=1.56) and SSL (*P*<0.001, *d*=2.27; *P*=0.005, *d*=1.2), but only males exhibited greater knee flexion moment with PSL compared to SSL (*P*=0.009, *d*=1.07).

Body borne load had a significant effect on PS hip flexion (*P*=0.001), but not knee flexion (*P*=0.165) or ankle dorsiflexion (*P*=0.078) moments (Figure 3.6). Specifically, PS hip flexion moment increased with 35 kg compared to 20 kg (*P*=0.007, *d*=0.76), but not between any other loads.

Stride length had a significant effect on PS hip (*P*<0.001) and knee (*P*<0.001) flexion, and ankle (*P*<0.001) dorsiflexion moments (Figure 3.9). Post-hoc analysis revealed both PS knee flexion and ankle dorsiflexion moments were larger with LSL

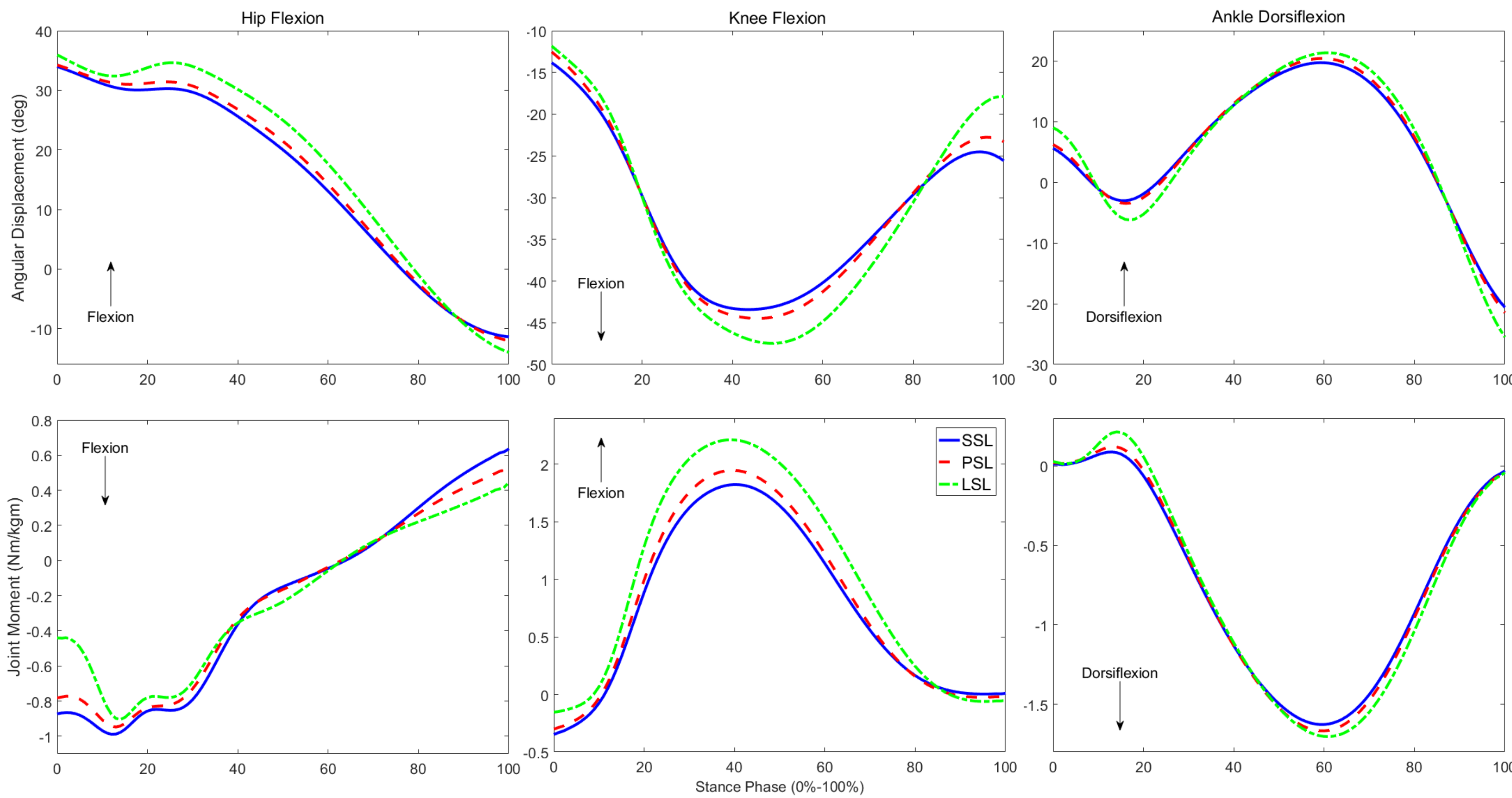

Figure 3.9 **Stance phase (0% - 100%) hip and knee flexion, and ankle dorsiflexion joint angles and moments exhibited when running with each stride length (SSL, PSL, and LSL).**

