Comparison of Lower Extremity Biomechanics and Effort in Male and Female Military Members During a Standardized Load Carriage Task

by

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Abstract

Objective: To understand the effect of load carriage on knee moment features linked to development/progression of knee osteoarthritis and whether there are differences between males/females.

Methods: 24 military members(14 male, 10 female) walked at self-selected speed (loaded/unloaded) and fixed speed(loaded) on an instrumented treadmill. Motion capture cameras and custom software captured, analyzed, and calculated discrete knee moment features. Repeated measures analysis of variance tested for between condition differences for the total sample. Confidence intervals(95%) were calculated to determine between sex and between conditions within sex differences.

Results: Greater knee moment features were found for loaded versus unloaded conditions. Three joint moment features were different between loaded conditions. One moment feature was different between sexes; males/females did not have the same between condition pairwise differences for three moment features.

Conclusion: Loaded marching and increased speed increased knee moment features. Preliminary data indicate differences in joint moment features for males and females.

List of abbreviations and symbols used

3D	Three-dimensional
ACL	Anterior Cruciate Ligament
ANOVA	Analysis of Variance
BMI	Body Mass Index
BMOQ	Basic Military Officer Qualification
BMQ	Basic Military Qualification
CAF	Canadian Armed Forces
CFB	Canadian Forces Base
CI	Confidence Intervals
EMG	Electromyography
FS	Fixed Speed
FSL	Fixed Speed Loaded
GRF	Ground Reaction Force
HLIS	Health and Lifestyle Information Survey
Hz	Hertz
ICC	Interclass Correlation Coefficient
KAM	Knee Adduction Moment
KE	Knee Extensor
KEM	Knee Extension Moment
KF	Knee Flexor
KFM	Knee Flexion Moment
kg	Kilogram

L	Loaded
M	Mean
m	Metre
MCU	Muscle Capacity Utilization
mm	Millimetre
MVIC	Maximal Voluntary Isometric Contraction
MRI	Magnetic resonance imaging
N	Newton
Nm	Newton Metre
OA	Osteoarthritis
P	p-value
ROM	Range of Motion
RPE	Rate of Perceived Exertion
s	Second
SD	Standard Deviation
SS	Self-Selected
SSL	Self-Selected Speed Loaded
SSU	Self-Selected Speed Unloaded
TKA	Total Knee Arthroplasty
U	Unloaded
UK	United Kingdom
US	United States

Chapter 1. Introduction

The overall goal of this thesis is to advance understanding of the risks for knee joint clinical osteoarthritis (OA) development and progression for military members by examining biomechanical adaptations during load carriage with operationally relevant load and speed and to provide preliminary data on potential differences in these adaptations between males and females. Chapter 1 describes the background and motivation for the thesis followed by the specific objectives, hypotheses, and thesis outline.

1.1 Background and motivation

Musculoskeletal injuries and associated conditions, such as osteoarthritis (OA), are a global concern for militaries including the Canadian Armed Forces (CAF) and those of allied nations such as the United Kingdom (UK), Australia, and the United States (US) [1-4]. Military personnel are routinely exposed to biomechanical factors associated with increased risk for musculoskeletal injury and OA [3, 5-8]. Therefore, it is not surprising that CAF members commonly report musculoskeletal injuries with chronic medical conditions involving the muscles and joints of the lower extremity being among those most commonly reported [3]. Injuries that are high risk for medial knee OA (e.g., ACL and meniscal injuries) [9] have been reported in the CAF and other militaries [10-13] with their incidence exceeding that of the general population in some military populations and occupations [10, 11, 14, 15]. Together, the exposure to biomechanical risk factors and injury may partly explain the incidence of chronic musculoskeletal conditions including OA in military populations [3, 14, 16]. While the relationship between injury and post-traumatic knee OA is more apparent [9, 17], the understanding of the relationship between common military tasks (e.g., load carriage) and knee OA development and progression is less well understood.

OA has been identified as a major public health challenge [18] with knee OA related disability being identified as a worldwide concern [19]. There is evidence that the burden of OA may be even greater for military populations compared to the general population [16, 20, 21]. US military members have been reported as being five times more likely to receive a diagnosis of post traumatic knee OA compared to the civilian population [14] and are more likely to be diagnosed with clinical OA as a young adult [20, 21]. Joint replacements, including knee arthroplasty, are common for military members [22] with the frequency of US military members reaching end-state knee OA exceeding that of civilians [21]. Additionally, the greater burden of OA associated with military service may impact female military members disproportionately. Despite comprising the minority of members, female members of the US military have a higher incidence of knee OA than male members [1, 23-25]. Similar CAF findings for female members indicate that they are more likely to report an injury or chronic musculoskeletal health condition compared to their male colleagues [3].

This study focuses on knee OA given the associated limitations in functional mobility, participation in occupational tasks, leisure, and activities of daily living [26, 27]. From an overall health perspective, knee OA has been associated with increased obesity rates (25% of CAF members reported obesity in 2014 [3]) [26], and lower quality of life for members of the general population [27]. The clinical diagnosis of knee OA is based on the presence of signs and symptoms (e.g., swelling, joint pain) plus evidence of knee joint structural degradation [28].

Structural OA changes only may be indicative of future clinical progression [29] but are not an accurate representation of an individual's clinical presentation as the degree of structural damage is not well correlated with the severity of symptomology [30, 31]. The diagnosis of clinical knee OA has additional implications for military personnel including potential for early, involuntary, release from service [1, 21, 25, 32, 33]. The requirement to manage clinical OA due to its associated signs and symptoms better reflects the burden of OA on the individual [34] and health care systems [35]. Importantly, musculoskeletal injuries and conditions, including knee OA, are a leading contributor to military health care costs [23, 32, 36] and are a threat to a military's ability to fulfill its operational mandates [1, 4].

There is evidence that military load carriage contributes to injuries to both the knee and supporting lower extremity structures that can lead to the development of knee OA [4, 7, 8, 37-41]. In 2014, CAF members reported 4.9 serious injuries per 1000 hours of loaded marching – the second highest rate of serious injury (i.e., injury that limited activity for at least one week) associated with a specific activity [3]. Factors identified as potential contributors to injury during load carriage include greater physiological demands [42-44] with some evidence of altered gait mechanics [40, 44, 45] and increased load on lower extremity joints and musculature [41, 44].

Studies included in recent systematic reviews [45, 46] of load carriage have typically utilized mixed civilian and military cohorts and, while valuable, the findings from these studies may have limited application to military load carriage. The nature of military experience, training, and work requirements with heavy load carriage has been cited as reasons why studies involving load carriage by civilians may not be directly relevant to military personnel [40]. For example, studies of load carriage with military members have identified changes to biomechanical and electromyography (EMG) features that were not consistently identified in civilian participants during load carriage [40, 45, 47, 48].

Data shows that the magnitude of the loads carried by military members has more than doubled since World War I [48-50]. Coinciding with the increase in loads carried by military members, a six-fold increase in disability rates in US Army members has been reported since the 1980's [48]. Unsafe practices relating to the mass and pace of load carriage have been reported by CAF members as factors contributing to acute and repetitive strain injuries [3]. While numerous studies have examined the effects of load carriage on spatio-temporal features and angular kinematics in both civilian and military populations [40, 45], few studies have examined the effect of load carriage on biomechanical and muscular stresses using standardized, operationally relevant, loads at operationally relevant fixed speed (FS) [40]. Typically, loads and speeds relative to the participant's body weight and self-selected (SS) walking speed have been used; relative loads and speeds may not accurately represent military operational or training requirements. FS walking has been associated with adaptations to stride length and frequency in order to maintain pace [33], with increased energy cost [51, 52], GRFs [40, 53], knee joint loads [40], and injury rates [3, 54]. These are important consideration for female military members as an operationally relevant load typically represents a greater percentage of a female's body mass [55-57]. Height has been linked to gait alterations [33] and the height of female military members is typically less than that of male military members [25, 57, 58]. Notably, a recent scoping review of the literature concerning sex differences during load carriage by military members identified that studies are either not including female military members or are grouping them with male military members and not conducting sex based comparisons [52].

Most studies related to biomechanical gait adaptations during military load carriage have focused on features related to injury prevention. This is important to post-traumatic OA development, however, the findings of a recent systematic review show that few studies have investigated the effects of cumulative load on the knee, or features of gait shown to be linked to knee OA development or clinical progression including structural and symptom worsening [40]. Furthermore, a scoping review that included 18 studies found that there has been limited comparison of biomechanical and physiological adaptations to load carriage between male and female military members [52].

Crucial to this study is evidence from studies of human biomechanical and muscle activation analysis of walking gait, summarized in a systematic review [59], that examined the role of joint motion/loading and muscle recruitment patterns to better understand risk factors related to the development and clinical progression of knee OA. These studies have identified key features of gait linked to development and progression of knee OA including higher external knee adduction moment (KAM) [60-62] impulse [63, 64], Peak KAM [59, 63, 65], peak knee flexion moment (KFM) [66], smaller KFM range [63] and more prolonged EMG patterns [67].

Since methods of directly measuring knee joint contact forces *in vivo* are challenging and not pragmatic for most participant populations (e.g., healthy military members), surrogate measures have been used to approximate knee joint loading [59, 68]. The KAM and the KFM features are the two most studied gait features related to knee joint OA [59, 69].The KAM is considered a surrogate for the ratio of medial to lateral knee joint loading [59]; KAM impulse captures the overall magnitude and duration of load whereas Peak KAM captures the highest magnitude loading at only one point in the gait cycle [59]. KAM impulse and Peak KAM have been associated with structural progression outcomes of knee OA [60-62, 64, 66] where the structural changes are a component of clinical progression [29]. KAM magnitudes are influenced by walking speeds with higher peaks reported at faster walking speeds [70]. Furthermore, higher KAM magnitudes have been associated with development of chronic knee pain [71], and have been reported in individuals with knee OA who are at higher risk for total knee arthroplasty (TKA) surgery [63]. Increased magnitudes of KAM features during load carriage may be a risk factor for knee OA development and clinical knee OA progression in military members but only a few military load carriage studies using FS and standardized loads have reported on KAM metrics (e.g., Peak KAM) [7, 72]. The relative speeds (e.g., 10% less than gait transitional velocity) and loads (e.g., 45% body weight) from these studies are not necessarily relevant to military operations [7] and the samples used were not military personnel [7, 72]. The effect of military load carriage with FS and standardized load on other key features of gait linked to OA progression (i.e., KAM impulse, Peak KAM, Peak KFM, and KFM-KEM) has not been addressed [40, 52].

The external KFM is a surrogate for the internal muscle moment at the knee joint and represents the torque produced by the knee extensors [73], but may be an underestimation if there is antagonist co-activity. Joint loading patterns (e.g., cyclic or sustained), not only magnitudes, play a key role in regulating joint physiology and pathology [74-76]. In clinical studies, higher KF muscle activity magnitudes [77] and increased co-activation of the KE and KF muscles have been associated with structural and clinical progression of knee OA *in vivo* [77, 78]. While the focus of the literature concerning features of gait and knee OA risk has been on KAM magnitude features [59], a decreased ability to unload the knee during gait was found to be predictive of clinical OA progression based on a higher overall KAM magnitude, smaller difference in early and midstance KAM magnitudes and smaller differences between early stance KFM and late stance phase knee extension moment (KEM) range [63]. This stiff gait pattern is characterized by decreased difference between early Peak KFM and late stance KEM (KFM-KEM), is supported by greater magnitude knee flexor (KF) muscle activity, and prolonged co-contraction of the KF and knee extensor (KE) muscles [63, 77] indicative of higher active stiffness. A better understanding of the effect of load carriage using operationally relevant loads and FS on features of gait linked to the clinical progression of knee OA would add to the knowledge concerning the development of medial knee joint OA in military members.

There is emerging evidence of sex differences in adaptations to military load carriage [52]. Differences in height, body mass, muscle strength, and aerobic fitness, have been reported between sexes in military populations [25, 33, 39, 57, 79]. While these differences have been cited as factors relating to discrepancies in load carriage adaptations between male and female military member [33, 55], whether stature or sex influences gait adaptations during FS load carriage is unclear [33]. Mixed military-civilian and civilian load carriage studies have found sex differences in lower limb angular displacement measures during load carriage [52, 57, 80] including in knee sagittal loading adaptations with females displaying less knee excursion during early stance [80], less medial-lateral centre of mass displacement than males [57], and males displaying increased knee ROM as load is increased [57].

Another biomechanical adaptation, forward trunk lean (i.e., forward spinal flexion), commonly reported as the peak trunk angle during load carriage, has been consistently reported in studies of military load carriage [40]. Studies found that females have greater peak trunk angle compared to males during load carriage with the same similar mass, in studies with mixed civilian-military cohorts [47, 48]. Forward lean alters spinal [81] and lower extremity kinematics, alters spinal and pelvic muscle activation patterns [48] and may alter the magnitude of knee joint moments (e.g., reduced KFM) and counterbalance posterior loads [82]. However, whether trunk lean is sex related or body mass related is unclear [55, 80] and this feature provides a potential explanatory measure for altered joint moments and differences in physiological demands.

Physiological factors (e.g., strength) and biomechanical factors (e.g., peak trunk angle) have the potential to increase both the effort required during load carriage and the risk of injury [33, 55, 56]; these factors may contribute to the increased incidence of knee OA in military members. Muscles are important to producing movements, joint stability, and joint loading, and are subjected to increased physiological demands during load carriage [83, 84] with muscle fatigue a potential risk for knee OA associated with load carriage. Fatigue of the KE muscles has been linked to increased KAM, changes to the knee loading environment [85], and injuries that are high risk for knee OA development [86, 87]. Importantly, there is evidence that females expend greater energy (effort) [43, 52, 88] compared to males for similar load carriage tasks potentially increasing their risk for knee OA development. Effort can be assessed qualitatively through self-report measures of perceived exertion (RPE) (i.e., Borg Scale) [89, 90] or quantitatively through measures of exertion, such as muscle capacity utilization (MCU) [90, 91] or EMG. RPE, measured using a Borg Scale, has been shown to be a reliable measure of effort with good-excellent levels of relationship to measures of cardiovascular effort (e.g., cardiovascular stress) [92]. At the knee joint, MCU has been used to assess the ability to utilize muscle strength to efficiently and effectively complete physical activities and is calculated as the joint moment measure during an activity relative to maximal capacity at the joint [90, 91, 93].

In summary, there are gaps in the load carriage literature related to the biomechanical adaptations for military members using operationally relevant load and FS while few studies have investigated the effects of cumulative load on the knee or features of gait shown to be linked to knee OA development or clinical progression for military load carriage. Furthermore, female military members have been underrepresented in biomechanical studies related to load carriage [40, 52] and there is emerging evidence of sex differences in biomechanical and physiological responses to load carriage [52]. Examining whether differences in knee joint moment adaptations linked to medial compartment knee OA development and progression exist between male and female military members could address knowledge gaps related to knee OA risk for both male and female military members.

1.2 Thesis aim

The primary aim of this thesis was to determine how walking speed, self-selected speed (SS) or fixed speed (FS), and standardized, operationally relevant, load (L) carriage affects the knee joint loading environment versus unloaded (U) walking at a SS speed in CAF members and if there was a difference in the knee joint loading environment between male and female CAF members. For the secondary aim, self-report and quantitative measures of exertion and trunk flexion were examined to better understand the differences in effort and trunk forward lean during walking gait among loaded and unloaded conditions and between sexes. This aim was addressed through two main objectives and four sub-objectives.

1.3 Objectives and sub-objectives

Objective 1: To determine whether there are differences among walking conditions (i.e., SSU, SSL, and FSL) for joint moment features related to the development and clinical progression of knee OA (i.e., KAM impulse, first Peak KAM, Peak KFM, and KFM-KEM).

> *Sub-objective 1a:* To quantify differences in exertion (effort) using a self-reported measure of exertion (i.e., RPE) and a quantitative measure of KE muscle exertion (i.e., MCU) between the walking conditions.
>
> *Sub-objective 1b:* To quantify differences in sagittal plane trunk angle (i.e., peak trunk angle) between walking conditions.

Given the gap in studies that compare between male and female military members during load carriage [52], and none specifically looking at features related to risk of knee OA, the following objectives and subobjectives are exploratory in nature.

Objective 2: To provide preliminary data on whether there are sex differences in joint moment features related to the development and clinical progression of knee OA (i.e., KAM impulse, first Peak KAM, Peak KFM and KFM-KEM) within and between walking conditions.

Sub-objective 2a: To provide preliminary data on potential differences between sexes for self-reported and quantitative measures of exertion (i.e., RPE and MCU respectively) within and between walking conditions.

Sub-objective 2b: To provide preliminary data on potential differences between sexes in sagittal plane trunk angle (i.e., peak trunk angle) within and between walking conditions.

1.4 Hypotheses

1.4.1 *Primary outcomes*

For Objective 1, the main hypothesis is that there will be significant differences among walking conditions with the FSL walking condition resulting in the highest KAM impulse, Peak KAM, Peak KFM, and the smallest KFM-KEM differences. The hypothesis for Objective 2 is that there will be significant differences between groups (i.e., sexes) with females having lower Peak KAM and KAM impulse, higher Peak KFM, and smaller KFM-KEM, for each walking condition. Furthermore, there will be differences within groups (i.e., sex) among walking conditions where females will have a smaller change in KAM impulse and Peak KAM, a greater change in Peak KFM, and greater KFM-KEM than males, for the loaded conditions and that both groups will have greater KAM impulse, Peak KAM, Peak KFM, and smaller KFM-KEM during FSL than SSL and SSU.

1.4.2 Secondary outcomes

For Sub-objective 1a, it was hypothesized that there would be significant differences in MCU and RPE between walking conditions (i.e., SSU, SSL, FSL) with FSL resulting in highest RPE and greatest MCU. The hypothesis for Sub-objective 1b was that there would be significant differences in peak trunk angle between walking conditions with FSL resulting in greatest peak trunk angle.

The hypothesis for Sub-objective 2a was that females will have greater RPE and MCU than males during each walking condition and that there will be a significant difference within groups with the FSL condition resulting in both the highest RPE and greatest MCU for both sexes. For Sub-objective 2b, it was hypothesized that females will have greater peak trunk flexion angle than males during walking conditions and that the greatest peak trunk angle for each sex would occur during the FSL condition.

1.5 Thesis outline

This master's thesis includes six chapters. Chapter 2 reviews the relevant background literature on OA, the burden of knee OA in civilian and military populations, features of gait linked to the progression of knee OA. Subsequently, the literature concerning biomechanical and physiological adaptations to load carriage in civilian and military populations was reviewed. Chapter 3 provides a detailed description of the study methodology including participant recruitment, procedure, and statistical analysis. Chapter 4 provides the key results related to Objective 1 and Sub-objectives 1a and 1b and discusses these findings. Chapter 5 provides the key results related to Objective 2 and Sub-objectives 2a and 2b and discusses these findings. A summary of key findings and discussion of the implications, limitations, and conclusions related to the overall goal of the thesis are presented in Chapter 6.

Chapter 2. Background literature

This chapter contains a review of the burden of knee OA in both the general population and military contexts. The literature relating to the primary outcomes, key features of gait related to the progression of knee OA, is reviewed. Next, the literature relating to military load carriage effects on gait biomechanics, physiological demands, and the primary and secondary outcomes is reviewed. A summary of the reviewed literature is presented in the last section of this chapter.

2.1 Osteoarthritis

Osteoarthritis (OA) is a chronic health condition characterized by the progressive degeneration of articular cartilage in synovial joints [94]. Signs and symptoms of OA include joint pain and stiffness, swelling, loss of function, impaired muscle strength, and reduced range of motion [95]. OA is a whole joint disease that affects joint tissues including its boney structure, cartilage, synovium, synovial fluid, joint capsule, ligaments/menisci, and supporting musculature [96]. Radiographically, OA related changes may be identified through the formation of osteophytes, cysts, narrowed joint space, and sclerosis of subchondral bone [97].

Negative findings on blood and joint fluid analysis may help clinicians differentiate OA from other types of arthropathy when there is a requirement to exclude conditions such as Rheumatoid Arthritis or infection [97]. There is no cure for OA; treatment options include non-pharmacologic (e.g., exercise), pharmacologic (e.g., non-steroidal anti-inflammatory medication), complimentary (e.g., acupuncture) and surgical interventions (e.g., TKA) [6]. The current strategy of symptom management and conservative care up until the point of TKA burdens affected individuals and places demands on health care systems [35].

OA is prevalent globally with incidence being highest in the US [18]. The global burden of OA has been theorized to be increasing due to the world wide trends of increased lifespan and obesity [18]; projections indicate that 25% of Canadians will be diagnosed with OA by the year 2030 [58]. The knee, particularly its medial compartment [98], is one of the joints commonly affected by OA and clinical knee OA has been recognized as a significant cause of pain and locomotor disability [19, 99]. Clinical knee OA may limit functional mobility (e.g., walking and stair climbing), participation in occupational tasks, leisure and activities of daily living, and is associated with increased obesity rates and lower quality of life [27]. Military populations have been reported as having higher incidences of OA than civilian populations [14-16] and the frequency of US military members reaching end-state knee OA is greater than that of the general population [21]; chronic lower extremity joint conditions are among the most common health concerns reported by CAF members [3]. Musculoskeletal conditions, including knee OA, are a leading contributor to military health care costs [23, 32, 36] and are a threat to a military's ability to fulfill its operational mandates [1, 4].

2.2 Historical perspective and risk factors for knee OA

Historically, OA has been viewed as a purely mechanically driven, or wear and tear, disease process with inflammatory factors considered to play a limited role or to be the consequence of altered biomechanical load on tissues; inflammation was not seen as having a significant role in the onset or progress of OA [5]. The etiology of OA was, and is still, not well understood [5, 95]. As with other chronic illnesses, a detailed understanding of its onset, progression, and potential preventative or mediating factors is limited by available experimental modelling and observational studies [95]. Risk factors for OA include previous injury to the joint or its supporting structures, obesity, congenital issues impacting cartilage or bone formation/alignment, repeated exposure to excess joint loading [100, 101], increased age, and female sex; other risk factors for OA have been identified as a genetic pre-disposition, and metabolic disease [5, 6]. While it may be possible that these other risk factors also impact an individual's joint biomechanics, there is evidence indicating that pro-inflammatory molecules (e.g., cytokines and leptin) play a role in the development of OA and that the associated pathogenesis and progression of OA is more complex than what was previously assumed [5, 94]. Some risk factors that were assumed to have an association with OA onset and progression based solely on their biomechanical influence (e.g., obesity) may have an influence through biochemical effects or combined biomechanical and biochemical interactions [102]. Injury, obesity, repeated exposure to excess joint loading, female sex, and age, are all knee OA risk factors that are reported by military members [3, 13, 20, 23-25, 55, 56, 103].

Knee alignment has been investigated as both a mechanism for knee injury and for its role in knee OA development and progression [104]. Greater quadriceps angle, associated with a more valgus alignment, has been reported in females than males [62], but no differences in quadriceps angle measurements have been reported between sexes in other studies [105, 106]. The gold standard for measuring knee alignment is full limb radiograph [107]; this method is costly, exposes the participant to radiation, and may not be readily accessible. The reliability and validity of other common quadriceps angle measurement procedures (e.g., goniometry) have been questioned [108]. Although greater quadriceps angles have been associated with increased rates of various musculoskeletal knee conditions, a recent review of 69 studies found that there was insufficient evidence to support an association between greater quadriceps angles and knee pathology [109]. While varus and valgus alignments have been associated with tibiofemoral OA development and progression [110-112], there is evidence suggesting that malalignment is a marker of knee OA disease progress and not a risk factor for knee OA development or progression [113].

2.3 Understanding knee OA through models and gait analysis

OA in humans is multifactorial [114]. In the absence of pathology, articular cartilage is able to tolerate repeated cyclical loading [115] provided the cyclical loads are not of excessive or injurious magnitudes [101]. Recent studies have observed that physiologic (moderate) levels of cyclic loading are associated with anabolic or anti-inflammatory responses in synovial joints and that hyper-physiologic loads and injurious loads lead to cell death and extracellular matrix metalloproteinases – supporting the concept of a complex interaction of biomechanical and biochemical factors affecting articular cartilage as the initial step in the onset of OA [94, 115]. Biochemical factors (e.g., cytokines) have been recognized as having a role in the onset and progression of OA [94] with disruption of chondrocyte metabolism leading to a net catabolism of articular cartilage and progressive joint degeneration [114], however, biomechanical factors also play a role in knee OA structural onset and clinical progression [69].

Animal models and explants have provided insights into the progression of OA in response to biomechanical loading *in vitro* [5, 102, 116], however, these types of studies are not well suited for human subjects. Human gait analysis, including kinetic and kinematic assessments, and EMG analysis, have provided insights into the role of biomechanics in the onset and progression of clinical knee OA [59, 67, 77, 117, 118]. Human gait analysis has been used as a model to better understand the local joint loading environment and knee OA processes [59, 119]. Both the effects of pathology on joint function and the effect of joint function during gait on knee OA processes have been examined using gait analysis.

2.4 Key features of gait linked to knee osteoarthritis progression

Differences in gait mechanics between individuals with and without knee OA have been reported in the literature [70, 120-124]. These gait changes include slower walking velocity and cadence, greater stance ratios and durations, smaller stride lengths [63, 70, 120], decreased knee flexion angle [121], increased KAM [120, 122], and decreased KFM [123] for those with knee OA while higher Peak KAM has been linked to onset of knee OA [59, 65]. Studies have identified key features of gait associated with structural [60, 64, 66] and clinical progression of knee OA [59, 63, 65, 67, 77, 117]. These features include kinetic measures (i.e., joint moments) and muscle activation patterns.

2.4.1 Gait mechanics and muscle activation patterns

The gait patterns of individuals with diagnosed moderate medial compartment knee OA that eventually require TKA have been observed to differ from those with a similar diagnosis that do not require surgery [63, 67, 118, 125]. Specifically, those receiving TKA have higher KAM magnitudes [118, 125], peaks [126, 127], less variation in KAM between early and midstance [63], smaller late-stance knee extension compared to early stance knee flexion moments [63] and increased KE and KF muscle activity during stance [67]. There is more load distributed to the medial compartment of the knee throughout the gait cycle for these individuals. While joint moments are not direct measures of joint contact forces, KAM is widely accepted as a surrogate measure of medial to lateral compartment loading ratio [59, 69].

The external KFM is a surrogate for the internal muscle moment at the knee joint and represents the torque produced by the KE muscles [68, 73, 128]. Although KFM has been inconsistently linked to knee OA progression in the literature [61, 66], potentially due differences in study samples, it contributes to increased knee joint loading [128] and may be underestimated if there is antagonist co-activity. Combined, KAM and KFM may account for up to 85% of the variance in medial knee joint loading during unloaded gait [128].

A stiff gait pattern has been observed following anterior cruciate ligament (ACL) injury and is associated with increased quadriceps and hamstring activity [117, 128]. The increased muscle activity and coactivation associated with stiff gait increases the stability of the knee joint, however, it also alters the cyclic loading and unloading pattern of normal gait and may lead to altered patterns of cartilage loading and imbalances in the joint's anabolism and catabolism of cartilage thereby contributing to the onset of OA. Repeated (i.e., 500ms on, 1500ms off), extended (i.e., 50 minutes), submaximal (i.e., 20% maximum isometric force), joint loading via muscular activation has been shown to increase chondrocyte death in animal models [75] and increased KF muscle activity magnitudes and increased co-activation of the KE and KF muscles have been associated with clinical progression of knee OA *in vivo* [77].

Only 2 of 20 studies included in a recent systematic review of military load carriage [40] examined features of gait linked to knee OA progression. Improved understanding of the effect of load carriage on key features of gait known to be linked to the onset and clinical progression of knee joint OA may provide insight into the higher incidence of knee OA in military personnel compared to the general population.

2.4.2 Sex differences in gait mechanics and muscle activity patterns

In a systematic review including 41 articles and more than 23 000 participants [129], similar over ground walking speeds were reported for males and females without pathology for each decade of life, from the third through seventh, and that females demonstrated greater decreases in walking speed than males during later decades. Similar walking speeds have been reported in participants with and without knee OA. Similar knee joint kinetics have been reported for males and females without knee OA [122, 130] while studies of knee joint kinematics and spatio-temporal parameters have yielded unclear results concerning potential sex differences [131, 132]. A study of participants with severe knee OA found greater KAM peaks and magnitudes for males compared to females [133, 134] and a study of patients with moderate knee OA found no between sex differences for KAM features [135]. Lower KFM-KEM differences, indicative of a stiff, less dynamic, gait have been identified in females with moderate and severe arthritis compared to males with similar knee OA severity [133, 135]. The literature is less clear concerning sex differences for KFM in individuals with knee OA. While some knee OA studies have reported lower KFM for females [135] other have reported greater magnitudes [133]. Overall, these findings are suggestive of a difference in later stage knee OA manifestation in males and females [133].

Few studies have investigated, or compared, features of gait related to knee OA during load carriage for females or males [52]. One recent study of recruit aged females identified a stiff gait during load carriage tasks using bodyweight relative loads (i.e., 25% and 45%) and speeds relative to gait transitional velocity [136]. However, this gait pattern was identified based on kinematic data and did not include measures of muscle activity. Stiff gait may be a sex specific load carriage gait adaptation and its presence during load carriage for female military members could partially explain sex differences in knee OA incidence in military personnel. Higher KAM percentages of knee joint total load were also identified in a separate study, using the same sample, during FS marching using relative loads [7]. No studies included in a review of military load carriage [40], or a recent scoping review on sex differences in load carriage biomechanics [52], compared male and female military members for features of gait related to clinical progression of OA. There is a need to better understand between sex differences in gait biomechanics for military load carriage [52] and its relationship to features of gait linked to clinical progression of knee OA.

2.5 Structural changes and clinical progression of knee OA

The degree of structural changes to a joint, often reported as Kellgren-Lawrence grades, does not necessarily match the symptoms reported by the individual [30, 31]. Clinical OA is defined as the existence of OA related symptoms (e.g., pain, stiffness) in the presence of structural changes on diagnostic imaging [28]; structural knee OA may exist in the absence of symptoms [137]. The lack of matching between symptom severity and structural changes has led to the differentiation between OA as an illness (i.e., clinical OA) and OA as an underlying disease (i.e., structural changes) [138, 139]. Although structural changes may be associated with future clinical progression [29], clinical knee OA is more likely to cause an individual to seek care and better represents the burden of OA on the person, health care systems, and society [35]. Given the difficulties and limitations in measuring symptoms and their lack of matching with structural changes, the requirement for TKA may be a more representative measure of clinical OA progression [34, 140]. From a military health care perspective, the symptoms, impairments, locomotor disability, and requirement for clinical management, potentially including TKA, associated with clinical knee OA are more likely to result in a member's release from service [21, 22], and contribute to health care costs [32], than OA related knee joint structural changes in isolation. Enhanced understanding of the role of common military tasks, such as load carriage, on the development and progression of clinical knee OA may assist to reduce its burden on all stakeholders.

2.6 Knee OA and military personnel

Military personnel are commonly exposed to risk factors for development of knee OA including several that relate to lower extremity joint loads and biomechanics, such as injury to joints and their supporting structures, and repeated exposure to excess joint loading [8, 21, 23, 33, 39, 90]. Post-traumatic knee OA resulting from battlefield injuries (e.g., blast injuries, intra-articular fractures, amputations) is a unique health concern for military members [20].

The most recent Health and Lifestyle Information Survey (HLIS) of CAF members indicated that three modifiable risk factors for OA are common in the CAF population including obesity, injury, and repetitive exposure to high levels of load [39]. The HLIS reported that 25% of CAF members were obese, that approximately one in five had sustained an acute injury within the twelve months leading up to the survey and that nearly one in three had sustained a repetitive strain injury; these injuries were most often associated with physical training including load carriage [39]. Injuries that are high risk for knee OA (e.g., ACL and meniscal injuries) [9] have been reported in the CAF and other militaries [10-13] with their incidence exceeding that of the general population in some military populations and occupations [10, 11, 14, 15].

Unsafe load carriage practices relating to both mass and speed at which loads were carried have been reported [39]; CAF members reported 4.9 serious injuries per 1000 hours of rucksack marching – the second highest rate of serious injury (i.e., an injury that limited activity for at least one week) associated with a specific activity [39]. With respect to chronic medical conditions amongst CAF members, the most commonly reported concerns were those effecting the muscles or joints of the lower extremity [39]. The available information from the US military, where the majority of military member related musculoskeletal injury research has been conducted [103], is more specific with respect to the burden of OA on its population. While knee OA is common in the general population, with approximately 10% of the population of the US being diagnosed with clinical knee OA by age 60 [141], the incidence of post traumatic OA in US military members is higher than the general population [14, 15]. Importantly, the rate of end state knee OA is greater among military members and veterans [21] and the incidence of OA increased in the US military from 2004-2014 [142]. Clinical knee OA is a significant public health concern globally [18] and military members may be more at risk and face a greater burden [14, 15, 21].

Musculoskeletal conditions, including chronic conditions such as knee OA, have been recognized as detrimentally impacting the health of military members [14, 21, 143], being a barrier to recruiting and retention [21, 22, 144], while significantly contributing to military health care costs [1, 4, 13, 39]. Clinical knee OA may result in career progress limitations or early, involuntary, release from military service [21, 22]; in Canada this means that the requirements associated with care are transferred to a provincial healthcare system in addition to contributing to federal Veterans' Affairs related costs [32]. Clinical knee OA is associated with negative impacts for military members, organizations, and society.

2.7 Military load carriage

2.7.1 Injury and knee OA

The requirement of military members to carry heavy loads, both during training and as an operational or occupational requirement, is a risk factor for both injury and development of knee OA [7, 24, 33, 37, 38, 41, 56]. There are a number of factors that can contribute to injury during load carriage including increased physiological demands [42, 43, 90] and altered gait mechanics [40, 45] placing increased loading on lower extremity joints and musculature [7, 41]. Sex differences, including variances in gait biomechanics, while completing load carriage tasks may contribute to differences in injury rates and knee OA development in female and male military members [8, 52, 56]. Most of the studies of military load carriage have focused on performance or injury prevention, which is relevant to post-traumatic knee OA, and understanding associated kinetic or spatial-temporal changes as opposed to the effects of cumulative load on the knee or features of gait shown to be linked to clinical progression of knee OA [40, 52]. There has also been limited study of the relationship between sex and gait biomechanics in military members when the speed of marching is controlled or fixed (i.e., self selection of speed is not permitted) [33, 40, 52].

Bodyweight relative loads, commonly used in military load carriage studies [40], may not accurately represent operational relevant or training loads, and FS walking may result in adaptations to stride length and frequency to maintain pace. Gait adaptations to maintain pace during FS walking have been associated with increased energy cost [51], GRFs [53], knee joint loads [145], and injury rates [54]. Recent reviews indicate that there is a gap in the literature on the effect of load carriage on biomechanical and muscular stresses using standardized loads at a FS for military members [40] and for potential differences in gait adaptations between males and females during military load carriage [52].

Load carriage with operationally relevant load and speed may result in the member walking at an unpreferred FS that often exceeds unloaded SS walking speeds [41, 129, 146-148] and is associated with altered gait biomechanics [33, 145]. With increased load, a common adaptation is to take shorter, more frequent steps [33, 55]. However, during FS marching this strategy may not be viable and military members may be required to overstride to keep pace [33]. Overstriding during forced marching has been observed as a strategy adopted to maintain the required speed; overstriding increases both patellofemoral and medial tibiofemoral joint loads [145] and has been suggested to be a contributor to injury during load carriage tasks [33].

Whether overstriding is related to sex or stature is unclear [33]. However, anthropomorphically matched (height and body mass) males and females have been reported to differ in both knee excursion during early stance phase and medial-lateral trunk displacement during load carriage [57]. The loading pattern described for females may be less cyclic and with increased load, and loading duration, may result in forces that exceed physiologic loading levels thereby leading to increased magnitude, more static, loads being applied to certain parts of the knee's articular cartilage. Static and excessive loads have been identified *in vitro* as contributing to increased degradation of articular cartilage [5, 74, 94, 102, 115]. Advanced understanding of the relationship between marching at FS with standardized loads and features of gait linked to clinical knee OA progression may provide insight into knee OA incidence rates for male and female military members. Given the Government of Canada's aim for the CAF to be comprised of 25% female members by 2026 [149], the lifting of area of service restrictions on female personnel across many militaries [55], and the increased burden of injury and knee OA being borne by female military members [1, 16, 25, 39], this issue can no longer be of secondary consideration.

2.7.2 Historical perspectives, trends, and impact

Loads carried by military members have been shown to have increased over the history of modern conflict [33, 48-50, 90, 150]. Despite recommendations that the load carried should be relative to the size of the soldier [33, 39, 151], the mass of the load carried is typically dictated by the task, environment, training standard, and operational requirement, as opposed to the size, strength, or fitness, of the member [33, 39]. Loads carried by military members may exceed 45% of the their body mass [150]. During the conflict in Afghanistan, dismounted members routinely carried 45 kilograms (kg) [48] and in some combat roles British Army members may carry as much as 70kg [33]. Following World War I, it was recommended that body borne loads not exceed one-third of the member's mass [49] while more recent recommendations suggest carried loads should be limited to 30-45% of body mass [90]; these recommendations have reportedly not been consistently adhered to and have been cited as a source of injury in CAF members [39]. Since the 1980's, which saw a large increase in the body borne loads of military members, there has been a six-fold increase in disability rates, attributable mostly to musculoskeletal injuries, amongst US Army personnel [48].

Military load carriage tasks are often required to be completed under controlled conditions during training (i.e., absolute load and fixed speed), as a measure of operational readiness, or to meet a physical performance standard. For example, the FORCE Combat™ Test is a standardized test of readiness to deploy used by the CAF; this test is commonly performed by members of Canadian Army units. The FORCE Combat™ Test consists of five tasks, one of which, the load carriage task, requires the candidate to complete a five-kilometer march in no less than fifty minutes, but no more than sixty minutes, while carrying a combination of equipment and pack weighing 35kg [146]. The acceptable range of speed for this task is 1.39m/s to 1.67m/s (i.e., 1.52m/s +/- 10%); this range of walking speed is comparable to many studies of military load carriage included in a systematic review excluding rucksack (i.e., loaded) running [40]. The FORCE Combat™ standards apply to all CAF members regardless of individual characteristics including sex. Although training loads and paces for the US military vary by unit, loads and distances ranging from 18 – 45kg and 3.2-26.4 km respectively with a pace of 1.77m/s have been reported [147]. The forced march speed used by the UK Army is 1.33m/s with a load of 25-70kg depending on the soldier's role; a faster paced insertion march (1.53m/s) is utilized by some UK Army units [148]. Australian Army marching speeds have been reported as 1.53m/s and 1.81m/s for administrative and approach marches respectively [41]. These requirements highlight the need to better understand the effects of standardized loads and FS on gait biomechanics. To better understand the relationship of military load carriage to knee OA development and clinical progression, key features of gait known to be associated with knee OA development and clinical progression (i.e., KAM impulse, Peak KAM, Peak KFM, KFM-KEM) should be investigated under these operationally relevant conditions.