compared to both PSL (P<0.001, d=1.11; P<0.001, d=0.44) and SSL (P<0.001, d=1.79; P<0.001, d=0.75), but not between the PSL and SSL (P=0.277, d=0.62; P=0.141, d=0.27). After correcting Type I error, no significant difference in PS hip flexion moment was evident between strides.

Males exhibited greater PS ankle dorsiflexion moment compared to females (P=0.017, d=1.06) (Table C.4), but no sex differences were observed for PS hip (P=0.172) or knee flexion (P=0.093) moments.

Joint Strength

Males exhibited greater hip and knee flexion (P=0.028, d=0.97; P=0.016, d=1.07), and knee extension (P=0.018, d=1.07) strength compared to females (Table 3.3), but no sex difference was observed for hip extension strength (P=0.051, d=0.85).

Table 3.3 Mean (SD) Isometric Strength Data by Sex

	Knee Strength (%BW)		**Hip Strength (%BW)**	
	Flexion*	**Extension***	**Flexion***	**Extension**
Male	48.8 (10.1)	71.3 (13.5)	64.1 (24.2)	64.6 (27.3)
Female	38.8 (8.9)	58.0 (12.0)	45.0 (11.4)	46.5 (6.1)

*Indicates a significant effect of sex (P<0.05)

Discussion

Large increases in leg stiffness when running with body borne load may contribute to the high incidence of MSI for military personnel. In agreement with previous literature[13], current participants exhibited a significant increase in leg stiffness when running with body borne load. But hypothesis 1.1, that each incremental addition of body borne load (from 20 kg to 35 kg) would produce a significant increase in leg stiffness during running, was only partially supported. Participants increased leg stiffness between 8 % and 12 % with the 5 kg through 15 kg addition of body borne load, but a statistically significant increase in stiffness was only observed between the 20 kg and 35 kg load conditions. The increased leg stiffness and risk of MSI may be attributed to the increase in peak vGRF from 2.75 BW to 3.00 BW between the 20 kg and 35 kg load conditions. During running, larger magnitudes of vGRF increase risk of MSI[77,78] by requiring the lower limb musculature to safely attenuate more force in order to prevent injury of the passive structures of the limb[79]. Subjecting these passive structures to excessive GRFs increases strain placed upon them and can result in injuries such as stress fractures or ruptured ligaments[60].

Running with longer strides did not further increase leg stiffness. Although participants exhibited a significant increase in peak vGRF with longer strides, they may have attenuated the associated injury risk by decreasing leg stiffness. In contradiction to hypothesis 2.1, participants exhibited a significant 15% and 12% decrease in leg stiffness as stride length increased from SSL to PSL, and PSL to LSL, respectively. To run with shorter strides, participants may need a stiffer leg to prevent collapse of the lower limb[23]; conversely, participants may be able to reduce leg stiffness by increasing lower limb flexion and producing a greater change in leg length with longer strides. The fact that current participants increased the change in leg length 22% and 19% with LSL compared to SSL and PSL supports this contention.

The ability to modulate leg stiffness and risk of MSI across stride lengths may differ between sexes. Male participants decreased leg stiffness by 7% and 15% with each incremental increase in stride length, while females exhibited no significant change in leg stiffness with longer strides. This may be attributed to a sex dimorphism in lower limb biomechanics observed across stride lengths. Females adopted a more flexed hip and knee than males with SSL. But, only male participants exhibited a significant increase in

PS hip and knee flexion with PSL and LSL. This sex dimorphism may stem from a difference in lower limb strength. In accordance with previous literature[42], males exhibited greater hip and knee strength than females. Strength differences, particularly that observed in the knee extensors, may allow male participants to attenuate the greater peak vGRF evident with longer strides[25,26,80] by increasing knee flexion. Increasing knee flexion may promote greater energy absorption of the GRF through eccentric contraction of the knee extensors and result in a reduction in the leg stiffness necessary to prevent collapse of the lower limb[34,79]. The fact males exhibited greater PS knee flexion moment compared to females with PSL and LSL supports this contention. Females, however, may not possess the strength to increase the hip and knee flexion necessary to decrease leg stiffness and promote greater absorption of the vGRFs. Female's inability to decrease leg stiffness when running with body borne load may result in higher risk of bony and ligamentous injuries[25,26,80], and stem from weakness of their lower limb musculature. But, considering participants increased knee flexion using LSL compared to PSL with 20 kg, 25 kg, and 30 kg, but not 35 kg, future work is warranted to determine whether lower limb strength, rather than sex, determines whether individuals can decrease leg stiffness, and subsequent risk of MSI.

Participants increased joint stiffness and potential risk of MSI running with body borne load. Specifically, participants exhibited a significant increase in knee and ankle, but not hip stiffness with the addition of load. In agreement with Holt et al.[31], the current participants increased knee joint stiffness between 7% and 17% with the addition of body borne load. During unloaded running, the knee is a major contributor to leg stiffness[81] and primarily responsible for energy absorption during weight acceptance[79]. With the addition of load, the current participants may have exhibited a significant reduction in PS knee flexion angle to increase stiffness and prevent collapse of the lower limb. However, the stiffer knee may increase risk of MSI by transmitting larger GRFs to the passive structures of the lower limb[24,82]. During running, the ankle also contributes to energy absorption during weight acceptance[34,79], potentially putting its passive structures at risk for injury with an increase in stiffness. However, after correcting for Type I error, precise differences in ankle stiffness between load conditions were not currently observed. Considering previous research[13] reported a significant effect of body borne load on PS ankle joint angles and moments, further study is warranted to determine the specific adaptations of ankle biomechanics necessary to maintain lower limb stability and forward progression when running with load.

Stride had an effect on ankle, but not hip or knee stiffness. Specifically, a significant decrease in ankle stiffness was seen with LSL compared to both PSL and SSL. Similar to walking with load[32], participants may have increased energy absorption at the ankle by increasing PS ankle dorsiflexion moment with longer strides. Decreased ankle stiffness may help prevent bony and ligamentous injuries from occurring at the joint, but increase the risk of injury for the associated musculature[32].

The current hip stiffness calculation may be a limitation. This stiffness calculation assumes the joint behaves like a torsional spring – where the moment compresses the joint. During unloaded running, the hip primarily functions as an energy producer[83], but reportedly increases energy absorption with the addition of load[84]. Considering participants exhibited greater PS hip flexion moment, but not a significant increase in PS hip flexion angle with the addition of load, future work is warranted to determine how the hip acts to maintain stability of the lower limb and provide forward propulsion of the center of mass during running with load. Another limitation may be the leg stiffness calculation. While this is a common method for calculating leg stiffness[73], the equation does not account for frontal plane GRF or leg length changes. These frontal plane measures may be of particular importance when comparing males and females, as a sex dimorphism in frontal plane biomechanics has been observed during locomotion[39,40] and warrants further study.

Conclusion

In conclusion, adding body borne load increased leg stiffness and potential risk of MSI during over-ground running. The increased leg stiffness may help attenuate the greater peak vGRFs and prevent collapse of the limb when running with body borne load. Running with longer strides did not further increase leg stiffness. Participants, in fact, decreased leg stiffness and potentially risk of MSI when running with longer strides. But, the ability to modulate leg stiffness across changes in stride length may differ between sexes. Males increased hip and knee flexion and reduced leg stiffness with longer strides; whereas, females did not increase lower limb flexion or decrease leg stiffness with longer strides. The addition of body borne load may further strain the passive structures of the knee and ankle during over-ground running. Both the knee and ankle exhibited a significant increase in joint stiffness with the addition of body borne load. The increase joint stiffness may be necessary to provide lower limb stability during weight acceptance, but may increase subsequent risk of MSI.

CHAPTER FOUR: CONCLUSION

This study sought to: (1) determine how altering stride length while running with military relevant body borne loads affects leg and joint (hip, knee, and ankle) stiffness of the lower limb; and (2) determine whether changes in leg and joint stiffness as a result of body borne load and stride length differ between males and females. The current outcomes support the hypothesis that larger body borne loads will lead to greater leg and joint stiffness, but similar increases in leg and joint stiffness were not observed for both sexes or as stride length increased.