Knee OA is prevalent in military populations [3, 14-16, 21]. While it is easy to identify potential risk factors for OA amongst military populations (e.g., injury rates, exposure to excess forces, and carrying heavy loads) the role of key features of gait linked to the onset and progress of both structural and clinical knee OA has not been extensively examined [40]. Gait analysis, particularly during high-risk activities such as load carriage tasks [8, 38, 39, 103], may identify potential changes in key features of gait by military members shown to be linked to knee OA development and clinical progression. Advancing the understanding of these features during load carriage may assist with the development of injury reductions strategies and potentially identify modifiable factors to assist with mitigation of the risk of knee OA in military members.

2.8 Load carriage effects on biomechanics of gait

2.8.1 Spatio-temporal parameters

Studies examining the effects of loaded walking on spatio-temporal gait variables (i.e., cadence, stride rate, stride length, and speed) have yielded inconsistent and even conflicting results [33, 45, 46]. The inconsistencies between studies may have been influenced by the wide range of loads carried and participant samples [40]. Some studies utilized relatively lighter loads (e.g., 9 kg) while others used presumably heavier loads (e.g., 65% of body mass) [33]. Studies included in two recent reviews on load carriage did not use samples drawn exclusively from military populations [45, 46]; experience with load carriage, such as is expected with military personnel, has been cited as a factor potentially influencing gait biomechanics during load carriage [40]. Mixed cohorts of military and civilian participants may account for inconsistencies identified in spatio-temporal measures of gait .

Boffey et al., (2019) noted that fixed-pace studies (i.e., FS) have been used to examine the effect of load on stride rate and stride length [46]; as the load carried was increased, stride length-decreased and step rate increased to maintain the assigned pace [82, 152]. This strategy has been observed most commonly in female participants, however, it may be related to stature as females are typically shorter than males [33] and preferred walking speed is related to stature and leg length [153]. Inconsistencies in findings with respect to adaptation of a shorter step length have been reported and attributed to population differences (e.g., experienced versus inexperienced carriers), a potential non-linear response to increased load, variability in loads carried, and samples over representing one sex [33]. A scoping review, published in 2022, indicated that the quantity of studies including between sex comparisons of load carriage by military members represented a gap in the military load carriage literature [52].

A recent systematic review by Walsh et al., (2021) [40] examined the effect of load carriage on spatio-temporal gait variables, lower extremity joint kinematics and kinetics, GRF, plantar pressures and lower extremity EMG. Unlike other reviews of load carriage and gait mechanics [45, 46], this review included studies with exclusively military members. The differences in experience with, and exposure to, load carriage and carrying of cumbersome equipment, use of military specific load carriage systems, and typically increased fitness levels of military members compared to the general civilian population, may limit the generalizability of the findings from previous reviews to military populations [40]. Liew et al., (2016) highlighted differences in gait mechanics between novice and skilled load carriers (e.g., military members) [45]. Skilled load carriers utilized lower magnitude braking GRF in response to load carriage [45]. Increased GRF with load carriage may contribute to injury of articular structures [41] and stress fractures [154]. Differences in GRF magnitudes between skilled (e.g., trained, experienced, military members) and novice load carriers may explain the high rates of injury during basic and formative military training, particularly for female military members [25, 155].

There were limitations to the review conducted by Walsh et al (2021). Firstly, the reviewed studies showed a large variability in the measures used, measures reported, measurement techniques, as well as loading and assessment protocols [40]. Secondly, only qualitative synthesis was performed due to the heterogeneity of the 20 studies included [40]. Thirdly, of the 20 studies reviewed, only one included female military members (n = 18) and no sex comparisons were performed [40]. Females have been noted to be under-represented in the literature related to load carriage by military members [33, 40, 52, 155] and in military related health research [156]. While there appears to be differences in spatio-temporal parameters between mixed civil-military cohorts [52] and military exclusive cohorts [40], the wide variety of protocols and load conditions, and the limited number of female subjects in the military only studies limit the generalizability and comparability of the findings. There are disparities in both the military load carriage literature and military musculoskeletal health literature concerning the effects of load carriage on the gait biomechanics of female military members compared to males [52].

2.8.2 Kinematics

Joint kinematics examine the relative motion between consecutive body segments. Examining alterations in kinematics under load may provide insight into both energy expenditure during load carriage tasks and possible injury mechanisms [4]. It is likely that movement pattern adaptations during load carriage tasks are required to both maintain an upright walking position and minimize the energy expended during the task [33]. A systematic review and preliminary meta-analysis found an association between increased load and increased range of motion in the sagittal plane at both the hip and ankle whereas no change in sagittal range of motion was reported at the knee or trunk [45]. These findings are inconsistent with other studies that have reported changes in knee ROM with increased load - both overall knee sagittal plane ROM and increased knee flexion with increased load have been observed elsewhere [84, 157]. Differences in knee ROM has been reported between height and body mass matched males and females during load carriage with males displaying increased knee ROM [57]. Other inconsistencies in the literature with respect to ankle kinematics have also been reported [33].

Increased forward trunk lean has consistently been observed as posterior load is increased in military load carriage studies [40]. Only one study included in the systematic review of military load carriage by Walsh et al. [40] did not identify an increased forward trunk lean during military load carriage. That study included only male participants with specific anthropometric and fitness values, carriage of a weapon system (in front of the body), and a backpack load (posteriorly) and was conducted on a treadmill at 1.34m/s. The carriage of the weapon system may have offset the requirement to adopt a forward trunk lean by creating less asymmetry between anterior and posterior loads [55]. Forward trunk lean leads to increased hip flexion in attempt to maintain the load over the base of support [33], alters spinal [81] and lower extremity kinematics, spinal and pelvic muscle activation patterns [48] and may alter the magnitude of knee joint moments (e.g., reduced KFM) and counterbalance posterior loads. It is unclear if male and female military members differ with respect to forward trunk lean or if this adaptation is body mass related [80].

2.8.3 *Kinetics*

Increased KAM and KFM metric percentages with increased load mass carried have been identified [37, 72, 80]. Krajewski (2020) [7], found that in recruit age females forced marching at a speed of 10% greater than their gait transitional velocity resulted in increased percentage of the knee joint total moment comprised by KAM and KFM magnitudes using relative loads of 25% and 45% of the participant's body weight. It was reported that the percentage KAM values were like those of individuals already suffering with knee OA [7]. This comparison was made to individuals with structural OA changes (Kellgren-Lawrence scores > 1) but whose clinical status was unclear.

In a related study using the same sample, it was observed that with loads of greater than 25% body mass, that recruit aged females adopted a stiff gait pattern similar to those with knee OA or following ACL injury [136]. These studies included civilian females with no military experience – the generalizability to military members is limited given the identification of military experience influencing biomechanical adaptations during load carriage tasks [46]. Furthermore, the participants were also fitted with new combat boots for this study; it was unclear if the time to break in and familiarize with the military style combat boots was sufficient [7, 136] as military footwear may influence postural stability and muscle recruitment [158] while lower extremity kinematics and kinetics may be influenced by boot stiffness [159]. Additionally, all carried loads and speeds for this study were relative; participants may have all completed the forced march at different speeds and with different mass loads. However, the finding that females exposed to fixed speed loaded conditions demonstrated gait characteristics consistent with ACL injured individuals and those with knee OA is of interest from a military health perspective with respect to both acute injury and the development of chronic conditions (e.g., knee OA).

A study of 12 male officer-cadets (i.e., untrained officers) found increased Peak KFM with the addition of 15 and 30% body weight loads prior to completion of a FS, 4-kilometer, march. Higher KFM has been associated with increased knee joint loading [128] and this increase was attributed to adaptation at the knee to attenuate shock or reduce load elsewhere [44]. A treadmill was subsequently used to maintain a marching FS of 1.67m/s for 40 minutes. Following the march, re-testing found that knee mechanics were not sustained as Peak KFM was lower than pre-march. This post FS, loaded, walking change was attributable to excessive KE muscle fatigue. KE muscle fatigue has been associated with changes to the knee joint loading environment, increased KAM [85] and increased risk for injuries associated with higher risk for knee OA development [86, 87].

There are few studies that directly compare gait mechanics under loaded conditions between males and females, particularly in military populations [33, 52, 146], and a lack of available female military members to conduct sex comparisons has been cited as a barrier to including both sexes in military load carriage studies [41]. In one study comparing gait biomechanical adaptations in a mixed military – civilian cohort of males and females using absolute loads (22kg) and fixed walking pace, it was observed that females and males adopted different strategies in the frontal plane at the hip and at the sagittal plane at the knee [80]. Females demonstrated decreased peak hip abduction moments (normalized to total mass) and reduced sagittal plane knee excursion during early stance. Males employed increased frontal plane hip excursion and peak hip adduction angle in response to various loaded walking conditions. The authors suggest that the differences in gait adaptations between males and females in response to loading walking may be a source of differences in injury prevalence [80]. Reduced knee excursion may be linked to features of gait also linked to clinical knee OA [121]; there is a requirement to identify the effect of operationally relevant, FS, load carriage tasks on key features of gait (i.e., KAM, KFM, KFM-KEM) related to the clinical progression of knee OA and for findings and to compare the findings for male and female military members .

2.9 Strength, aerobic fitness, and stature

Differences in strength and aerobic fitness have been observed between sexes [55] including in CAF members [90]; these differences have been cited as possible factors relating to the discrepancy in injury rates between male and female military personnel [1, 155]. When differences in aerobic fitness are accounted for in military members, the disparity in injury rates between males and females is reduced considerably, with the notable exception being the difference in stress fractures remaining higher in females [155]. This does not fully account for the increased incidence of knee OA in female military members. Bone stress reactions are more common in female military members [25, 155]; such reactions in subchondral bone may be a possible mechanism [95, 160] for OA development.

2.10 Physiological demands

Loads borne by military members have increased; heavier loads necessitate increased muscle [84] and physiological effort to carry them [55] and FS load carriage tasks require increased energy expenditure compared to SS speed tasks [46, 51]. Fatigue of the KE muscles has been associated with changes in gait biomechanics [85], including increased KAM, and these changes were similar to those reported for the gait of individuals with moderate knee OA [161]. Alterations to the knee joint loading environment due to KE fatigue may increase risk for OA by increasing knee joint loading, shifting the path of loading to less well conditioned cartilage [69, 101, 162], and increasing the risk for knee injuries that are high risk for knee OA [86, 87].

Sex differences in physiological demands during unloaded [163-165] and loaded walking have been reported [52]. Females in both military and civilian samples have been reported as having lower average strength and aerobic capacity compared to males [79, 166, 167]. Sex comparisons of muscle activity in healthy participants during unloaded walking have reported greater [164, 165] and more complex [163] muscle activity in females than males.

The smaller, on average, stature of females compared to males means that females carrying absolute loads at FS will be carrying relatively greater loads and expending relatively greater energy and effort to do so compared to a male [55]. This concept is supported by reported improved performances in load carriage tasks by taller, heavier, females compared to those that are shorter and lighter [55]. The increased physiological demands experienced by females during load carriage tasks [52] is also reflected by observations that female participants have been found to typically work at a higher percentage of their maximum aerobic capacity than their male counterparts when carrying the same absolute loads at the same intensity [55]. Physiological factors (e.g., stature, strength) and biomechanical factors (e.g., forward trunk lean) have the potential to increase both the effort required during load carriage and the risk of injury [55]; these factors may contribute to the increased incidence of knee OA in military members and the differences in its incidence between male and female members. Strategies to address both relative strength and aerobic capacity in female military members have been suggested as means to reduce the physiological burden associated with load carriage [55].

2.11 Summary

Clinical knee OA is a significant health concern for military organizations due to its detrimental effects on the health, employment, and quality of life, of military members and its associated negative impact on military readiness. Furthermore, clinical knee OA places demands on health care systems and contributes to veterans' health related expenditures. Load carriage tasks are physically demanding, with evidence of adaptations in gait biomechanics, and are associated with risk factors for knee OA including injury and repeated exposure to increased magnitude loads. Load carriage tasks have been investigated with respect to injury prevention, kinematic, and spatial-temporal features, but the effect of cumulative load and features of gait shown to be associated with the progression of clinical knee OA have not been investigated to a similar extent. It is unclear whether males and females adapt gait biomechanics differently in response to fixed speed load carriage tasks. There is a gap in the literature on the effect of load carriage on biomechanical and muscular stresses using operationally relevant load and speed. Additionally, there are gaps in both the military load carriage literature and military musculoskeletal health literature concerning the effects of common military tasks (e.g., load carriage) on female members.

The stresses during standardized load carriage tasks and their relationship to biomechanical and muscle activation factors associated with knee OA development and clinical progression in military members have not been extensively investigated. Examining key features of gait linked to clinical knee OA progression during operationally relevant load carriage tasks in male and female military members could improve the understanding of the effect of these tasks on the knee loading environment and be used to inform injury reduction strategies, rehabilitation protocols, and physical training plans. Additionally, understanding differences in effort during operationally relevant load carriage tasks may inform physical training plans, equipment design, and injury prevention strategies.

Chapter 3. Methods

This chapter provides details for the methods used in this study. First, participant recruitment processes and inclusion criteria are detailed followed by an overview of how the study procedure was conducted. Secondly, data processing and analysis procedures are described. Next is a description of how each outcome was measured including discrete knee moment features related to the development and progression of knee OA, self-reported and quantitative measures of effort, and peak forward trunk angle during the walking conditions. Next, data processing and analysis is described. Lastly, the statistical analysis used for Objective 1, Sub-objective 1a, 1b, and Objective 2, Sub-objective 2a, 2b are described. The study protocol was approved by the Dalhousie University Research Ethics Board (file #2022-6378).

3.1 Participant recruitment

Current, healthy, CAF members (regular or reserve force) between the ages of 20-50 years, that had completed Basic Military Qualification (BMQ) or Basic Military Officer Qualification (BMOQ), were recruited from units located in Halifax, Nova Scotia. An upper age limit of 50 years was selected as decline in the muscle strength of both the KE and KF muscles has been reported in males and females after age 50 [168] and age related gait changes have been identified in adults beginning in the sixth decade of life [169]. Decline in KE and KF muscle strength and age-related gait changes after age 50 may be independently associated with alterations in knee joint moments or dynamic changes to gait. Experience has been identified as a factor influencing load carriage biomechanics [40, 46]; establishing the lower end of the age range at 20 years was selected to increase the likelihood of recruiting those with experience in military load carriage. To ensure safety, and reduce potential confounding variables, participants were included if they did not have neurological, cardiovascular, or musculoskeletal conditions that could be exacerbated or cause a loss of balance, while walking with operationally relevant load and speed and did not have military medical employment limitations precluding lifting/carrying of 35kg or restricting moderate/vigorous exercise (Appendix A).

For Objective 1 and Sub-objectives 1a, 1b, results of an *a priori* power analysis [170] indicated that the required sample size to achieve 80% power for detecting a large effect [171] with α set at 0.05 was n=12 for a one-way repeated measures ANOVA. As Objective 2 and Sub-objectives 2a and 2b of this study were exploratory, and aimed at determining the feasibility/requirement of conducting the protocol with a larger sample, a target sample of 30 participants (15 male and 15 female) was determined as the goal [172]. The aim of selecting a sample size of 30 participants was to provide sufficient quantitative data to perform sample size estimates for sex analyses, identify trends in data to be explored in testable hypotheses, and address the necessity for, as well as feasibility of, the protocol including recruitment, conditions, and variables to include in future studies.

Participants were recruited using e-mails and posters (Appendix B) displayed at CFB Halifax and at other Halifax area CAF units assessed as most likely to employ members with experience with military load carriage (e.g., Canadian Army). Members that were recruited via email advertisements and posters contacted the research coordinator (AH) directly by email. The research coordinator then arranged a phone interview with the participant. Using a standardized script, all participants were provided an overview of the study to determine their interest and obtain verbal consent to answer a general health screening questionnaire (Appendix C) that served to determine the participant's eligibility to participate. If they met the inclusion criteria, participants were asked if they were willing to attend a data collection session at the Joint Action Research Laboratory (JAR Lab) at Dalhousie University. The research coordinator scheduled the time and date, provided the participant with a standardized email (Appendix D) including study information, confirmation of date/time, directions to the JAR Lab, arrival instructions, and a letter to the participant's Chain of Command (CoC) (Appendix E) to obtain permission for the participant to attend the study during standard working hours. All participants obtained permission from their CoC to be eligible for this study; this ensured that they were fully compensated for their time. All participants reviewed and signed the informed consent form before participating. Participants that chose to withdraw prior to data collection were removed from the study and all corresponding documentation was destroyed. The age, sex, and reason for participant exclusion or withdrawal was recorded. Figure 3.1 outlines the recruitment and informed consent processes.

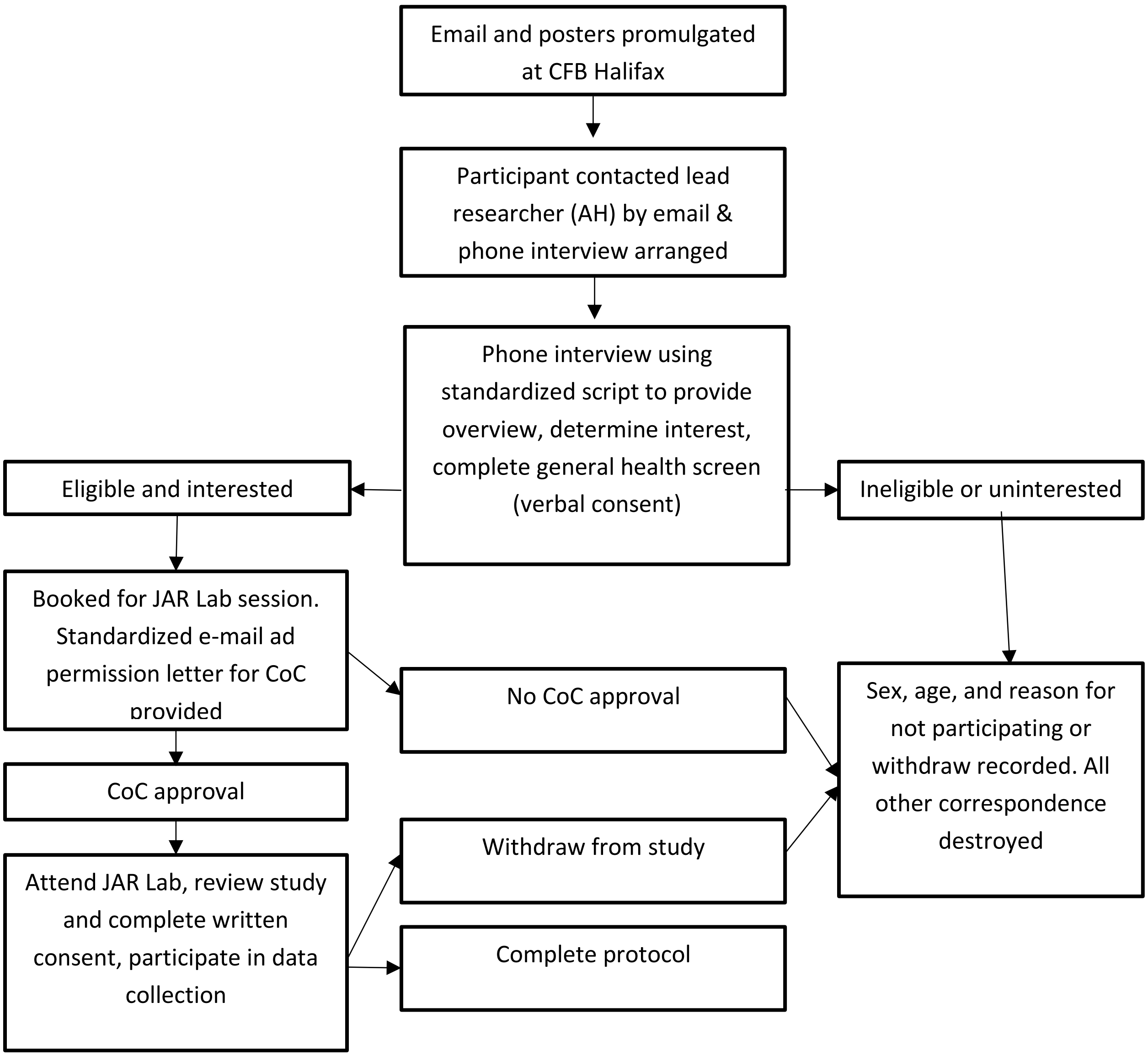


Figure 3.1: ***Participant recruitment and informed consent process.***

3.2 Procedure

This section describes participant preparation, gait analysis, progression through the walking conditions, and strength testing. An overview of participant progression through the study's procedure is presented in Figure 3.6 at the end of section 3.2

3.2.1 Participant preparation

The JAR Lab, located on the 3rd Floor of the Forrest Building, was used for all data collection procedures. On the day of testing, participants were asked to provide the CoC permission letter (Appendix E) then review and sign the Dalhousie Health Research Ethics Board approved informed consent form (Appendix F). All participants received a brief orientation to the JAR Lab and a review of the equipment used during data collection. Participants had the opportunity to ask questions about the study procedures/objectives. Participants wore a t-shirt, tight-fitting shorts, socks, and their personal military combat boots during testing procedures. Demographic information including sex, age (years), height (m), mass (kg), and military service (years) were collected and recorded. Body mass index (BMI) (kg/m^2) was calculated. Measurements of waist, hip, and bilateral thigh and shank circumference were collected and recorded; these measurements were included in the biomechanical analysis models to calculate knee joint moments.

Although not pertinent to address the objectives for this thesis, participants were prepared for surface EMG collection using standard sensor locations [173] for seven lower limb muscle sites (three KE, two KF, and two plantar flexor (PF) muscles) bilaterally using protocols that have been shown to be reliable [174, 175]. EMG was collected during the walking conditions and strength testing but will not be described or discussed in this thesis.

The GaitRITE™ pressure sensitive walkway, valid and reliable for measuring gait speed in adults [176, 177], was used to determine the participant's SS walking speed. Participants walked back and forth across the system at least ten times. Five trials were randomly recorded and averaged (rounded to the nearest hundredth of a second) to determine the SS walking speed of each participant. Next, retro-reflective markers were attached to the participant over key anatomical landmarks (Table 3.1) using Velcro straps and adhesive tape based on standard published protocols [178-180] with high day to day reliability for knee joint frontal and sagittal plane moments [178].

Table 3.1: ***Retro-reflective marker type and location***

Type of Retro-reflective Marker	Location
Set of Four (headband)	Head (forehead, anterolateral and posterolateral bilaterally)
Set of Four (rigid)	Pelvis (sacrum) Bilateral lateral femurs (mid-thigh) Bilateral lateral tibias (mid-lower leg)
Set of Three (rigid)	Foot (lateral midfoot)
Single	Lateral humeral epicondyle Ulnar styloid process Bilateral lateral shoulders Seventh Cervical Vertebrae Bilateral greater trochanters Bilateral medial femoral epicondyles Bilateral lateral femoral epicondyles Bilateral lateral tibial condyles Bilateral medial tibial condyles Bilateral lateral malleolus Bilateral medial malleolus Bilateral head of first metatarsal Bilateral head of second metatarsal Bilateral head of fifth metatarsal Bilateral posterior calcaneus

Prior to gait analysis, a kinematic model calibration was completed with the participant standing upright on the treadmill deck. This included a standing calibration trial, a virtual sternum location trial, two virtual anterior superior iliac spine location trials, and bilateral hip joint center of rotation trials. The retro-reflective markers over the greater trochanters, medial femoral epicondyles, medial and lateral tibial condyles, medial malleoli, and bilateral first and fifth metatarsal, were removed prior to treadmill walking. Participants wore an upper body safety harness while walking on the treadmill that permitted unrestricted lower body movement while ensuring safety in the event of a loss of balance. Figure 3.2 illustrates the location of the retro-reflective markers and the safety harness during treadmill walking.

Figure 3.2: ***Retro-reflective marker locations during treadmill walking for key lower extremity and pelvic and spinal markers. Combat boots not pictured. (From Rutherford et al. [181] with permission Sage, 2023)***

3.2.2 Gait analysis

The retro-reflective spheres were tracked using eight Qualisys® OQUS 500 motion analysis cameras at 100 Hz. (Gothenburg, Sweden). Three-dimensional ground reaction forces (GRF) were sampled at 2000Hz from bilateral force plates in the dual-belt instrumented treadmill (R-Mill, Motek Forcelink, The Netherlands) simultaneously with surface EMG signals at the same rate and synchronized with the motion data. All captured analog signals were analog to digital converted using an analog to digital converted (16bit, +/-5V), and synchronized using Qualisys® Track Manager V2.10 software.

3.2.3 Walking conditions – speed and load

As described above, the speed for SSU and SSL for each participant was determined from participants' GaitRITE™ trials. The speed for FSL was initially set at 1.52m/s for all participants. Participants were required to maintain a speed +/-10% of the initial setting (i.e., 1.37m/s – 1.67m/s) during FSL. The speed for FSL was determined to be operationally relevant based on its association with the FORCE COMBAT™ test [146].

No additional load was carried for SSU. The standardized 35kg load, used for both loaded conditions (i.e., SSL and FSL), was distributed between a loaded weight vest (XM Fitness Commercial Weighted Adjustable Vest), and a CAF standard issue small pack (NATO Stock Number 8465-20-000-2774) (Figure 3.3). The pack's straps were adjusted to fit comfortably around the participant's shoulders, chest, and waist. The weighted vest accounted for 18.2kg and the pack load for 16.8kg in total [146]. The mass and distribution of the load for the loaded conditions was determined to be operationally relevant based on their association with the FORCE COMBAT™ test [146].

Figure 3.3: ***Rear, side, and frontal view with standardized load of 35kg***

3.2.4 Walking conditions – order of testing and data collections

Following preparation and calibration, all participants completed a 5-minute dual belted treadmill walking familiarization protocol [179, 182] at their SS walking speed previously determined using GaitRITE™. Participants were requested to remain in the centre of the treadmill and to walk with each foot on its respective side's belt. The order of the walking conditions was the same for all participants; 1) SSU, 2) SSL, and 3) FSL.

Participant progression through the walking conditions is visually presented in Figure 3.4. Following the 5-minute treadmill familiarization [179, 182], participants completed a sixth minute of continuous walking during which the 20s collection of data for SSU was completed. Fitting of the participant with the standardized load (i.e., weighted vest and small pack) occurred following SSU. Fitting required a minimum of 5-minutes to complete and occurred on the treadmill deck due to the position of the safety harness attachment (i.e., posteriorly between the participant's thorax and the small pack above the weighted vest). Both loaded conditions (i.e., SSL and FSL) were 5-minutes in duration. To minimize the effects of learning and fatigue, the 20s data collection from the fifth minute of SSL and FSL were used for the analysis of each respective condition. Participants were blinded to collection intervals for all conditions. A 10-minute rest period, during which the load was removed, the safety harness detached, and the participant dismounted the treadmill, was provided following completion of the SSL walking condition (i.e., prior to starting the FSL condition). Participants had the option to sit, stand, drink water, or move about the Jar Lab during the rest period. The EMG electrodes, leads, and retroreflective markers remained in situ during the rest period. The speed for the FS condition was initially set at 1.52m/s. Participants were advised, prior to starting the condition, of the minimum and maximum speeds (i.e., 1.52m/s +/- 10%) and their relation to the Combat FORCE™ Test. Participants were asked if they wanted to increase or decrease the speed during the first minute of the FSL condition and were requested to maintain the same speed throughout the FS condition.

Participants completed a BORG Scale (RPE) (Sub-objective 1a, 2a) at the end of each walking condition (Appendix G). The BORG 15-point scale includes ratings of 6, no exertion, to 20, maximal exertion and is strongly, positively, correlated to physiological measures of exertion (e.g., heart rate) [92]. After completing FSL, the load and all retro-reflective skin surface markers were removed.

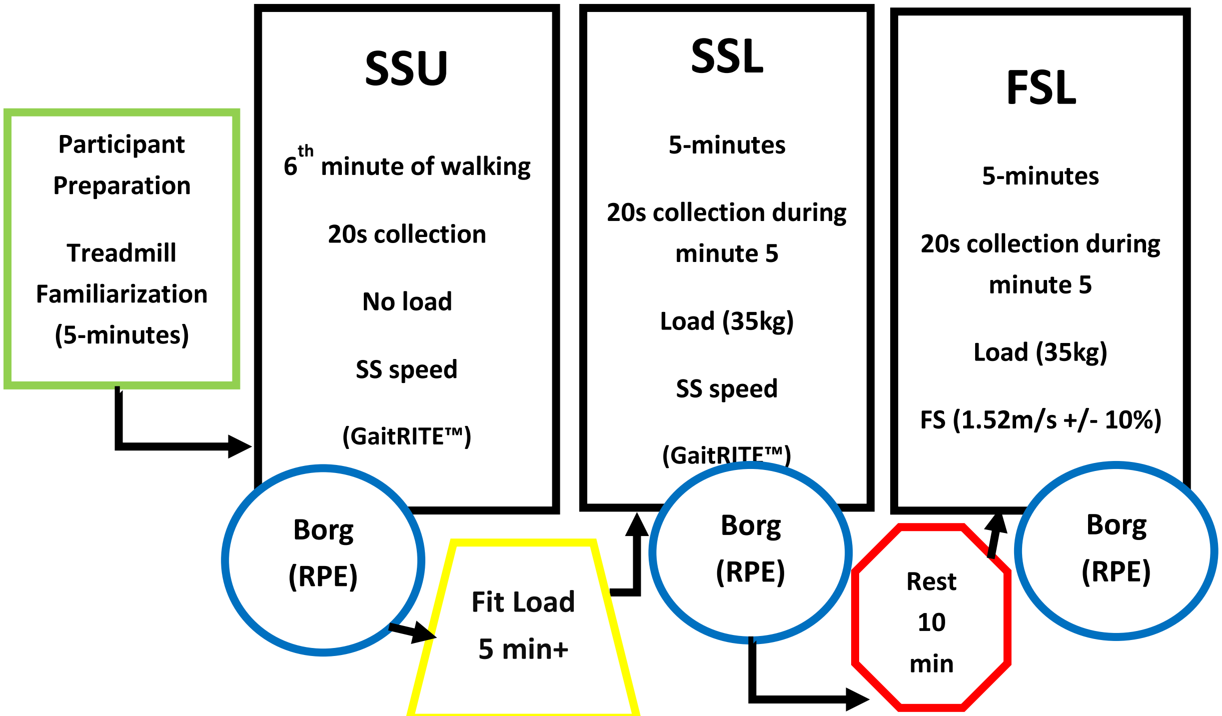


Figure 3.4: ***Participant progression through walking conditions***

3.2.5 Muscle strength testing

Maximum voluntary isometric contractions (MVIC) were collected for the KE and KF muscles using a Humac Norm Isokinetic Dynamometer (Computer Sports Medicine Inc., USA) and published, standardized, procedures [179]. Torque data collected using isokinetic dynamometry, the gold standard for muscle strength testing [183], was recorded for all exercises and the order of testing was the same for all participants (i.e., right KE, right KF, left KE, left KF). Surface EMG were simultaneously recorded but not reported in this study. The normalized to body mass torques were calculated as a relative strength measure for each muscle group (Nm/kg).

Figure 3.5: ***Positioning of participant and set-up of isokinetic dynamometer for MVIC testing of KE and KF muscles***

As mentioned above, an overview of participant progression through the study's procedure is presented in Figure 3.6.

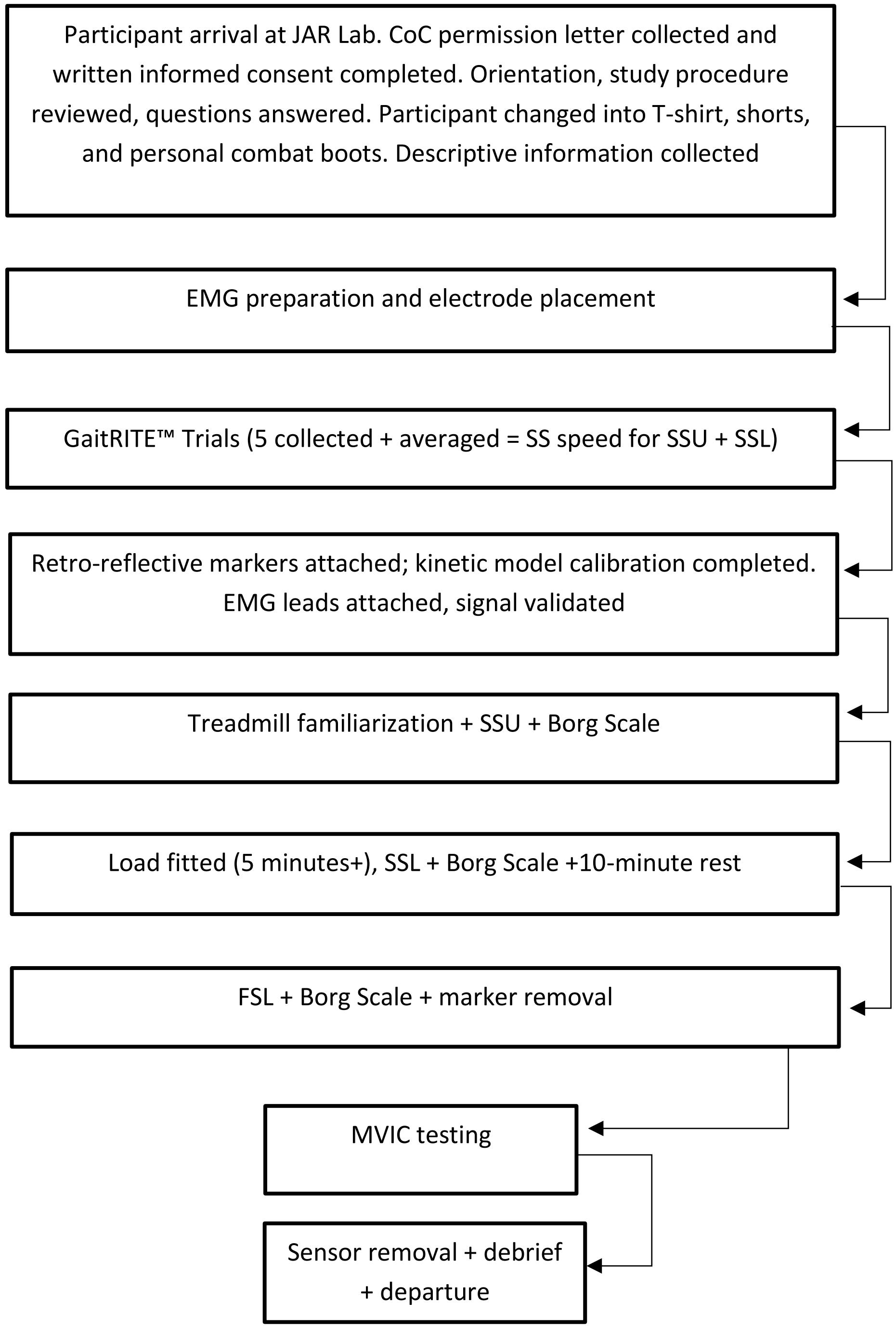


Figure 3.6: ***Outline of participant progression through the testing procedure***

3.3 Data Processing and analysis

3.3.1 Kinematics

A three dimensional Cartesian coordinate system was used to calculate trunk and knee joint angles [184]. Kinematic data was low pass filtered (Butterworth 4th order, Fc 6Hz recursive) and processed using custom software written in MatLab™ 2016b (The Mathworks Inc., USA). Technical and local anatomical bone embedded pelvis, thigh, shank, and foot coordinate systems were derived from physical markers and virtual points. A vector between the medial and lateral femoral condyles was used to define the axis for flexion/extension. The orthonormal shank coordinate systems were defined with a fixed medial-lateral axis while the anterior-posterior and proximal-distal axes were derived from cross-products [185]. Positive motion is described as flexion, adduction, and medial rotation about the knee joint with the distal segment's motion being described [186]. The reliability for knee motion outcomes in the sagittal and frontal planes using the methods described above are excellent (ICC= 0.85-0.94) [178]. To compare findings to other military load carriage studies, peak trunk angle was measured and reported during the loaded walking conditions [40, 157, 187].

3.3.2 Gait waveform analysis

While joint motion and moment waveforms are often time normalized to represent 100% of the gait cycle (heel strike to heel strike ipsilaterally), the joint motion waveforms were time normalized to represent 100% of stance phase (heel strike to toe off ipsilaterally). Heel strike was identified when a GRF exceeded a 30 Newton(N) threshold while toe off was identified when GRFs passed below 30N; these events were confirmed by kinematic association with heel strike or toe off [179, 188]. Key events for stance phase and the gait cycle are illustrated in Figure 3.7. Ensemble averaged frontal and sagittal plane knee moment waveforms were calculated from each condition's respective 20s data collection.