Key Findings

The addition of body borne load increased leg stiffness during over-ground running. The increased leg stiffness may be due to the larger peak vGRFs evident with the addition of body borne load. But with longer strides, participants decreased leg stiffness, despite concurrent increases in peak vGRF. Participants may have decreased leg stiffness with the longer strides by using greater lower limb flexion. However, the ability to increase PS knee flexion angle with the longer strides may be dependent on body borne load. Body borne load also led to significant increases in knee and ankle stiffness. However, significant differences in ankle stiffness were not seen between load conditions after correcting for Type 1 error. Participants decreased ankle stiffness, but not hip or knee stiffness with longer strides. The reduction in ankle stiffness may be attributed to an increase in PS ankle dorsiflexion angle with the longer strides, despite a concomitant increase in PS ankle dorsiflexion moment.

A sex dimorphism in lower limb biomechanics was evident while running with body borne load. Specifically, males decreased leg stiffness with longer strides, but similar changes in leg stiffness were not evident for female participants. Females exhibited greater lower limb flexion compared to males with the shorter strides, but did not increase lower limb flexion with longer stride lengths like the male participants. Furthermore, males increased peak knee flexion moment with each incremental increase in stride length, while females only increased knee moment between the preferred and long strides.

Limitations

This study may be limited by the methods of calculating leg and joint stiffness. Leg stiffness was calculated using sagittal plane measures of GRF and leg length. The leg stiffness calculation did not include mediolateral components of GRF or frontal plane measures of leg length, but a sex dimorphism may be evident for these frontal-plane measures during locomotion[39,40]. Adding those measures to the leg stiffness calculation may be warranted. The joint stiffness calculation assumes that the joint behaves like a torsional spring – where the joint moment compresses the spring. During unloaded running, the hip's primary function is energy production[83] and the moment may not act to compress the joint. However, the hip reportedly increases its contribution to energy absorption with the addition of body borne load[84], where the moment would act to compress the joint, and thus, may require greater stiffness during weight acceptance when carrying heavy loads.

Another limitation may be the military relevance of the chosen method for applying body borne load and running speed. Currently, the body borne load was applied with a weighted vest. The weighted vest allows for systematic adjustment of load applied to the torso of each participant, but may not have the same load distribution as the equipment donned by military personnel during training or operational exercises. Further, all participants ran at 4 m/s during testing. While this allows testing to be consistent, it may not be representative of the running speed chosen during military operations, potentially limiting the ability to generalize the current findings to all military activities. Finally, each participant's preferred stride length was calculated for each body borne load and may limit the direct effect of load.

Future Work

Future research is warranted to determine if strength, rather than sex, impacts the ability to adjust leg stiffness when running with body borne load. Each incremental addition of body borne load increases peak GRFs and thus, adequate muscle strength may allow participants to safely attenuate the large GRFs. A prospective study analyzing how a strength training program, focused on the lower limb, prior to carrying heavy body borne loads affects the attenuation of ground reaction forces and injury rates of military personnel would be beneficial. Further, research may be necessary to determine the ankle's role during the weight acceptance of running with the body borne load. Targeting ankle strength during training may be helpful for adequately attenuating GRFs during weight acceptance of running with body borne load, and may reduce injury risk of military training.

Future work to improve the method for calculating hip stiffness may be warranted. Considering the hip is an important energy producer during running and may aid with energy absorption during load carriage, a better, more accurate, method for calculating stiffness at the joint may improve understanding of hip function during load carriage leading to more appropriate training of its associated musculature.

Finally, differences in frontal plane biomechanics are documented between males and females during running[39]. A 3D calculation of leg stiffness that accounts for these frontal plane differences may be a more appropriate method of analyzing a potential sex dimorphism during load carriage.

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APPENDIX A

Pre-participation Questionnaire

1. Have you suffered an injury to your hip, knee, or ankle in the past 6 months?

YES NO

If yes, please describe: ____

2. Have you undergone surgery to your hip, knee, or ankle?

YES NO

If yes, please describe: ____

3. Are you currently undergoing rigorous physical training or do you plan to start a rigorous training program in the next 3 months?