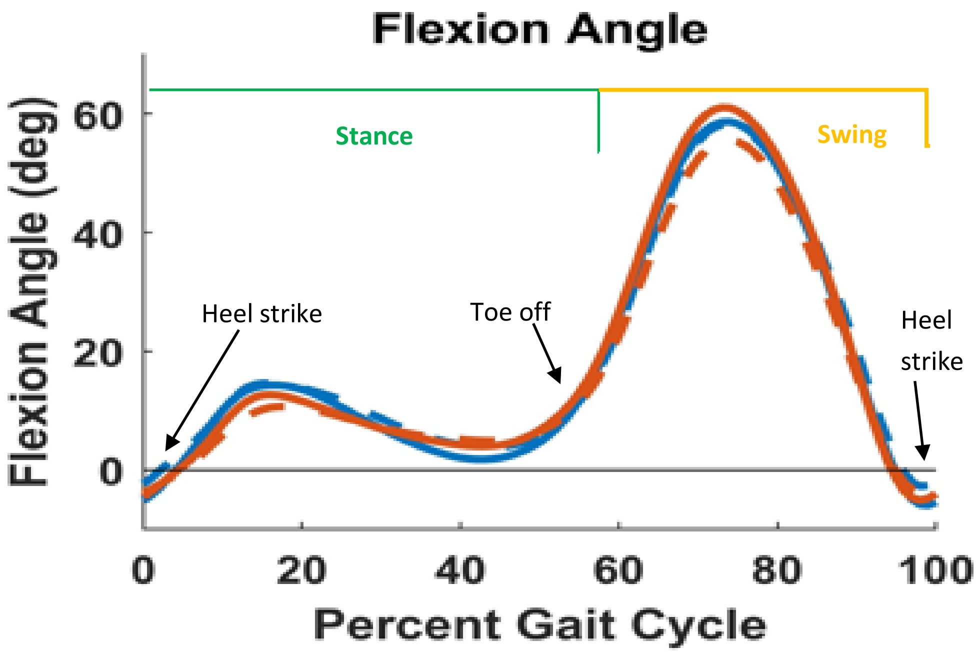


Figure 3.7: Knee motion waveform for 100% of gait cycle – key events and phases labelled. Data from Dynamics of Human Movement Laboratory (adapted with permission from Hubley-Kozey)

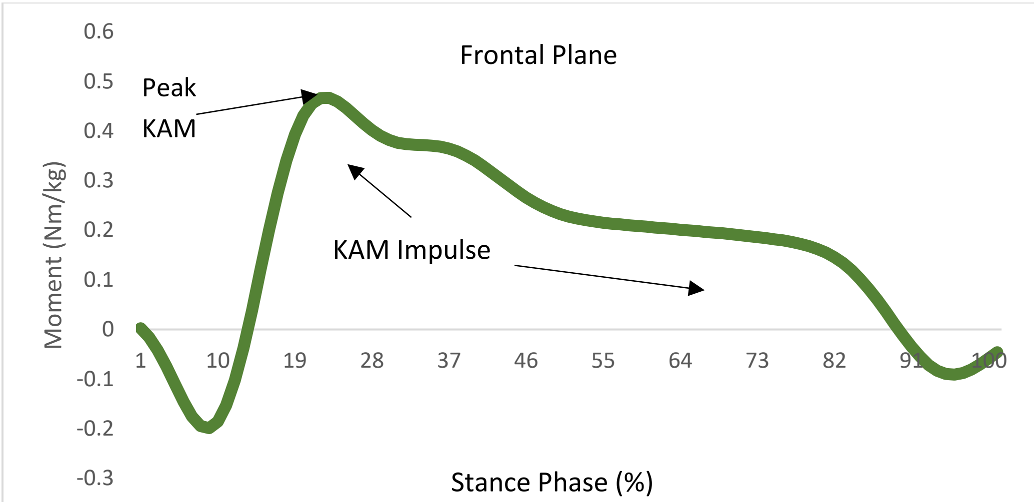
3.3.3 Kinetics

Three dimensional GRF and moments were generated from the six sensors imbedded in the dual-belt treadmill. The treadmill's force plate system was aligned with the global coordinate system used to capture motion as previously described. Custom software written in MatLab 2016b (The Mathworks Inc., USA) was used to process the GRF and moment data after it had been lowpass filtered (Butterworth 4^{th} order, Fc: 30Hz recursive). Using an inverse dynamic model, external joint moments were calculated from GRFs, linear/angular accelerations, and normalized to participant body mass [189-191]. Three dimensional moments and joint forces were calculated and projected on to the joint coordinate system. A lowpass filter (Butterworth 4th order, Fc: 10Hz recursive) was applied to the moment data and the moments normalized to body mass (Nm/kg). The reliability for knee moment outcomes in the sagittal and frontal planes using these methods are high-excellent [178, 192].

The primary outcomes, including four discrete knee joint moment features related to knee OA clinical progression, were calculated, and compared between walking conditions (Objective 1) and sexes (Objective 2). These features included KAM impulse [64], first Peak KAM [60], Peak KFM [66], and the difference between the first Peak KFM and late phase knee extension moment (KEM) (KFM-KEM) [63]. Descriptions of the calculation for each feature is presented in Table 3.2 and visual representations are displayed in Figure 3.8. All knee joint moments reported for this study were calculated using custom written software (Matlab 2016b, Mathworks Inc., USA).

Table 3.2: ***Description of discrete knee joint moment features***

Knee joint moment feature	Calculation
KAM Impulse (Nms/kg)	Integral of KAM over stance phase in seconds
Peak KAM (Nm/kg)	Highest magnitude KAM first 40% stance phase
Peak KFM (Nm/kg)	Highest magnitude KFM first half of stance
KFM-KEM (Nm/kg)	Difference between Peak KFM (early stance) and Peak KEM (late stance)



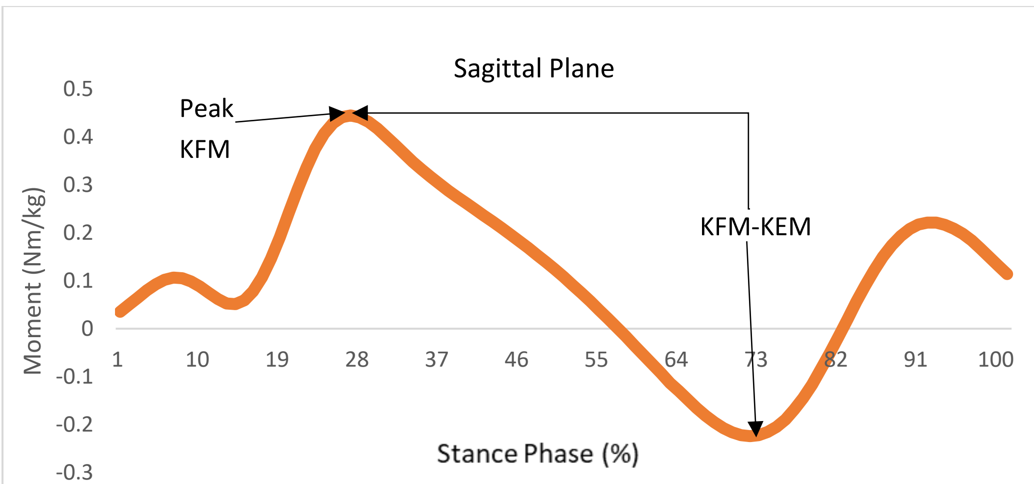


Figure 3.8: ***Visual representation of discrete knee joint moment features.*** ***Peak KAM and KAM Impulse (above), Peak KFM and KFM-KEM (below)***

3.3.4 Muscle strength

The MVIC data collected using the Humac Norm Isokinetic Dynamometer was processed using software written in MatLab 2016b (The Mathworks Inc., USA). The maximum amplitude torque from all trials per muscle group was calculated using a 500ms moving average window. These values were normalized to participant body mass to calculate a relative strength measure for each muscle group (Nm/kg) and an average value was calculated for male and female participants. The MCU for the KE muscle was calculated as the ratio of Peak KFM to KE muscle strength normalized to body mass (Sub-objectives 1a, 2a) [93]. All secondary outcomes are described in Table 3.3.

3.3.5 Peak sagittal plane trunk angle

The peak sagittal plane trunk angle (i.e., peak trunk angle) was calculated from a vector created from the pelvis to the C7 spinous process with respect to the vertical using the cartesian coordinate system (Figure 3.9).

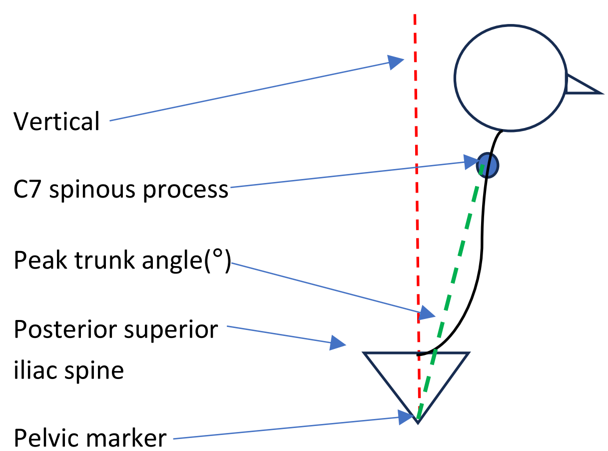


Figure 3.9: ***Peak trunk angle schematic***

The measured peak trunk angles were corrected to the mean trunk angle calculated during the sternum virtual point trial (up right standing). Positive angles indicate flexion and negative angles indicate extension relative to standing. Peak trunk angle during each walking condition was identified as the maximum trunk angle measured relative to the mean trunk angle calculated during the sternum virtual point trial (Sub-objective 1b, 2b) (Table 3.3).

Table 3.3: Description of secondary outcomes

Secondary outcome	Sub-objectives	Calculation
Rate of perceived exertion (ordinal)	1a, 2a	15-point BORG scale (6-20)
Muscle capacity utilization (unitless)	1a, 2a	Peak KFM/KE body mass normalized strength
Peak trunk angle (degrees)	1b, 2b	Maximum trunk angle measured relative to the mean trunk angle calculated during the sternum virtual point trial

3.4 Statistical analysis

Motion, ground reaction forces, and muscle strength data were collected from the left and right leg of each participant and the data from one leg was randomly selected for analysis (13 left, 11 right). All statistical analyses were conducted in SPSS 27.0.1.0 (IBM Corp., Chicago, IL). Descriptive statistics including means (M), standard deviation (SD), and 95% confidence intervals (CI) for demographic characteristics including age (years), height (m), mass (kg), BMI (kg/m^2), and military service (years) were calculated for the total sample (Objective 1) and each sex separately (Objective 2). The same descriptive statistics were calculated for non-normalized (Nm) and body mass normalized (Nm/kg) KE and KF muscle strength, SS and FS overground walking speed (m/s), and for stance phase duration during each walking condition for the total sample and each sex separately. Differences between male and female participants for the demographic characteristics (Objective 2); measures of strength, walking speed, and stance phase duration were tested using Independent T-tests.

3.4.1 Hypothesis testing - Objective 1 and Sub-objectives 1a, 1b

One-way repeated analysis of variance (ANOVA) tested for differences in primary measures, stance phase duration and secondary measures among conditions. For Objective 1, one-way repeated measures ANOVA models tested the null hypothesis that there would be no differences among walking conditions (i.e., SSU, SSL, FSL) in knee joint moment features related to the clinical progression of knee OA (i.e., KAM impulse, Peak KAM, Peak KFM, and KFM-KEM) for the total sample. For Sub-objectives 1a and 1b, the ANOVA models tested the null hypotheses that there was no difference in RPE, MCU, and peak trunk angle, among walking conditions for the total sample.

Prior to running the ANOVAs, assumptions of normality were examined using three criteria, 1) Shapiro-Wilk, 2) graphical analysis of histograms, and 3) values for skewness and kurtosis. The assumption of sphericity was tested using Mauchly's test and a Greenhouse-Geisser correction was applied where this assumption was violated.

Post hoc analyses of significant effects were calculated using the Bonferroni correction. Trend lines were included in figures to provide visualization of the data and to show the significant pairwise differences.

Variables with non-normal distributions, based on violation of two or more criteria, were first transformed (Lg10) to determine if transformation resulted in a normal distribution. Normally distributed transformed data was analyzed, and results were compared to those of the original data. If the results were the same (i.e., same significant findings for omnibus test and pairwise comparisons) then the original data's analysis was presented and analysis of the transformed data was reported in an appendix. A sensitivity analysis was conducted with outlier data removed for variables where transformation did not result in a normal distribution. As before, if no differences were found when comparing the results for the original and outlier removed data then the analysis of the original data was presented, and the results of the sensitivity analysis were placed in an appendix.

3.4.2 Confidence interval for mean differences - Objective 2 and Sub-objectives 2a, 2b

The total sample was divided into two groups based on sex to explore potential between and within sex differences among walking conditions for the primary and secondary outcomes. For Objective 2, mean differences and 95% CI of the mean differences [193] were used to compare between male and females during each walking condition and to compare within each sex during walking conditions. Significant differences were determined based on CI's consisting of only positive or negative values (i.e., CI did not include 0). The analysis for Objective 2 was selected due to the group sizes (14 males, 10 females), the exploratory nature of the objective, and concerns regarding assumptions of normality and sphericity.

To address Sub-objectives 2a and 2b, mean differences and 95% CI [193] of the differences were used to compare the RPE, MCU, and peak trunk angles between males and females. Within group comparisons of RPE, MCU, and peak trunk angle were also completed for each sex during each walking condition using mean differences and 95% CI.

Trend lines were included in figures to provide visual indication of potential interaction effects for the sex analysis in Objective 2, Sub-objective 2a, and 2b. A post-hoc analysis was conducted to determine achieved statistical power and to estimate sample size requirements for non-significant findings for the male and female comparison.

Chapter 4. Differences in joint moments features among walking conditions

This chapter provides a summary of the results for Objective 1, to determine differences in knee joint moment features related to knee OA development and progression among the three walking conditions (i.e., SSU, SSL and FSL). The results for Sub-objectives 1a, to determine differences in RPE and MCU among walking conditions, and 1b, to determine trunk angle differences among the walking conditions, are also summarized. Discussion of the interpretation and context of the results is found in Section 4.6.

4.1 Participant enrollment, exclusion, and withdrawal

In total, 33 CAF members inquired about the study and 25 participants (76%) were enrolled over 13 weeks (February 1 - May 3, 2023). The overall rate of recruitment was just under 2 participants/week. Figure 4.1 details participant inquiries, enrollment, reason for exclusion, and withdrawal; more details are found in Appendix H. One participant withdrew prior to testing and did not complete the study. There was no missing data, and no data was removed from the analysis. No adverse events were reported.

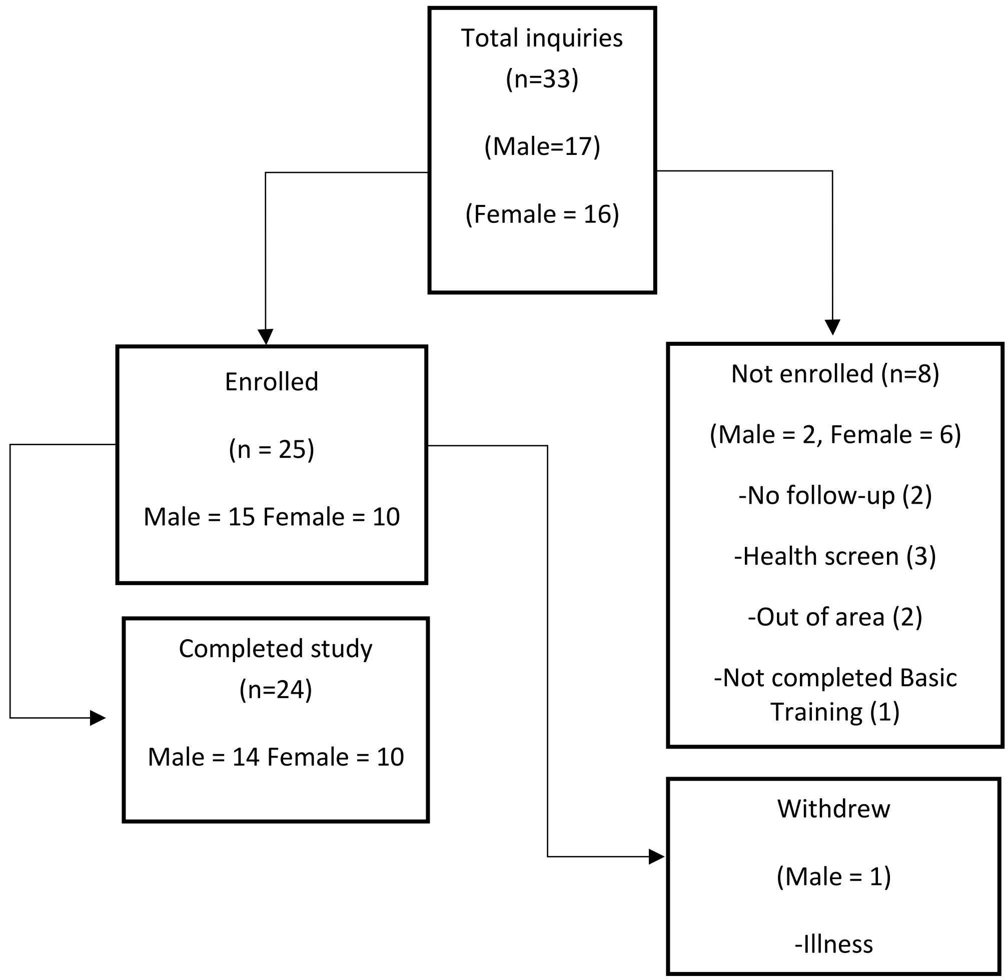


Figure 4.1: ***Participant inquiries, enrollment, exclusion, and withdrawal***

4.2 Demographic characteristics of participants

Descriptive statistics (i.e., mean, standard deviation, and 95% CI) of the participants' demographic characteristics including sex, age (years), height (m), body mass, BMI (kg/m^2), military service (years), are found in Table 4.1.

Table 4.1: ***Participants' demographic characteristics***

Characteristic	Total sample (n=24)	95% CI Lower	Upper
Sex - male/female* (number)	14/10		
Age (years)	29.1 (6.9)	26.2	32.0
Height (m)	1.8 (0.1)	1.7	1.8
Mass (kg)	76.3 (11.8)	71.3	81.3
BMI (kg/m^2)	25.01 (3.5)	23.6	26.6
Military Service (years)	9.0 (6.9)	6.1	11.9

*All data presented as mean (standard deviation) except for sex

Descriptive statistics for absolute KE and KF muscle strength, and body mass normalized KE and KF muscle strength, are found in Table 4.2. The absolute and normalized to body mass KE muscle strength were greater than that of the KF muscles.

Table 4.2: ***Participant muscle strength and body mass normalized muscle strength values for knee extensor and knee flexor muscles***

Muscle Strength Measure Type	Muscle Group	M (SD)	95% CI Lower	Upper
Absolute (Nm)	KE	148.7 (35.6)	133.7	163.8
	KF	97.4 (30.6)	84.5	110.4
Normalized (Nm/kg)	KE	2.0 (0.5)	1.8	2.2
	KF	1.3 (0.4)	1.1	1.5

4.3 Knee joint moment features linked to knee OA clinical progression across walking conditions

Twenty-four CAF members (14 male, 10 female) completed the study protocol. All participants completed the three walking conditions and were able to maintain both SS speed and FS walking on the treadmill for the duration of each condition (i.e., SSU, SSL, and FSL). SS walking speed, from the GaitRITE™ trials (Chapter 3, Section 3.2), was used for both SSU and SSL. Four participants (17%) walked at a FS of greater than 1.52m/s (i.e., the initial speed setting for FSL). Table 4.3 presents the descriptive statistics for participant SS and FS walking for each walking condition.

Table 4.3: ***Participant walking speed descriptive statistics for self-selected and fixed speed walking***

Speed Type	Total Sample (n=24)	95% CI Lower	95% CI Upper
SS	1.30 (0.13)	1.24	1.35
FS	1.55 (0.06)	1.52	1.56

All data presented as mean (standard deviation) except where noted

The descriptive statistics and one-way repeated measures ANOVA results comparing stance phase duration for each walking condition are found in Table 4.4. The stance duration data did not meet the assumption of sphericity, $X^2(2)=0.62$, $p=0.005$; a Greenhouse-Geisser correction was applied and is reported in Table 4.4.

Table 4.4: ***Descriptive statistics and one-way repeated measures ANOVA results for stance phase duration across conditions***

Walking Condition	M (SD)	95% CI Lower	95% CI Upper	p
SSU (s)	0.69 (0.05)	0.67	0.71	***<.001****
SSL (s)	0.70 (0.05)	0.68	0.72	
FSL (s)	0.65 (0.03)	0.64	0.66	

*Greenhouse-Geisser correction applied

A Bonferroni post hoc analysis of the statistically significant difference (p<0.05) (Table 4.5) for stance phase found pairwise differences among all walking conditions. FSL had shorter stance phase duration than the two SS speed conditions, and SSL had a longer duration stance phase than SSU.

Table 4.5: ***Pairwise comparisons for stance phase duration across conditions***

Compared Walking Conditions	M Difference	95% CI of the Difference Lower	95% CI of the Difference Upper	p
SSU-SSL	-0.02	-0.03	-0.01	***<.001***
SSU-FSL	0.04	0.02	0.06	***<.001***
SSL-FSL	0.06	0.04	0.07	***<.001***

The ensemble averaged waveforms (Nm/kg) of the entire sample's frontal plane (KAM) and sagittal plane (KFM) moment data, normalized to 100% of stance phase, for SSU, SSL, and FSL, are presented in Figures 4.2. and 4.3, respectively. The frontal plane waveform for FSL was characterized by a distinct peak in the first half of stance phase whereas the SSU waveform had a less distinct peak and was comparatively flat throughout this portion of stance phase. All three frontal plane waveforms were similar throughout mid-late stance. The sagittal plane ensemble averaged waveform for FSL was characterized by two peaks in the first half of stance phase making it visually distinct from the waveforms for SSU and SSL. The magnitudes of the

KEM in the second half of stance phase for both loaded conditions were greater than the unloaded condition with the greatest KEM magnitude observed for the FSL waveform. The loaded conditions' (i.e., SSL and FSL) waveforms were characterized visually by a greater magnitude late stance KFM.

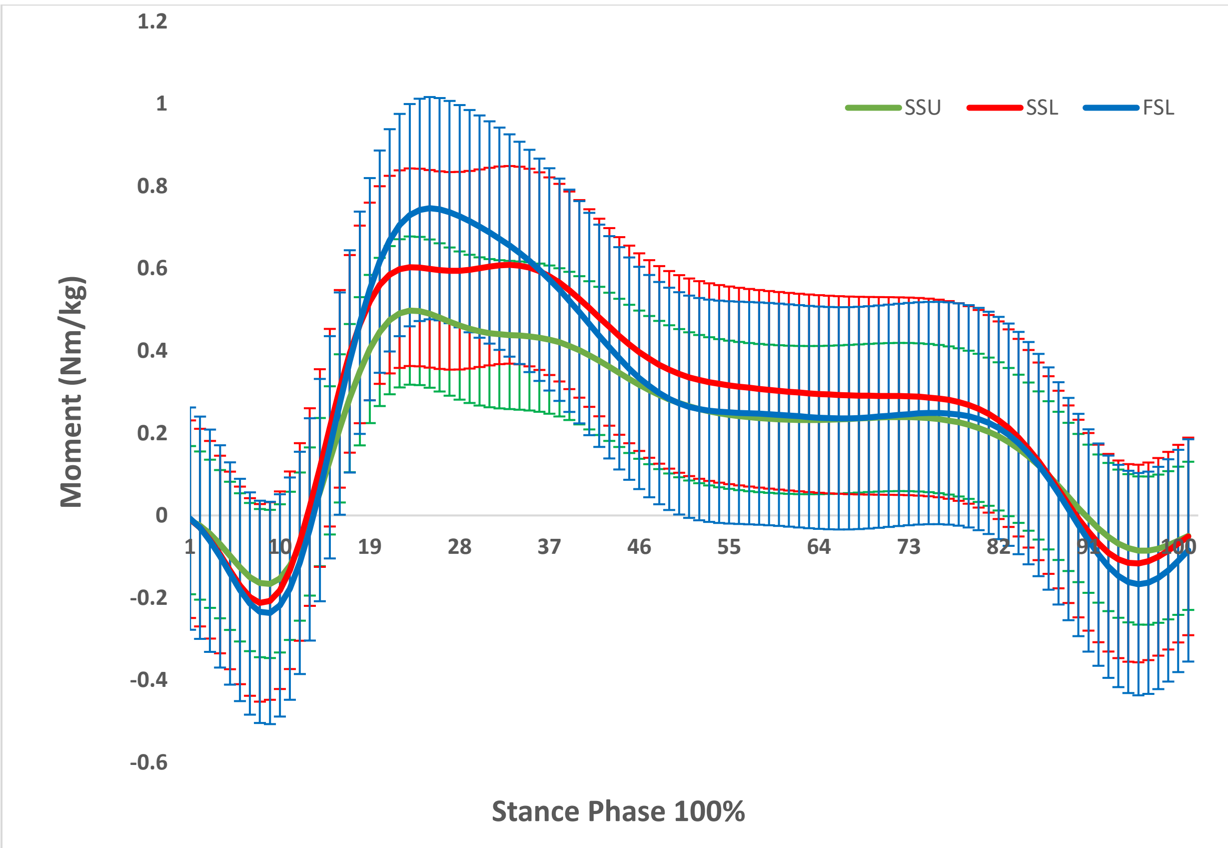


Figure 4.2: ***Stance phase frontal plane (KAM) ensemble averaged waveforms for SSU, SSL, and FSL. Positive values are KAM and negative values are knee abduction moments (error bars - standard deviation). Note curves are normalized to stance phase whereas the KAM impulse was calculated using time in seconds***

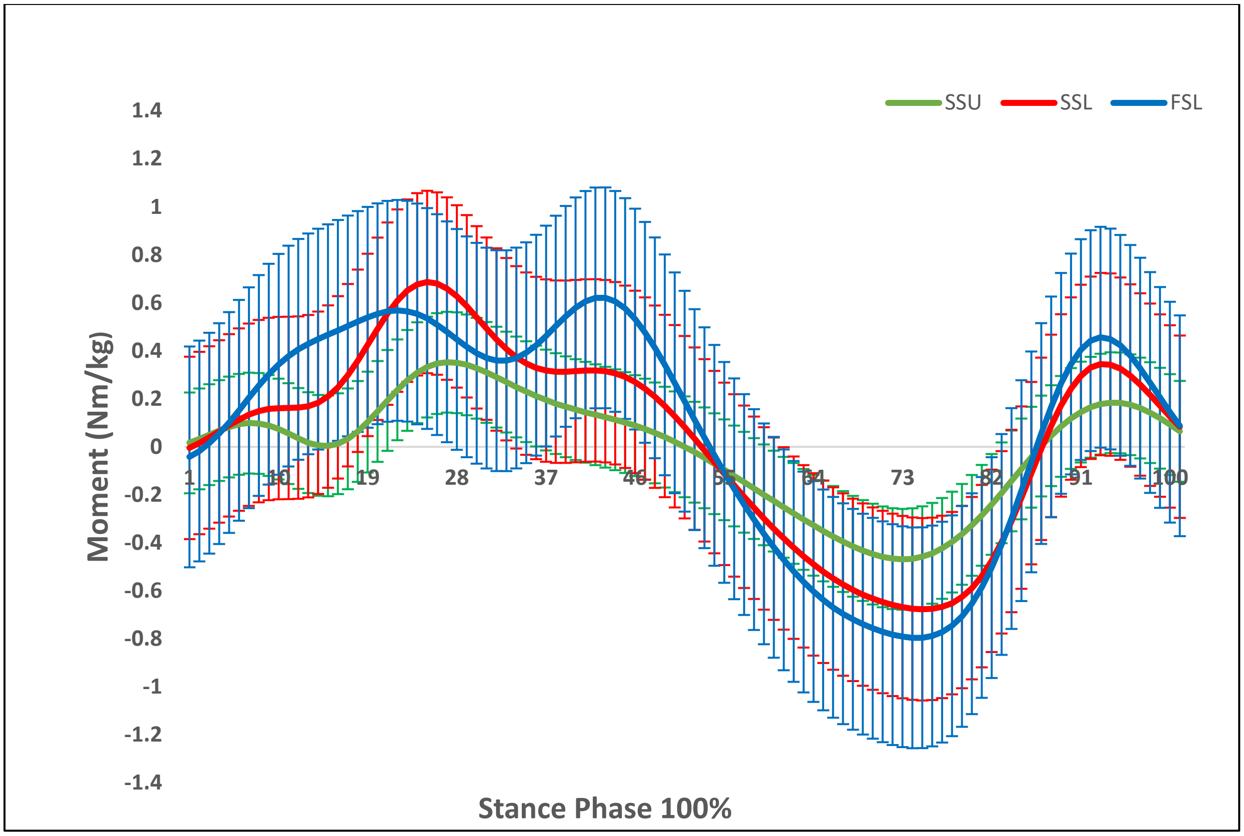


Figure 4.3: Stance phase *sagittal plane (KFM) ensemble averaged waveforms for SSU, SSL, and FSL. Positive values are KFM, and negative values are KEM (error bars - standard deviation)*

The mean and standard deviation for each knee joint moment feature and the p values for each one-way repeated measures ANOVA across walking condition are found in Table 4.6. P values for the four discrete moment features were statistically significant ($p<0.05$) indicating differences across walking conditions.

Table 4.6: ***One-way repeated ANOVA results for knee joint moment features related to knee OA clinical progression during walking conditions***

Knee Moment Measure	**Walking Conditions** **SSU**	**SSL**	**FSL**	**p**
KAM Impulse (Nms/kg)	0.16 (0.04)	0.21 (0.07)	0.19 (0.61)	***< .001***
Peak KAM (Nm/kg)	0.55 (0.15)	0.70 (0.24)	0.79 (0.27)	***< .001***
Peak KFM (Nm/kg)	0.43 (0.29)	0.77 (0.67)	0.82 (0.56)	***< .001***
KFM-KEM (Nm/kg)	0.92 (0.27)	1.50 (0.40)	1.69 (0.36)	***< .001***

All data presented as mean (standard deviation) except where noted
Bold indicate significant differences between walking conditions condition (p<0.05)

4.3.1 Frontal plane knee moment features

Both KAM features met the assumptions for ANOVA including normality and sphericity. A Bonferroni post hoc analysis of the statistically significant difference ($p<0.05$) (Table 4.6) for KAM impulse found pairwise differences (Appendix I) among the walking conditions. The pairwise comparisons showed that KAM impulse for SSL was greater than both FSL and SSU and that KAM impulse was greater during FSL than SSU. Figure 4.4 presents the pairwise comparisons for KAM impulse for each walking condition and pairwise differences among conditions.

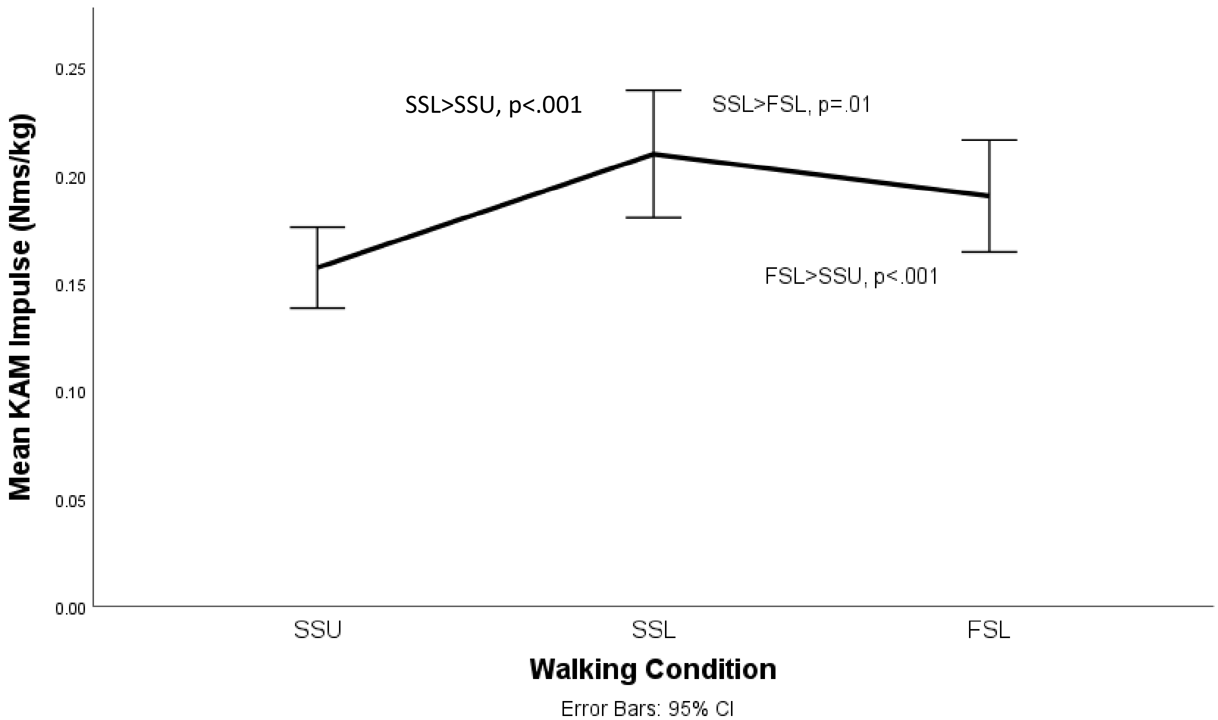


Figure 4.4: ***KAM Impulse – pairwise comparisons for walking conditions (95% error bars). Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

The pairwise comparisons for the Peak KAM (Appendix I) among walking conditions are illustrated in Figure 4.5. The mean Peak KAM for FSL was greater than for the SSL and SSU conditions and the SSL condition was greater than the SSU condition.

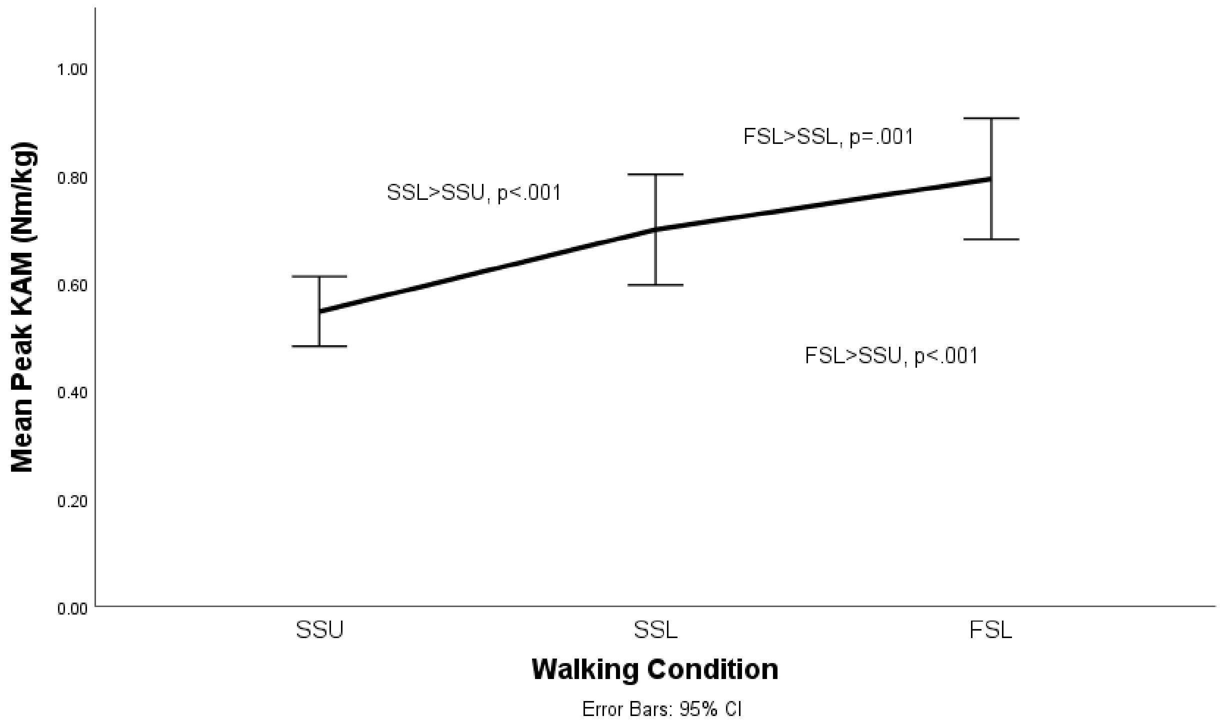


Figure 4.5: ***Peak KAM – pairwise comparisons for walking conditions (95% error bars) Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

4.3.2 Sagittal plane knee moment features

For the sagittal plane moment features, the assumption of normality was violated for both the Peak KFM and KFM-KEM data. For Peak KFM, data transformation (Lg10) resulted in normally distributed data. One-way repeated measures ANOVA findings were similar across the data sets irrespective of which form of the data was analysed (i.e., untransformed or transformed) where there was a statistically significant ($p<0.05$) effect among walking conditions for Peak KFM. Since there were no differences in ANOVA results or pairwise comparisons, and the transformed data are not easily interpretable or comparable to other data [194, 195], the discussion will focus on the non-transformed data: the untransformed, Peak

KFM data, is presented here. Details of the normality tests and analysis of the transformed Peak KFM data can be found in Appendix J.

The Bonferroni pairwise comparisons for the Peak KFM (Appendix I) found that the mean for both FSL and SSL were significantly greater than the mean for SSU (Figure 4.6).

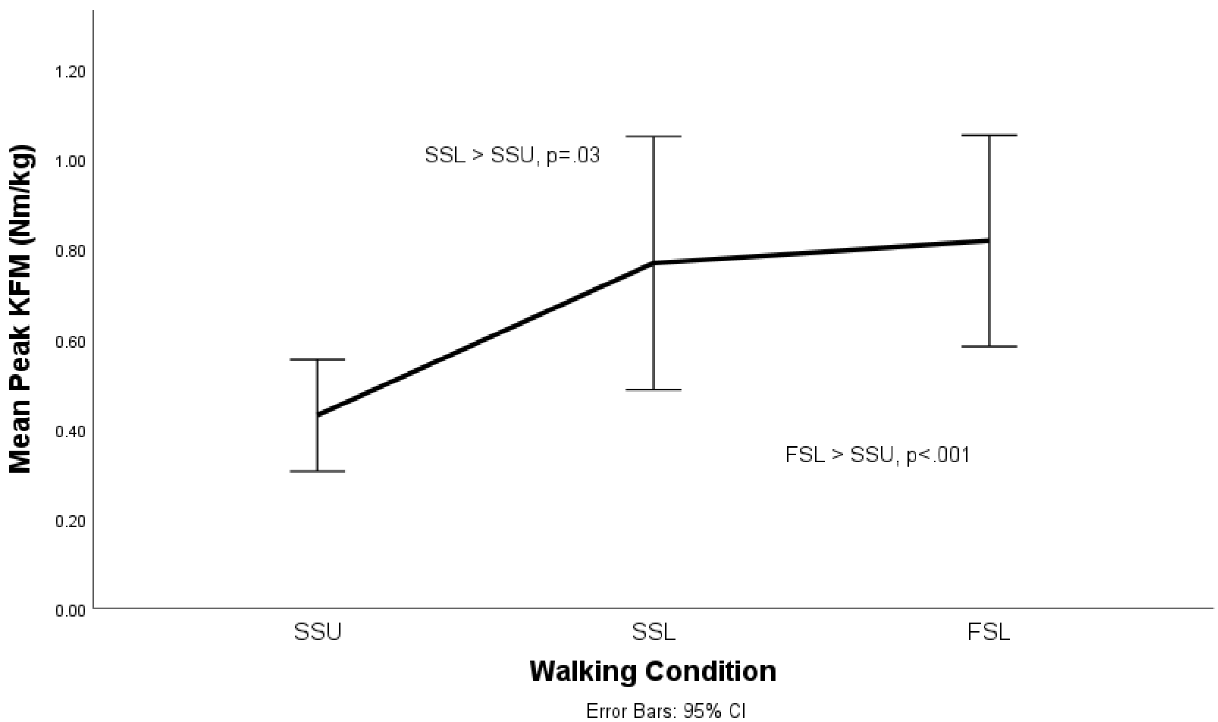


Figure 4.6: ***Peak KFM*** **pairwise comparisons** ***for walking conditions (95% error bars)*. *Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

The KFM-KEM data was not normally distributed, and transformation did not result in a normal distribution. Removal of one outlier resulted in a normalized distribution; the one-way repeated measures ANOVA of the complete sample's KFM-KEM data (n=24) and the analysis of the KFM-KEM data with one outlier removed (n=23) both found a statistically significant ($p<0.05$) effect for walking condition. There was no difference in the pairwise comparisons of the original (n=24) or outlier removed (n=23) data. Based on similar omnibus tests and pairwise comparisons for the data sets, the full sample's (n=24) KFM-KEM data is presented here. The normality tests and analysis of the KFM-KEM data with the outlier removed (n=23) can be found in Appendix K. Mauchley's Test indicated no violations of the assumption of equal variance for both Peak KFM and KFM-KEM data.

The statistically significant difference in KFM-KEM among walking conditions determined from the Bonferroni pairwise comparisons (Appendix I) are illustrated in Figure 4.7. The KFM-KEM means for FSL were greater than both SSL and SSU and KFM-KEM was greater during SSL than SSU.

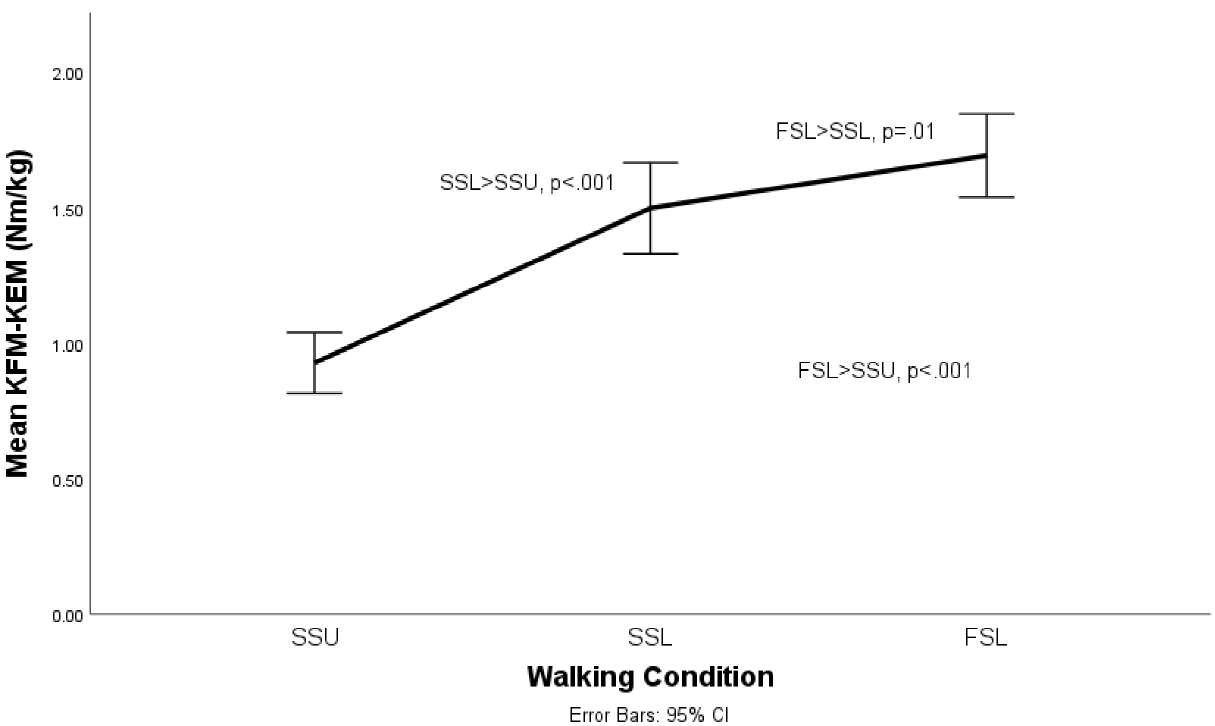


Figure 4.7: ***KFM-KEM*** **pairwise comparisons for walking conditions (95% error bars).** ***Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

4.4 *Walking condition and participant RPE, MCU, and peak trunk angle*

One-way repeated measures ANOVAs tested for the effect of walking condition on mean values for RPE, MCU, and peak trunk angle measures (Table 4.7). The assumption of sphericity was violated for all secondary outcomes; Greenhouse-Geisser corrected p-values are reported.

Table 4.7: ***One-way repeated measures ANOVA results for rate of perceived exertion, muscle capacity utilization, and peak trunk angle during walking conditions***

Measure	Walking Condition	M (SD)	95% CI Lower	95% CI Upper	p
RPE	SSU	7.5 (1.7)	6.8	8.2	***<.001****
	SSL	11.4 (1.9)	10.6	12.2	
	FSL	12.3 (2.1)	11.4	13.2	
MCU (unitless)	SSU	0.2 (0.2)	0.2	0.3	***<.001****
	SSL	0.4 (0.3)	0.3	0.6	
	FSL	0.4 (0.3)	0.3	0.6	
Peak Trunk Angle (degrees)	SSU	5.5 (3.2)	4.2	6.9	***<.001****
	SSL	15.6 (5.3)	13.4	17.9	
	FSL	16.2 (4.8)	14.1	18.2	

*****Greenhouse-Geisser correction applied

RPE was significantly different among walking conditions. Pairwise comparisons (Appendix L) showed that the RPE means for FSL were greater than both SSL and SSU and RPE was greater during SSL than SSU. The statistically significant difference in RPE among walking conditions determined from the Bonferroni pairwise comparisons are illustrated in Figure 4.8.

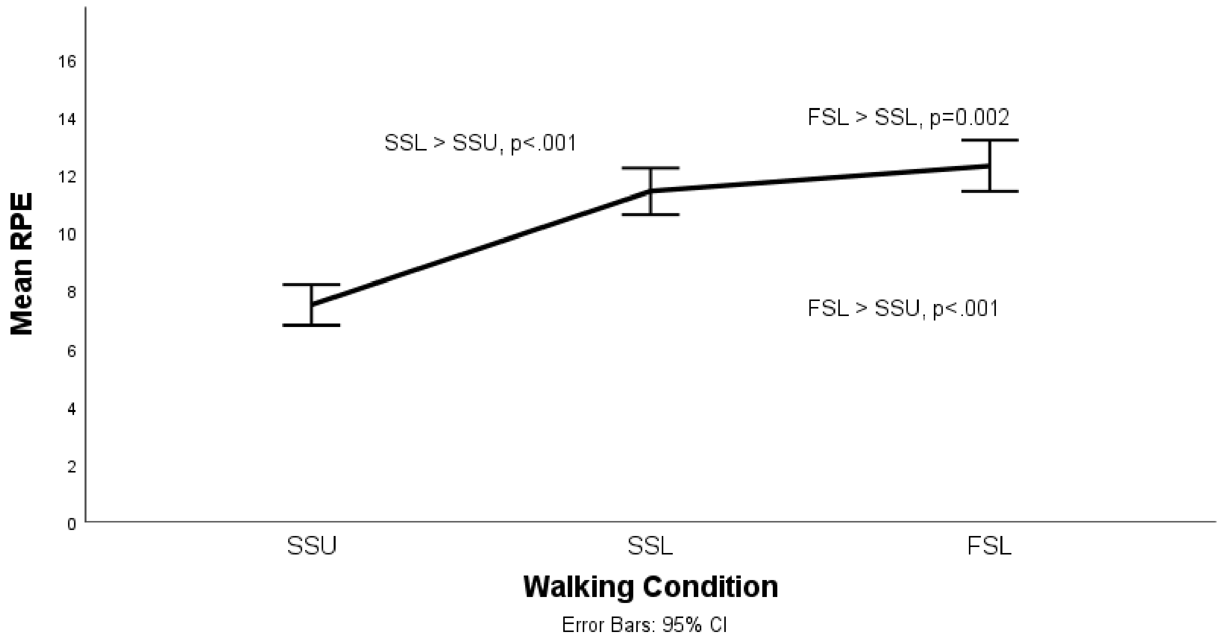


Figure 4.8: ***RPE*** **pairwise** ***comparisons for walking conditions (95% error bars). Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

MCU was significantly different among walking conditions. Pairwise comparisons (Appendix L) showed that the MCU means for FSL and SSL were greater than SSU. The statistically significant difference in MCU among walking conditions determined from the Bonferroni pairwise comparisons are illustrated in Figure 4.9.

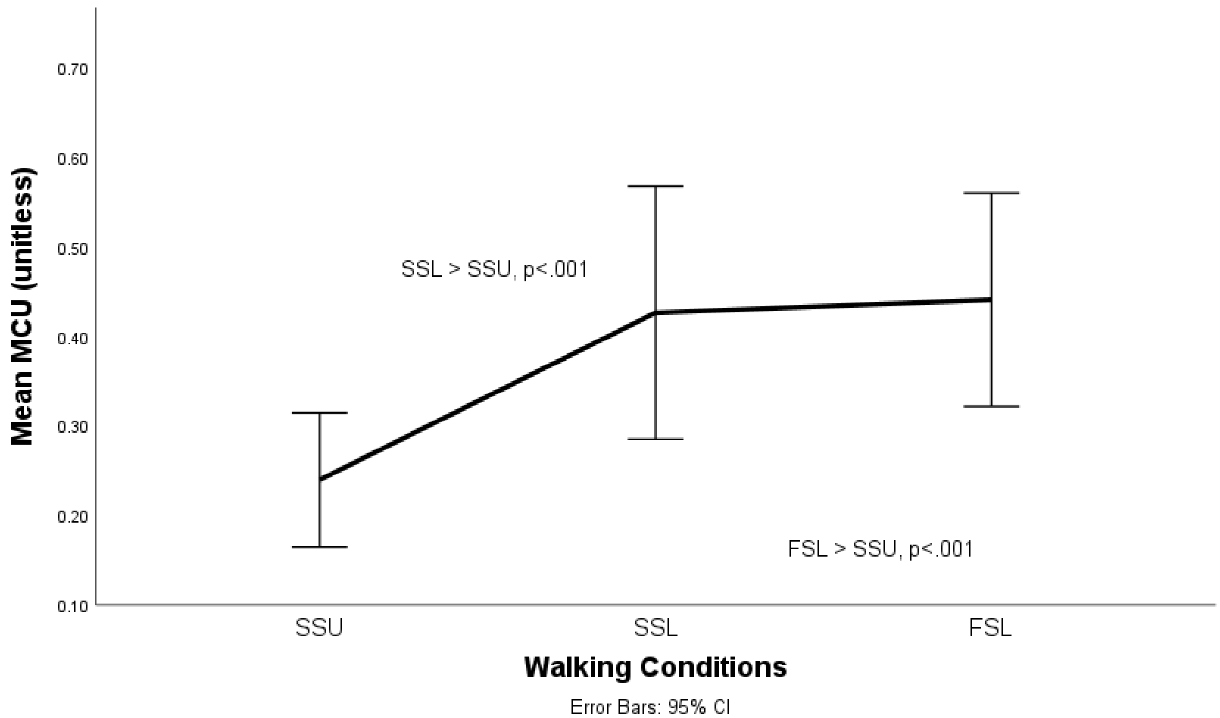


Figure 4.9: ***MCU pairwise comparisons for walking conditions (95% error bars). Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

Peak trunk angle was significantly different among walking conditions. Pairwise comparisons (Appendix L) showed that the peak trunk angle means for FSL and SSL were greater than SSU. The statistically significant difference in peak trunk angle among walking conditions determined from the Bonferroni pairwise comparisons are illustrated in Figure 4.10.

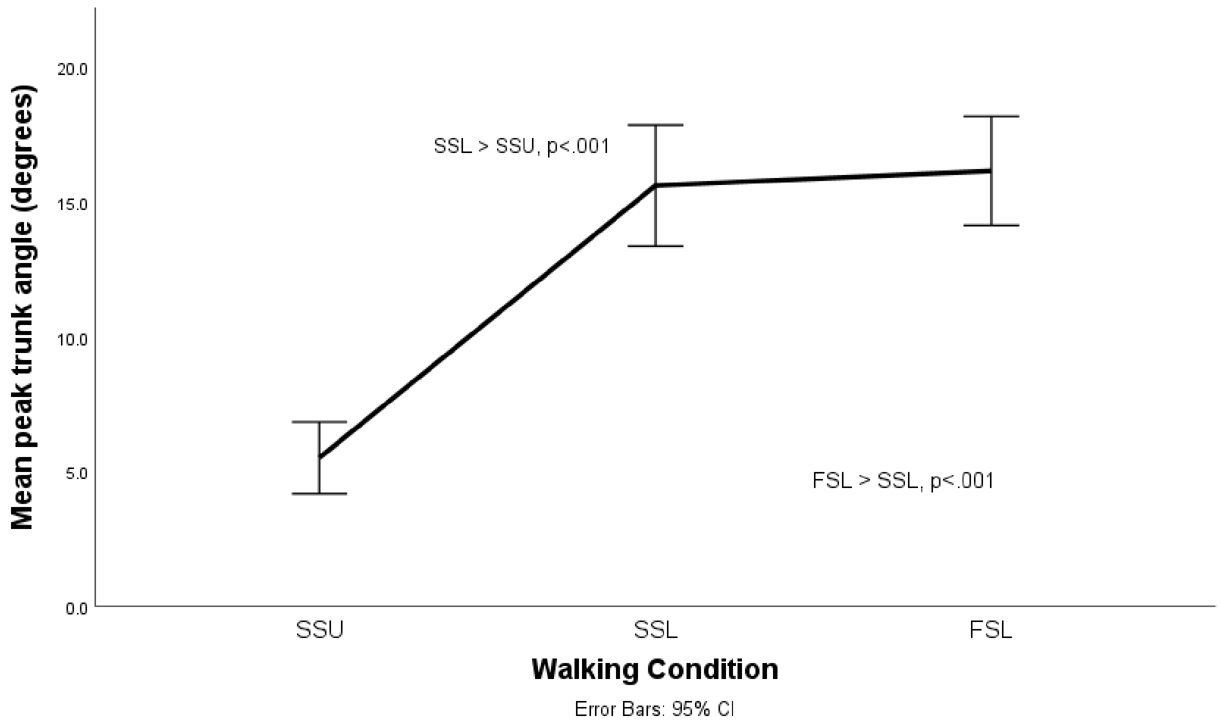


Figure 4.10: ***Peak trunk angle pairwise comparisons for walking conditions (95% error bars). Significant differences between conditions are labelled on the graph with associated p values for each pairwise comparison***

4.5 Summary of results

4.5.1 Summary for primary outcomes – Objective 1

In summary, the results found significant differences among walking conditions for all primary outcomes. For the frontal plane moment features (i.e., KAM impulse, Peak KAM), the unloaded condition (SSU) values were lower for both features than the loaded conditions, but for the loaded condition the KAM impulse was greatest during SSL and the Peak KAM was greatest during FSL. For the sagittal plane moment features (i.e., Peak KFM and KFM-KEM), the unloaded (SSU) values were lower than the two loaded conditions, with a significantly greater KFM-KEM for the FSL compared to the SSL condition.

4.5.2 Summary for secondary outcomes – Sub-objective 1a,1b

The highest RPE occurred during the FSL condition, and the RPE for the SSL condition was greater than the SSU condition. The MCU for the two loaded conditions (i.e., SSL and FSL) were greater than for the unloaded condition (i.e., SSU). The peak trunk angle showed that the two loaded conditions (i.e., SSL, FSL) had greater trunk flexion than the unloaded condition (SSU).

4.6 Discussion

The demographic information and years of military service of the study sample indicates that they were experienced military members based on a mean age of 29 years with 9 years of service. The sample of 24 members, including 10 females, is a larger sample than that of 17 of the 20 studies included in a recent systematic review of load carriage by military members [40]; only one of the studies cited in that review included female participants. The mean age and BMI of the study sample was similar to that of the larger studies included in that systematic review that investigated gait features during military load carriage [40]. In the current study, participants walked on the treadmill at their SS speed, which on average was 1.30 m/s, and all achieved the minimum FS walking speed. The SS speed for this study was similar to other findings for healthy adults reported in a descriptive meta-analysis [129] and was consistent with the SS speeds reported in military load carriage studies [40].

The overall goal of this research was to advance understanding of the risks for clinical knee OA development and progression for military members by examining biomechanical adaptations during load carriage with operationally relevant load and speed. The significant differences found in the joint moment features across walking conditions partially support the study's primary hypotheses. The differences between loaded (FSL, SSL) and unloaded (SSU) conditions indicate, as expected, that the standardized load influenced all joint moment features. The differences found between the two loaded conditions for KAM impulse, Peak KAM, and KFM-KEM, provide evidence of a speed effect that has been less well studied in the military load carriage literature [40].

The objective measure of KE muscle effort differed among walking conditions, with the two loaded conditions having the greatest MCU whereas the self-reported effort was greatest for the FSL and lowest for the unloaded condition. Together, these findings partially support this sub-objective's hypotheses. For the final sub-objective, the two loaded conditions resulted in greater forward trunk angle than the unloaded condition with no additional effect for the FSL condition. The discussion below places the results into context related to the current literature and highlights the new knowledge gained on the effect of load carriage on risk factors for medial knee joint OA.

4.6.1 Knee joint moment features linked to knee OA progression

The higher KAM impulse and higher Peak KAM for the two loaded versus the unloaded condition (Figures 4.4 and 4.5) indicate a higher medial to lateral joint loading ratio over stance phase and during early stance when carrying the 35kg load. These differences were hypothesized, as they are influenced by the load. KAM impulse is sensitive to changes in gait speed [126, 196] and the shorter stance phase duration in the FSL condition partially explains why the KAM impulse was lower than for the SSL condition. The higher Peak KAM for FSL compared to the SSL walking conditions is consistent with previous findings of unloaded walking where increased walking speed was associated with increased Peak KAM in participants with and without knee OA [70, 197]. Despite this higher Peak KAM, the combination of the shorter stance phase duration and lower KAM magnitude later in stance phase for FSL versus SSL (Figure 4.2) suggests a lower overall loading exposure [196] in the FSL condition than the SSL condition. The hypotheses for the KAM features were partially supported.

Meta-analysis identified that higher KAM impulse and Peak KAM are associated with knee OA development and progression [59, 65] and both are significantly correlated with overall KAM magnitude [125] which has been linked to clinical progression to TKA [63, 118, 125-127]. Although there are some discrepancies in the literature concerning these measures' respective association with structural versus clinical medial compartment knee OA progression [198, 199], the lower KAM impulse and greater Peak KAM with higher speed suggest an effect, albeit in different directions, of speed on both KAM features in addition to load. These differences in responses highlight the importance of considering both features when examining the knee joint loading environment during load carriage.

There has been limited study of KAM features during military load carriage [40]. One study [7] that examined KAM features during load carriage reported that with increased load the KAM contributed to a greater percentage of the knee joint total moment for recruit aged females during forced marching (i.e., FS walking) and running. However, only percentage values were provided, making direct comparisons with the current findings difficult, and the knee joint total moments were measured at specific gait events (i.e., heel strike and mid-stance) [7], where neither have been linked with knee OA progression limiting interpretation with respect to knee OA risk. Furthermore, the current study included experienced military members of both sexes and used operationally relevant load and speed whereas Krajewski et al. [7] used a sample of civilian females, body weight relative loads, and speeds relative to participants gait transitional velocity.

A second study of FS marching while carrying loads of 15kg and 30 kg at speeds of 1.53m/s and 1.83m/s found increased medial tibiofemoral joint contact forces of approximately 10% and 20% respectively, compared to no load, for male military members [41]. The authors concluded that carrying loads greater than 15kg for prolonged periods may present a greater risk for knee musculoskeletal injury due to the associated increase in joint contact forces and that high contact forces associated with load carriage may explain increased incidence of knee OA in military members [41]. The KAM findings from the current study are consistent with the general findings from the second study of increased medial knee joint loading. Methodological differences, such as the reporting of joint contact forces, percentages of total knee joint load, and the selection of a faster speed (1.83m/s) make direct comparison of the two studies difficult, however, the slower speed (i.e., 1.53m/s) is consistent with the operationally relevant speed used for the current study.

The sagittal plane moment features (i.e., Peak KFM and KFM-KFM) provide a surrogate measure of the total joint load and the KE muscle force across walking conditions. As hypothesized, both loaded conditions (i.e., FSL and SSL) had significantly greater mean Peak KFM magnitudes than SSU based on the addition of load and associated increase in the GRFs.

Increased Peak KFM with military load carriage is consistent with other findings for military members. Quesada et al. [44] reported an increase in Peak KFM for male military members using 30% body weight relative loads compared to 15% body weight and no load. The authors suggested that the increase in KFM was attributed to KE muscle functioning as a shock absorber during early stance due to increased load [44]. Lenton et al. [41] found increased demands on the KE muscles with load carriage using operationally relevant load and speed concluding that they contributed substantially to medial knee joint contact forces. The military experience of the sample used by Quesada et al. [44] (i.e., officer cadets, age 18-26) indicates caution for generalization of the findings to a more experienced military population [40].

Differences in frontal plane (Figure 4.2) and sagittal plane moment waveforms (Figure 4.3) for each condition provide data on load and speed effects during load carriage. The Peak KAM had a systematic increase from the unloaded (SSU) to the loaded (SSL) condition and the fixed speed condition (FSL) had the highest peak. These findings are consistent with findings for unloaded walking that showed higher Peak KAM with greater walking speed [70, 197].

As expected, the two loaded conditions had greater peaks for their respective KFM waveforms than the unloaded condition in the first half of stance, but there were no visually appreciable differences in the peaks for the SSL and FSL. The findings for the sagittal plane waveform for FSL (Figure 4.3) show that increased speed of load carriage is associated with double peaks in early stance and a third peak in late stance. The individual data for FSL showed that some participants had their greatest Peak KFM (Figure 4.3) just before mid-stance (approximately 44% of stance phase), during single limb support, whereas it has been reported to occur earlier in stance phase (i.e., 22%) during unloaded walking [128]. This was an unexpected finding as it was not found to have been previously described in the literature. This feature should be further explored concerning its potential impact on knee OA risk.

Exposure to three separate points of KFM peaks should be further explored as a greater rate of cyclic loading during FS walking and the greater magnitude KFM and KEM, illustrated in the sagittal plane ensemble averaged waveforms for SSL and FSL, indicate a dynamic loading pattern with increased joint loads during load carriage due to changes in muscle activation patterns (Figure 4.3). Prolonged (i.e., 50 minutes), cyclic loading via submaximal muscle contractions resulted in increased chondrocyte death *in vitro* [75] and epidemiological studies have linked higher occupational related knee joint loading and excessive loading with the initiation of knee OA [100] while structural progression of OA in the patellofemoral joint has been linked to increased KFM magnitudes in late stance [66]. Visual comparison of the late stance (second half) KFM from the sagittal plane ensemble averaged waveform for FSL indicated a Peak KFM greater than that reported for those with patellofemoral OA that progressed over 1-year [200]. While there is discordance between structural changes and the severity of signs and symptoms of knee OA, higher knee joint loads due to higher muscular activation patterns during military load carriage may play a role in the onset and progression of structural knee OA.

The larger KFM-KEM difference, as opposed to the hypothesized smaller KFM-KEM difference indicative of stiff gait linked to clinical progression of knee OA [118] and progression to TKA [67], is due to both higher Peak KFM from the statistical comparisons and visually greater

Peak KEM. The findings from the current study, increased KFM-KEM with higher speed, may be explained by studies of load carriage that have identified an increased role of KE muscles during load carriage and of the PF muscles during the propulsive phase of gait [40, 44, 45, 84, 90]. While other studies have identified increased PF activity during military load carriage [40, 90] this study is the first to report larger KFM-KEM during military load carriage at FS compared to SS speeds. It is important to note that KFM-KEM may be underestimated as the inverse dynamics model does not consider antagonist muscle group activity. The EMG for the PF muscles was collected for this study but, due its scope, these findings were not reported.

The other notable observation from the sagittal plane waveform (Figure 4.3) was the systematic increase in the late Peak KEM among the conditions with the FSL having the greatest peak and the SSU the smallest peak. While the statistical analysis of the primary outcomes (i.e., the four knee moment features linked to knee medial compartment OA progression) contributes to a better understanding of the risks for clinical knee OA development and progression associated with load carriage using operationally relevant load and speed for military members, there are qualitative waveform features that warrant future study.

4.6.2 Self-reported and quantitative measure of effort

The higher self-reported effort (RPE) for the two loaded conditions compared to the unloaded was expected due to the addition of the load [42, 46, 56] as was the greatest RPE for the FS condition due to the fixed pace [46, 51]. The greater MCU for the KE muscles in this study is consistent with previous reports of higher KE muscle activity for load carriage tasks [40, 90]. In addition to a higher KE muscle contribution based on the higher Peak KFM, the greater RPE for

FSL may be due to increases in PF muscle activation associated with higher load carriage [84] and walking speeds [40] contributing to a greater cardiovascular stress [84]. While MCU only provides an indication of muscle effort at one point during stance phase, it has been used at the knee to assess the ratio of body mass normalized KE muscle strength utilized to complete physical activities [90, 93]. The findings for this study suggest that the KE muscles are more sensitive to load than walking speed.

4.6.3 Peak trunk angle

The greater peak forward trunk lean with the addition of load in the current study is consistent with 4 of 5 studies that examined this feature and were included in a recent systematic review of military load carriage [40]. Although differences in landmarks used and angle calculations limit direct comparisons of trunk lean values [40], the review's authors suggested that peak trunk flexion is likely influenced by the distribution of the load with backpack loads, as used for this study, resulting in greater peak forward trunk lean than evenly distributed loads. The presence of a hand carried rifle, not included for this study, but included in the single study that found no difference in trunk lean, was suggested by the authors as resulting in less impact on centre of mass position due to the addition of mass anteriorly thereby offsetting the posteriorly positioned backpack load [40]. The findings for this study, where the same mass was used for both loaded conditions, suggests that forward trunk lean is more sensitive to the mass of the load than walking speed. Forward trunk lean may reduce KFM by bringing the centre of mass closer to the knee joint's axis of rotation. However, forward trunk lean is associated with increased posterior muscle group activity [40] (e.g., hip extensors) that may lead to an underestimation of the KEM and alterations to the net KFM do not necessarily correspond with an alteration in joint contact forces.

4.6.4 Limitations

As with any study there are potential limitations that need to be considered when designing a study and interpreting the data. The sample size, while adequate based on the significant differences found and compared to other load carriage gait studies, had some variables that were not normally distributed. These variables had to be transformed before conducting the statistical analysis but, importantly, the findings were not altered with the transformed data.

The order of the study's protocol was the same for each participant to account for treadmill familiarization for loaded and unloaded conditions. While the order of exposure could affect results due to learning and fatigue, steps were taken to minimize these effects. The effects of learning were minimized with the use of a treadmill familiarization protocol [179, 182] and by analyzing the data collected during the final trial of each condition (Chapter 3, Section 3.4). Furthermore, recovery of the KE muscles following a high intensity fatigue protocol was reported following 5-minutes of rest for sedentary males and females aged 19-35 years [85]; fatigue effects would likely by minimal given the lower intensity walking tasks in this healthy, relatively, fit sample. However, to minimize potential effects of fatigue, a rest period was provided between loaded conditions (10-minutes) and there was a rest period (5-minutes minimum) prior to strength testing.

The absolute load carried for this study was equal to the load associated with the FORCE Combat™ test [146] but, due to pragmatic reasons, the load did not include a C7 rifle, rubber training rifle, or substitute, and no helmet was worn. The load associated with the helmet and rifle, approximately 4.5kg [146], was distributed evenly to the weighted vest and small-pack. Carriage of a rifle during loaded walking may result in higher KE and PF muscle activity [40]. Replication of the exact load distribution (i.e., inclusion of rifle and helmet) used during the FORCE Combat™ test may be considered for future studies to optimally replicate operational and training requirements.

The KAM and KFM features are not direct measures of joint contact loads. The method of measuring knee joint loads *in vivo* includes instrumented knee joint replacements, which is not feasible for measuring knee joint loads in healthy military personnel. Modeling knee joint contact forces is an alternative [41], but this method requires numerous assumptions and is computationally time consuming [201]. The moments were interpreted with the knowledge that inverse dynamics calculates the net joint moment and does not consider the presence of antagonist muscle activity. While not direct measures, the moment features included do have predictive validity for progression of knee medial compartment OA. The discrete metrics used for this study do not capture the entire ensemble averaged waveform [202], subsequently, there are other features such as the KFM double peak that were not examined and could potentially be captured using more sophisticated techniques (e.g., principle component analysis).

The MCU, as a measure of KE effort, is the ratio of the Peak KFM and the maximum torque (strength) [93] generated by the KE muscles [93] and, as above, assumes no antagonist activity. MCU has been used in the literature to assess utilization of muscle strength [90, 93], but muscle activity data would improve interpretation of the KE muscle effort during load carriage.

4.6.5 Conclusion

This study provides novel data that help to advance understanding of the risks for clinical medial compartment knee OA development and progression during load carriage with operationally relevant load and speed in military members. Carrying a standardized 35kg load at either SS or FS was associated with greater KAM impulse, Peak KAM, and Peak KFM, compared to unloaded walking (SSU), consistent with a higher risk for knee OA development and progression. The greater KFM-KEM difference for the loaded conditions was not consistent with a stiff knee gait pattern linked to medial compartment knee OA progression. The differences between the two loaded conditions in KAM impulse, Peak KAM, and KFM-KEM supports that a speed effect exists that may alter knee OA risk. Both loaded conditions had greater exertion measures (RPE, MCU) and greater trunk flexion than the unloaded condition, with the only difference between loaded conditions being a greater perceived exertion for the FSL condition. These findings provide evidence suggesting differences in perceived effort between FS and SS speed load carriage but that KE effort and trunk flexion adaptation between the two loaded conditions were not different.

Chapter 5. Results and discussion: Comparison of male and female participants' joint moment features among walking conditions

This chapter provides a summary of the results for Objective 2, Sub-objective 2a, and 2b. Objective 2 aimed to provide preliminary data on whether there are differences in the joint moment features (i.e., primary outcomes) related to development and progression of knee OA between male and female military members for each walking condition or within sex differences among the three walking conditions. Sub-objective 2a and 2b aimed to determine if there are differences in measures of effort (i.e., MCU, RPE) and forward trunk angle (i.e., peak trunk angle) between sexes for each walking condition or within sex differences among the three walking conditions.

5.1 Male and female demographic characteristics

A total of 24 participants, 14 males and 10 females, completed the study protocol over the 13-week study period. Male participants were recruited at a rate of 1.1/week and females at a rate of 0.8/week. The demographic statistics and independent T-test results comparing the two groups are found in Table 5.1. Males in the sample were on average taller than the females while females had greater mean BMI based on significant independent T-test ($p < 0.05$) results (Table 5.1). Four participants, 3 females (30%) and 1 male (7%), had BMI values of greater than 29.9kg/m^2 (obese). Six participants had BMI values between 25 - 29.9kg/m^2 (overweight); of the participants with BMI values considered overweight, 4 were female (40%) and 2 were male (14%).

Table 5.1: ***Descriptive statistics and T-test p values for male and female participant demographic characteristics***

			95% CI		
Characteristic	**Sex**	**M(SD)**	**Lower**	**Upper**	**p**
Age (years)	Male	28.4 (7.9)	23.8	32.9	.53
	Female	30.2 (5.2)	26.5	33.9	
Height (m)	Male	1.8 (0.1)	1.8	1.9	**<.001**
	Female	1.7 (0.1)	1.6	1.7	
Mass (kg)	Male	78.1 (12.5)	70.9	85.4	.37
	Female	73.7 (10.6)	66.1	81.3	
BMI (kg/m^2)	Male	23.9 (2.9)	22.2	25.5	**.04**
	Female	26.8 (3.7)	24.2	29.5	
Military Service (years)	Male	8.1 (7.9)	3.5	12.6	.43
	Female	10.4 (5.7)	6.3	14.5	

Bold indicates significant differences between sex ($p \leq 0.05$)

KE and KF muscle strength, both absolute (Nm) and body mass normalized (Nm/kg), for males and females are found in Table 5.2 as are the independent T-test results for between sex differences. As a group, the males' normalized and absolute KE muscle strength were significantly ($p<0.05$) greater than the values for females. There was a statistically significant ($p<0.05$) difference for absolute KF muscle strength alone, with males having greater absolute strength than females.

Table 5.2: ***Descriptive statistics and T-test p values for male (n=14) and female(m=10) absolute and body mass normalized knee extensor and knee flexor muscle strength***

Muscle strength value	Muscle group	Sex	M(SD)	95% CI Lower	95% CI Upper	p
Absolute (Nm)	KE	Male	165.0 (25.9)	150.1	179.9	***.01***
		Female	125.9 (35.9)	100.3	151.6	
	KF	Male	108.7 (28.6)	92.2	125.3	***.03***
		Female	81.6 (27.1)	62.1	101.0	
Body mass normalized (Nm/kg)	KE	Male	2.2 (0.4)	1.9	2.4	***.05***
		Female	1.8 (0.6)	1.3	2.2	
	KF	Male	1.4 (0.3)	1.2	1.6	.15
		Female	1.2 (0.5)	0.8	1.5	

Bold indicates significant differences between sex ($p \leq 0.05$)

Descriptive statistics for SS and FS walking speeds, along with independent T-test results, are presented in Table 5.3. All participants (100%) achieved and maintained their SS walking speed based on the mean speed from their Gait Rite trials for SSU and SSL walking conditions. All participants (100%) walked at a FS equal to, or greater than, 1.52m/s. There were no significant differences between male and female participants for SS or FS walking speeds ($p>0.05$).

Table 5.3: ***Descriptive statistics and T-test p values*** **for m*ale (n=14) and female (n=10) self-selected and fixed speed walking***

Type of speed (condition)	Sex	M (SD)	95% CI Lower	Upper	p
SS (m/s)	Male	1.29 (0.14)	1.21	1.37	.68
	Female	1.31 (0.13)	1.22	1.40	
FS (m/s)	Male	1.55 (0.06)	1.52	1.59	.48
	Female	1.54 (0.05)	1.50	1.57	

Descriptive statistics and 95% CI of the mean differences for stance phase duration for males and females are in Table 5.4. There were no significant between sex within walking condition pairwise differences for stance phase duration (s) based on the 95% CI of the mean differences.

Table 5.4: ***Descriptive statistics of stance duration for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking Condition	Sex	M (SD)	95% CI Lower	95% CI Upper	M Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU (s)	Male	0.69 (0.05)	0.66	0.72	0.01	-0.03	0.05
	Female	0.68 (0.04)	0.65	0.71			
SSL (s)	Male	0.70 (0.05)	0.68	0.73	0.00	-0.04	0.04
	Female	0.70 (0.05)	0.67	0.74			
FSL (s)	Male	0.65 (0.03)	0.63	0.66	0.00	-0.03	0.02
	Female	0.65 (0.03)	0.63	0.68			

No significant between sex differences based on CI of the mean difference

Descriptive statistics and 95% CI of the mean differences for stance phase duration for males and females are in Table 5.5. There were significant between walking condition pairwise differences for stance phase duration for males and females based on the 95% CI of the mean differences for each walking condition.

Table 5.5: ***Confidence intervals for the mean pairwise differences in stance duration among walking conditions for male (n=14) and female (n=10) participants***

Sex	**Compared Conditions**	**Mean Difference**	**95% CI of the Difference**	
			Lower	**Upper**
Male	SSU-SSL	-0.01	-0.03	0.00
	SSU-FSL	0.04	***0.01***	***0.08***
	SSL-FSL	0.06	***0.03***	***0.08***
Female	SSU-SSL	-0.03	***-0.04***	***-0.01***
	SSU-FSL	0.03	0.00	0.06
	SSL-FSL	0.05	***0.02***	***0.08***

Bold CI indicate significant differences between conditions within sexes

5.2 Male and female frontal plane ensemble averaged waveforms and knee moment features linked to knee OA clinical progression

Stance phase frontal plane ensemble averaged moments over 100% stance phase for male and female participants separately for SSU, SSL, and FSL walking conditions are shown in Figure 5.1.