YES NO

If yes, please describe: ____

4. Are you currently experiencing knee pain?

YES NO

5. Are you currently suffering from or have you ever suffered from a heart condition?

YES NO

If yes, please describe: ____

6. Do you know of any reason why you cannot participate in this study?

YES NO

If yes, please explain: ____

I certify that the information I provided above is accurate.

Subject's Signature: ___________________________ Date: ______________

Subject's Name (Print): _________________________

Parent/Legal Guardian Signature: ___________________ Date: ___________

Parent/Legal Guardian Name (Print): _______________________

APPENDIX B

Footedness Questionnaire

Instructions: Answer each of the following questions as best you can. If you always use one foot to perform the described activity, circle **Ra** or **La** (for right always or left always). If you usually use one foot circle **Ru** or **Lu**, as appropriate. If you use both feet equally often, circle **Eq**.

Please do not simply circle one answer for all questions, but imagine yourself performing each activity in turn, and then mark the appropriate answer. If necessary, stop and pantomime the activity.

1. Which foot would you use to kick a stationary ball at a target straight in front of you?

La **Lu** **Eq** **Ru** **Ra**

2. If you had to stand on one foot, which foot would it be?

La **Lu** **Eq** **Ru** **Ra**

3. Which foot would you use to smooth sand at the beach?

La **Lu** **Eq** **Ru** **Ra**

4. If you had to step up onto a chair, which foot would you place on the chair first?

La **Lu** **Eq** **Ru** **Ra**

5. Which foot would you use to stomp on a fast-moving bug?

La **Lu** **Eq** **Ru** **Ra**

6. If you were to balance on one foot on a railway track, which foot would you use?

La **Lu** **Eq** **Ru** **Ra**

7. If you wanted to pick up a marble with your toes, which foot would you use?

La **Lu** **Eq** **Ru** **Ra**

8. If you had to hop on one foot, which foot would you use?

La **Lu** **Eq** **Ru** **Ra**

9. Which foot would you use to help push a shovel into the ground?

La **Lu** **Eq** **Ru** **Ra**

10. During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first?

La **Lu** **Eq** **Ru** **Ra**

11. Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities?

Yes **No**

12. Have you ever been given special training or encouragement to use a particular foot for certain activities?

Yes **No**

13. If you have answered YES for either question 11 or 12, please explain:

APPENDIX C

Table C.1 Mean (SD) Leg Stiffness, Peak vGRF, and Change in Leg Length Exhibited by Males and Females with Changes in Stride Length and Load

		20 kg		25 kg		30 kg		35 kg	
		Male	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**
Leg Stiffness[a,b,e]	**SSL**	30.05 (7.89)	26.32 (6.76)	32.93 (8.69)	27.48 (6.88)	33.66 (9.15)	27.54 (5.15)	33.05 (7.45)	28.19 (5.39)
	PSL	28.03 (5.85)	25.59 (4.37)	28.93 (7.03)	28.84 (8.18)	30.80 (6.56)	28.05 (7.50)	32.80 (7.09)	28.78 (5.60)
	LSL	23.95 (5.87)	22.81 (4.59)	25.68 (6.34)	25.55 (6.41)	25.62 (5.63)	26.28 (7.11)	27.02 (6.26)	26.27 (6.19)
Peak vGRF (BW)[a,b,c,e]	**SSL**	2.74 (0.31)	2.60 (0.15)	2.90 (0.33)	2.60 (0.22)	2.91 (0.27)	2.69 (0.24)	3.00 (0.31)	2.78 (0.21)
	PSL	2.91 (0.39)	2.52 (0.16)	2.98 (0.32)	2.61 (0.27)	3.06 (0.35)	2.65 (0.25)	3.15 (0.36)	2.80 (0.28)
	LSL	2.97 (0.40)	2.76 (0.18)	3.12 (0.39)	2.87 (0.19)	3.15 (0.43)	2.90 (0.21)	3.25 (0.39)	3.02 (0.23)
Change LL (%)[e]	**SSL**	9.53 (2.90)	9.96 (1.74)	10.06 (5.25)	9.74 (2.00)	9.02 (2.44)	9.56 (1.64)	10.27 (5.52)	9.72 (1.53)
	PSL	10.64 (3.11)	9.86 (1.71)	10.55 (2.93)	9.26 (2.25)	9.90 (1.97)	9.90 (2.50)	9.78 (2.70)	9.64 (1.68)
	LSL	12.71 (3.77)	11.91 (2.15)	12.11 (3.05)	10.92 (1.88)	12.24 (2.60)	10.84 (2.61)	12.36 (4.55)	11.16 (2.24)