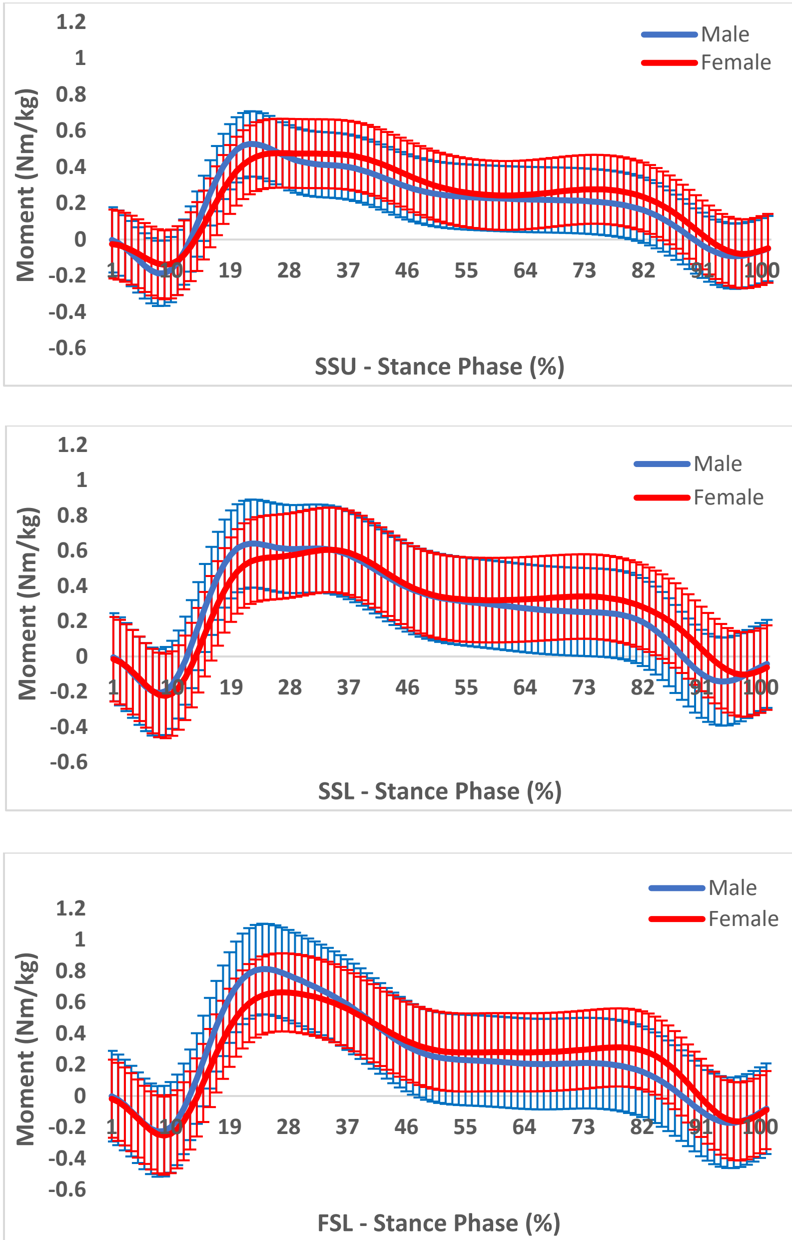


Figure 5.1: ***Stance phase frontal plane (KAM) ensemble averaged waveforms for male (n=14) and female (n=10) participants for SSU (top), SSL (middle), and FSL (bottom). Positive values are KAM and negative values are knee abduction moments (error bars – standard deviation)***

Descriptive statistics for the KAM impulse for males and females for each walking condition are presented in Table 5.6. There were no significant pairwise differences between sex within walking conditions based on the 95% CI of the mean differences in Table 5.6.

Table 5.6: ***Descriptive statistics of KAM Impulse for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking Condition	Sex	Mean (SD)	95% CI Lower	95% CI Upper	Mean Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU	Male	0.15 (0.05)	0.12	0.18	-0.01	-0.05	0.03
(Nms/kg)	Female	0.14 (0.04)	0.14	0.19			
SSL	Male	0.21 (0.08)	0.16	0.25	0.00	-0.07	0.06
(Nms/kg)	Female	0.21 (0.06)	0.17	0.25			
FSL	Male	0.19 (0.06)	0.15	0.22	0.00	-0.05	0.05
(Nms/kg)	Female	0.19 (0.06)	0.14	0.23			

No significant between sex differences based on CI of the mean difference

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.7. For male participants, there were statistically significant differences across conditions based on the 95% CI, with the mean FSL and SSL KAM impulse values greater than the mean value for SSU. Female participants had greater mean KAM impulse values for SSL than SSU only. These significant differences are graphically illustrated in Figure 5.2.

Table 5.7: ***Confidence intervals for the mean pairwise differences in KAM Impulse among walking conditions for male (n=14) and female (n=10) participants***

Sex	Compared Conditions *(Nms/kg)*	Mean Difference	95% CI of the Difference Lower	Upper
Male	SSU-SSL	-0.06	***-0.09***	***-0.27***
	SSU-FSL	-0.04	***-0.06***	***-0.01***
	SSL-FSL	0.02	-0.01	0.04
Female	SSU-SSL	-0.05	***-0.08***	***-0.02***
	SSU-FSL	-0.03	-0.07	0.01
	SSL-FSL	-0.02	0.00	0.04

Bold CI indicate significant differences between conditions within sex

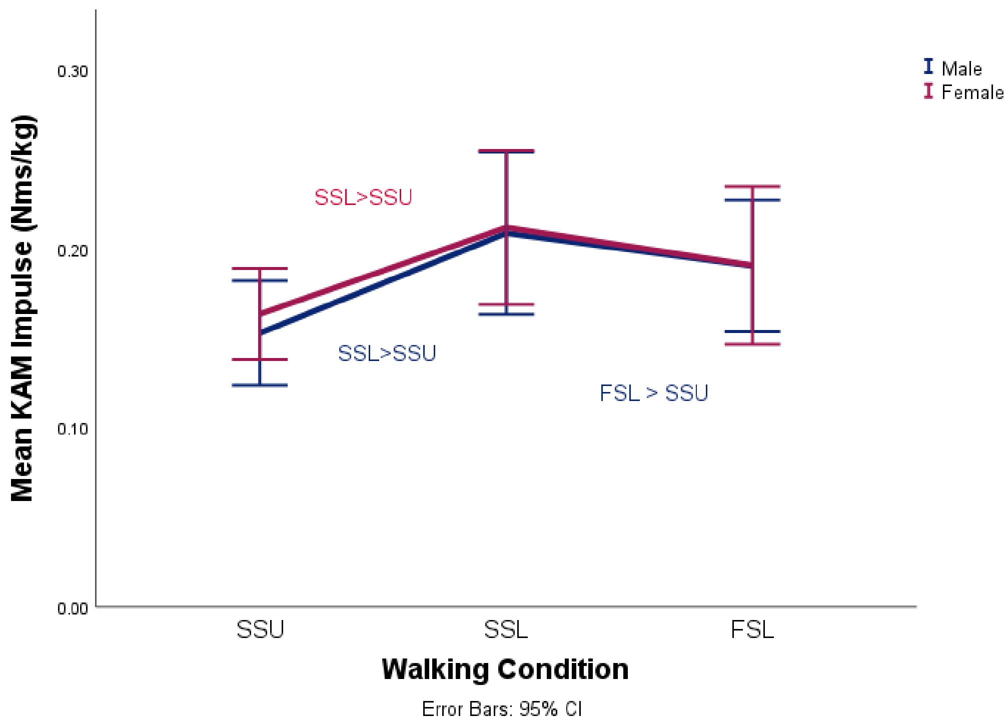


Figure 5.2: ***Mean KAM Impulse for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females***

Descriptive statistics for the Peak KAM for males and females for each walking condition are presented in Table 5.8. The male Peak KAM was 0.01 to 0.08 Nm/Kg higher than females but there were no significant pairwise differences between sex within walking conditions based on the 95% CI of the mean differences (Table 5.8).

Table 5.8: ***Descriptive statistics of Peak KAM for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking Condition	Sex	M (SD)	95% CI Lower	95% CI Upper	Mean Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU	Male	0.56 (0.16)	0.47	0.65	0.03	-0.10	0.17
(Nm/kg)	Female	0.53 (0.15)	0.42	0.63			
SSL	Male	0.70 (0.25)	0.56	0.85	0.01	-0.20	0.23
(Nm/kg)	Female	0.69 (0.24)	0.62	0.86			
FSL	Male	0.82 (0.27)	0.67	0.98	0.08	-0.15	0.31
(Nm/kg)	Female	0.74 (0.27)	0.55	0.94			

No significant between sex differences based on CI of the mean difference

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.9. For male participants, there were statistically significant differences across conditions based on the 95% CI, FSL Peak KAM was greater than for SSL and SSU; the Peak KAM for SSL was greater than for SSU. Female participants had greater Peak KAM for FSL and SSL than for SSU (Table 5.9). These significant differences are graphically illustrated in Figure 5.3.

Table 5.9: ***Confidence intervals for the mean pairwise differences in Peak KAM among walking conditions for male (n=14) and female (n=10) participants***

Sex	Compared Conditions (Nm/kg)	Mean Difference	95% CI of the Difference	
			Lower	Upper
Male	SSU-SSL	-0.14	***-0.22***	***-0.06***
	SSU-FSL	-0.27	***-0.37***	***-0.16***
	SSL-FSL	-0.12	***-0.04***	***- 0.21***
Female	SSU-SSL	-0.16	***-0.27***	***-0.05***
	SSU-FSL	-0.21	***-0.34***	***-0.09***
	SSL-FSL	-0.05	-0.14	0.04

Bold CI indicate significant differences between conditions within sex

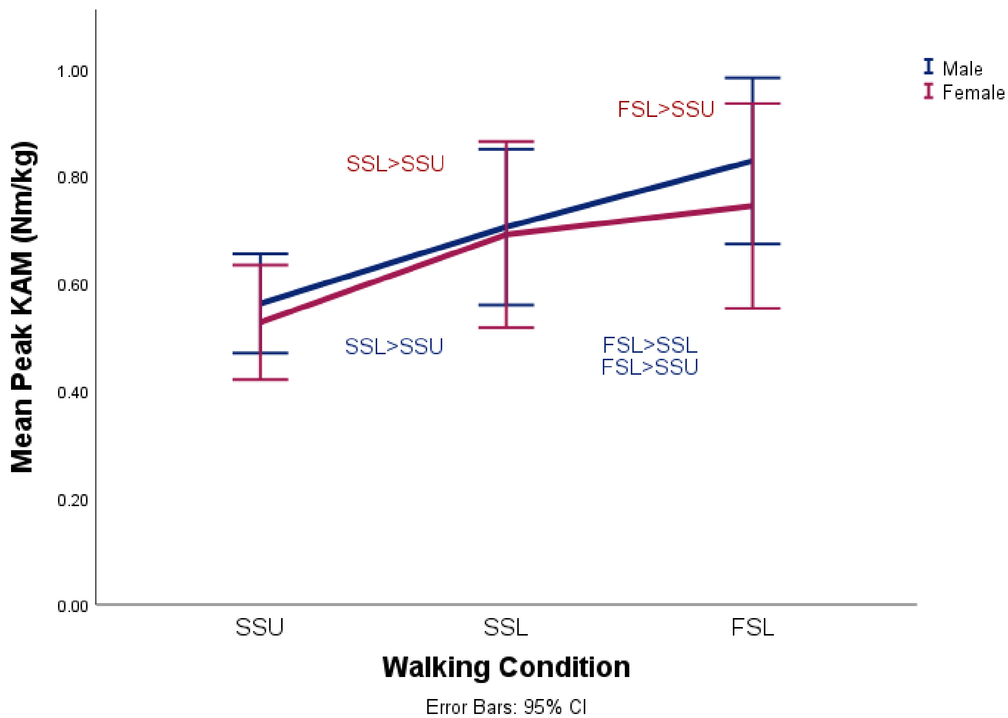


Figure 5.3: ***Mean Peak KAM for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females***

5.3 *Male and female sagittal plane ensemble averaged waveforms and knee moment features linked to knee OA progression*

Stance phase sagittal plane ensemble averaged moments over 100% stance phase for male and female participants separately for SSU, SSL, and FSL walking conditions are shown in Figure 5.4.

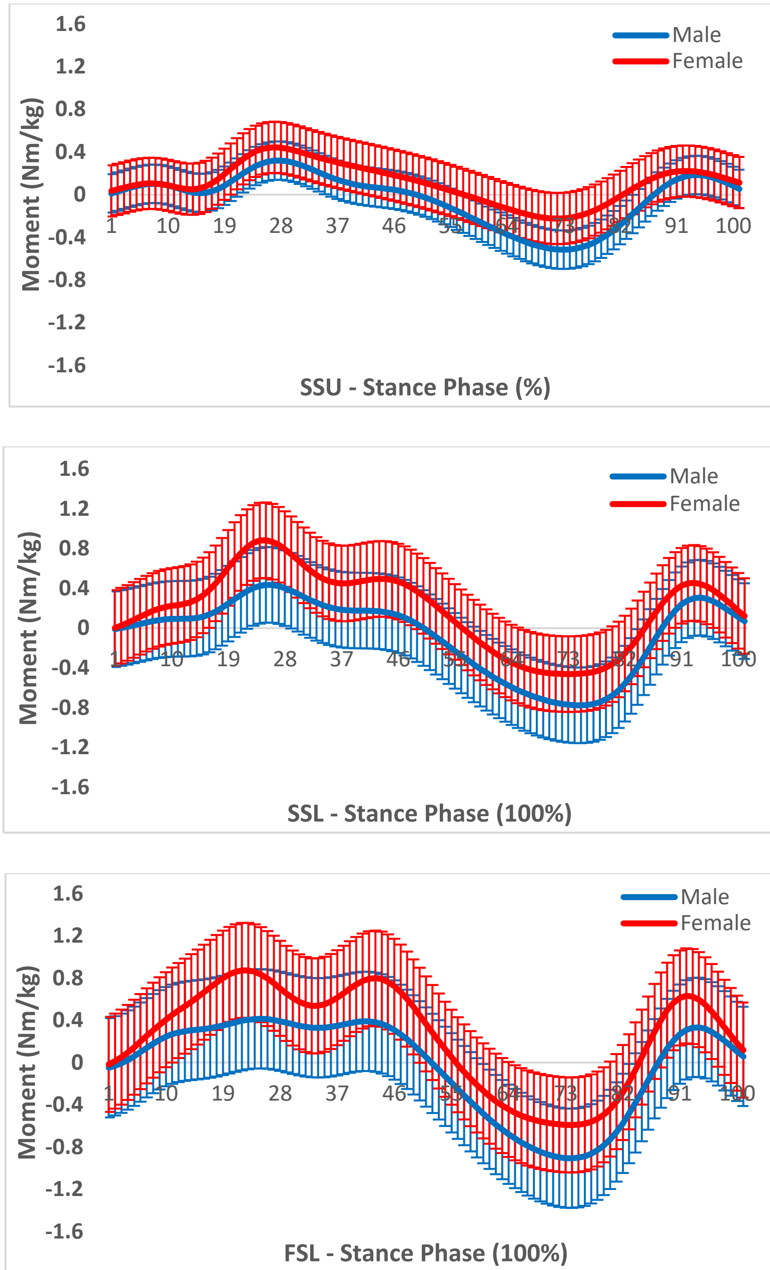


Figure 5.4: ***Stance phase sagittal plane (KFM) ensemble averaged waveforms for male (n=14) and female (n=10) participants for SSU (top), SSL (middle), and FSL (bottom). Positive values are KFM, and negative values are KEM (error bars – standard deviation)***

Descriptive statistics for the Peak KFM for males and females for each walking condition are presented in Table 5.10. There were significant pairwise differences between sex within walking conditions based on the 95% CI of the difference. The female mean Peak KFM was greater than the mean for males within the SSU and FSL conditions (Table 5.10). Between sex differences are illustrated in Figure 5.5.

Table 5.10 ***Descriptive statistics of Peak KFM for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking Condition	Sex	M (SD)	95% CI Lower	95% CI Upper	Mean Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU	Male	0.33 (0.20)	0.21	0.44	-0.24	***-0.48***	***-0.01***
(Nm/kg)	Female	0.57 (0.33)	0.32	0.83			
SSL	Male	0.57 (0.37)	0.35	0.78	-0.48	-1.03	0.06
(Nm/kg)	Female	1.05 (0.89)	0.42	1.69			
FSL	Male	0.60 (0.39)	0.38	0.83	-0.52	***-0.95***	***-0.08***
(Nm/kg)	Female	1.12 (0.63)	0.67	1.57			

Bold CI indicate significant differences between sex within condition

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.11. For male participants, there were statistically significant differences across conditions based on the 95% CI, with the mean Peak KFM for SSL and FSL being greater than the mean for SSU. Female participants had greater mean Peak KFM for FSL and than SSU. These significant pairwise differences are graphically illustrated in Figure 5.5.

Table 5.11: ***Confidence intervals for the mean pairwise differences in Peak KFM among walking conditions for male (n=14) and female (n=10) participants***

Sex	Compared Conditions *(Nm/kg)*	Mean Difference	95% CI of the Difference Lower	Upper
Male	SSU-SSL	-0.24	**-0.43**	**-0.05**
	SSU-FSL	-0.28	**-0.49**	**-0.07**
	SSL-FSL	-0.04	-0.18	0.10
Female	SSU-SSL	-0.48	-1.04	0.09
	SSU-FSL	-0.55	**-0.96**	**-0.14**
	SSL-FSL	-0.07	-0.45	0.32

Bold CI indicate significant differences between conditions within sex

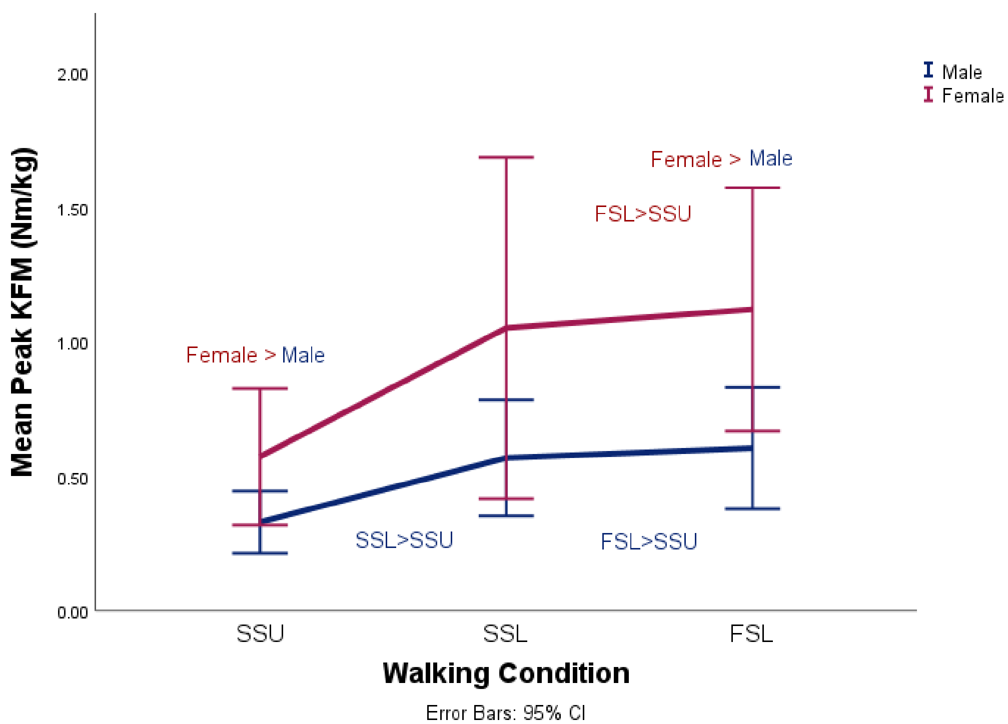


Figure 5.5: ***Mean Peak KFM for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females. Significant between sex within condition differences are indicated***

Descriptive statistics for the KFM-KEM for males and females for each walking condition are presented in Table 5.12. There were no significant pairwise differences between sex within walking conditions based on the 95% CI of the mean differences in Table 5.12.

Table 5.12: ***Descriptive statistics of KFM-KEM for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking Condition	Sex	M (SD)	95% CI Lower	95% CI Upper	Mean Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU	Male	0.93 (0.27)	0.78	1.09	0.01	-0.22	0.24
(Nm/kg)	Female	0.92 (0.28)	0.72	1.12			
SSL	Male	1.44 (0.38)	1.22	1.66	-0.14	-0.48	0.21
(Nm/kg)	Female	1.58 (0.43)	1.27	1.89			
FSL	Male	1.61 (0.36)	1.40	1.82	-0.20	-0.51	0.10
(Nm/kg)	Female	1.81 (0.35)	1.57	2.06			

No significant between sex differences based on CI of the mean difference

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.13. For male and female participants, there were statistically significant differences based on the 95% CI, within conditions with the mean KFM-KEM for FSL and SSL being greater than the mean for SSU (Table 5.13). These significant differences are graphically illustrated in Figure 5.6.

Table 5.13: ***Confidence intervals for the mean pairwise differences in KFM-KEM among walking conditions for male (n=14) and female (n=10) participants***

Sex	Compared Conditions *(Nm/kg)*	Mean Difference	95% CI of the Difference Lower	Upper
Male	SSU-SSL	-0.51	***-0.69***	***-0.34***
	SSU-FSL	-0.68	***-0.87***	***-0.48***
	SSL-FSL	-0.16	-0.33	0.01
Female	SSU-SSL	-0.66	***-1.01***	***-0.26***
	SSU-FSL	-0.89	***-1.27***	***-0.52***
	SSL-FSL	-0.24	-0.56	0.08

Bold CI indicate significant differences between conditions within sex

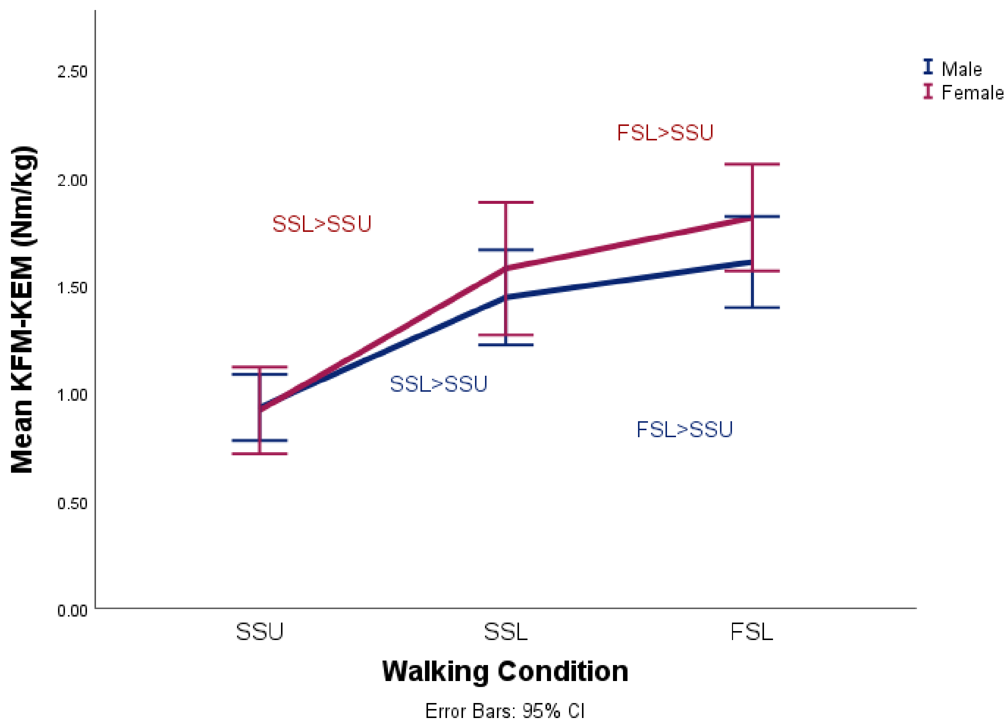


Figure 5.6: ***Mean KFM-KEM for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females***

5.4 Male and female RPE, MCU, and peak trunk angle during walking conditions

Descriptive statistics for RPE for males and females for each walking condition are presented in Table 5.14. There were no significant pairwise differences between sex within walking conditions based on the 95% CI of the mean differences in Table 5.14.

Table 5.14: ***Descriptive statistics for RPE for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking condition	Sex	M (SD)	95% CI Lower	95% CI Upper	Mean Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
FSL	Male	7.86 (2.07)	6.66	9.05	0.86	-0.56	2.27
	Female	7.00 (0.67)	6.52	7.48			
SSL	Male	11.21 (2.26)	9.91	12.52	-0.49	-2.15	1.18
	Female	11.70 (1.34)	10.74	12.66			
FSL	Male	11.86 (2.48)	10.43	13.29	-1.04	-2.82	0.74
	Female	12.90 (1.29)	11.98	13.82			

No significant between sex differences based on CI of the mean difference

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.15. For male participants, there were statistically significant differences, based on the 95% CI, across conditions with the mean RPE for SSL and FSL being greater than the mean for SSU (Table 5.15). Female participants had greater mean RPE for FSL than SSL and SSU; the female mean RPE for SSL was greater than for SSU (Table 5.15). These significant differences are graphically illustrated in Figure 5.7.

Table 5.15: ***Confidence intervals for the mean pairwise differences in RPE among walking conditions for male (n=14) and female (n=10) participants***

Sex	Compared Conditions	Mean Difference	95% CI of the Difference Lower	Upper
Male	SSU-SSL	-3.36	***-4.34***	***-2.38***
	SSU-FSL	-4.00	***-5.15***	***-2.85***
	SSL-FSL	-0.64	-1.33	0.04
Female	SSU-SSL	-4.70	***-5.94***	***-3.46***
	SSU-FSL	-5.90	***-7.17***	***-4.63***
	SSL-FSL	-1.20	***-2.34***	***-0.60***

Bold CI indicate significant differences between conditions within sex

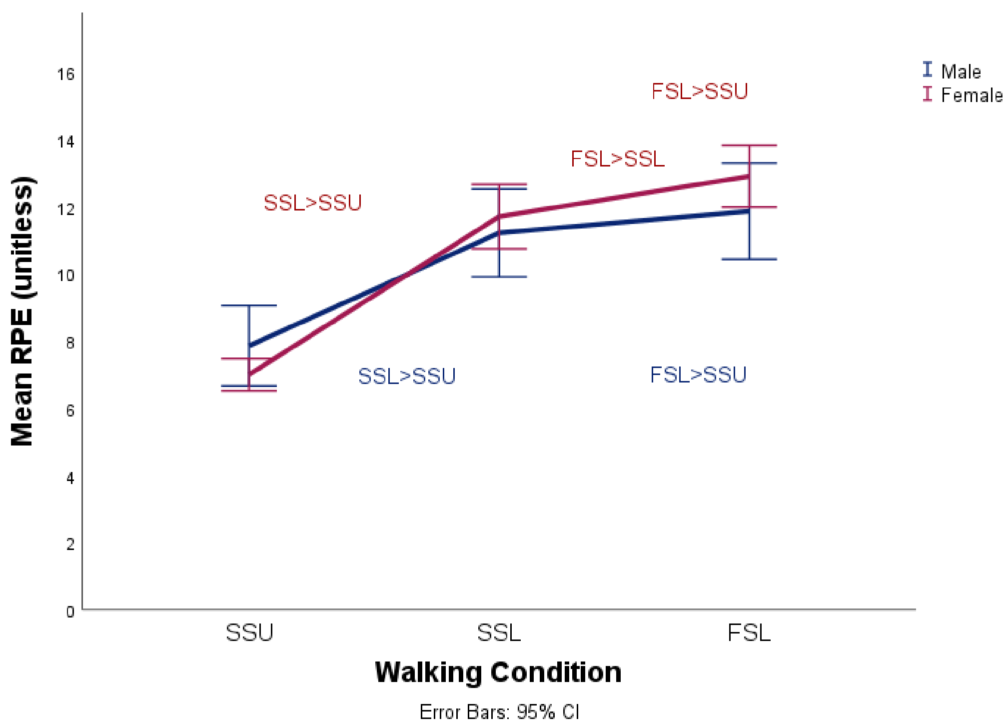


Figure 5.7: ***Mean RPE for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females***

Descriptive statistics for MCU for males and females for each walking condition are presented in Table 5.16. There were significant pairwise differences between sex within each walking condition (i.e., SSU, SSL, and FSL) based on the 95% CI of the mean differences in Table 5.16. Females had significantly larger MCU than males for all three conditions with values over two times greater than the male value for the same condition. Significant differences are displayed in Figure 5.8.

Table 5.16: ***Descriptive statistics for MCU for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking condition	Sex	M (SD)	95% CI Lower	95% CI Upper	Mean Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU (unitless)	Male	0.16 (0.11)	0.10	0.23	-0.18	***-0.31***	***-0.05***
	Female	0.34 (0.20)	0.20	0.49			
SSL (unitless)	Male	0.29 (0.23)	0.16	0.42	-0.33	***-0.58***	***-0.07***
	Female	0.62 (0.37)	0.35	0.88			
FSL (unitless)	Male	0.30 (0.21)	0.17	0.42	-0.34	***-0.54***	***-0.15***
	Female	0.64 (0.24)	0.47	0.82			

Bold CI indicate significant differences between sex within condition

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.17. For male and female participants, there were statistically significant based on the 95% CI differences across conditions with the mean MCU for SSL and FSL greater than the mean for SSU (Table 5.17). These significant differences are graphically illustrated in Figure 5.8.

Table 5.17: ***Confidence intervals for the mean pairwise differences in MCU among walking conditions for male (n=14) and female (n=10) participants***

Sex	Compared Conditions (unitless)	Mean Difference	95% CI of the Difference Lower	Upper
Male	SSU-SSL	-0.13	***-0.24***	***-0.02***
	SSU-FSL	-0.13	***-0.24***	***-0.03***
	SSL-FSL	-0.01	-0.08	0.07
Female	SSU-SSL	-0.27	***-0.51***	***-0.03***
	SSU-FSL	-0.20	***-0.45***	***-0.14***
	SSL-FSL	-0.03	-0.20	0.15

Bold CI indicate significant differences between conditions within sex

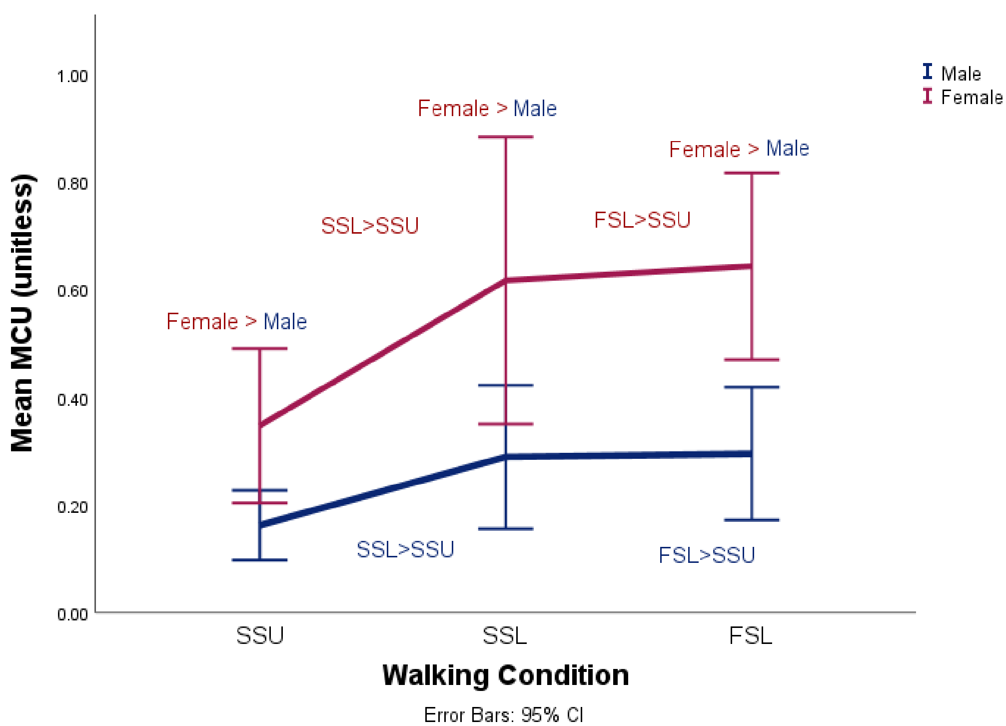


Figure 5.8: ***Mean MCU for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females. Significant between sex within condition differences are indicated***

Descriptive statistics for peak trunk angle for males and females for each walking condition are presented in Table 5.18. There was a significant pairwise difference between sex within walking condition based on the 95% confidence intervals; females had a greater mean peak trunk angle during SSL than males. This significant difference is displayed in Figure 5.9.

Table 5.18: ***Descriptive statistics for peak trunk angle for male (n=14) and female (n=10) participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions***

Walking condition	**Sex**	**M (SD)**	**95% CI Lower**	**95% CI Upper**	**Mean Difference**	**95% CI of the Difference Lower**	**95% CI of the Difference Upper**
SSU (degrees)	Male	4.6 (2.5)	3.2	6.0	-2.2	-4.8	0.5
	Female	6.8 (3.7)	4.1	9.4			
SSL (degrees)	Male	13.7 (3.9)	11.4	15.8	-4.6	***-8.8***	***-0.4***
	Female	18.3 (6.1)	13.0	22.6			
FSL (degrees)	Male	16.3 (5.4)	13.1	19.4	0.2	-4.0	4.4
	Female	16.0 (4.0)	13.1	18.9			

Bold CI indicate significant differences between sex within condition

The 95% CI for the within sex pairwise comparisons among walking conditions are in Table 5.19. For male and female participants, there were statistically significant differences based on the 95% CI, across conditions with the mean peak trunk angle for SSL and FSL greater than the mean for SSU (Table 5.19). These significant differences are graphically illustrated in Figure 5.9.

Table 5.19: ***Confidence intervals for the mean pairwise differences in peak trunk angle among walking conditions for male (n=14) and female (n=10) participants***

Sex	**Compared Conditions (degrees)**	**Mean Difference**	**95% CI of the Difference**	
			Lower	**Upper**
Male	SSU-SSL	-9.1	***-10.7***	***-7.4***
	SSU-FSL	-11.6	***-15.1***	***-8.2***
	SSL-FSL	-2.6	-5.4	0.3
Female	SSU-SSL	-11.5	***-16.9***	***-6.2***
	SSU-FSL	-9.2	***-13.4***	***-5.1***
	SSL-FSL	2.3	-5.8	10.4

Bold CI indicate significant differences between conditions within sex

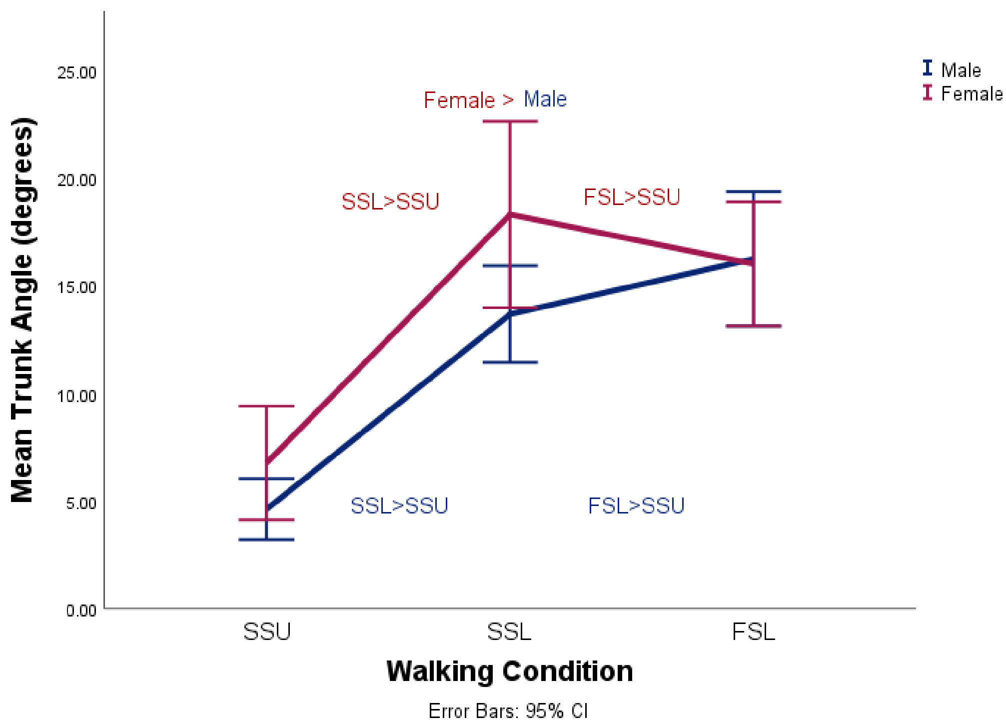


Figure 5.9: ***Peak trunk angle (degrees) for males (n=14) and females (n=10) for each walking condition. Significant pairwise differences among conditions are indicated for males and females. Significant between sex within condition differences are indicated***

5.5 Summary – sex comparisons – primary and secondary outcomes

5.5.1 Summary for primary outcomes – Objective 2

In summary, both KAM features were not significantly different between sexes based on the CI of the mean differences. There were significant between condition differences for both KAM Impulse and Peak KAM but the between condition differences were not the same for males and females. The KAM impulse for both loaded conditions were greater than the unloaded condition in males, but only the SSL was greater than the SSU for female participants. For the Peak KAM, males had significant differences among all conditions whereas the females had greater values for the two loaded conditions versus the unloaded condition, but not between the two loaded conditions. There were significant differences between sex and between conditions for the Peak KFM. Females had greater Peak KFM values for SSU and FSL than males, but males had greater Peak KFM for both loaded conditions than the unloaded condition whereas only the FSL condition was greater than the unloaded condition in females. The KFM-KEM feature had significant differences between conditions within sex only and these were the same for both sexes where the loaded conditions were greater than the unloaded conditions.

5.5.2 Summary for secondary outcomes – Sub-objective 2a and 2b

The RPE values did not differ between sexes but the differences between conditions were not the same for males and females. For both males and females, the loaded conditions had greater ratings than the unloaded conditions but only females rated the FSL greater than the SSL condition.

The MCU values were significantly different between sexes for all walking conditions with females having over two times greater mean MCU than males within each condition. There were significant within sex differences between conditions and these differences were the same for males and females where the two loaded conditions were higher than the unloaded condition.

The peak trunk angles were significantly different between sexes for the SSL walking condition where females had a greater peak trunk angle indicative of more trunk flexion than males. Between conditions both males and females had significantly greater forward flexion for the two loaded conditions than the unloaded condition.

5.6 Discussion

The aim of Objective 2 was to provide preliminary data on potential differences in biomechanical adaptations during load carriage with operationally relevant load and speed between male and female military members that help to advance understanding of the risks for clinical knee OA development and progression for military members. The recruitment aims for this objective (i.e., recruit 15 males and 15 females) were not achieved within the study period, however, significant findings were found that partially support the study hypotheses, and the data can inform sample size and power calculations for future studies on sex differences in response to the load carriage tasks. An overview of participant enrollment, exclusion, and withdrawal is in Appendix H. The CAF is comprised of 85% males and 15% females, but, despite this population difference, the rates of recruitment for males and females were similar at just over one per week for males and just under one per week for females. In a recent systematic review of 20 military load carriage studies, only one study included female military members (n=18) and no between sex comparisons were made [40]. In one study [41], the authors stated that it was impossible to recruit sufficient numbers of female members on which to draw robust scientific conclusions indicating that enrollment rates of female members may limit their inclusion in military load carriage studies for some nations. While a smaller number of females were recruited, the recruitment rates in the current study were encouraging given the population size, distribution, and interest in the study shown by all who inquired. This study is one of the first to compare differences in load carriage between male and female military members and the first, based on the current literature, to investigate differences in knee joint moment features related to the progression of medial compartment knee OA using an operationally relevant load and walking speed.

Both male and female participants in this study were experienced military members based on mean age and years of military service. Consistent with the general population [3, 33] male participants were on average taller than females (Table 5.1). The only other demographic feature that differed between sexes was the higher BMI values in females than males. While statistically significant, the BMI for females was only slightly above the healthy BMI range [3]. The BMI values calculated for males were consistent with those previously reported for CAF members while the BMI calculated for females was slightly higher [3]. Higher BMI values have been associated with negative health outcomes and chronic conditions, however, categories for BMI should be interpreted with caution as BMI has limitations in accurately reflecting percent fat in certain populations including those with lean or muscular builds [3]. Due to the physical fitness requirements of military service, interpretation of BMI scores may be limited for this study's participants. To better understand the implications of BMI scores for future studies, collecting participant annual fitness test incentive level scores could be considered. Incentive level scores provide a percentile ranking of an individual CAF members' fitness level compared to an overall standard and may provide additional context to better understand potential between and within sex differences related to load carriage.

Consistent with previous literature [90, 135, 203, 204], males had greater absolute and body mass normalized KE muscle strength and greater absolute KF muscle strength than females, although normalized KF strength was not different between sexes. These differences in height, BMI, and muscle strength, could be potential covariates for some of the variables examined. For example, the normalized KE muscle strength difference between sexes is used in the calculation of MCU to assess quantitative effort [93] during the walking conditions and height has been linked to changes in step length during load carriage [33]. Given the sample size for each sex in this study, CIs were used to examine pairwise comparisons and covariates cannot be easily accounted for in this analysis. Inclusion of covariates for future studies could be considered, but with caution, as they can mask true differences in tested variables between groups [205].

SS walking speed for male and female participants in the current study was not different, consistent with a descriptive meta-analysis by Bohannon et al. [129] that found no between sex differences for walking speed. All participants were able to maintain the fixed speed, and there was no difference in the duration of stance phase between sexes during each condition. Since walking speed has been linked to changes in both frontal and sagittal plane moment features [70], and KAM impulse calculation includes the duration of stance phase [59], it was important for interpreting the joint moment results to determine whether differences in

SS walking speed or stance phase duration between sexes were present. Based on these results, walking speed and stance phase duration would not explain sex differences in primary or secondary outcomes.

Few studies have included female military members to examine the effects of military load carriage on biomechanical features and comparisons between sex have mostly focused on spatio-temporal and kinematic features of gait [40, 52]. The current study provides preliminary data supporting sex differences for joint moment features related to the clinical progression of knee OA, self-reported and quantitative effort, and peak trunk angle, during load carriage with operationally relevant load and speed. Overall, this provides evidence for including male and female military members in load carriage studies and conducting a sex analysis in studies that examine responses in gait biomechanics features linked to risk of medial compartment knee OA during load carriage tasks by military members. The discussion below places the results into context related to the current literature and highlights the new knowledge gained on the effect of load carriage on risk factors for medial knee joint OA.

5.6.1 Knee adduction joint moment features linked to knee OA progression

The KAM impulse and Peak KAM for each condition (Table 5.6 and 5.8) were not different between sex based on the 95% CI of the mean difference. These findings for no between sex differences during unloaded walking for KAM features are consistent with reported findings for healthy individuals [122] and those with moderate knee OA [133-135] whereas males have been shown to have greater Peak KAM and KAM magnitudes than females for participants with severe knee OA [133, 134]. The distribution of the KAM impulse CIs for the between sex comparisons for all three conditions were equally distributed around the mean difference; the combination of magnitude and stance duration resulted in a similar loading exposure between sexes.

Some of the between walking condition comparisons for KAM features that were not the same for males and females (i.e., KAM impulse for SSU-FSL and Peak KAM for SSL-FSL) did not have CI distributions that were equally distributed about the mean. Sample size calculations, using α set at 0.05, power at 0.8, were performed for these features. These calculations showed samples sizes ranging from 20 females, to detect differences in KAM impulse for the SSU-SSL comparison, to 199 for Peak KAM to detect a significant difference for females between the loaded conditions (SSL-FSL). The findings for the KAM features partly support the hypothesis for Objective 2 and provide mean difference and variance data to calculate effect size and sample sizes for future studies.

The male Peak KAM values were slightly higher than the female values for the SS speed conditions but were almost 10% higher for the FS condition and the CI was not equally distributed around the mean difference (Table 5.8). The greater increase in Peak KAM for the faster walking speed for males suggests a different biomechanical response which was a systematic increase across conditions (Figure 5.3). In contrast, the load (SSL) increased the Peak KAM in females with only a small increase related to increased walking speed (FSL) (Table 5.8). A difference in response to load and speed between sexes has not been previously reported in military load carriage literature [11, 12]. Examining whether males have greater Peak KAM than females for the FSL condition would provide insight on risk for medial compartment knee OA initiation and progression associated with the addition of speed in males.

5.6.2 Knee flexion joint moment features linked to knee OA progression

The only significant difference between sex in knee joint moment features linked to medial compartment knee OA progression was the greater Peak KFM in females compared to males for the unloaded (SSU) and the FSL conditions (Table 5.10) and there were differences among conditions within sex between males and females; these findings partially support the hypothesis for Objective 2. For females, only the FSL condition had a higher Peak KFM than the unloaded condition whereas males had higher values for both loaded conditions, suggesting that the added speed resulted in a slightly larger response in female members. The Peak KFM was almost two times greater in the females compared to males for all conditions and the non-significant finding for the SSL condition might be related to the small sample size and high variability resulting in low statistical power. A sample size calculation (α=0.05, power=0.8), using the mean difference and variance between sex for Peak KFM for the SSL condition, determined that 28 participants per group (56 total) were needed to detect a significant difference between males and females for that condition. For the SSL-SSU comparison of Peak KAM for females, a sample size calculation indicated that 20 females would be required to detect a significant difference based on 95% CI of the mean difference. Greater Peak KFM indicates greater KE muscle force for females, compared to males during operationally relevant load carriage, and may represent a higher risk for knee medial compartment knee OA development in female military members. The KE MCU values below (Section 5.6.3) support that females utilize greater relative KE muscle effort compared to males to produce the needed early stance Peak KFM.

A finding of greater Peak KFM during the first half of stance, for females compared to males, during unloaded walking differs from other studies [122, 130, 132] of healthy participants and a study of participants with moderate knee OA [135] that found that females had lower KFM amplitude compared to males. A 2022 scoping review [52] of 18 articles examining the biomechanical effects for female military member load carriage did not include any studies that compared KFM magnitudes between female and male military members using operationally relevant load and speed. This study is one of the first to examine this KFM feature in male and female military members.

While the relationship between Peak KFM and structural medial compartment knee OA progression is less clear in the literature [61, 66], KFM magnitude is a predictor of overall joint loads, provides an indication of the KE muscle moment [128] and is associated with worse cartilage health for the patellofemoral joint [206]. Higher magnitude loading due to KE muscle activity can lead to imbalance in cartilage metabolism resulting in increased catabolism [5, 101]. Repeat loading via muscular contractions are associated with increase chondrocyte death *in vitro* [75] with explant studies showing that repeated exposure to high magnitude loading negatively influences biosynthesis, enhances degradation, and yields pro-inflammatory responses in cartilage [94]. Eccentric cyclic loading, as with the KE muscles during the first half of stance, has been linked with increased cell death compared to concentric cyclic loading [75]. However, the association between knee joint OA prevalence and repetitive loading during activity *in vivo* (e.g., running) is not clear [207, 208]. These preliminary data support that Peak KFM should be examined in load carriage studies and that a sex analysis would be appropriate.

Although a less dynamic gait pattern with smaller KFM-KEM [125] was hypothesized for operationally relevant load carriage, the KFM-KEM differences were greater for the loaded than unloaded conditions for both males and females indicating that both sexes adopted more dynamic gait patterns in response to operationally relevant loads. Findings of between sex differences in KFM-KEM have been reported in populations with moderate and severe knee OA and a lower difference is predictive of a higher risk of clinical progression [63]. Participants with lower KFM-KEM differences may not have been included in this study based on criteria with respect to age, musculoskeletal health, and military medical employment limitations. The non-significant between sex comparison for FSL may be the result of the small sample size and low power. Sample size calculations for sex differences provided a range indicating that 40 males and 40 females are required to detect a significant between sex difference for FSL and 122 per sex group for the SSL condition.

For the between walking condition comparison for KFM-KEM the CIs for SSL-FSL of both males and females were not equally distributed about the mean difference. Sample size estimate calculations determined that 36 males and 36 females would be required to detect a significant difference for this comparison. The findings for the KFM-KEM features partly support the hypothesis for Objective 2 and provide mean difference and variance data to calculate effect size and sample sizes for future studies.

The sagittal plane Peak KFM means in Table 5.10 and the ensemble averaged waveform (Figure 5.4) show that the KFM magnitude for females is higher than that of males while the KEM magnitude for males is greater than that of females. The net KFM-KEM differences between sexes does not capture this nuance for the between sex differences. This increased KFM-KEM difference with load and speed for both sexes is consistent with studies included in the systematic review by Walsh et al. [40] and a study by Silder et al. [84] that reported increased KE muscle (early Peak KFM) and PF muscle activity (late Peak KEM) during load carriage. While a study of a mixed civilian-military sample of males and females identified reduced sagittal plane knee excursion in females [80] using FS and operationally relevant load, this is the first study to identify between sex differences in sagittal knee joint moment features during load carriage with operationally relevant load and speed.

A scoping review by Wendland et al. [52] found limited comparison of male and female military members during load carriage. None of the 18 studies included in that review compared KAM and KFM features in males and female military members during operationally relevant load carriage. Two studies that are most similar to the current study include one that examined effect of load and speed on joint moments measures in civilian females [7] and the other examined male members of the Australian military [41]; neither made between sex comparisons. In the former [7], there was an increase in the KAM and KFM percent contribution to the knee joint total load calculation when loads were increased based on percent body mass during forced marching at a FS that was relative to their gait transitional speed to a run [7]. Only general comparisons can be made with this study due to reporting relative and not absolute values for the KAM. Additionally, the FS walking speeds used were relative, presumably, total joint load would include an increase in Peak KFM as found in the current study. The current study provides the comparative data between sexes.

Only the Peak KFM in the first half of stance was measured in this study, given it is the most common KFM variable reported in the OA literature [59, 68]. However, as seen in Figure 5.4, there were qualitative differences in the sagittal plane ensemble averaged waveforms for males and females. The FSL ensemble averaged waveform (Figure 5.4) for females was characterized by two distinct peaks prior to midstance and a distinctive late stance peak that was not as visually evident in the ensemble average waveform for males. The implication of two similar peaks during the first half of stance is that females may be exposed to their maximal and near maximal Peak KFM twice prior to mid-stance, which could lead to more sustained loads that can lead to an imbalance in cartilage metabolism resulting in increased cartilage catabolism [5, 101]. In addition to the second peak in the first half of stance, a third Peak KFM, observed during late stance for the FSL sagittal plane waveform of females, as depicted in Figure 5.4 for FSL, has been associated with progression of patellofemoral OA on MRI [200]. These waveform peaks suggest that more robust forms of waveform pattern analysis (e.g., principal component analysis) would help put the joint moment findings into perspective in relation to risk for knee OA among female military personnel. Principal component analysis is a statistical pattern recognition technique that has been used to examine features of joint moment waveforms [117, 125, 133, 209].

This is one of the first studies to compare between sex for military members using operationally relevant load and speed. The between sex and between condition differences provide preliminary data on sex differences in joint moment features related to the clinical progression of knee OA (i.e., KAM impulse, Peak KAM, Peak KFM and KFM-KEM) within and between walking conditions.

5.6.3 Self-reported and quantitative measures of effort

The findings for the secondary measures of self-reported and quantitative effort partially support hypothesis 2a. The self-reported RPE increased in a systematic manner with load and walking speed for females but only with load for males (Table 5.13). Despite similar self-reports of effort between sexes, the KE muscle effort (MCU) for females was almost double that of males for all conditions (Table 5.15). The body mass normalized KE muscle strength was significantly higher for males than for females and the higher KFM in females contribute to the MCU sex differences. This finding supports previous reports indicating that load carriage was more physiologically demanding for females [55]. The between condition differences for MCU were the same for males and females (i.e., loaded conditions greater than unloaded). KE muscle effort was influenced by the load carried while increased walking speed did not require an increase in MCU. While a previous systematic review of military load carriage included studies that identified increased KE muscle activity using EMG [40] this is one of the first studies to identify between sex differences for KE muscle effort during military load carriage using an objective measure of KE muscle effort [52].

MCU has been used as an objective measure of KE muscle effort for military load carriage and calculates effort at the point in the gait cycle where Peak KFM occurs [93]. In a CAF report by Hebert et al. [90], CAF members had muscle recruitment utilization (i.e., MCU) value ranges that were the same at the lower end but lower at the upper end of range than the range of values for this study. Differences in KE muscle strength collection limit comparisons between the two studies.

The greater magnitude of the male KEM in response to load and speed (FSL) suggests that KE muscle MCU may not capture the overall quantitative effort of males due to differences in muscle patterns during load carriage with operationally load and speed. As discussed above, the ensemble averaged waveform for females walking with load at an operationally relevant speed (i.e., FSL) was associated with a double peak for KFM. Higher MCU at a single time point in stance phase may not reflect the overall demands on the KE muscles during load carriage and greater MCU in one muscle group may not reflect overall effort. Monitoring muscle activity signals from multiple muscle groups during load carriage by surface EMG may provide a useful alternative for assessing effort objectively [210].

In this study, MCU for the KE muscles was approximately doubled for males and females during the loaded conditions compared to unloaded walking while the self-reported effort for males and females during loaded conditions (Table 5.13) reflect a perceived effort that is between light and somewhat hard based on the BORG scale of RPE (Appendix G). A discordance between RPE and quantitative effort in CAF members during load carriage tasks has been previously observed [90], however, based on the method used to measure quantitative effort for this study, drawing conclusions about between sex differences in discordance of self-reported and quantitative efforts is limited. Hébert et al., [90] postulated that CAF members may have difficulty distinguishing between factors that may influence how they perceive their effort during load carriage (e.g., pain, shortness of breath, exercise intensity) and may focus more or less on one aspect of effort or that unknown factors may contribute to discordance between RPE and objective measures of effort in CAF members. Loaded marching is a common military task that typically involves longer duration load carriage, without a break, than what was used for this study. The short duration of the study's loaded conditions, compared to typical training and operational tasks, may have influenced the participants' RPE responses. The increased MCU for females compared to males and the trend of CAF members under reporting effort [90] are important considerations for planning physical training, designing rehabilitation programs, and for military leadership, when evaluating the physical demand of training and operational tasks.

5.6.4 Peak trunk angle

In this study, a difference in peak trunk angle between males and females was observed for SSL (mean speed 1.30m/s) with females having a greater forward trunk lean than males during that condition. The between condition differences were the same for males and females with peak trunk angle being greater for the loaded than for the unloaded condition. These findings partly support hypothesis 2b. This forward trunk lean adaptation, thought to be a response to posterior load that enables the centre of mass to be maintained in position [40] has been reported consistently in military load carriage studies and is associated with increased posterior muscle activity (e.g., erector spinae, hip extensor, PF) [40]. The findings for this study are not entirely consistent with other studies' findings of greater peak forward trunk angle for females compared to males while carrying equal mass loads [47]. Sex differences in trunk angle measures during load carriage have been attributed to a typically lower body mass for females [47, 55], however, in this study there was no difference in body mass between males and females. Krupenevich et al., [47] reported that females adopted greater forward trunk angles compared to males while carrying load at speeds similar to the FSL condition (i.e., both were at approximately 1.50m/s) and discussed that further analysis of participant gait and anthropometric features showed that forward trunk lean was inversely related to participant body mass [47]. The results from Krupenevich et al. [47], in part explain the findings for the faster speed loaded condition but do not account for the between sex difference for the slower speed loaded condition. There were methodological differences in the Krupenevich et al. [47] study as they used a mixed cohort of military and non-military participants and load carriage was conducted on a raised walkway with force plates as opposed to an instrumented treadmill. Other methodological differences related to trunk angle calculations limit comparisons between studies.

With consideration given to the limitations below, the results of this comparison indicate that there are differences between sexes and between conditions within sexes for specific primary and secondary outcomes across the load carriage conditions with operationally relevant load and speed. Significant between sex differences for Peak KFM during load carriage with operationally relevant load and speed, and during unloaded walking at self-selected speed, were identified. A larger sample size would likely have found significant between sex differences for Peak KFM for all conditions. There were differences in the between condition comparisons that were not the same for both sexes. One of these comparisons, Peak KAM for females for SSL-FSL, was unlikely to be detected as significant with a larger sample size. The systematic increase in Peak KAM across conditions for males provides evidence that male military members may have a different risk for structural progression of knee OA than females [133].

The analysis of secondary outcomes found that females had greater utilization of the KE muscles across all walking conditions and that female participants reported greater RPE during load carriage with operationally relevant speed. Furthermore, some of the differences displayed in the ensemble averaged waveform shapes for males and females with operationally relevant load and speed (FSL) were not fully explained using discrete measures. There is evidence, based on findings of higher KFM, lower absolute and normalized KE muscle strength, and higher MCU and RPE during loaded walking, that females may be at greater risk for KE muscle fatigue than males during operationally relevant load carriage. In addition to increased overall knee joint loads with higher KFM [66, 128], KE muscle fatigue has been associated with increased KAM, changes to the knee joint loading environment that may increase risk for OA development [85], and is a risk factor for injuries [86, 87] that are high-risk for knee OA [9, 85].

These findings add to the current knowledge on sex differences in knee joint moment features linked to medial compartment knee OA development and progression. This comparison supports a fully powered study comparing sex and load carriage with operationally relevant load at self-selected and operationally relevant speed. Consideration should be given to including evaluation using other methods (e.g., PCA), in addition to discrete measures, and measuring muscle activity (EMG) for KE and PF muscles.

5.6.5 Limitations

In addition to the limitations discussed in Chapter 4, Section 4.6.3, there were limitations related to the male – female comparison during loaded walking with operationally relevant load and speed. This study was exploratory; the sample was not based on a sample size estimate as there were no previous studies to utilize to estimate mean differences and variance in the data to perform this estimate.

5.6.6 Conclusion

These preliminary data provide evidence of sex differences in joint moment features previously linked to medial compartment knee OA progression. The mean difference and variance estimates support a study on a larger sample to examine between sex and condition differences and interactions in joint moments related to operationally relevant load carriage tasks. For the secondary outcomes, the key finding is that the KE muscle effort (MCU) was greater for females than males across all conditions and both increase MCU in response to load but not speed.

Chapter 6. Conclusion

Chapter 6 provides a summary of the key findings, impact, significance for stakeholders, and an overall conclusion related to the overall goal and specific objectives associated with the findings presented in Chapters 4 and 5.

6.1 Summary of key findings

The overall goal of this thesis was to advance understanding of the risks for medial compartment knee OA development and clinical progression by examining the biomechanical adaptations associated with an operationally relevant load carriage task in military members and to provide preliminary data on whether differences in these adaptations exist between male and female military members. To achieve the study's goal, specific objectives were established to determine whether there are differences among unloaded and loaded walking conditions on joint moment features related to the progression of knee OA (Objective 1) and to provide preliminary data on sex differences (Objective 2). Sub-objectives were established to better understand the effect of walking condition on exertion (Sub-objective 1a) and sagittal plane trunk angle (Sub-objective 1b), and to provide preliminary data on sex differences in exertion and trunk angle (Sub-objectives 2a and 2b).

These findings provide evidence that knee moment features related to medial compartment knee OA development and progression, measures of effort, and peak-trunk angle, are altered for military members walking with an operationally relevant load and some features are altered differently with a fixed, faster, operationally relevant, walking speed. Secondly, this study provides preliminary data that specific joint moment features differ between male and female military members during load carriage tasks and that some responses differed among walking conditions between sexes.

6.1.1 Summary of key findings Chapter 4 (Objective 1, Sub-objectives 1a and 1b)

A summary of the key findings from Chapter 4 is presented in Table 6.1. All primary and secondary outcomes were greater for the loaded conditions compared to the unloaded condition. All the primary joint moment were greater with load, but only the Peak KAM and KFM-KEM were greater with the addition of an operationally relevant speed and KAM impulse was lower versus the self-selected speed loaded condition. For the secondary outcomes, KE muscle effort (MCU) and peak trunk angle were greater with load and did not increase at the fixed speed. Only perceived exertion (RPE) systematically increased across conditions and was greatest with the addition of an operationally relevant speed. All reported values are significant ($p<0.05$).

Table 6.1: ***Chapter 4 – summary of significant pairwise comparisons for primary and secondary outcomes***

Outcomes	Pairwise comparison	p
Primary outcomes		
KAM impulse (Nms/kg)	SSL > SSU	**< 0.001**
	FSL > SSU	**0.01**
	FSL < SSL	**< 0.001**
Peak KAM (Nm/kg)	SSL > SSU	**< 0.001**
	FSL > SSU	**< 0.001**
	FSL > SSL	**0.001**
Peak KFM (Nm/kg)	SSL > SSU	**0.03**
	FSL > SSU	**< 0.001**
KFM-KEM (Nm/kg)	SSL > SSU	**< 0.001**
	FSL > SSU	**0.01**
	FSL > SSL	**< 0.001**
Secondary outcomes		
RPE (unitless)	SSL > SSU	**< 0.001**
	FSL > SSU	**< 0.001**
	FSL > SSL	**0.002**
MCU (unitless)	SSL > SSU	**< 0.001**
	FSL > SSU	**< 0.001**
Peak trunk angle (degrees)	SSL > SSU	**< 0.001**
	FSL > SSU	**< 0.001**

6.1.2 Summary of key findings Chapter 5 (Objective 2, Sub-objectives 2a and 2b)

Summaries of the key sex comparison findings from Chapter 5 are presented in Table 6.2 and Table 6.3. For the primary outcomes, there were significant between sex differences for KFM during unloaded and FS load carriage and there were significant differences between conditions that were not the same for males and females. For the secondary outcomes (Table

6.3), there were significant between sex differences with females having greater MCU than males within all conditions, and females having greater peak trunk angle for the SSL condition. There were significant between condition differences that were not the same for males and females within each condition.

Table 6.2: ***Chapter 5 - summary* of *significant pairwise comparisons for between sex within condition and between condition within sex based on 95% CI of the mean difference for primary outcomes***

Primary outcome	**Pairwise comparison**	
	Within	**Between**
KAM impulse (Nms/kg)	Male	SSL > SSU*****
		FSL > SSU*****
	Female	SSL > SSU*****
Peak KAM (Nm/kg)	Male	SSL > SSU*****
		FSL > SSU*****
		FSL > SSL*****
	Female	SSL > SSU*****
		FSL > SSU*****
Peak KFM (Nm/kg)	SSU	Female > Male******
	FSL	Female > Male******
	Male	SSL > SSU*****
		FSL > SSU*****
	Female	FSL > SSU*****
KFM-KEM (Nm/kg)	Male	SSL > SSU
		FSL > SSU
	Female	SSL > SSU
		FSL > SSU

***** Indicates males and females had different between condition differences
****** Indicates between sex differences

For the secondary outcomes, there were significant between sex differences with females having greater MCU than males within all conditions, and females having greater peak trunk angle for the SSL condition. There were significant between condition differences that were not the same for males and females within each condition.

Table 6.3: ***Chapter 5 - summary* of *significant pairwise comparisons for between sex within condition and between condition within sex based on 95% CI of the mean difference for secondary outcomes***

	Pairwise comparison	
Secondary outcome	**Within**	**Between**
RPE (unitless)	Male	SSL > SSU*
		FSL > SSU*
	Female	SSL > SSU*
		FSL > SSU*
		FSL > SSL*
MCU (unitless)	SSU	Female > Male**
	SSL	Female > Male**
	FSL	Female > Male**
	Male	SSL > SSU
		FSL > SSU
	Female	SSL > SSU
		FSL > SSU
Peak trunk angle	SSL	Female > Male**
	Male	SSL > SSU
		FSL > SSU
	Female	SSL > SSU
		FSL > SSU

*   Indicates males and females had different between condition differences
** Indicates between sex differences

6.2 Implications

The biomechanical adaptations to load carriage found in this study include greater magnitude frontal and sagittal plane knee moment features supporting greater medial compartment and overall joint loading exposure [196]. Overall, these adaptations provide evidence of knee biomechanical outcomes that are linked to increased risk of knee OA development and progression [59, 66] for walking with operationally relevant load and speed. In addition, they provided preliminary data on differences in these biomechanical features between male and female military members during load carriage.

For the total sample, all knee joint moment features, and KE muscle effort, were increased with the addition of an operationally relevant load. Two of the joint moment features, Peak KAM and KFM-KEM, were further increased with the addition of an operationally relevant speed that was 0.25m/s greater than the average SS speed for the sample (Chapter 4). These increases in knee joint moment magnitudes while walking with load are consistent with gait features indicative of an increased risk for knee OA development and progression compared to walking with no load. Greater KE muscle effort (i.e., MCU) with loaded walking indicates higher physiological demand compared to walking with no load. The additional increases in Peak KAM and KFM-KEM indicates that speed of load carriage alters risk for knee OA development and progression. Both the mass of the load and speed of carriage are factors to consider when designing training and rehabilitation programs based on their effect on both knee joint moments and muscle utilization.

From Chapter 5, the findings for the knee joint moment features show that only Peak KAM and Peak KFM for male participants had the same between condition significant differences as those found for the total sample and that the findings for males and females were not the same. The only identified, significant, between sex difference for these features was for KFM (SSU and FSL). For the secondary measures, females had greater KE muscle effort (MCU) than the total sample and males for all conditions and the significant within condition comparisons were the same. All between condition differences for the total sample, males, and females, were the same for perceived exertion except the RPE for males was affected by load but not speed. Drawing general conclusions about joint moments, effort, or risk of development and progression of knee OA during load carriage tasks from studies including only one sex, mixed sex samples, or without analysing whether there are potential interactions, should be done cautiously.

The evidence from this study suggests exposure to high magnitude KAM, linked to knee OA clinical progression [63], and increases in magnitudes of features of gait linked with the development progression of knee OA [59, 65] during load carriage with operationally load and speed. In the absence of pathology, repeated cyclical loading can be beneficial to articular cartilage health [115] and moderate magnitude cyclic loading has been associated with anabolic responses in cartilage explants [94, 115]. However, studies have also demonstrated that applying either hyper-physiologic cyclical loading, or injurious levels of load, negatively influences biosynthesis, enhances degradation, and yields pro-inflammatory responses in cartilage [94, 115]. The findings from studies of the response of articular cartilage to various types, and magnitudes, of load, and the increase in the magnitudes of gait features related to knee OA clinical progression found in this study, indicate a risk for knee OA development and progression associated with operationally relevant load carriage.

An aspect of military load carriage that requires consideration, based on the biomechanical and physiological responses identified in this study, is the frequency of exposure to load carriage for military personnel [55]. While load carriage is commonly performed by members of the combat arms (e.g., infantry) and may be included regularly in the scope of their physical training, military training, and other occupational demands, members of supporter occupations (e.g., logistics) may not regularly include load carriage as part of their physical training. These members may be expected to perform load carriage with operationally relevant load and speed as part of their military duties or training without adequate time to comprehensively prepare, recover, and for body tissues to adapt. Operationally relevant load and speed may expose some military members to hyper-physiologic or injurious levels of cyclic knee joint loading [41, 83]. Furthermore, although military members may have higher levels of aerobic fitness and muscle strength that enable successful completion of load carriage tasks, despite insufficient training periods, specific body tissues (e.g., cartilage, bone) may not be prepared to adapt positively to higher magnitude knee moments [162]. Importantly, female military members are most often represented in units and military occupations less likely to regularly engage in load carriage; only 4.3% of CAF combat arms personnel were reported as being female in 2018 [211].

The physiological demands of load carriage, including greater muscle effort (e.g., KE muscles) due to load, speed, and biomechanical adaptation (e.g., forward trunk lean) may predispose some members to increased risk of muscle fatigue [44] which has been associated with changes in knee loading (e.g., increased KAM magnitude) related to knee OA development and progression [85, 212] or injuries that are high risk for future development of knee OA [85].

The total duration of the loaded conditions (i.e., SSL and FSL) for this study was 10 minutes whereas the duration of the Combat FORCE™ test is 5-6 times longer and does not include a break. Other training and operational load carriage requirements necessitate even greater durations of load carriage [50, 150]. The effect of longer duration, operationally relevant, load carriage on knee joint moment features and effort could be considered for future studies.

Overall, based on significant pairwise differences within sex across conditions for both sagittal and frontal plane measures, there is evidence that the biomechanical responses to increased load and speed are not the same for males and females. Notably, the findings for SS walking speed, stance duration, age, military experience, and body mass indicate that these variables do not account for the sex differences found in this study. While differences in height and muscle strength may influence biomechanical adaptations, these same sex differences are widely represented in both the military and civilian populations [3, 25, 33, 79].The findings of this study support other findings of sex differences in terms of biomechanical and physiological adaptations by male and female military members during load carriage [52].

The loads carried by military members result in increased knee joint load magnitudes, increased magnitudes of features of gait linked to clinical knee OA progression, and are physiologically demanding [55, 146]. The findings from this study, of increased magnitude frontal and sagittal knee joint moments and of features of gait linked to clinical knee OA progression, partly explain the occupational risk for knee OA associated with military service. The sex differences in sagittal plane findings and muscle utilization from this study may partly explain the differences in knee OA incidence between male and female military members. Military leaders should be cognizant of the loads borne by military members as part of operational planning and military training design and, in addition to ensuring adequate physical preparation, take steps to ensure the loads reflect operational necessity.

6.3 Limitations and Considerations

As with any study, interpreting the results must consider the effect of limitations. Knee joint moments are based on an inverse dynamic model that relies on calculations from external forces. Direct comparisons to internal joint contact forces are limited [189] as the inverse dynamic model assumes no antagonist muscle activity; without including muscle forces, joint moments may result in underestimation of joint contact forces. However, alternative methods are impractical (e.g., internal knee joint sensors) or require cumbersome calculations and assumptions (e.g., estimation of joint contact forces). The primary measures used for this study provide surrogate measures of joint loading [59, 68], are the most common metrics assessed [59] and have predictive validity [59, 63] for knee OA development and progression and as such are indicative of biomechanical risk factors for OA.

The findings indicate differences in muscle effort between conditions and between sexes. Although KE, KF, and PF, muscle activity was recorded using EMG during the data collection, the scope of this master's thesis was on the knee joint mechanics and reported on the MCU of the KE muscles, a value supported as an indicator of muscle effort [93]. More comprehensive analysis of muscle activity may provide additional insight on the effect of operational loads and speed on gait and differences in joint loading and muscle activation patterns between males and females during military load carriage.

The total sample resulted in statistical comparisons among conditions using a one-way repeated measures ANOVA model. The sample size limited the statistical analysis for sex comparisons using a two-factor ANOVA model so that both main effects (sex and condition) and interactions could be tested. Given the exploratory nature of this study, the sample size was large enough to detect significant differences, based on CI of the mean difference, using pairwise comparisons indicative of both between and within sex differences. With a larger sample size, the SSU-SSL comparisons of KAM impulse and Peak KAM for females would be significant (i.e., the same as for the male comparisons for these conditions). These sample size calculations support pairwise comparison between sex and between loaded conditions. For example they indicate that larger sample sizes could result in significant between sex differences for Peak KFM (SSL) and for KFM-KEM (FSL) supporting inclusion of sex analysis These preliminary findings also provide mean differences and variance estimates to calculate sample size estimates for designing a more robust study that examines main effects (sex and condition) and interaction effects to determine if responses to the loading conditions differ between males and females and better understand sex differences in risk of knee OA associated with load carriage.

The use of a treadmill limits the generalizability of findings to operational conditions and the overall duration of the walking conditions was shorter than that of standardized CAF testing and what participants would experience during military training and operations. Longer duration loaded conditions may have resulted in further alterations to biomechanical adaptations and increased physiological demands. However, longer duration testing, or testing on multiple days, was not feasible within the scope of this master's level thesis project. These findings illustrate specific adaptations, some of which increase the risk for knee OA, and other studies of prolonged walking or muscular fatigue show that some biomechanical variables such as peak knee adduction moment [85, 213] and muscle co-activity increase with fatigue [214].

6.4 Future Research

Future research could utilize the data from the current study to examine loaded and fixed speed walking conditions during overground walking and to monitor changes over a relevant duration (e.g., 50-60 minutes).

The discrete measures used in this study were picked *a priori* based on the literature related to medial compartment knee OA. As mentioned, some potentially interesting features were observed in the ensemble averaged joint moment waveforms (i.e., double peak prior to mid-stance and late stance third Peak KFM). Principal component analysis (PCA) has been used in numerous gait studies (REF) or other statistical pattern recognition techniques could be used to of examine the entire joint moment waveforms in future studies. PCA captures the key variations among waveforms and some that were observed and discussed were related to patellofemoral OA, not just medial compartment OA.

The statistical analysis for the sex comparisons was, in part, limited by the sample size. The preliminary findings from this study support a fully powered study to further explore sex differences and interactions in operationally relevant load carriage. A larger sample, with more males and females, would permit hypothesis testing of both main effects (condition and sex) and analysis of interaction effects. Based on partial η^2 values, walking condition had a large effect size (all η_p^2 values >0.14) for all primary and secondary measures. For primary outcomes, sex had a medium effect size for Peak KFM (η_p^2=0.13) and KFM-KEM (η_p^2=0.06) while a small effect size was found for KAM impulse and Peak KAM (η_p^2= 0.01). There were large effect sizes for condition for all primary outcomes and for sex on RPE (η_p^2>0.14). Medium effect sizes were identified for sex for MCU (η_p^2=0.13) and peak trunk angle (η_p^2=0.11).

While the findings from this thesis provide evidence of increased biomechanical risk for knee OA during load carriage and some differences between sexes, a longer duration study could identify if, and when, changes take place and whether differences between male and female military members are consistent across time in performance of this task.

6.5 Conclusion

This study provides evidence that military load carriage with operationally relevant load and speed alter knee joint moment features previously linked to medial compartment development and progression of knee OA, self-reported and objective effort, and peak trunk angle. Specifically, operationally relevant load (i.e., 35kg) was associated with increases in all primary outcomes (i.e., knee joint moment features, and secondary outcomes (i.e., measures of effort and trunk) while operationally relevant speed was associated with further increases in the magnitude of Peak KAM and KFM-KEM joint moment features and self-reported effort. The increases in magnitudes of knee joint moment features with operationally relevant load and speed provides evidence of the occupational risk for knee OA progression in military members during load carriage.

The preliminary analysis provides evidence of between sex differences for military load carriage with operationally relevant load and speed. The Peak KFM for females was greater compared to males, and, together with the lower knee extensor muscle strength, resulted in a greater KE muscle effort in response to operationally relevant load and speed compared to males. Females also had different responses to load than males in three joint moment features providing evidence of a potential sex by condition interaction that may indicate a difference in occupational risk for knee OA development and progression between male and female military members during load carriage tasks.

References

1. Molloy, J.M., et al., *Musculoskeletal injuries and United States Army readiness part I: overview of injuries and their strategic impact.* Military medicine, 2020. **185**(9-10): p. e1461-e1471.

2. Sharma, J., et al., *Musculoskeletal injuries in British Army recruits: a prospective study of diagnosis-specific incidence and rehabilitation times.* BMC musculoskeletal disorders, 2015. **16**(1): p. 1-7.

3. Thériault, F., K. Gabler, and K. Naicker, *Health and Lifestyle Information Survey of Canadian Forces Personnel 2013/2014-Regular Force Report*, D.o.N. Defense, Editor. 2016.

4. Orr, R.M., et al., *Soldier occupational load carriage: a narrative review of associated injuries.* International journal of injury control and safety promotion, 2014. **21**(4): p. 388-396.

5. Guilak, F., et al., *The Role of Biomechanics and Inflammation in Cartilage Injury and Repair.* Clinical Orthopaedics and Related Research®, 2004. **423**: p. 17-26.

6. Sinusas, K., *Osteoarthritis: diagnosis and treatment.* American family physician, 2012. **85**(1): p. 49-56.

7. Krajewski, K.T., et al., *Load carriage magnitude and locomotion strategy alter knee total joint moment during bipedal ambulatory tasks in recruit-aged women.* Journal of biomechanics, 2020. **105**: p. 109772.

8. Orr, R., et al., *Soldier Load Carriage, Injuries, Rehabilitation and Physical Conditioning: An International Approach.* International Journal of Environmental Research and Public Health, 2021. **18**(8): p. 4010.

9. Whittaker, J.L., et al., *Risk factors for knee osteoarthritis after traumatic knee injury: a systematic review and meta-analysis of randomised controlled trials and cohort studies for the OPTIKNEE Consensus.* British Journal of Sports Medicine, 2022.

10. Owens, B.D., et al., *Incidence of anterior cruciate ligament injury among active duty US military servicemen and servicewomen.* Military medicine, 2007. **172**(1): p. 90-91.

11. Jones, J.C., et al., *Incidence and risk factors associated with meniscal injuries among active-duty US military service members.* Journal of athletic training, 2012. **47**(1): p. 67-73.

12. Orr, R., B. Schram, and R. Pope, *Sports injuries in the Australian regular Army.* Safety, 2020. **6**(2): p. 23.

13. Wilkinson, D.M., et al., *Injuries and injury risk factors among British army infantry soldiers during predeployment training.* Injury Prevention, 2011. **17**(6): p. 381-387.

14. Cameron, K., T. Shing, and J. Kardouni, *The incidence of post-traumatic osteoarthritis in the knee in active duty military personnel compared to estimates in the general population.* Osteoarthritis and Cartilage, 2017. **25**: p. S184-S185.

15. Cameron, K.L., J.B. Driban, and S.J. Svoboda, *Osteoarthritis and the tactical athlete: a systematic review.* Journal of athletic training, 2016. **51**(11): p. 952-961.

16. Cameron, K.L., et al., *Incidence of physician-diagnosed osteoarthritis among active duty United States military service members.* Arthritis & Rheumatism, 2011. **63**(10): p. 2974-2982.