[a] Denotes statistically significant effect of load
[b] Denotes statistically significant effect of stride length
[c] Denotes statistically significant effect of sex
[d] Denotes statistically significant interaction between sex and load
[e] Denotes statistically significant interaction between sex and stride length
[f] Denotes statistically significant interaction between load and stride length

Table C.2 **Mean (SD) Peak Stance Hip Joint Stiffness, Flexion Angle, and Flexion Moment Exhibited by Males and Females with Changes in Stride Length and Load**

		20 kg		25 kg		30 kg		35 kg	
		Male	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**
Hip Stiffness (N/Kg°)	**SSL**	0.14 (0.07)	0.58 (1.07)	0.20 (0.23)	0.28 (0.20)	0.34 (0.72)	3.24 (9.68)	0.33 (0.85)	0.13 (0.06)
	PSL	0.20 (0.19)	0.24 (0.24)	0.68 (1.91)	0.87 (1.97)	0.44 (0.77)	0.11 (0.05)	0.26 (0.43)	0.61 (0.80)
	LSL	4.47 (14.64)	0.34 (0.54)	0.31 (0.36)	0.17 (0.08)	0.46 (0.46)	0.54 (1.15)	0.59 (1.16)	0.49 (0.67)
Hip Flx. Angle (deg)[b,e]	**SSL**	29.27 (4.16)	33.12 (6.61)	29.36 (4.80)	33.87 (6.62)	29.15 (5.03)	35.44 (5.65)	30.58 (4.48)	35.45 (5.56)
	PSL	31.03 (5.00)	33.37 (6.33)	31.19 (6.01)	33.21 (5.09)	31.82 (4.87)	34.27 (5.44)	31.20 (7.40)	34.69 (5.30)
	LSL	32.79 (5.79)	35.57 (6.64)	32.78 (5.03)	34.39 (6.11)	32.85 (6.92)	35.76 (4.95)	33.41 (5.78)	32.76 (12.57)
Hip Flx. Moment (Nm/Kgm)[a,b]	**SSL**	1.04 (0.20)	1.03 (0.19)	1.09 (0.18)	1.18 (0.16)	1.12 (0.18)	1.19 (0.10)	1.16 (0.22)	1.25 (0.11)
	PSL	1.00 (0.21)	1.08 (0.20)	1.00 (0.16)	1.13 (0.14)	1.08 (0.24)	1.14 (0.21)	1.08 (0.20)	1.21 (0.13)
	LSL	0.98 (0.21)	0.99 (0.18)	1.01 (0.22)	1.06 (0.21)	1.02 (0.23)	1.12 (0.19)	1.09 (0.29)	1.20 (0.28)

[a] Denotes statistically significant effect of load
[b] Denotes statistically significant effect of stride length
[c] Denotes statistically significant effect of sex
[d] Denotes statistically significant interaction between sex and load
[e] Denotes statistically significant interaction between sex and stride length
[f] Denotes statistically significant interaction between load and stride length

Table C.3 Mean (SD) Peak Stance Knee Joint Stiffness, Flexion Angle, and Flexion Moment Exhibited by Males and Females with Changes in Stride Length and Load