17. Roos, E.M., *Joint injury causes knee osteoarthritis in young adults.* Current opinion in rheumatology, 2005. **17**(2): p. 195-200.

18. Safiri, S., et al., *Global, regional and national burden of osteoarthritis 1990-2017: a systematic analysis of the Global Burden of Disease Study 2017.* Annals of the Rheumatic Diseases, 2020. **79**(6): p. 819-828.

19. Lespasio, M.J., et al., *Knee osteoarthritis: a primer.* The Permanente Journal, 2017. **21**.

20. Rivera, C.J.D., et al., *Posttraumatic osteoarthritis caused by battlefield injuries: the primary source of disability in warriors.* The Journal of the American Academy of Orthopaedic Surgeons, 2012. **20**(0 1): p. S64.

21. Murtha, A.S., et al., *Total knee arthroplasty for posttraumatic osteoarthritis in military personnel under age 50.* Journal of Orthopaedic Research, 2017. **35**(3): p. 677-681.

22. Talbot, M., *Joint replacements in the Canadian Armed Forces.* Canadian Journal of Surgery, 2020. **63**(5): p. E409.

23. Rhon, D.I., et al., *Much work remains to reach consensus on musculoskeletal injury risk in military service members: a systematic review with meta-analysis.* European Journal of Sport Science, 2022. **22**(1): p. 16-34.

24. Lovalekar, M., et al., *Musculoskeletal injuries in military personnel—descriptive epidemiology, risk factor identification, and prevention.* Journal of Science and Medicine in Sport, 2021. **24**(10): p. 963-969.

25. Blacker, S.D., et al., *Risk factors for training injuries among British Army recruits.* Military medicine, 2008. **173**(3): p. 278-286.

26. Fautrel, B., et al., *Impact of osteoarthritis: results of a nationwide survey of 10,000 patients consulting for OA.* Joint Bone Spine, 2005. **72**(3): p. 235-240.

27. Sutbeyaz, S.T., et al., *Influence of knee osteoarthritis on exercise capacity and quality of life in obese adults.* Obesity, 2007. **15**(8): p. 2071-2076.

28. Altman, R., et al., *Development of criteria for the classification and reporting of osteoarthritis: classification of osteoarthritis of the knee.* Arthritis & Rheumatism: Official Journal of the American College of Rheumatology, 1986. **29**(8): p. 1039-1049.

29. Cicuttini, F.M., et al., *Rate of cartilage loss at two years predicts subsequent total knee arthroplasty: a prospective study.* Annals of the rheumatic diseases, 2004. **63**(9): p. 1124-1127.

30. Hannan, M.T., D.T. Felson, and T. Pincus, *Analysis of the discordance between radiographic changes and knee pain in osteoarthritis of the knee.* The Journal of rheumatology, 2000. **27**(6): p. 1513-1517.

31. Barker, K., et al., *Association between radiographic joint space narrowing, function, pain and muscle power in severe osteoarthritis of the knee.* Clinical rehabilitation, 2004. **18**(7): p. 793-800.

32. Trudel, M.R.D., *JCSP 47 Master of Defence Studies.* 2021.

33. Gill, N., et al., *Role of sex and stature on the biomechanics of normal and loaded walking: implications for injury risk in the military.* BMJ Mil Health, 2021.

34. Gossec, L., et al., *The role of pain and functional impairment in the decision to recommend total joint replacement in hip and knee osteoarthritis: an international cross-sectional study of 1909 patients. Report of the OARSI-OMERACT Task Force on total joint replacement.* Osteoarthritis and Cartilage, 2011. **19**(2): p. 147-154.

35. Costello, K.E., *Overall Knee Joint Loading Exposure And Clinical Progression Of Knee Osteoarthritis*, in *Faculty of Graduate Studies Online Theses*. 2018, Dalhousie University.

36. Grimm, P.D., T.C. Mauntel, and B.K. Potter, *Combat and noncombat musculoskeletal injuries in the US military.* Sports medicine and arthroscopy review, 2019. **27**(3): p. 84-91.

37. Drew, M.D., *Effects of Prolonged Load Carriage of Knee Adduction Biomechanics*. 2020: Boise State University Theses and Dissertations.

38. Schram, B., et al., *Risk factors for development of lower limb osteoarthritis in physically demanding occupations: A narrative umbrella review.* Journal of occupational health, 2020. **62**(1): p. e12103.

39. Thériault, F.L., *Health and lifestyle information survey of Canadian Armed Forces personnel 2013/2014-Regular Force report*. 2016: National Defence= Défense nationale.

40. Walsh, G.S. and D.C. Low, *Military load carriage effects on the gait of military personnel: A systematic review.* Applied ergonomics, 2021. **93**: p. 103376.

41. Lenton, G.K., et al., *Tibiofemoral joint contact forces increase with load magnitude and walking speed but remain almost unchanged with different types of carried load.* PloS one, 2018. **13**(11): p. e0206859.

42. Koerhuis, C.L., et al., *Predicting Marching Capacity While Carrying Extremely Heavy Loads.* Military Medicine, 2009. **174**(12): p. 1300-1307.

43. Li, S.S., et al., *Gender differences in energy expenditure during walking with backpack and double-pack loads.* Human factors, 2019. **61**(2): p. 203-213.

44. Quesada, P.M., et al., *Biomechanical and metabolic effects of varying backpack loading on simulated marching.* Ergonomics, 2000. **43**(3): p. 293-309.

45. Liew, B., S. Morris, and K. Netto, *The effect of backpack carriage on the biomechanics of walking: a systematic review and preliminary meta-analysis.* Journal of applied biomechanics, 2016. **32**(6): p. 614-629.

46. Boffey, D., et al., *The physiology and biomechanics of load carriage performance.* Military medicine, 2019. **184**(1-2): p. e83-e90.

47. Krupenevich, R., et al., *Males and females respond similarly to walking with a standardized, heavy load.* Military medicine, 2015. **180**(9): p. 994-1000.

48. Seay, J.F., *Biomechanics of load carriage—Historical perspectives and recent insights.* The Journal of Strength & Conditioning Research, 2015. **29**: p. S129-S133.

49. Cathcart, E., D. Richardson, and W. Campbell, *Army Hygiene Advisory Committee Report No. 3.: On the Maximum Load to be Carried by the Soldier.* BMJ Military Health, 1923. **41**(3): p. 161-178.

50. Orr, R., *The history of the soldier's load.* Australian army journal, 2010. **7**(2): p. 67-88.

51. Hreljac, A., *Preferred and energetically optimal gait transition speeds in human locomotion.* Medicine and Science in Sports and Exercise, 1993. **25**(10): p. 1158-1162.

52. Wendland, R., L. Bossi, and M. Oliver, *Biomechanical and physiological effects of female soldier load carriage: A scoping review.* Applied Ergonomics, 2022. **105**: p. 103837.

53. Martin, P.E. and A.P. Marsh, *Step length and frequency effects on ground reaction forces during walking.* Journal of biomechanics, 1992. **25**(10): p. 1237-1239.

54. Seay, J.F., et al., *Lower extremity mechanics during marching at three different cadences for 60 minutes.* Journal of applied biomechanics, 2014. **30**(1): p. 21-30.

55. Orr, R.M., et al., *Load carriage for female military personnel.* Strength & Conditioning Journal, 2020. **42**(4): p. 50-58.

56. Schram, B., et al., *Risk factors for injuries in female soldiers: a systematic review.* BMC sports science, medicine and rehabilitation, 2022. **14**(1): p. 1-24.

57. Bode, V.G., et al., *Spatiotemporal and Kinematic Comparisons Between Anthropometrically Paired Male and Female Soldiers While Walking With Heavy Loads.* Military Medicine, 2021. **186**(3-4): p. 387-392.

58. Bombardier, C., D. Mosher, and G. Hawker, *The impact of arthritis in Canada.* 2016.

59. D'Souza, N., et al., *Are biomechanics during gait associated with the structural disease onset and progression of lower limb osteoarthritis? A systematic review and meta-analysis.* Osteoarthritis and Cartilage, 2021.

60. Miyazaki, T., et al., *Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis.* Annals of the rheumatic diseases, 2002. **61**(7): p. 617-622.

61. Chang, A.H., et al., *External knee adduction and flexion moments during gait and medial tibiofemoral disease progression in knee osteoarthritis.* Osteoarthritis and cartilage, 2015. **23**(7): p. 1099-1106.

62. Chang, A., et al., *The relationship between toe-out angle during gait and progression of medial tibiofemoral osteoarthritis.* Annals of the rheumatic diseases, 2007. **66**(10): p. 1271-1275.

63. Hatfield, G.L., W.D. Stanish, and C.L. Hubley-Kozey, *Three-dimensional biomechanical gait characteristics at baseline are associated with progression to total knee arthroplasty.* Arthritis care & research, 2015. **67**(7): p. 1004-1014.

64. Bennell, K.L., et al., *Higher dynamic medial knee load predicts greater cartilage loss over 12 months in medial knee osteoarthritis.* Annals of the rheumatic diseases, 2011. **70**(10): p. 1770-1774.

65. Erhart-Hledik, J., et al., *Longitudinal changes in tibial and femoral cartilage thickness are associated with baseline ambulatory kinetics and cartilage oligomeric matrix protein (COMP) measures in an asymptomatic aging population.* Osteoarthritis and cartilage, 2021. **29**(5): p. 687-696.

66. Chehab, E.F., et al., *Baseline knee adduction and flexion moments during walking are both associated with 5 year cartilage changes in patients with medial knee osteoarthritis.* Osteoarthritis and cartilage, 2014. **22**(11): p. 1833-1839.

67. Hubley-Kozey, C., G. Hatfield, and W. Stanish, *Muscle activation differences during walking between those with moderate knee osteoarthritis who progress to total knee arthroplasty and those that do not: a follow up study.* Osteoarthritis and Cartilage, 2013. **21**: p. S38.

68. Creaby, M., *It's not all about the knee adduction moment: the role of the knee flexion moment in medial knee joint loading.* Osteoarthritis and Cartilage, 2015. **23**(7): p. 1038-1040.

69. Andriacchi, T.P., et al., *A framework for the in vivo pathomechanics of osteoarthritis at the knee.* Annals of biomedical engineering, 2004. **32**(3): p. 447-457.

70. Landry, S.C., et al., *Knee biomechanics of moderate OA patients measured during gait at a self-selected and fast walking speed.* Journal of biomechanics, 2007. **40**(8): p. 1754-1761.

71. Amin, S., et al., *Knee adduction moment and development of chronic knee pain in elders.* Arthritis care & research, 2004. **51**(3): p. 371-376.

72. Salverda, G.J., et al., *Prolonged Load Carriage Impacts Magnitude and Velocity of Knee Adduction Biomechanics.* Biomechanics, 2021. **1**(3): p. 346-357.

73. Schipplein, O. and T. Andriacchi, *Interaction between active and passive knee stabilizers during level walking.* Journal of orthopaedic research, 1991. **9**(1): p. 113-119.

74. Griffin, T.M. and F. Guilak, *The role of mechanical loading in the onset and progression of osteoarthritis.* Exercise and sport sciences reviews, 2005. **33**(4): p. 195-200.

75. Horisberger, M., et al., *The influence of cyclic concentric and eccentric submaximal muscle loading on cell viability in the rabbit knee joint.* Clinical biomechanics, 2012. **27**(3): p. 292-298.

76. Horisberger, M., et al., *Long-term repetitive mechanical loading of the knee joint by in vivo muscle stimulation accelerates cartilage degeneration and increases chondrocyte death in a rabbit model.* Clinical Biomechanics, 2013. **28**(5): p. 536-543.

77. Hatfield, G.L., et al., *Baseline Gait Muscle Activation Patterns Differ for Osteoarthritis Patients Who Undergo Total Knee Arthroplasty Five to Eight Years Later From Those Who Do Not.* Arthritis care & research, 2021. **73**(4): p. 549-558.

78. Hodges, P.W., et al., *Increased duration of co-contraction of medial knee muscles is associated with greater progression of knee osteoarthritis.* Manual Therapy, 2016. **21**: p. 151-158.

79. Conkright, W.R., et al., *Sex differences in the physical performance, physiological, and psycho-cognitive responses to military operational stress.* European Journal of Sport Science, 2022. **22**(1): p. 99-111.

80. Loverro, K.L., L. Hasselquist, and C.L. Lewis, *Females and males use different hip and knee mechanics in response to symmetric military-relevant loads.* Journal of biomechanics, 2019. **95**: p. 109280.

81. Meakin, J.R., et al., *The effect of axial load on the sagittal plane curvature of the upright human spine in vivo.* Journal of biomechanics, 2008. **41**(13): p. 2850-2854.

82. Harman, E., et al., *The effects of backpack weight on the biomechanics of load carriage*. 2000, Army Research Inst Of Environmental Medicine Natick Ma Military Performancediv.

83. Orr, R.M., et al., *Load carriage: Minimising soldier injuries through physical conditioning-A narrative review.* Journal of military and veterans health, 2010. **18**(3): p. 31-38.

84. Silder, A., S.L. Delp, and T. Besier, *Men and women adopt similar walking mechanics and muscle activation patterns during load carriage.* Journal of biomechanics, 2013. **46**(14): p. 2522-2528.

85. Murdock, G.H. and C.L. Hubley-Kozey, *Effect of a high intensity quadriceps fatigue protocol on knee joint mechanics and muscle activation during gait in young adults.* European journal of applied physiology, 2012. **112**: p. 439-449.

86. Bonci, C.M., *Assessment and evaluation of predisposing factors to anterior cruciate ligament injury.* Journal of athletic training, 1999. **34**(2): p. 155.

87. Shimokochi, Y. and S.J. Shultz, *Mechanisms of noncontact anterior cruciate ligament injury.* Journal of athletic training, 2008. **43**(4): p. 396-408.

88. Bhambhani, Y. and R. Maikala, *Gender differences during treadmill walking with graded loads: biomechanical and physiological comparisons.* European journal of applied physiology, 2000. **81**(1): p. 75-83.

89. Borg, G.A., *Psychophysical bases of perceived exertion.* Medicine & science in sports & exercise, 1982.

90. Hébert, L.J., S. Nadeau, and D. Gravel, *Quantification of muscle and joint requirements during the weight load march in the Canadian Forces: characterization of the factors limiting performance.* , D.o.N. Defense, Editor. 2006: Ottawa.

91. Brady, A.O., C.R. Straight, and E.M. Evans, *Body composition, muscle capacity, and physical function in older adults: an integrated conceptual model.* Journal of aging and physical activity, 2014. **22**(3): p. 441-452.

92. Chen, M.J., X. Fan, and S.T. Moe, *Criterion-related validity of the Borg ratings of perceived exertion scale in healthy individuals: a meta-analysis.* Journal of sports sciences, 2002. **20**(11): p. 873-899.

93. Tung, E.V., et al., *The relationship between muscle capacity utilization during gait and pain in people with symptomatic knee osteoarthritis.* Gait & Posture, 2022. **94**: p. 58-66.

94. Piscoya, J.L., et al., *The influence of mechanical compression on the induction of osteoarthritis-related biomarkers in articular cartilage explants.* Osteoarthritis and Cartilage, 2005. **13**(12): p. 1092-1099.

95. Cope, P.J., et al., *Models of osteoarthritis: the good, the bad and the promising.* Osteoarthritis and Cartilage, 2019. **27**(2): p. 230-239.

96. Zaki, S., C.L. Blaker, and C.B. Little, *OA Foundations–Experimental models of Osteoarthritis.* Osteoarthritis and Cartilage, 2021.

97. Taruc-Uy, R.L. and S.A. Lynch, *Diagnosis and treatment of osteoarthritis.* Primary care, 2013. **40**(4): p. 821-36, vii.

98. Felson, D.T., et al., *Physical activity, alignment and knee osteoarthritis: data from MOST and the OAI.* Osteoarthritis and cartilage, 2013. **21**(6): p. 789-795.

99. McAlindon, T.E., et al., *OARSI guidelines for the non-surgical management of knee osteoarthritis.* Osteoarthritis and cartilage, 2014. **22**(3): p. 363-388.

100. Englund, M., *The role of biomechanics in the initiation and progression of OA of the knee.* Best practice & research Clinical rheumatology, 2010. **24**(1): p. 39-46.

101. Guilak, F., *Biomechanical factors in osteoarthritis.* Best Practice & Research Clinical Rheumatology, 2011. **25**(6): p. 815-823.

102. Griffin, T.M., et al., *Diet-induced obesity differentially regulates behavioral, biomechanical, and molecular risk factors for osteoarthritis in mice.* Arthritis Research & Therapy, 2010. **12**(4): p. R130.

103. Sammito, S., et al., *Risk factors for musculoskeletal injuries in the military: a qualitative systematic review of the literature from the past two decades and a new prioritizing injury model.* Military Medical Research, 2021. **8**(1): p. 1-40.

104. Tillman, M., et al., *Differences in lower extremity alignment between males and females: potential predisposing factors for knee injury.* Journal of sports medicine and physical fitness, 2005. **45**(3): p. 355.

105. Merchant, A.C., et al., *A reliable Q angle measurement using a standardized protocol.* The Knee, 2020. **27**(3): p. 934-939.

106. Grelsamer, R., A. Dubey, and C. Weinstein, *Men and women have similar Q angles: a clinical and trigonometric evaluation.* The Journal of Bone & Joint Surgery British Volume, 2005. **87**(11): p. 1498-1501.

107. Goulston, L.M., et al., *A comparison of radiographic anatomic axis knee alignment measurements and cross-sectional associations with knee osteoarthritis.* Osteoarthritis and cartilage, 2016. **24**(4): p. 612-622.

108. Smith, T.O., N.J. Hunt, and S.T. Donell, *The reliability and validity of the Q-angle: a systematic review.* Knee Surgery, Sports Traumatology, Arthroscopy, 2008. **16**: p. 1068-1079.

109. Sharma, R., et al., *A systematic review on quadriceps angle in relation to knee abnormalities.* Cureus, 2023. **15**(1).

110. Sharma, L., et al., *Varus and valgus alignment and incident and progressive knee osteoarthritis.* Annals of the rheumatic diseases, 2010. **69**(11): p. 1940-1945.

111. Chapple, C.M., et al., *Patient characteristics that predict progression of knee osteoarthritis: a systematic review of prognostic studies.* Arthritis care & research, 2011. **63**(8): p. 1115-1125.

112. Palazzo, C., et al., *Risk factors and burden of osteoarthritis.* Annals of physical and rehabilitation medicine, 2016. **59**(3): p. 134-138.

113. Hunter, D.J., et al., *Knee alignment does not predict incident osteoarthritis: the Framingham osteoarthritis study.* Arthritis & Rheumatism, 2007. **56**(4): p. 1212-1218.

114. Chu, C.R. and T.P. Andriacchi, *Dance between biology, mechanics, and structure: a systems-based approach to developing osteoarthritis prevention strategies.* Journal of orthopaedic research, 2015. **33**(7): p. 939-947.

115. Sanchez-Adams, J., et al., *The mechanobiology of articular cartilage: bearing the burden of osteoarthritis.* Current rheumatology reports, 2014. **16**(10): p. 451.

116. Moran, C.J., et al., *The benefits and limitations of animal models for translational research in cartilage repair.* Journal of Experimental Orthopaedics, 2016. **3**(1): p. 1.

117. Hatfield, G.L., et al., *The association between knee joint muscle activation and knee joint moment patterns during walking in moderate medial compartment knee osteoarthritis: implications for secondary prevention.* Archives of Physical Medicine and Rehabilitation, 2021.

118. Hatfield, G., W. Stanish, and C. Hubley-Kozey, *Does prolonged muscle activity during gait explain a decreased ability to unload the knee joint in those with medial compartment knee osteoarthritis?* Osteoarthritis and Cartilage, 2015. **23**: p. A91-A92.

119. Andriacchi, T.P., S. Koo, and S.F. Scanlan, *Gait mechanics influence healthy cartilage morphology and osteoarthritis of the knee.* The Journal of Bone and Joint Surgery. American volume., 2009. **91**(Suppl 1): p. 95.

120. Baliunas, A., et al., *Increased knee joint loads during walking are present in subjects with knee osteoarthritis.* Osteoarthritis and cartilage, 2002. **10**(7): p. 573-579.

121. Childs, J.D., et al., *Alterations in lower extremity movement and muscle activation patterns in individuals with knee osteoarthritis.* Clinical biomechanics, 2004. **19**(1): p. 44-49.

122. Hurwitz, D.E., et al., *Dynamic knee loads during gait predict proximal tibial bone distribution.* Journal of biomechanics, 1998. **31**(5): p. 423-430.

123. Kaufman, K.R., et al., *Gait characteristics of patients with knee osteoarthritis.* Journal of biomechanics, 2001. **34**(7): p. 907-915.

124. Rutherford, D.J., et al., *Neuromuscular alterations exist with knee osteoarthritis presence and severity despite walking velocity similarities.* Clinical Biomechanics, 2011. **26**(4): p. 377-383.

125. Hatfield, G.L., W.D. Stanish, and C.L. Hubley-Kozey, *Relationship between knee adduction moment patterns extracted using principal component analysis and discrete measures with different amplitude normalizations: Implications for knee osteoarthritis progression studies.* Clinical biomechanics, 2015. **30**(10): p. 1146-1152.

126. Thorp, L.E., et al., *Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis.* Arthritis & Rheumatism: Official Journal of the American College of Rheumatology, 2006. **54**(12): p. 3842-3849.

127. Thorp, L.E., et al., *Relationship between pain and medial knee joint loading in mild radiographic knee osteoarthritis.* Arthritis care & research, 2007. **57**(7): p. 1254-1260.

128. Manal, K., et al., *A more informed evaluation of medial compartment loading: the combined use of the knee adduction and flexor moments.* Osteoarthritis and cartilage, 2015. **23**(7): p. 1107-1111.

129. Bohannon, R.W. and A.W. Andrews, *Normal walking speed: a descriptive meta-analysis.* Physiotherapy, 2011. **97**(3): p. 182-189.

130. Kerrigan, D.C., et al., *Knee joint torques: a comparison between women and men during barefoot walking.* Archives of physical medicine and rehabilitation, 2000. **81**(9): p. 1162-1165.

131. Cho, S.-H., J.M. Park, and O.Y. Kwon, *Gender differences in three dimensional gait analysis data from 98 healthy Korean adults.* Clinical biomechanics, 2004. **19**(2): p. 145-152.

132. Kerrigan, D.C., M.K. Todd, and U.D. Croce, *Gender differences in joint biomechanics during walking normative study in young adults*. 1998, LWW. p. 2-7.

133. Wilson, J.L.A., M.J. Dunbar, and C.L. Hubley-Kozey, *Knee joint biomechanics and neuromuscular control during gait before and after total knee arthroplasty are sex-specific.* The Journal of arthroplasty, 2015. **30**(1): p. 118-125.

134. Sims, E.L., et al., *Sex differences in biomechanics associated with knee osteoarthritis.* Journal of women & aging, 2009. **21**(3): p. 159-170.

135. McKean, K.A., et al., *Gender differences exist in osteoarthritic gait.* Clinical Biomechanics, 2007. **22**(4): p. 400-409.

136. Dever, D.E., et al., *Increases in Load Carriage Magnitude and Forced Marching Change Lower-Extremity Coordination in Physically Active, Recruit-Aged Women.* Journal of Applied Biomechanics, 2021. **1**(aop): p. 1-8.

137. Kraus, V.B., et al., *Call for standardized definitions of osteoarthritis and risk stratification for clinical trials and clinical use.* Osteoarthritis and cartilage, 2015. **23**(8): p. 1233-1241.

138. Zhang, Y. and J. Niu, *Shifting gears in osteoarthritis research towards symptomatic osteoarthritis.* Arthritis & rheumatology (Hoboken, NJ), 2016. **68**(8): p. 1797.

139. Bastick, A.N., et al., *Prognostic factors for progression of clinical osteoarthritis of the knee: a systematic review of observational studies.* Arthritis research & therapy, 2015. **17**(1): p. 1-13.

140. Maillefert, J.-F. and M. Dougados, *Is time to joint replacement a valid outcome measure in clinical trials of drugs for osteoarthritis?* Rheumatic Disease Clinics, 2003. **29**(4): p. 831-845.

141. Losina, E., et al., *Lifetime risk and age at diagnosis of symptomatic knee osteoarthritis in the US.* Arthritis care & research, 2013. **65**(5): p. 703-711.

142. Showery, J.E., et al., *The rising incidence of degenerative and posttraumatic osteoarthritis of the knee in the United States military.* The Journal of arthroplasty, 2016. **31**(10): p. 2108-2114.

143. Boyer, K.A., *Biomechanical response to osteoarthritis pain treatment may impair long-term efficacy.* Exercise and sport sciences reviews, 2018. **46**(2): p. 121-128.

144. Guérin, E. and J. Laplante, *Understanding risk and protective factors of injuries: Insights from Canadian recruits during basic military training.* Journal of Military, Veteran and Family Health, 2022(aop): p. e20210113.

145. Willy, R.W., et al., *Effects of load carriage and step length manipulation on Achilles tendon and knee loads.* Military medicine, 2019. **184**(9-10): p. e482-e489.

146. Reilly, T., E. Walsh, and B. Stockbrugger, *Reliability of FORCE COMBAT™: A Canadian army fitness objective.* Journal of science and medicine in sport, 2019. **22**(5): p. 591-595.

147. Hauschild, V., et al., *Foot marching, load carriage, and injury risk*. 2016, Army Public Health Center Aberdeen Proving Ground-Edgewood Area United States.

148. Vine, C.A., et al., *Accuracy of metabolic cost predictive equations during military load carriage.* Journal of Strength and Conditioning Research, 2022. **36**(5): p. 1297-1303.

149. Shields, L.-C.N. and S. Flight, *CANADIAN ARMED FORCES 25% WOMEN BY 2026: ATTAINABLE GOAL OR PIPE DREAM?*

150. Knapik, J.J., K.L. Reynolds, and E. Harman, *Soldier load carriage: historical, physiological, biomechanical, and medical aspects.* Military medicine, 2004. **169**(1): p. 45-56.

151. van Dijk, J., *Common military task: marching.* Optimizing Operational Physical Fitness, 2009.

152. Martin, P.E. and R.C. Nelson, *The effect of carried loads on the walking patterns of men and women.* Ergonomics, 1986. **29**(10): p. 1191-1202.

153. Kung, S.M., et al., *What factors determine the preferred gait transition speed in humans? A review of the triggering mechanisms.* Human movement science, 2018. **57**: p. 1-12.

154. Unnikrishnan, G., et al., *Effects of body size and load carriage on lower-extremity biomechanical responses in healthy women.* BMC Musculoskeletal Disorders, 2021. **22**(1): p. 1-11.

155. Middleton, K., et al., *Mechanical differences between men and women during overground load carriage at self-selected walking speeds.* International Journal of Environmental Research and Public Health, 2022. **19**(7): p. 3927.

156. Englert, R.M. and A.M. Yablonsky, *Scoping review and gap analysis of research related to the health of women in the US military, 2000 to 2015.* Journal of Obstetric, Gynecologic & Neonatal Nursing, 2019. **48**(1): p. 5-15.

157. Majumdar, D., M.S. Pal, and D. Majumdar, *Effects of military load carriage on kinematics of gait.* Ergonomics, 2010. **53**(6): p. 782-791.

158. Kodithuwakku Arachchige, S.N., et al., *Muscle Activity during Postural Stability Tasks: Role of Military Footwear and Load Carriage.* Safety, 2020. **6**(3): p. 35.

159. Cikajlo, I. and Z. Matjačić, *The influence of boot stiffness on gait kinematics and kinetics during stance phase.* Ergonomics, 2007. **50**(12): p. 2171-2182.

160. Goldring, M.B. and S.R. Goldring, *Articular cartilage and subchondral bone in the pathogenesis of osteoarthritis.* Annals of the New York Academy of Sciences, 2010. **1192**(1): p. 230-237.

161. Astephen, J.L., et al., *Gait and neuromuscular pattern changes are associated with differences in knee osteoarthritis severity levels.* Journal of biomechanics, 2008. **41**(4): p. 868-876.

162. Andriacchi, T.P. and A. Mündermann, *The role of ambulatory mechanics in the initiation and progression of knee osteoarthritis.* Current opinion in rheumatology, 2006. **18**(5): p. 514-518.

163. Chung, M.-J. and M.-J.J. Wang, *The change of gait parameters during walking at different percentage of preferred walking speed for healthy adults aged 20–60 years.* Gait & posture, 2010. **31**(1): p. 131-135.

164. Chumanov, E.S., C. Wall-Scheffler, and B.C. Heiderscheit, *Gender differences in walking and running on level and inclined surfaces.* Clinical biomechanics, 2008. **23**(10): p. 1260-1268.

165. Chiu, M.-C. and M.-J. Wang, *The effect of gait speed and gender on perceived exertion, muscle activity, joint motion of lower extremity, ground reaction force and heart rate during normal walking.* Gait & posture, 2007. **25**(3): p. 385-392.

166. Allison, K.F., et al., *Musculoskeletal, biomechanical, and physiological gender differences in the US military.* US Army Medical Department Journal, 2015.

167. McArdle, W.D., F.I. Katch, and V.L. Katch, *Exercise physiology: nutrition, energy, and human performance*. 2010: Lippincott Williams & Wilkins.

168. Haynes, E.M., et al., *Age and sex-related decline of muscle strength across the adult lifespan: a scoping review of aggregated data.* Applied Physiology, Nutrition, and Metabolism, 2020. **45**(11): p. 1185-1196.

169. Verlinden, V.J., et al., *Gait patterns in a community-dwelling population aged 50 years and older.* Gait & posture, 2013. **37**(4): p. 500-505.

170. Faul, F., et al., *G* Power 3: A flexible statistical power analysis program for the social, behavioral, and biomedical sciences.* Behavior research methods, 2007. **39**(2): p. 175-191.

171. Cohen, J., *Statistical power analysis for the behavioral sciences*. 2013: Academic press.

172. Hertzog, M.A., *Considerations in determining sample size for pilot studies.* Research in nursing & health, 2008. **31**(2): p. 180-191.

173. Hermens, H.J., et al., *Development of recommendations for SEMG sensors and sensor placement procedures.* Journal of electromyography and Kinesiology, 2000. **10**(5): p. 361-374.

174. Hubley-Kozey, C., et al., *Neuromuscular alterations during walking in persons with moderate knee osteoarthritis.* Journal of Electromyography and Kinesiology, 2006. **16**(4): p. 365-378.

175. Hubley-Kozey, C.L., et al., *Reliability of surface electromyographic recordings during walking in individuals with knee osteoarthritis.* Journal of Electromyography and Kinesiology, 2013. **23**(2): p. 334-341.

176. McDonough, A.L., et al., *The validity and reliability of the GAITRite system's measurements: A preliminary evaluation.* Archives of physical medicine and rehabilitation, 2001. **82**(3): p. 419-425.

177. Bilney, B., M. Morris, and K. Webster, *Concurrent related validity of the GAITRite® walkway system for quantification of the spatial and temporal parameters of gait.* Gait & posture, 2003. **17**(1): p. 68-74.

178. Rutherford, D.J., et al., *High day-to-day repeatability of lower extremity muscle activation patterns and joint biomechanics of dual-belt treadmill gait: A reliability study in healthy young adults.* Journal of Electromyography and Kinesiology, 2020. **51**: p. 102401.

179. Rutherford, D., et al., *Dual-belt treadmill familiarization: Implications for knee function in moderate knee osteoarthritis compared to asymptomatic controls.* Clinical Biomechanics, 2017. **45**: p. 25-31.

180. Dunphy, C., et al., *Contralateral pelvic drop during gait increases knee adduction moments of asymptomatic individuals.* Human movement science, 2016. **49**: p. 27-35.

181. Rutherford, D.J., J. Moreside, and I. Wong, *Differences in hip joint biomechanics and muscle activation in individuals with femoroacetabular impingement compared with healthy, asymptomatic individuals: is level-ground gait analysis enough?* Orthopaedic Journal of Sports Medicine, 2018. **6**(5): p. 2325967118769829.

182. Zeni Jr, J.A. and J.S. Higginson, *Gait parameters and stride-to-stride variability during familiarization to walking on a split-belt treadmill.* Clinical biomechanics, 2010. **25**(4): p. 383-386.

183. Stark, T., et al., *Hand-held dynamometry correlation with the gold standard isokinetic dynamometry: a systematic review.* PM&R, 2011. **3**(5): p. 472-479.

184. Wu, G. and P.R. Cavanagh, *ISB recommendations for standardization in the reporting of kinematic data.* Journal of biomechanics, 1995. **28**(10): p. 1257-1262.

185. Baker, M.D., *UNDERSTANDING SELF-REPORTED INSTABILITY USING GAIT OUTCOMES AND WALKWAY SURFACE TRANSLATIONS IN THOSE WITH KNEE OSTEOARTHRITIS.* 2020.

186. Grood, E.S. and W.J. Suntay, *A joint coordinate system for the clinical description of three-dimensional motions: application to the knee.* Journal of biomechanical engineering, 1983. **105**(2): p. 136-144.

187. Attwells, R.L., et al., *Influence of carrying heavy loads on soldiers' posture, movements and gait.* Ergonomics, 2006. **49**(14): p. 1527-1537.

188. Zeni Jr, J., J. Richards, and J. Higginson, *Two simple methods for determining gait events during treadmill and overground walking using kinematic data.* Gait & posture, 2008. **27**(4): p. 710-714.

189. Vaughan, C.L.D.B.L.O.C.J.C., *Dynamics of human gait.* 1999.

190. DeLuzio, K.J., et al., *A procedure to validate three-dimensional motion assessment systems.* Journal of biomechanics, 1993. **26**(6): p. 753-759.

191. Li, J., et al., *An integrated procedure to assess knee-joint kinematics and kinetics during gait using an optoelectric system and standardized X-rays.* Journal of biomedical engineering, 1993. **15**(5): p. 392-400.

192. Zahradka, N., et al., *An evaluation of three kinematic methods for gait event detection compared to the kinetic-based 'gold standard'.* Sensors, 2020. **20**(18): p. 5272.

193. Gardner, M.J. and D.G. Altman, *Confidence intervals rather than P values: estimation rather than hypothesis testing.* Br Med J (Clin Res Ed), 1986. **292**(6522): p. 746-750.

194. Bland, J.M. and D.G. Altman, *Statistics notes: Transforming data.* Bmj, 1996. **312**(7033): p. 770.

195. Bland, J.M. and D.G. Altman, *Transformations, means, and confidence intervals.* BMJ: British Medical Journal, 1996. **312**(7038): p. 1079.

196. Robbins, S.M. and M.R. Maly, *The effect of gait speed on the knee adduction moment depends on waveform summary measures.* Gait & posture, 2009. **30**(4): p. 543-546.

197. Mündermann, A., et al., *Potential strategies to reduce medial compartment loading in patients with knee osteoarthritis of varying severity: reduced walking speed.* Arthritis & Rheumatism: Official Journal of the American College of Rheumatology, 2004. **50**(4): p. 1172-1178.

198. Kean, C.O., et al., *Comparison of peak knee adduction moment and knee adduction moment impulse in distinguishing between severities of knee osteoarthritis.* Clinical biomechanics, 2012. **27**(5): p. 520-523.

199. Hutchison, L., et al., *Relationship Between Knee Biomechanics and Pain in People With Knee Osteoarthritis: A Systematic Review and Meta-Analysis.* Arthritis Care & Research, 2023. **75**(6): p. 1351-1361.

200. Teng, H.-L., et al., *Higher knee flexion moment during the second half of the stance phase of gait is associated with the progression of osteoarthritis of the patellofemoral joint on magnetic resonance imaging.* journal of orthopaedic & sports physical therapy, 2015. **45**(9): p. 656-664.

201. Lloyd, D.G. and T.F. Besier, *An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo.* Journal of biomechanics, 2003. **36**(6): p. 765-776.

202. Deluzio, K. and J. Astephen, *Biomechanical features of gait waveform data associated with knee osteoarthritis: an application of principal component analysis.* Gait & posture, 2007. **25**(1): p. 86-93.

203. Sisante, J.F., et al., *Influence of antagonistic hamstring coactivation on measurement of quadriceps strength in older adults.* PM&R, 2020. **12**(5): p. 470-478.

204. Bacon, K.L., et al., *Concurrent change in quadriceps strength and physical function over five years in the Multicenter Osteoarthritis Study.* Arthritis care & research, 2019. **71**(8): p. 1044-1051.

205. Wilson, J.L.A., *Challenges in dealing with walking speed in knee osteoarthritis gait analyses.* Clinical biomechanics, 2012. **27**(3): p. 210-212.

206. Teng, H.-L., et al., *Associations between patellofemoral joint cartilage T1ρ and T2 and knee flexion moment and impulse during gait in individuals with and without patellofemoral joint osteoarthritis.* Osteoarthritis and cartilage, 2016. **24**(9): p. 1554-1564.

207. Burfield, M., M. Sayers, and R. Buhmann, *The association between running volume and knee osteoarthritis prevalence: A systematic review and meta-analysis.* Physical Therapy in Sport, 2023.

208. Timmins, K.A., et al., *Running and knee osteoarthritis: a systematic review and meta-analysis.* The American journal of sports medicine, 2017. **45**(6): p. 1447-1457.

209. Robbins, S.M., et al., *Reliability of principal components and discrete parameters of knee angle and moment gait waveforms in individuals with moderate knee osteoarthritis.* Gait & posture, 2013. **38**(3): p. 421-427.

210. Cifrek, M., et al., *Surface EMG based muscle fatigue evaluation in biomechanics.* Clinical biomechanics, 2009. **24**(4): p. 327-340.

211. McCristall, P. and K. Baggaley, *The progressions of a gendered military: A theoretical examination of gender inequality in the Canadian military.* Journal of Military, Veteran and Family Health, 2019. **5**(1): p. 119-126.

212. BALOGH, P., *Gray Zone Activities–with a Focus on the Social Domain1.* Conflicts in the Gray Zone A Challenge to Adapt: p. 15.

213. Drew, M.D., S.M. Krammer, and T.N. Brown, *Effects of prolonged walking with body borne load on knee adduction biomechanics.* Gait & posture, 2021. **84**: p. 192-197.

214. Peterson, D.S. and P.E. Martin, *Effects of age and walking speed on coactivation and cost of walking in healthy adults.* Gait & posture, 2010. **31**(3): p. 355-359.