		20 kg		25 kg		30 kg		35 kg	
		Male	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**
Knee Stiffness (N/Kg^o)[a]	**SSL**	0.07 (0.01)	0.07 (0.01)	0.07 (0.01)	0.08 (0.01)	0.08 (0.03)	0.08 (0.01)	0.08 (0.02)	0.09 (0.01)
	PSL	0.07 (0.01)	0.08 (0.01)	0.07 (0.01)	0.08 (0.01)	0.08 (0.01)	0.08 (0.01)	0.08 (0.01)	0.08 (0.01)
	LSL	0.07 (0.01)	0.07 (0.01)	0.07 (0.01)	0.08 (0.01)	0.08 (0.01)	0.08 (0.02)	0.08 (0.01)	0.08 (0.01)
Knee Flx. Angle (deg)[a,b,c,e,f]	**SSL**	40.80 (3.71)	49.22 (6.79)	41.53 (4.96)	48.90 (6.11)	41.34 (5.13)	47.51 (5.69)	41.04 (3.96)	47.94 (5.98)
	PSL	43.12 (5.23)	49.73 (6.48)	43.58 (5.71)	47.69 (6.89)	44.35 (5.01)	48.70 (7.29)	42.79 (5.11)	46.87 (6.36)
	LSL	47.22 (5.59)	51.38 (7.50)	47.20 (5.35)	50.48 (6.88)	48.30 (5.67)	49.38 (6.88)	46.18 (4.12)	45.85 (10.85)
Knee Flx. Moment (Nm/Kgm)[b,d,e]	**SSL**	1.68 (0.28)	1.91 (0.26)	1.87 (0.21)	1.84 (0.29)	1.85 (0.24)	1.81 (0.29)	1.91 (0.25)	1.92 (0.24)
	PSL	1.92 (0.26)	1.83 (0.29)	2.13 (0.31)	1.84 (0.26)	2.15 (0.27)	1.80 (0.21)	2.01 (0.37)	1.90 (0.23)
	LSL	2.14 (0.31)	2.17 (0.24)	2.36 (0.32)	2.11 (0.28)	2.41 (0.29)	2.12 (0.16)	2.38 (0.32)	2.08 (0.43)

[a] Denotes statistically significant effect of load
[b] Denotes statistically significant effect of stride length
[c] Denotes statistically significant effect of sex
[d] Denotes statistically significant interaction between sex and load
[e] Denotes statistically significant interaction between sex and stride length
[f] Denotes statistically significant interaction between load and stride length

Table C.4 Mean (SD) Peak Stance Ankle Joint Stiffness, Dorsiflexion Angle, and Dorsiflexion Moment Exhibited by Males and Females with Changes in Stride Length and Load

		20 kg		25 kg		30 kg		35 kg	
		Male	**Female**	**Male**	**Female**	**Male**	**Female**	**Male**	**Female**
Ankle Stiffness (N/Kg°)[a,b]	**SSL**	0.07 (0.02)	0.07 (0.02)	0.08 (0.02)	0.07 (0.01)	0.08 (0.02)	0.07 (0.01)	0.08 (0.02)	0.07 (0.01)
	PSL	0.07 (0.02)	0.07 (0.01)	0.077 (0.02)	0.070 (0.01)	0.08 (0.02)	0.06 (0.01)	0.08 (0.02)	0.07 (0.01)
	LSL	0.07 (0.01)	0.06 (0.01)	0.07 (0.02)	0.07 (0.01)	0.07 (0.01)	0.06 (0.01)	0.07 (0.01)	0.06 (0.01)
Ankle Dflx. Angle (deg)[b]	**SSL**	14.39 (6.05)	18.76 (6.97)	12.20 (6.08)	17.43 (9.02)	13.39 (6.61)	17.23 (6.52)	12.80 (6.07)	17.60 (9.07)
	PSL	16.20 (7.07)	19.39 (8.21)	13.54 (6.29)	16.11 (10.33)	15.51 (6.27)	17.81 (7.13)	15.29 (6.84)	16.40 (9.38)
	LSL	17.77 (7.53)	19.02 (13.75)	15.99 (6.05)	17.30 (15.71)	17.68 (6.36)	21.86 (9.13)	16.02 (5.87)	21.48 (8.88)
Ankle Dflx. Moment (Nm/Kgm)[b,c]	**SSL**	1.32 (0.38)	1.07 (0.17)	1.35 (0.33)	1.15 (0.15)	1.37 (0.31)	1.15 (0.23)	1.38 (0.28)	1.20 (0.22)
	PSL	1.44 (0.35)	1.15 (0.23)	1.42 (0.27)	1.07 (0.21)	1.51 (0.31)	1.17 (0.27)	1.52 (0.35)	1.21 (0.29)
	LSL	1.49 (0.36)	1.27 (0.25)	1.58 (0.35)	1.32 (0.22)	1.58 (0.37)	1.37 (0.26)	1.60 (0.33)	1.38 (0.22)

[a] Denotes statistically significant effect of load
[b] Denotes statistically significant effect of stride length
[c] Denotes statistically significant effect of sex
[d] Denotes statistically significant interaction between sex and load
[e] Denotes statistically significant interaction between sex and stride length
[f] Denotes statistically significant interaction between load and stride length