Appendix A

Participant inclusion and exclusion criteria

Inclusion	Exclusion
• Member of the CAF (regular or reserve force)	• Neurological, cardiovascular, or musculoskeletal condition that could be exacerbated or pose safety risk
• Age 20-50 years	• Military medical employment limitations limiting lifting or performing moderate/vigorous exercise
• Completed Basic Military Qualification or Basic Military Officer Qualification	• History of major spinal or knee surgery (e.g., spinal fusion, discectomy, laminectomy, foraminotomy, TKA, uni-compartmental knee replacement, high-tibial osteotomy)
	• Injection for pain to one or both knees within last 2-years

Appendix B

Base wide recruitment email and poster

Title: Dalhousie Research Study Seeking Participants to Understand Effect of Loaded Marching on Leg Muscles and Walking Patterns

Researchers at Dalhousie University are conducting a study to understand how leg muscles and walking patterns change while marching with a standard load and walking pace. The hope is to better understand if changes could influence the risk of knee osteoarthritis (joint disease) and if males and females have similar changes with the standard load tasks. To do this we will look at tests of how you walk at your own pace, with and without a load, and at a set pace with load. We will contact you after completion of the study to share our findings.

We are looking for healthy volunteers who are active members in the Canadian Armed Forces (Regular Force or Reserve Force). If you are between the ages of 20-50 years old you may be eligible for this study.

The study will take place at the School of Physiotherapy at Dalhousie University. Compensation for incidentals (parking and travel to Dalhousie University) will be provided.

If you are interested in this study, please contact the Lead Researcher (Adam Hannaford) at ahannafo@dal.ca.

Appendix C

Phone interview script: general study information and health screen

Hello, my name is: Adam Hannaford, I am conducting research at Dalhousie University.

Thank-you for connecting with me and your interest in my research study looking at how walking with load affects leg muscle function and walking patterns. By doing this study we hope to identify factors that contribute to injury and osteoarthritis, a common chronic joint disease in military members. This information may be used in designing ways to reduce the risk of injuries and developing osteoarthritis to better inform physical training plans. For this study we are going to compare your typical walking speed (self-paced) with and without load, to a standard pace as in forced marching at a pace like the FORCE Combat™ Test loaded march. The testing will be done on a treadmill. While you are walking, we will collect signals from your leg muscles from surface electrodes, which are small, self-sticking discs, applied to your skin above the seven leg muscles of interest. We will also use a motion capture system to record how your trunk and legs move during walking. This testing will require one visit to the lab that will take approximately 2.5-3 hour to complete the testing. We understand this is a large time commitment, but, if you are interested, we will provide you with a letter to deliver to your chain of command so you can perform this study during working hours. However, you will need their consent to participate in this study. Participating in this study will have no impact on your career. Do you think you might be interested?

If no, Thank-you for your interest….

If yes, Thank-you for your interest. I would like to conduct a short interview including a general health screen, to see if you are eligible for this study. If you are eligible, I would like to schedule a time for you to participate in this study. This screen should take at most 5-7 minutes. Would now be a good time for you to participate in this interview?(*If "yes" continue, if "No". When would be a good time for me to contact you regarding this interview?)*

____

To minimize your risk of harms if you participate in this study, I must first ask you a few health-related questions. You may discontinue answering the questions at any time and terminate this phone conversation. If you choose to discontinue during this interview, or do not meet the criteria for the study, any information we record from this survey will be immediately destroyed. These questions will take about 3 minutes. Do you agree to answer the following health-related questions completely and to the best of your knowledge? (*If "NO" to the above verbal consent, the conversation will be terminated with "thank you for your time." If "YES," the following questions will be asked).*

General Health Screening

In this section, we will ask you eight "yes or no" questions about any current medical conditions, to determine your general health. First, please listen to all these questions without telling me your answers. At the end of these questions, I will ask if you answered "yes" to any of them. If so, you do not have to tell me which question was answered with a "yes." This will secure your privacy by ensuring I do not know which of the conditions you may have. However, if you do have any questions regarding your specific situation and would feel comfortable asking a question so I can provide clarity, do not hesitate to ask. To protect your privacy no information will be recorded from this conversation.

1. Do you currently have medical employment limitations that restrict your ability to engage in moderate/vigorous exercise, walk, lift, or carry weight?
2. Currently, is there any reason, or have you been told by a health care provider, that you should avoid or restrict moderate/vigorous exercise, lifting or carrying a moderate to heavy weight (30-40Kg)?
3. Have you ever had major knee or spinal surgery? (i.e., high tibial osteotomy, total or uni-compartmental joint replacement) or spinal surgery (i.e., spinal fusion, discectomy, laminectomy, or foraminotomy)
4. Have you had an injection for pain to one, or both, knees, or your spine, in the past two years?

5. Do you have high blood pressure that is not controlled?
6. Do you have any problems with your heart that interfere with your day-to-day activities or your ability to exercise or perform physical activity such as fast heart rate (arrhythmias), a past heart attack, angina, or irregular heartbeats?
7. Do you have any lung conditions or breathing difficulties that interfere with your day-to-day activities or your ability to exercise?
8. Have you ever experienced any problems with your nervous system that have left you with resulting muscle weakness or loss of skin sensation, balance problems (i.e., loss off feeling)?

Potential participants will be <u>excluded</u> if they answer "YES" to any of the questions. If participant is not excluded, proceed by forwarding the standardized e-mail, CoC permission letter, and booking a data collection session.

If excluded, read the following:

"You do not meet all the criteria for our study but thank you for your time. We record the reasons why individuals were not eligible, without identifying information, and then shred the information at the end of the study."

If included, read the following:

"You are eligible to participate in the study. Would you like to book a time to come to Dalhousie to participate now? (If yes, book time and proceed if not now, see below). I will send you a confirmatory e-mail that includes the appointment time, arrival instructions, required clothing, and permission letter to be signed by your chain of command." (Confirm preferred e-mail for participant)."

If not prepared to book:

" If now is not a good time, when can I contact you to book a time?" Arrange follow-up to book.

Appendix D

Confirmation email and arrival instructions

Title: CONFIRMATION: Dalhousie University Research Study Session Date/Time and Chain of Command Letter

Thank-you for agreeing to participate in our research study. Your session is scheduled for (Date/Time). Please review, and have your Supervisor/Chain of Command sign, the attached permission letter for you to participate during working hours.

The study will take place at the Forrest Building, Dalhousie University School of Physiotherapy, 5869 University Avenue, Halifax. A researcher will meet you in the lobby inside the University Avenue entrance (Dalhousie University School of Dentistry) and escort you to the Joint Action Research Laboratory.

Please bring shorts, t-shirt, and your personal combat boots. Please ensure your boots are cleaned before arriving for the study.

If you have any concerns or questions, please contact the research coordinator by e-mail at ahannafo@dal.ca or phone (587-341-6634)

Adam

Adam Hannaford, PT, FCAMPT

Dalhousie University School of Physiotherapy

Appendix E

Chain of command permission letter

School of Physiotherapy
Forrest Building
5869 University Avenue
Halifax NS
B3H 3J5

Date

Address of recipient

<u>REQUEST FOR (Participant) TO PARTICIPATE IN RESEARCH AT DALHOUSIE UNIVERSITY</u>

1. This is a request for (participant name) to participate in a research project that is ongoing at Dalhousie University. This research is endorsed by the Surgeon General's Research Program.

2. The purpose of this project is to understand how military members adapt their walking and muscle patterns while forced marching with load. To do this we will measure leg muscle function and walking mechanics using a battery of tests that can accurately measure how individuals walk and how their muscles work. To achieve this, we will test Canadian Armed Forces (CAF) members.

3. Acute leg muscle/joint injuries and chronic conditions are commonly reported by CAF members. Determining how military members adapt their walking and muscle recruitment patterns while carrying loads may help to inform physical training programs and injury reduction strategies. This could provide cost savings to the Canadian Forces Health Services as well as enhance the ability to maintain a healthy, deployable, population.

4. Participation in this study will require (participant name) _to take part in one session. To participate, (participant name) will travel to Dalhousie University and take part in a two-and-a-half-hour data collection session at the Joint Action Research Laboratory. (Participant name) has indicated that they are interested in participating in this project and we are requesting your approval as their chain of command for them to complete this project during work hours.

5. Allowing the participant to complete this study during work hours will maximize our ability to recruit participants so that we can complete the study in a timely manner and report our findings to the CAF. If you feel that the participant cannot be reassigned from their regular duties during the requested times, we can work with you to find a time you feel the participant can perform the study. If you feel that reassignment cannot be accommodated at all, please let us know as soon as possible.

6. By signing this request, you are indicating that (participant) is reassigned from regular duties to participate in this project.

7. Thank-you for your time and consideration in this matter. If you have any questions or concerns, please contact the undersigned.

A.J. Hannaford, CD, PT, FCAMPT

Major
Physiotherapy Officer
(587) 341-6634
ahannafo@dal.ca

I, ___________________________________ (print name, rank) provide permission for

_____________________________________ to participate in the study named above at

Dalhousie University on____________________________________ (Time/Date).

Signature of Member's Supervisor:_____________________________________

Return to member for presentation at time of study participation.

Appendix F

Participant consent form

CONSENT FORM

COMPARING OBJECTIVE BIOMECHANICAL AND ELECTROMYOGRAPHICAL MEASURES DURING AN ABSOLUTE LOAD CARRIAGE TASK IN CANADIAN ARMED FORCES MEMBERS

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Contact Person

Please contact Adam Hannaford (contact information above), Dr. D. Rutherford or Dr. C. Kozey in the event of any unusual occurrences or difficulties related to the research, or to receive more information or clarification about the study procedure at any time.

Introduction

We invite you to take part in a research study at Dalhousie University. Taking part in this study is voluntary and you may withdraw from the study at any time. Participation in this study will have no effect on career progression and your personal data will not be shared with any third party. The study is described below. This tells you what you will be asked to do, and the potential risks, inconvenience, or discomfort you might experience. Participating in the study might not help you directly, but we hope to learn information that will help prevent others from developing osteoarthritis of those with osteoarthritis from having rapid worsening of pain and structural tissue damage . You should discuss any questions you have about this study with the researchers.

Purpose of the study

Knee injuries and osteoarthritis are common in military members. It is costly to the injured person as well as the health care system. Many injured members restrict their leisure activity, experience weight gain, and report lower quality of life. The reasons for injury and development of osteoarthritis are complicated, but scientists have provided evidence to show that changes to walking patterns during tasks such as weight load carriage (i.e., rucksack marching) may play a role. The purpose of this study is to gather information that can help to develop physical training programs and injury reduction strategies to reduce knee injuries and development of osteoarthritis. There are three main goals we want to do: i) identify how a person moves (biomechanics) and uses their muscles (electromyography) while carrying a standardized (35 Kg) load during walking at their self-selected pace; ii) identify if how a person moves (biomechanics) and uses their muscles (electromyography) changes when walking with the same standardized load at a specified pace (about 5.5km/hour); and iii) compare these findings between male and female Canadian Armed Forces members to see if there is a difference. Females have been identified as having increased injury and osteoarthritis rates compared to male members and some aspects of the way a person moves or uses their muscles have been associated with the development and progression of osteoarthritis. In studying the way people move with load at different paces we hope to be able to identify ways to reduce injury and reduce factors that may lead to osteoarthritis. This will help health professionals i) identify key factors related to injury and osteoarthritis; ii) develop comprehensive rehabilitation programs for injured members; and iii) assist with the planning of physical training programs.

Study Design

This study will compare how people move and how muscles work between people carrying the same 35 kg load (the Combat FORCE™ load)) at their preferred pace and at the pace of the Combat FORCE™ loaded march while walking on a treadmill. We will measure muscle activity using sensors (electrodes) placed over fourteen different points on your legs (back and front); these are called electromyograms (EMG). We will measure these EMG signals for each participant and then average these signals together for each condition and for each group (male and female) and this will help us determine how hard the muscles are working. These group averages will be compared to determine whether there are differences in muscle activity between the two groups and between these tasks. We will use small light-weight reflective spheres attached to twenty points on your body to capture three-dimensional images of how your trunk and legs move. We will use this information to determine the loads impacting your knees. We will also collect measurements of muscle strength from your leg muscles, this information will help us to understand how hard your leg muscles are working while carrying the load in these two walking conditions.

Who can participate in the Study?

Male and female Canadian Armed Forces members (regular or reserve force) between the ages of 20-50 years are eligible to participate. You should be healthy and have no problems with your nervous system, your muscles, joints, and bones or with your heart and breathing that would make it difficult for you to complete the test tasks safely, as indicated in questions asked in the telephone interview.

Who will be conducting the Research?

The principal investigators of this study are Dr. Cheryl Kozey, a professor in the School of Physiotherapy and Adam Hannaford, a CAF physiotherapist who is currently completing his Masters of Rehabilitation Research at Dalhousie University in the School of Physiotherapy. Student research assistants will also help with data collections.

What you will be asked to do

If you agree to participate, you will be asked to come to a single session at the Joint Action Research Laboratory at Dalhousie University. The session will last approximately *2.5 - 3 hours*.

You must bring your personal combat boots. If you do not have gym gear with you, it can be provided; changing facilities are available. Your height, weight, age, and years of military service will be recorded. To attach the little metal discs (electrodes) to record you muscle activity your skin will be lightly rubbed with an alcohol/water swab and if necessary, shaved with a hand-held razor at the locations where the EMG electrodes will be attached. If your skin is sensitive to alcohol, please advise us and we will use a water-solution to clean off your skin. Fourteen electrodes (10 mm each) will be placed on the skin overlying the leg muscles and two electrodes for references on the shin bone. Then twenty small sensors that track your trunk, pelvis, hip, knee, ankle, and foot motion will be attached to your skin, using double sided tape. These sensors will be placed on your: upper and lower legs, hips, upper and lower back, shoulders, feet, and neck to record motion using a camera system. The camera system records body motion but not your personal image.

You will be asked to walk along a pressure sensitive mat ten times. Measurements of your walking speed will be taken and averaged to determine your preferred walking speed. You will then be asked to walk at that speed on a dual belted treadmill for six minutes. Measurements about how you move and how your muscles work will be collected. You will then be fitted with a weight vest and small pack with a total weight of 35 kilograms. You will be asked to walk at your self-selected speed on the treadmill while carrying the load and then you will rest for 10 minutes to recover. You will then be asked to walk on the treadmill at the standardized speed of 1.52 meters per second (about 5.5 Km/hour) while carrying the same load.

The load will be removed, and you will be asked to lay on your back on a treatment bed and completely relax while your baseline muscle activity is measured. You will be asked to do five different exercises where you will push as hard as you can against non-elastic straps for the count of three. We will record how strong your leg muscles are using a strength-measuring device. These will be done twice with a 2-minute rest between trials. All of this will take approximately 2.5-3 hours.

The findings of the study will be shared with you once all data has been collected, analyzed, and the project is complete.

Possible Risks and Discomforts

There is minimal risk of harm associated with the testing session. If you have sensitive skin, the alcohol or electrode tape may cause irritation, please let the researcher know and we can dilute the alcohol wipe. If this occurs, it should not last more 3 days. There is always a small risk of electric shock when using any electrical device however the equipment used reduces this risk since the EMG unit to which you are attached is battery operated, the lab has hospital grade grounds, and the EMG system meets the standards of the Canadian Standards Association. Depending on your level of fitness, you may experience some post exercise muscle soreness, which should not last more than 3 days. While walking on the treadmill, an upper body safety harness will be connected to you to minimize risk of fall if you lose balance or trip. If you experience any discomfort during testing, report this immediately to the tester. Please remember that you may withdraw from the study at any time even after testing has begun. The health screening questions were used to identify potential problems that could increase risk or discomfort associated with participation. If you do feel muscle soreness or pain after the sessions, you may want to apply heat or ice to the area, but also try to continue with normal activities. If the problem persists for more than 3-days, then you should contact your primary health care professional.

Possible Benefits

We do not expect this study to have any direct benefit to you. However, from this study, we hope to develop a better understanding of how people move and how they use their muscles while carrying operationally relevant loads. We also hope to gain more insight into why some people get injured or develop knee osteoarthritis. This information may help clinicians and physical training staff better design rehabilitation programs and plan physical training. If you express interest, you will also be invited to a debriefing session once we have finished collecting and analyzing the data. This session will provide you with some of the key results of our study. In the past, we have found that many participants attend such sessions to learn about the results of the study, and to ask questions.

Compensation

A letter will be provided for you to complete for your chain of command approval if you wish to complete this study during working hours. You will be compensated in the amount of $20.00 for transportation and parking costs. Participants that opt to withdraw will still be compensated.

Confidentiality

Your privacy will be protected at all times. Your name and contact information will be kept secure by the research team. It will not be shared with others without your permission. Your name will not appear in any report or article published in relation to this study. Information collected will be kept for 7 years after the final publication. All participant information (e.g., telephone screening, age, mass, questionnaires) will be kept in a locked cabinet. The locked cabinet and the computer storing the information (i.e., database, EMG, motion, questionnaire scores) are in the Joint Action Research Laboratory in the Forrest Building on Dalhousie University Campus. This is a key locked room inside a room that has a keypad entrance and keyed outer lab door with limited access. The computer requires a password to access, and a backup disc is stored in the Principal Investigator's office (Dr. Kozey) which is key locked and inside a keypad locked entrance. To make sure that the participant's identity remains anonymous beyond the research team, each participant will be given a participant code and number, and that number will be used to link all data that we collect. Identifying information about you will not be used in any way. All data presented in publications, presentations, and in presentations to other participants will be about groups of people, not individuals.

Questions

If you have any questions regarding the study, please do not hesitate to contact Adam Hannaford at (587) 341-6634.

Signature

I have read the above description of this study. I have been given the opportunity to discuss it and my questions have been answered to my satisfaction. I hereby consent to take part in this study. I realize that my participation is voluntary and that I am free to withdraw from the study at any time, even after testing has begun. Should I decide to withdraw, I will inform Adam Hannaford of my decision.

I am aware that I am at work/on duty while participating in this research project if it is during work hours. The signature of my commanding officer or immediate supervisor on the attached request indicates that I have been reassigned from regular duties to participate in the project.

Signature_____________________________

Date ______________________

Witness____

In the event that you have any difficulties with, or wish to voice concern with any aspect of your participation in this study, you may contact Catherine Connors (ethics@dal.ca) the Director Research Ethics at Dalhousie University's Office of Human Research Ethics and Integrity for assistance: (902) 494-1462.

Please sign below to indicate agreement with each statement**.**

I agree to allow the data from my testing session to be used in future studies that would compare my gait pattern, effort, and muscle activation measures, to other groups' tests or to a larger sample to make the findings more applicable to the military population.

Signature ___________________________________

I give permission to the researchers to re-contact me to participate in future research studies related to load carriage involving military members.

Signature ___________________________________

I would be interested in being contacted in the future with regards to attending a debriefing session on the results of this study, and/or receive a lay summary of the results.

Signature ___________________________________

Appendix G

BORG rate of perceived exertion scale (RPE)

On a scale of 6 to 20, where 6 is no exertion at all and 20 is maximal exertion, what is the number from the scale below that best describes your feeling of effort and exertion right now?

Exertion	RPE
no exertion at all	6
extremely light	7
	8
very light	9
	10
light	11
	12
somewhat hard	13
	14
hard (heavy)	15
	16
very hard	17
	18
extremely hard	19
maximal exertion	20

Appendix H

Overview of participant enrollment, exclusion, and withdrawal

In total, 33 CAF members, 17 males and 16 females, inquired about the study and 25 participants (76%) were enrolled over 13 weeks (February 1 - May 3, 2023). Eight members (24%) that inquired were not enrolled in the study. Reasons for not enrolling included exclusion based on the general health screen (9%, 1 male, 2 female), not having completed BMOQ or BMQ (3%, 1 female), and not being in the Halifax area for the duration of the study (6%, 2 female). Two of the CAF members that inquired about the study did not respond to follow-up communications (6%, 1 male, 1 female). Of the 25 participants enrolled, 1 participant withdrew from the study (4%, 1 male). The reason for withdrawal was reported as illness on the day of testing; the participant could not be rescheduled prior to conclusion of the study. Figure 4.1 details participant inquiries, enrollment, exclusion, and withdrawal.

Appendix I

Post hoc analysis – one-way repeated measures ANOVA (Chapter 4)

The pairwise comparisons from the one-way repeated measures ANOVA (Chapter 4) of the knee joint moment features linked to knee OA clinical progression across walking conditions are displayed in Table I1.

Table I1: Pairwise comparisons for *knee joint moment features linked to knee OA clinical progression across walking conditions*

Knee Moment Measurement	Walking Condition	Compared Walking Condition	Mean Difference	P	95% CI Lower	Upper
KAM Impulse (Nms/kg)	FSL	SSU	0.03	**<.001**	0.01	0.05
		SSL	-0.19	**.007**	-0.03	-0.05
	SSL	SSU	0.05	**<.001**	0.03	0.07
Peak KAM (Nm/kg)	FSL	SSU	0.25	**<.001**	0.17	0.32
		SSL	0.09	**.001**	0.03	0.15
	SSL	SSU	0.15	**<.001**	0.09	0.21
Peak KFM (Nm/kg)	FSL	SSU	0.39	**<.001**	0.19	0.59
		SSL	0.05	1.00	-0.11	0.21
	SSL	SSU	0.34	**.003**	0.10	0.58
KFM-KEM (Nm/kg)	FSL	SSU	0.77	**<.001**	0.59	0.95
		SSL	0.19	**.007**	0.05	0.34
	SSL	SSU	0.57	**<.001**	0.40	0.75

Bold indicates significant differences between walking conditions (p≤0.05)

Appendix J

Normality tests for Peak KFM data – Lg 10 transformed

Table J1 presents the findings of the Shapiro-Wilk (S-W) test for normality results for the Lg 10 transformed Peak KFM data of the entire sample for all walking conditions.

Table J1: ***S-W Test results Lg10 Transformed Peak KFM data***

Walking condition	Statistic	df	p
SSU	0.98	24	.82
SSL	0.97	24	.71
FSL	0.98	24	.93

Figure J1 contains the histograms displaying the distribution of Lg 10 transformed Peak KFM data for each walking condition.

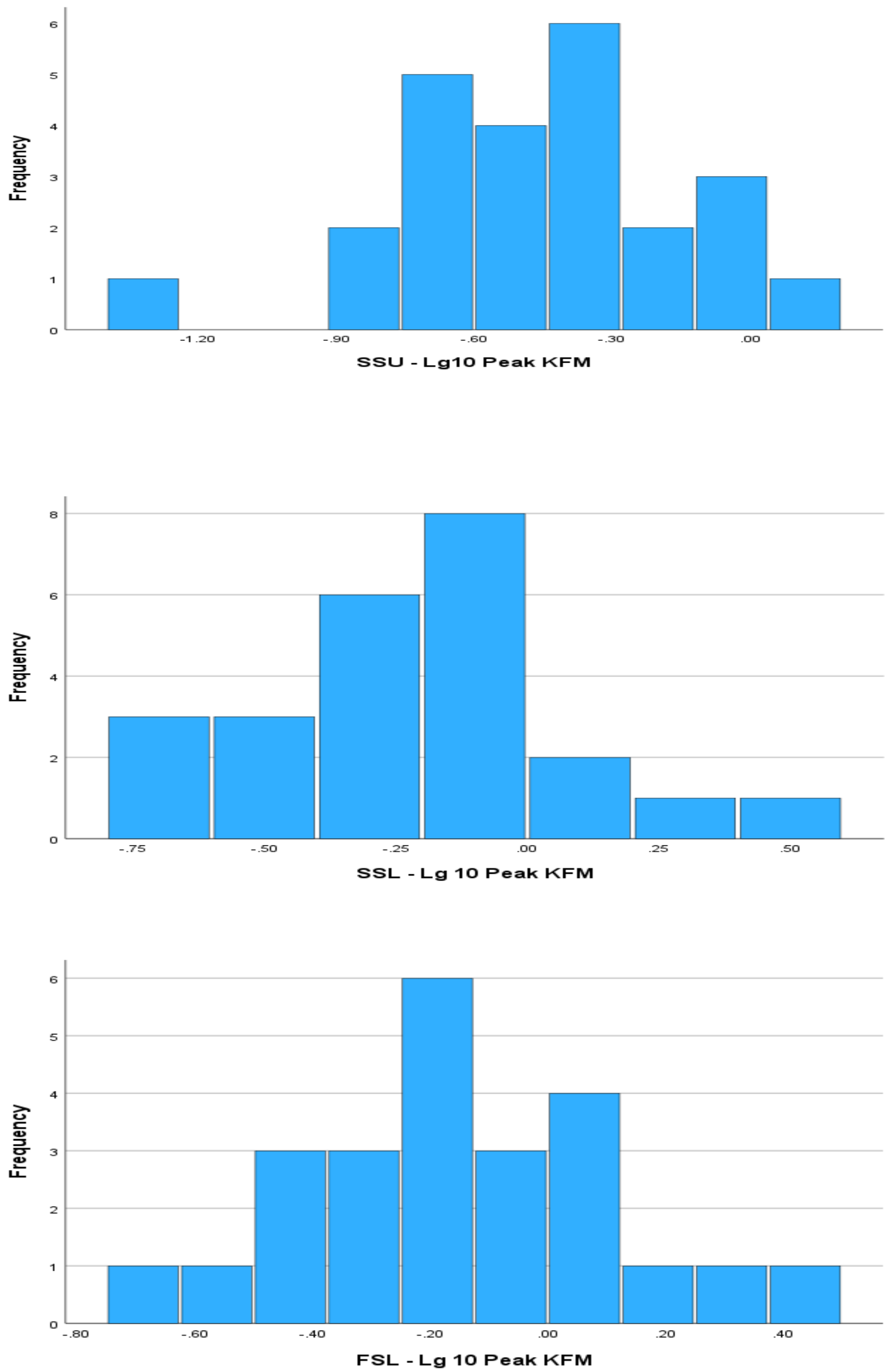

Figure J1: ***Distributions for Lg 10 transformed Peak KFM data for walking conditions, SSU (above), SSL (middle), and FSL (below)***

One-way repeated measures ANOVA found a statistically significant effect for walking condition, $F(2,46)= 24.72$, $p<.001$, $\eta^2=0.52$. Pairwise comparisons for the log 10 transformed Peak KFM data for SSU, SSL, and FSL are presented in Table J2.

Table J2: ***Pairwise comparisons of Peak KFM log 10 transformed data for walking conditions***

Knee Moment Measurement	Walking Condition	Compared Walking Condition	Mean Difference	P	95% CI Lower	95% CI Upper
Peak KFM	FSL	SSU	0.30	**< .001**	0.16	0.43
(Lg10)	SSL	SSU	0.25	**<.001**	0.13	0.37
	FSL	SSL	0.05	0.42	-0.34	0.14

Bold indicates significant differences between walking condition (p≤0.05)

Appendix K

Normality tests for KFM-KEM data – outlier removed

Table K1 presents the findings of the Shapiro-Wilk (S-W) test for normality results for the Lg10 transformed Peak KFM data of the entire sample for all walking conditions.

Table K1: ***S-W Test results for KFM-KEM data – outlier removed.***

Walking condition	Statistic	df	p
SSU	0.95	23	.31
SSL	0.94	23	.22
FSL	0.97	23	.68

Figure K1 contains the histograms displaying the distribution of KEM-KFM data with the outlier data removed.

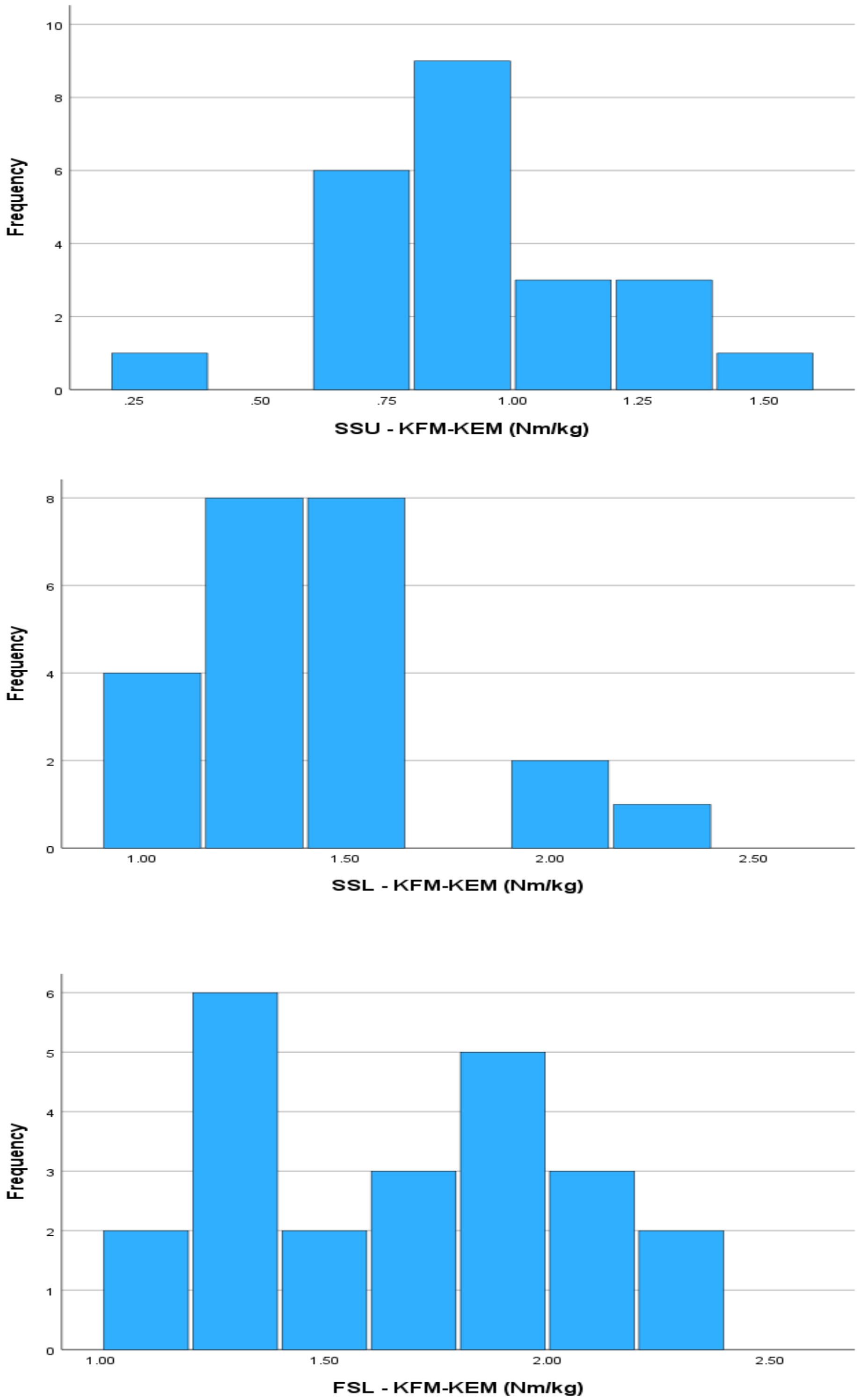

Figure K1: ***Distributions for KFM-KEM data with outlier removed for walking conditions.***

One-way repeated measures ANOVA found a statistically significant effect for walking condition, $F(2,44)= 93.02$, $p<.001$, $\eta^2=0.81$. Pairwise comparisons for the outlier removed KFM-KEM data for SSU, SSL, and FSL are presented in Table K2.

Table K2: ***Pairwise comparisons of outlier removed KFM-KEM data for walking conditions***

Knee Moment Measurement	**Walking Condition**	**Compared Walking Condition**	**Mean Difference**	**P**	**95% CI Lower**	**95% CI Upper**
KFM-KEM (Nm/kg)	FSL	SSU	0.73	**<.001**	0.57	0.91
	SSL	SSU	0.52	**<.001**	0.40	0.64
	FSL	SSL	0.22	**.002**	0.08	0.36

Bold indicates significant differences between walking conditions (p≤0.05)

Appendix L

Pairwise comparisons of secondary outcomes for entire sample

(RPE, MCU, Peak Trunk Angle)

Table L1 presents the pairwise comparisons for RPE for walking conditions.

Table L1: ***Pairwise comparisons of mean RPE during walking conditions***

Walking Condition	Compared Condition	M Difference	p	95% CI of the Difference	
				Lower	Upper
SSU	SSL	-3.92	***<.001***	-4.69	-3.14
	FSL	-4.79	***<.001***	-5.71	-3.87
SSL	FSL	-0.88	***.002***	-1.44	-0.31

Bold indicates significant differences between walking conditions (p≤0.05)

Table L2 presents the pairwise comparisons for MCU for walking conditions.

Table L2: ***Pairwise comparisons of mean MCU during walking conditions.***

Walking Condition	Compared Condition	M Difference	p	95% CI of the Difference	
				Lower	Upper
SSU	SSL	-0.19	***<.001***	-0.30	-0.08
	FSL	-0.20	***<.001***	-0.29	-0.11
SSL	FSL	-0.02	1.00	-0.09	0.06

Bold indicates significant differences between walking conditions (p≤0.05)

Table L3 presents the pairwise comparisons for mean peak trunk angle for walking conditions.

Table L3: ***Pairwise comparisons of mean peak trunk angle during walking conditions***

Walking Condition	**Compared Condition**	**M Difference**	**p**	**95% CI of the Difference**	
				Lower	**Upper**
SSU	SSL	-10.1	***<.001***	-12.3	-7.9
	FSL	-10.6	***<.001***	-13.1	-8.2
SSL	FSL	-0.5	1.00	-4.0	3.0

Bold indicates significant differences between walking conditions ($p \leq 0.05$)

Appendix M

Sensitivity analysis- sagittal plane knee moment features

Peak KFM and KFM-KEM– outlier removed (FAG)

Sensitivity analysis conducted by removing the Peak KFM data for one outlier (FAG) resulted in a significant between sex difference within the FSL condition (Table M1).

Table M1: ***Descriptive statistics of Peak KFM for male and female participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions with outlier data removed***

Walking Condition	Sex	M (SD)	95% CI Lower	95% CI Upper	M Difference	95% CI of the Difference Lower	95% CI of the Difference Upper
SSU	Male	0.33 (0.20	0.21	0.44	-0.17	-0.38	0.04
	Female	0.50 (0.28)	0.28	0.72			
SSL	Male	0.57 (0.37)	0.35	0.78	-0.20	-0.53	0.09
	Female	0.77 (0.31)	0.54	1.03			
FSL	Male	0.60 (0.39)	0.38	0.83	-0.35	***-0.67***	***-0.02***
	Female	0.95 (0.33)	0.70	1.20			

Bold CI indicate significant differences between sex within condition

The within sex pairwise comparisons and CI for Peak KFM are shown in Table M2. Removal of the outlier data resulted in an additional significant ($p<0.05$) pairwise comparison between conditions within sex (i.e., female, SSU-SSL).

Table M2: ***Confidence intervals for the mean pairwise differences in Peak KFM for males and females with outlier removed***

Sex	Compared Conditions	M Diff	95% CI of the Difference Lower	Upper
Male	SSU-SSL	-0.24	***-0.43***	***-0.05***
	SSU-FSL	-0.28	***-0.49***	***-0.07***
	SSL-FSL	-0.04	-0.18	0.10
Female	SSU-SSL	-0.29	***-0.41***	***-0.16***
	SSU-FSL	-0.45	***-0.78***	***-0.12***
	SSL-FSL	-0.16	-0.48	0.16

Bold CI indicate significant differences between conditions within sex

Results of the sensitivity analysis, conducted by removing the KFM-KEM data for one outlier (FAG), is shown in Table M3 for between sex comparisons.

Table M3: ***Descriptive statistics of KFM-KEM for male and female participants for each walking condition and confidence intervals for the pairwise between sex mean differences within walking conditions with outlier data removed***

Walking Condition	Sex	M (SD)	95% CI Lower	95% CI Upper	M Difference	95% CI of the Difference Lower	Upper
SSU	Male	0.93 (0.27)	0.78	1.09	0.00	-0.24	0.25
	Female	0.93 (0.30)	0.70	1.16			
SSL	Male	1.44 (0.38)	1.22	1.66	-0.02	-0.32	0.28
	Female	1.46 (0.24)	1.27	1.65			
FSL	Male	1.61 (0.36)	1.40	1.82	-0.16	-0.47	0.16
	Female	1.76 (0.33)	1.51	2.01			

The within sex pairwise comparisons and CI for KFM-KEM are shown in Table M4. Removal of the outlier data resulted in an additional significant ($p<0.05$) pairwise comparison between conditions within sex (i.e., female, SSL-FSL).

Table M4: ***Confidence intervals for the mean pairwise differences in KFM-KEM for males and females with outlier removed***

Sex	**Compared Conditions**	**M Diff**	**95% CI of the Difference Lower**	**Upper**
Male	SSU-SSL	-0.51	***-0.69***	***-0.34***
	SSU-FSL	-0.68	***-0.87***	***-0.48***
	SSL-FSL	-0.16	-0.33	0.01
Female	SSU-SSL	-0.53	***-0.71***	***-0.36***
	SSU-FSL	-0.84	***-1.22***	***-0.45***
	SSL-FSL	-0.30	***-0.59***	***-0.01***

Bold CI indicate significant differences between conditions within sex

For between sex comparisons, removal of the outlier data for Peak KFM resulted in the between sex comparison for the SSU condition being non-significant while the result for FSL (i.e., significant difference) was unchanged. The between sex comparison for males and females for FSL remained significant based on the 95% CI of the mean difference (i.e., no change with outlier removed. Removal of the outlier data resulted in the SSU-SSL pairwise comparison for Peak KAM and the SSL-FSL pairwise comparison for KFM-KEM for females being significant. The key finding from the sensitivity analysis for removal of the outlier data is that there was no change to the finding of a between sex difference for males and females for FSL based on 95% CI of the mean difference. This suggests a between sex difference in response to increased, fixed, speed load carriage.

Appendix N

Table N1 presents the descriptive statistics and 95% CI of the mean difference for frontal and sagittal plane knee moments for males and females for all walking conditions.

Table N1: ***Descriptive Statistics and 95% CI Difference for frontal and sagittal plane knee moments for males and females for walking conditions***

Plane	Sex	M (SD)	Mean Difference	95% CI of the Difference Lower	Upper
Frontal (Nm/kg)	Male	0.22 (0.24)	-0.03	-0.01	0.04
	Female	0.25 (0.23)			
Sagittal (Nm/kg)	Male	-0.04 (0.34)	-0.22	**0.13**	**0.32**
	Female	0.18 (0.34)			

Bold CI indicate significant differences between sex within plane (p<0.05)